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1pEAb10. The design of ultrasonic lead magnesium niobate-lead titanate (PMN-PT) composite transducers for power and signal delivery to implanted hearing aids

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We present a system for efficiently powering implanted hearing aids by transmitting an ultrasonic signal across the skin. The use of ultrasound as method for power and signal transfer is known for embedded systems in industrial applications, and has more recently been investigated for use with other medical implants. In our application, ultrasonic transducers are investigated as they offer substantially reduced size relative to traditional magnetic induction coil power delivery. The developed transducers use PMN-PT (lead magnesium niobate-lead titanate) piezoelectric material in a 1-3 composite formulation. PMN-PT offers an electromechanical coupling factor (kt, an indicator of maximum efficiency) that is up to 60% greater than traditional piezoceramics, while the use of composite transducers removes geometric constraints that can limit the achieved efficiency. The fabrication methods for the transducers are detailed. Experimental results are presented to show the composite transducers achieve a kt of 0.77 (out of 1.00), and a power transmission efficiency of 45%.

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INTRODUCTION

The use of ultrasound as method for wireless power and signal transfer is known for embedded industrial applications\(^1\) and has also been investigated for use with implanted medical devices\(^2\text{-}^3\text{-}^4\). However, for implanted medical devices electromagnetic coil induction is currently the accepted technology when wireless power transmission is required. Despite this, there are potential advantages from ultrasound systems that warrant their further development. Importantly, ultrasonic systems based on piezoelectric transducers do not necessarily contain ferrous materials. This means implant recipients can maintain good compatibility with magnetic imaging (MRI) for their medical diagnoses. Considering energy transfer performance there are both advantages and disadvantages to ultrasonic systems. For relatively large transducers over short transmission distances electromagnetic coils do outperform ultrasonic systems. However, when the sizes of transducers are reduced, electromagnetic coils suffer significantly greater efficiency losses than ultrasonic systems\(^5\). Furthermore, ultrasonic transmission beams potentially allow for energy transmission to greater depths within tissue\(^5\). Practically, these performance properties can result in smaller devices with greater surgical placement options when using ultrasound systems.

One application where electromagnetic wireless systems are prominently used is implanted hearing aids in order to transfer both signal and power across the skin. These devices include both cochlear implant devices for sensory-neural hearing loss patients and implanted mechanical transduction devices for conductive and mixed hearing loss patients. The work presented here investigates the use of ultrasound as an alternative technology to electromagnetic coils for powering these types of medical implants. Additionally, developments discussed on ultrasonic transducer efficiency improvements are extendable to other types of medical devices.

Piezoelectric disc transducers have previously been used to generate and receive the ultrasonic waves for the purposes of power transmission. In the work by Ozeri and Shmilovitz\(^3\) (and advanced in Ozeri et al.\(^4\)), piezoelectric disc transducers fabricated from lead zirconate titanate (PZT), with appropriate the acoustic matching layers for tissue interfacing, were used to achieve very good results. Their peak transmission efficiency through tissue was 27% including the losses in the supporting electrical system. In their system the majority of losses occur during electromechanical energy conversion and the acoustic transmission. As a result, improvements to the overall power transfer efficiency of ultrasonic systems can be most readily achieved through improvements to the piezoelectric disc transducers.

To meet this objective two techniques were considered. First, single-crystal lead magnesium niobate-lead titanate (PMN-PT) piezoelectric material was used. PMN-PT can possess a much higher piezoelectric coefficients than traditional PZT materials, having an electromechanical coupling that is close to the theoretical maximum \((k_{33}>0.92)\), and piezoelectric voltage-to-strain coefficients that are up to a factor of five greater\(^6\text{-}^7\). Electromechanical coupling \((k_{33})\) is the ratio of the mechanical energy generated in each cycle (oscillation) to the input electrical energy during that cycle.

The second technique used to improve efficiency is the application of a composite transducer design. Referring to Figure 1, a composite piezoelectric disc designed an acoustic projector by its thickness mode oscillations will consist of piezoelectric pillars surrounded by a relatively soft polymer matrix.

![Figure 1](https://example.com/figure1.png)

**FIGURE 1.** Photograph of a composite piezoelectric disc as used with the ultrasonic power delivery system. In the figure the disc is shown from the top, where square pillars of PMN-PT piezoelectric material are surrounded by an epoxy matrix.

A good overview of the key principals related to piezoelectric composites and a concise analytical model for predicting their aggregate properties has been provided by Smith and Auld\(^8\). In solid piezoelectric discs the achieved
electromechanical coupling, \( k_t \), is reduced from the materials \( k_{33} \) value due to the materials lateral stiffness, which impedes the thickness mode oscillations. In composite transducer the lateral stiffness is greatly reduced by the soft polymer and the piezoelectric pillars are able to operate nearer the maximum \( k_{33} \) value\(^8\). The use of piezoelectric composites is now well established for medical imaging applications, where their use has produced transducers with excellent efficiency and bandwidth\(^9\). In this work the techniques previously applied to imaging transducers are used to achieve similar performance improvements in ultrasonic power delivery.

**TRANSDUCER DESIGN AND FABRICATION**

The piezoelectric composite were fabricated using PMN-PT(30%) (manufactured by TRS Technologies Inc.) and EpoTek 301 epoxy as the polymer fill material. Relevant properties of these materials are provided in Table 1. The design parameters for the composite were determined using recommendations by previous researchers to achieve a high electromechanical coupling coefficient.

In Smith and Auld\(^8\) a pillar thickness to width aspect ratio of 10 is suggested to achieve ideal composite performance. Subsequent analysis by Hayward and Bennett\(^10\) found that composite performance within 80% of the maximum \( k_{33} \) could be achieved using pillar aspect ratios as low as 3.0. As high aspect ratio pillars are more likely to be damaged during the fabrication process the present design tends towards the lower end of this range and uses an aspect ratio of 3.5.

From Smith and Auld\(^8\) it is also known that good performance can be achieved from composites having a piezoelectric volume fraction ranging from 40% to 80%. In composites the electrical impedance is proportional to the piezoelectric volume fraction while the characteristic mechanical impedance is proportional to the volume fraction of polymer fill material. For application in a low voltage system, such as the hearing aid systems considered here, both low electrical and low mechanical impedances are desirable. In the current design a 60% piezoelectric volume fraction is used as a compromise solution.

Finally, within a composite transducer spurious resonances that remove energy from thickness mode oscillations can occur due to the periodicity of the kerfs. To avoid this condition the kerf’s resonance should at minimum be double the operating frequency\(^9\). Following this guideline the maximum kerf width is: \( l_k \leq \frac{v_c}{4\sqrt{2}f_o} \), where \( l_k \) is the kerf width, \( v_c \) is the shear wave speed in the filler material, and \( f_o \) is the transducers operating frequency. Based on these principles a 1.2 mm thick transducer with suitable pillar properties was designed to achieve a 1.0 MHz operating frequency. Additional details on the design are provided in Table 1.

The composite transducers were fabricated using the dice and fill technique\(^9\). First, solid samples of PMN-PT were adhered to a substrate and were cut into pillars using a micro-dicing saw (Disco DAD3220, Disco Hi-Tech, Japan). EpoTek 301 epoxy was then poured between the pillars and allowed to cure. Once cured, the filled composites were lapped to the designed thickness and chromium and gold electrode layers were deposited using a thermal evaporation process. Finally the composite discs were packaged within a pair of Delrin rings around the discs perimeter. The backing rings were left empty to provide a low impedance backing, thus allowing for maximum energy output in the forward direction. The front rings were filled with EpoTek 301, cured, and lapped to a thickness of 0.800 mm to act as an acoustic matching layer between the composite discs and tissue. This layer thickness, which is not a quarter wave length at the operating frequency, was selected based on the composite’s predicted impedance, the known impedance of 301 epoxy, and calculation results from the Smith and Auld\(^8\) composite model in conjunction with a KLM transducer model\(^12\). In Figure 2 a photograph of a completed power link transducer is provided.

**FIGURE 2.** Photograph of a completed power link transducer.
<table>
<thead>
<tr>
<th>PMN-PT</th>
<th>Value</th>
<th>EpoTek 301</th>
<th>Value</th>
<th>Composite</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>$Z$</td>
<td>33.6 MRayl</td>
<td>$Z$</td>
<td>3.048 MRayl</td>
<td>Thickness</td>
<td>1.200 mm</td>
</tr>
<tr>
<td>$\rho$</td>
<td>8060 kg/m$^3$</td>
<td>$\rho$</td>
<td>1150 kg/m$^3$</td>
<td>Diameter</td>
<td>5.00 mm</td>
</tr>
<tr>
<td>$\epsilon_{33}$</td>
<td>846</td>
<td>$\nu_f$</td>
<td>2650 m/s</td>
<td>Kerf width</td>
<td>0.100 mm</td>
</tr>
<tr>
<td>$c_{33}$</td>
<td>116.4 GPa</td>
<td>$\nu_s$</td>
<td>1230 m/s</td>
<td>Pillar pitch</td>
<td>0.440 mm</td>
</tr>
<tr>
<td>$k_t$</td>
<td>0.518</td>
<td></td>
<td></td>
<td>Aspect ratio</td>
<td>3.5</td>
</tr>
</tbody>
</table>

**RESULTS AND DISCUSSION**

The achieved electromechanical coupling of the composite transducer can be determined according to

$$k_t^2 \leq \frac{\pi}{2} \frac{f_s}{f_p} \tan \left( \frac{\pi (f_p - f_s)}{2 f_p} \right)$$

where $f_s$ is the series resonance and $f_p$ is the parallel resonance of the unloaded transducer. These values can be readily determined from electrical impedance measurements of the transducers, shown in Figure 3. For the fabricated transducers $f_s = 1.0$ MHz and $f_p = 1.50$ MHz, giving a coupling factor of $k_t = 0.77$. This value represents an improvement of 50% when compared to the coupling coefficient of the original solid PMN-PT disc.

![FIGURE 3. Measured and calculated electrical impedance of the composite transducer.](image)

The power transmission efficiency of the transducers was evaluated in a simple water test tank. The transmitting transducer was connected to an Agilent 33210A signal generator and the receive transducer was loaded using a 950 $\Omega$ resistor. After application of the acoustic matching layer it was found the operating frequency shifted to 1.07 MHz from the 1.00 MHz unloaded resonant frequency, and the electrical impedance increased to 470 $\Omega$ due to better acoustical matching, so the efficiency tests were carried out accordingly. The transducers were manually aligned until a position was found for maximum power transfer and the efficiency was recorded. From this procedure a peak power transfer efficiency of 45% was found. While this value was obtained in a water test tank improved performance is still expected when transmitting through tissue, additional testing for which is planned. A summary of the coupling factor findings and efficiency findings are provided in Table 2.

Overall results with the composite transducer have been positive. The high efficiency achieved with the relatively small transducers (1.2 mm by 5 mm) is promising for use with low power and low voltage implanted hearing aids. As previously mentioned, the use of these piezoelectric systems allows for greater MRI comparability in implant recipients. Additionally, the use of composite piezoelectric provides thinner transducers than solid piezoelectric discs, further improving surgical placement options when device thickness is a concern.
TABLE 2. Summary of ultrasonic power deliver system performance when using composite PMN-PT transducers.

<table>
<thead>
<tr>
<th>Property</th>
<th>Result</th>
</tr>
</thead>
<tbody>
<tr>
<td>Operating Frequency</td>
<td>1.07 MHz</td>
</tr>
<tr>
<td>Electromechanical coupling</td>
<td>0.78</td>
</tr>
<tr>
<td>Improvement vs. non-composite</td>
<td>51%</td>
</tr>
<tr>
<td>Maximum power transfer efficiency</td>
<td>45%</td>
</tr>
</tbody>
</table>

REFERENCES