Packaging Methods for Magnetoelectric Transducers Used as Wireless Power Receivers

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***Abstract*—Magnetoelectric (ME) transducers, composed of piezoelectric and magnetostrictive layers, have recently been demonstrated as a receiver in wireless power transfer (WPT) to miniature implantable medical devices (IMDs). Due to their complex mechano-magnetic-electric interactions, methods for biocompatible coating and packaging of ME transducers within a small IMD need to be investigated to achieve optimal ME response. This paper describes the effects of biocompatible coating and packaging using Polydimethylsiloxane (PDMS) on ME response. Three bar-shaped ~ 5×1×1 mm^3 ME transducers were fabricated with a 508 µm thick piezoelectric layer. In air, adding ~ 35 µm of PDMS coating reduced the received power (P_L) by 19.4% (131.04 µW vs. 105.62 µW). The ME transducer were mounted on a printed circuit board (PCB) and inside a custom 3D-printed holder. The PCB mounting had no significant effect on ME transducer P_L, but the 3D-printed holder considerably reduced P_L from 112.89 µW to 39.02 µW, when measured in a water medium.**

Keywords—Magnetoelectric, wireless energy, implantable biomedical devices, Polydimethylsiloxane (PDMS) coating.

I. Introduction

Magnetoelectric (ME) transducers as a power receiver are becoming attractive for wireless powering of miniature implantable medical devices (IMDs) using low-frequency magnetic fields [1]-[4]. This is because the conventional techniques for wireless power transfer (WPT) to miniature IMDs, including inductive (or RF), capacitive, and ultrasonic links, suffer from shortcomings. For inductive/RF links to be efficient in WPT to IMDs, they need to operate at very high frequencies (hundreds of MHz to a couple of GHz) [5]-[7], which limits the amount of transmitted power due to the specific absorption rate (SAR) safety limits [8]. Capacitive WPT suffers from the limited implant depth and large transmitter (Tx) and receiver (Rx) size [9]. The drawbacks of ultrasound technology are its extreme sensitivity to Tx and Rx alignment, and the significant attenuation in air and bone media due to their acoustic impedance mismatch [10]-[13].

Compared to coils, small ME transducers (composed of piezoelectric and magnetostrictive layers) are more efficient in converting the low frequency (hundreds of kHz) magnetic fields into electrical power in a two-step process [4]. The magnetostrictive layer generates a strain in response to the incident alternating magnetic field, which is directly coupled to the piezoelectric layer. In proportion to the applied strain, the piezoelectric layer generates an electric voltage. Fig. 1 shows a generic WPT link with the ME transducer as the wireless power receiver. The power management circuitry and implant core are modeled as the AC load, R_L.

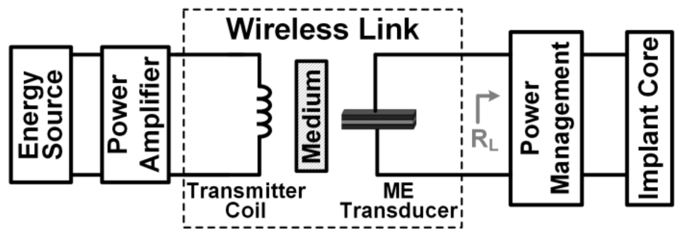

Fig. 1. Generic block diagram of a wireless power transfer link for powering an implant with the ME transducer as a power receiver.

ME transducers have several key advantages in comparison to inductive coils. Since they operate at low frequencies, higher transmitted power is achievable within the imposed safety limit. They also have better tolerance to misalignment in comparison to inductive coils [1], [4]. Thanks to these advantages, several biomedical applications with ME transducers as power receivers have already been demonstrated, including neural stimulation and recording (magnetic sensing) [2], [3].

In the previous works, the ME transducer has mainly been coated with a thin layer of Parylene-C. Our previous work with Parylene-C coating of ME transducers also showed that their response does not change as the coating is very thin. But the ME transducer could only operate for a short period of time in the water as the thin-film coating failed over time. Therefore, for the long-term and robust operation of an implant in a realistic environment, alternative methods for biocompatible coating and packaging of ME transducers within a small IMD needs to be investigated. To the best of our knowledge, there is no reported work that discusses the effect of the coating on the performance of ME transducers as wireless power receivers.

In this paper, three bar-shaped 5×1×1 mm^3 (length l = 5 mm, width w = 1 mm, thickness h = 1 mm) ME transducers, as listed

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TABLE I. SPECIFICATIONS OF THE ME TRANSDUCERS, COATING THICKNESS, AND MEASURED CHARACTERISTICS FOR DIFFERENT TYPES OF MOUNTING

ME Transducer	**ME Transducer Dimension (*l*×*w*×*h*) (mm)**	**Mounting Type**	**‡Coating Thickness (µm)**	**Optimal H_{DC} (Oe)**			**Optimal R_L (kΩ)**			***Received P_L (µW)**		
				Without PDMS Coating	**†PDMS Coated**		**Without PDMS Coating**	**†PDMS coated**		**Without PDMS Coating**	**†PDMS Coated**	
					In Air	**In Water**		**In Air**	**In Water**		**In Air**	**In Water**
ME_1	5.01×1.02×0.91	None	~ 35	410	410	400	1.8	1.9	2.5	132.44	106.1	79.13
ME_2	5.07×1.11×1.05	PCB	~ 36	440	430	430	2.2	2.2	2.5	98.55	73.32	54.7
ME_3	5.05×1.05×1.05	3D-Printed Holder	~ 250	390	400	390	3	3	2.5	112.89	43.01	39.02

‡Average measured thickness close to the two edges along the length of the ME sample. *Measured under an H_{AC} of 1 Oe rms.
†Measured when the ME transducer was hand-coated with PDMS alongside the mounting type mentioned.

in Table I and shown in Fig. 2, were fabricated by sandwiching a piezoelectric layer (PZT-5A) in between two magnetostrictive layers (Galfenol) to study the effects of biocompatible coating using Polydimethylsiloxane (PDMS or Sylgard; Fig. 2a) and packaging of ME transducers on their received power. Compared to Parylene-C, PDMS is preferable as it is more robust and transcutaneous (easily penetrated through the skin) for implantation [14]. To complete the packaging of the IMD, two mounting methods are also demonstrated: 1) using a printed circuit board (PCB) as a substrate (Fig. 2b), and 2) with a custom-designed 3D-printed holder (Fig. 2c).

The remainder of this paper is organized as follows. Section II explains the method of PDMS coating and its effects on the ME transducer's impedance, quality factor (Q), and received power (P_L). Section III details mounting methods for the ME transducer on a PCB and in a custom 3D-printed holder, and their effects on the above-mentioned ME properties. Discussion and future work are provided in Section IV with conclusions in Section V.

II. PDMS COATING OF THE ME TRANSDUCER

A. Method of ME Transducer Coating with PDMS and Experimental Setup

The ME transducers were meticulously fabricated as detailed in our previous work [4]. Each of the Galfenol layers were ~ 241 µm thick while the PZT-5A layer was ~ 508 µm thick. The dimensions of the fabricated ME transducers are given in Table I. The PDMS (184 Silicone Elastomer, Dow Industries, Midland, MI) was prepared by hand mixing the two parts (the elastomer base and the curing agent) in a weight ratio of 10:1. The mixed parts were allowed to rest for 30 minutes to remove any air bubbles. Next, a thin layer of the mixed portion was hand coated onto the ME transducer using a toothpick. It was then allowed to air-dry for 24 hours. An image of the ME transducer with PDMS coating is shown in Fig. 2a. After curing, the measured thickness of the PDMS coating was ~ 35 µm.

We followed the procedure reported in our previous work [4] to study the effect of the DC bias field (H_{DC}) on the ME quality factor (Q), and the effects of the operation frequency (f_p), H_{DC}, and load resistance (R_L) on the received power (P_L) of the ME transducer with and without PDMS coating. Note that for optimal operation, ME transducers require a bias H_{DC} in addition to the excitation AC magnetic field (H_{AC}). Hence, the experimental setup consisted of an electromagnet for H_{DC} generation and a custom-made 15-turn Helmholtz coil for

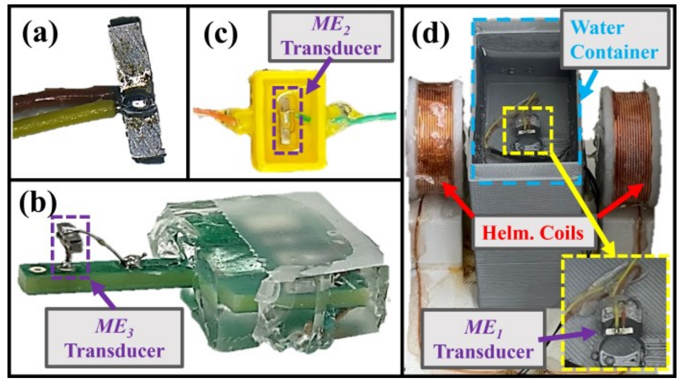

Fig. 2. (a) Image of the 5x1x1 mm³ ME_1 hand-coated with PDMS. (b) ME_2 mounted on a PCB. (c) ME_3 placed inside a custom 3D-printed holder. (d) Image of the modified Helmholtz coils with the water container in the center for measurements in water medium.

uniform H_{AC} generation. More details about the experimental setup can be found in [4]. Since water is a good representative fluid of the human body, the received power of ME transducers was measured in both air and water media. For measurements in water, a Helmholtz coil with a water container in the center was used, as shown in Fig. 2d.

B. Effect of PDMS Coating on ME's Impedance, Q, and P_L

The impedance profile of ME_1 before and after coating with PDMS (both in air and water) was measured with a network analyzer (E5071C, Tektronix, Beaverton, OR), which is shown in Fig. 3a. The magnitude of the impedance dropped with PDMS coating and further decreased when measured in a water medium. The measured impedance magnitude in air were 9.46 kΩ and 8.34 kΩ without and with PDMS coating at the anti-resonance frequencies of 308.6 kHz and 308.3 kHz, respectively. Further, in a water medium, the measured impedance magnitude was 3.4 kΩ at 305 kHz.

The effect of the PDMS coating on the Q of ME_1 was also studied. The open-circuit voltage across ME_1 was measured while performing a two-dimensional sweep of H_{DC} and f_p. At each H_{DC}, the Q of the device was calculated by taking the ratio of the peak voltage to its half-power bandwidth. The trend of the measured Q vs. H_{DC} is shown in Fig. 3b. The PDMS coating mechanically loaded the device while slightly restricting its freedom to vibrate, resulting in a reduction of Q. For example, without any coating at ~ 410 Oe of H_{DC}, the Q of ME_1 was 40.7 that reduced to 35.45 after coating with PDMS. With the surrounding medium as water, the Q was further reduced to 16.5 due to the higher mechanical loading.

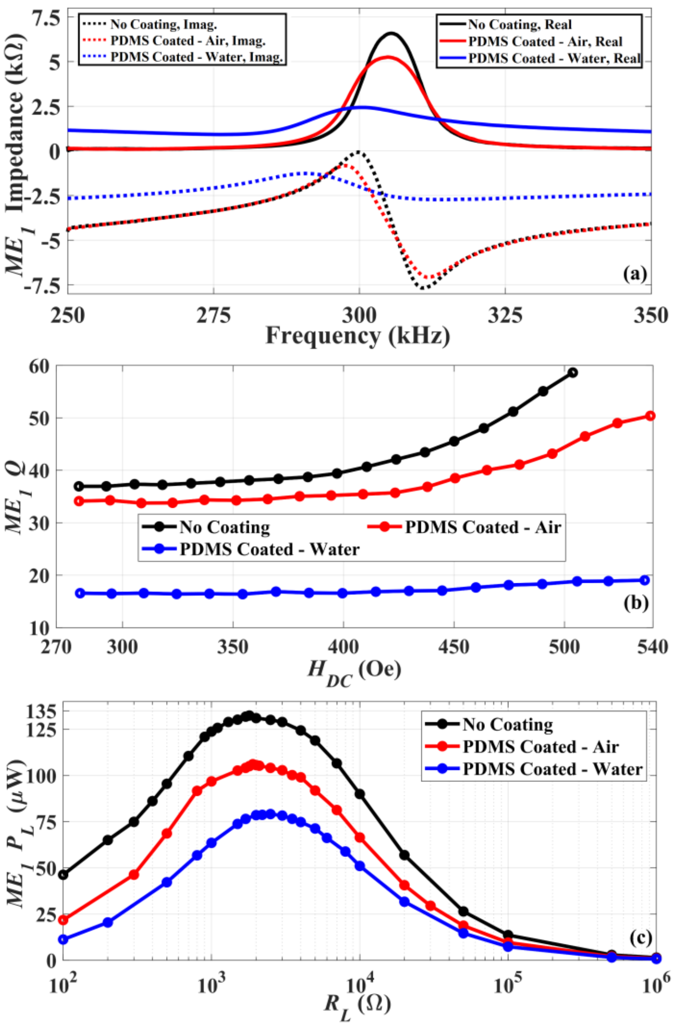

Fig. 3. (a) Measured impedance profile at different frequencies, (b) variation of Q with respect to H_{DC}, and (c) measured P_L of ME_1 transducer (shown in Fig. 2a) with and without PDMS coating in air and water medium.

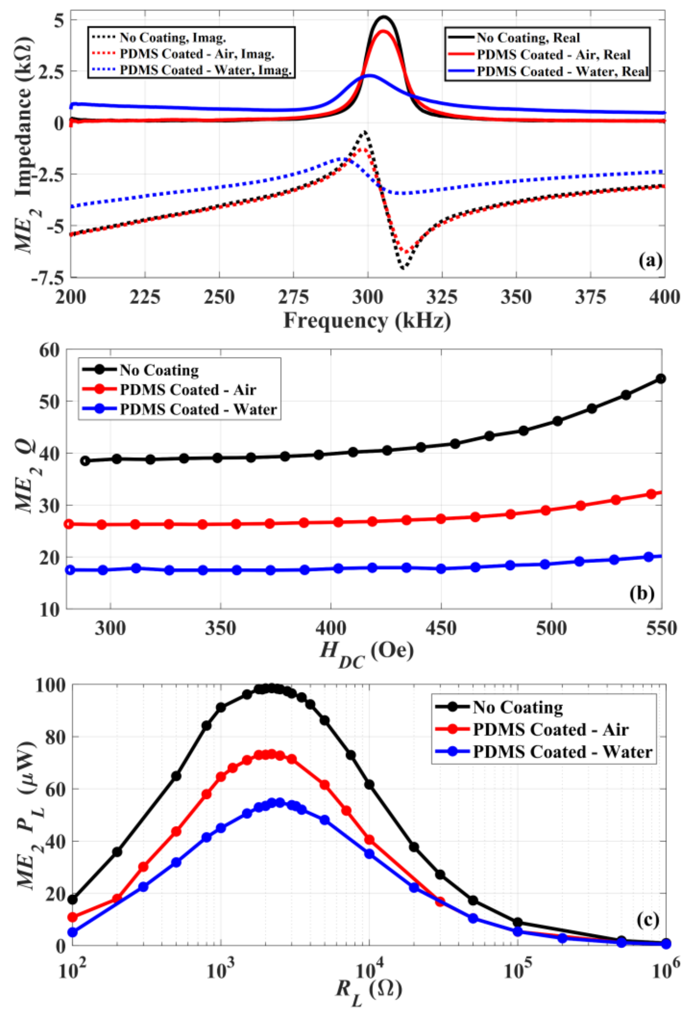

Fig. 4. (a) Measured impedance profile, (b) Q variation with respect to H_{DC}, and (c) measured P_L of ME_2 transducer (shown in Fig. 2b) with and without PDMS coating mounted on a PCB in air and water medium.

To evaluate the effect of PMDS coating on the P_L of ME_1, we measured P_L for a wide range of R_L. The optimal H_{DC} was found as described in [4], and the measured values are listed in Table I. For each R_L, a small frequency range (~ 7 kHz bandwidth) was swept and the maximum power within the measured bandwidth is shown in Fig. 3c. At all frequencies, H_{AC} was set to 1 Oe rms by controlling the current flowing through the Helmholtz coils. For all the measurements in Fig. 3c, the optimal H_{DC} was ~ 410 Oe. As expected, P_L reduced after PDMS coating due to the mechanical loading of the PDMS layer on the ME transducer (similar to reduction in impedance and Q). For example, P_L decreased by 19.4% from 131.04 µW to 105.62 µW at R_L = 2 kΩ in air due to the PDMS coating. The water medium further decreased P_L to 78.54 µW.

III. Packaging of the ME transducer using PCB and 3D Printed Holder

To ensure long-term implant durability, it is essential to package these ME transducers onto a holder or a substrate. In this paper, two types of mounting methods were investigated: 1) on a PCB as a substrate; and 2) on a custom-designed 3D-printed holder. After mounting the devices, they were coated with PDMS as described previously and their effects were studied, which are detailed hereon.

A. Mounting on a PCB

Fig. 2b shows the ME_2 transducer soldered onto a custom-designed PCB with a small air gap in-between. A thin layer of PDMS was then hand-coated, followed by the characterization of the ME transducer (as done previously). The resulting impedance profile, Q, and P_L trends are shown in Fig. 4. In water, at H_{AC} of 1 Oe rms and H_{DC} of ~ 430 Oe, ME_2 was able to receive 54.7 µW at 296 kHz for an R_L of 2.5 kΩ, which is a 25.4% drop in P_L compared to that in air with PDMS coating (73.32 µW). The reduction in P_L is similar to that when no mounting of the device was performed as the ME has the same degree of freedom for vibration. The key advantages of this mounting method are 3-fold. 1) The ME transducer's freedom of vibration is not impacted by the PCB or the mounting procedure. 2) The points where wires are soldered onto the ME transducer have more structural stability compared to the previous setup in Fig. 2a. 3) Since the ME transducer is on the PCB, it is easy to integrate it with other electronics to realize a complete IMD [1]-[3].

B. *Mounting inside a Custom 3D-Printed Holder*

One widely adopted packaging method for IMDs is to place the transducer inside a holder and coat it with a biocompatible material [13], [15]. Based on this approach, we custom-designed and 3D-printed a holder for the ME_3 transducer, as shown in Fig. 2c. The ME_3 transducer would conveniently sit at the center of this holder while the wires are routed through the sides. The top section of this holder is uncovered to assist with the mounting and coating of the ME transducer. It should be noted that ~ 250 µm of PDMS was coated on the top of the ME transducer.

The measured impedance, Q, and P_L of ME_3 transducer for this mounting method are shown in Fig. 5. At an R_L of 3 kΩ, ME_3 could achieve a P_L of only 43.01 µW with PDMS coating in air (vs. 112.9 µW without coating), corresponding to 2.62 times drop in P_L. Two reasons can be attributed to this drastic reduction in the received power. 1) The higher thickness of coated PDMS increased the mechanical loading. Note that when coated ME_3 was tested in water with additional mechanical loading, P_L slightly reduced to 39.02 µW (only ~ 9% drop) for an R_L of 2.5 kΩ. The mechanical loading of water was not highly pronounced as the thick PDMS has already dampened the ME transducer vibrations. 2) The perfect bonding between PDMS and the 3D-printed holder's walls significantly damped the vibrations. Therefore, one should be cautious either while restricting the ME transducer's freedom for vibration or mechanically loading it as it has a considerable impact on its performance.

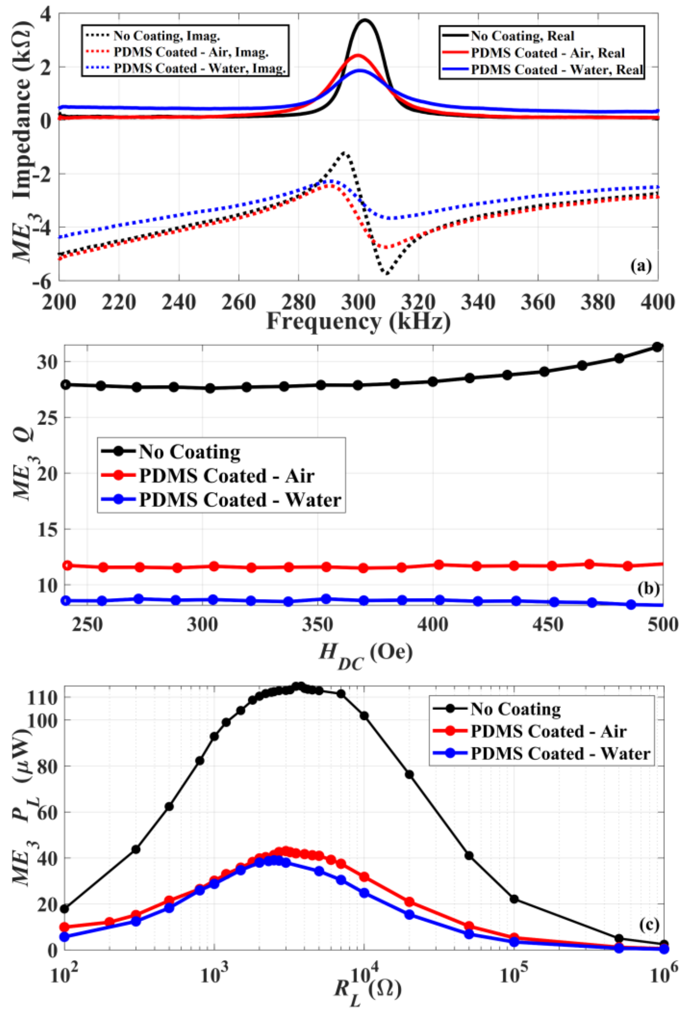

Fig. 5. (a) Measured impedance profile, (b) variation of Q with respect to H_{DC}, and (c) measured P_L of ME_3 transducer (shown in Fig. 2c) with and without PDMS coating mounted in a custom 3D-printed holder. Measurements were conducted in both air and water medium.

IV. Discussion And Future Work

It is important to ensure that IMDs are functioning well within the recommended safety limits imposed for human exposure to electromagnetic fields [8]. Based on the IEEE standard, for f_ps of $ME_{1\text{-}3}$, an $H_{AC} \leq 5.5$ Oe rms (at an implant depth of 30 mm) and $H_{DC} \leq 1670$ Oe are safe [4]. For all the measurements presented, the H_{DC} is well below this limit (Table I), and the H_{AC} was ensured to be 1 Oe rms. If H_{AC} is increased 5.5 times (ensuring operation under safety limit), the expected increase in P_L is 30.25 times as P_L increases proportionally to the squared value of H_{AC} [4]. This indicates that for an R_L of 2.5 kΩ, the P_L of ME_1 and ME_2 in the water medium can be increased from 79.13 µW to 2.39 mW and from 54.7 µW to 1.65 mW, respectively.

The measurements in this paper were conducted in water. We have already shown in [4] that tissue (e.g., chicken breast) only slightly reduces the received power at such low-frequency magnetic fields (i.e., < 10%). Therefore, our future work includes the integration of such ME transducers with electronics for sensing/stimulation applications such as gastric slow-wave recording [16]. Since all our ME transducers were hand-coated with PDMS, it was impossible to precisely control the thickness of the coating. However, sophisticated spin coating techniques can be used in the future to accurately control the thickness of the coated PDMS layer [14]. Finally, lead-free, biocompatible piezoelectric materials can be explored in developing ME transducers in the future [17].

V. Conclusion

A study on the effects of biocompatible coating (using PDMS) and packaging of ME transducers on their performance, in terms of their received power in a WPT link, was presented. Two mounting schemes involving PCB as the substrate and custom 3D-printed holders were also discussed, demonstrating their impact on the ME transducer's impedance, quality factor, and received power. The ME transducer characteristics were measured in both air and water medium with and without PDMS coating to better differentiate an actual implant scenario from benchtop testing. The measured results were compared in different conditions indicating that mechanical loading on the ME transducer should be minimized during encapsulation and device packaging to avoid drastic received power reduction. Sophisticated spin-coating techniques can be employed in the future to control the thickness of the PDMS coating.

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