2021 IEEE 20th International Conference on Micro and Nanotechnology for Power Generation and Energy Conversion Applications (PowerMEMS) | 978-1-6654-2218-5/21/$31.00 ©2021 IEEE | DOI: 10.1109/PowerMEMS54003.2021.9658362

A Dynamic Transmit Coil for Wirelessly Powering Small ME Transducer based Biomedical Implants

Erik Andersen
Mechanical Engineering
University of Utah
Salt Lake City, USA
erik.andersen@utah.edu

Orpita Saha
Electrical and Computer Engineering
University of Utah
Salt Lake City, USA
orpita.saha@utah.edu

Shad Roundy
Mechanical Engineering
University of Utah
Salt Lake City, USA
shad.roundy@utah.edu

***Abstract*—Magnetoelectric transducers (ME) wireless power transfer systems (WPTS) offer a way to power small biomedical implants. However, if the ME receiver becomes misaligned the wireless power delivered to the load can be dramatically reduced. A dynamic transmit coil using actuators and physically rotating or moving the transmit coil reduces the misalignment between the transmitter and the receiver. We model the expected power gains a WPTS has using a dynamic transmitter versus a static transmitter (a coil that does not move or rotate). We experimentally show that adding a single servo motor to make a dynamic transmit coil increases the power available to load by a factor of 2.4 over an otherwise identical static transmit coil for a given misaligned ME receiver in a WPTS.**

Keywords—Magnetoelectric, wireless power transfer, optimization, dynamic transmitter.

I. Introduction

Wirelessly powering biomedical implants is an attractive solution for eliminating the need for battery-powered implants and thus making the devices much smaller. However, it does introduce some potential challenges for successfully powering the implant. One such problem is misalignment between the transmitter and the receiver. Biomedical implants can easily shift and change their location/orientation inside the body over time. This shift cannot be easily seen or corrected once the biomedical receiver is implanted, but this change of receiver location/orientation can have a significant impact on the wireless power transfers system's (WPTS) overall efficiency.

Nearly all WPTS operate best when the receiver is exposed to a large magnetic field (B-field) that is aligned with the receiver's orientation. The system efficiency decreases when misalignment occurs. This decrease in efficiency occurs in both magnetoelectric transducer (ME) WPTS and inductive coil WPTS [1], [2].

There are several possible ways to reduce the misalignment between the transmit coil and the receiver. Research has been done on omnidirectional WPT transmit coils capable of changing the orientation of the B-field at a point in space. These transmitters can be either a planar array of coils [3] or a non-planar array of coils [4]. The most common non-planar configuration is that of three orthogonal solenoid coils.

However, using a coil array to alter the B-field has drawbacks, especially when powering biomedical implants. Violating the magnetic field safety constraints inside the body is a major concern with WPTS for biomedical implants. Two agencies govern the use of magnetic fields inside the human body, IEEE and ICNIRP [5], [6]. Both agencies set exposure limits for the human body based on the strength and frequency of the magnetic field. While both the planar and omnidirectional WPTS control the B-field strength and orientation at specific points in space, they cannot control the B-field gradient around those points. Thereby, it is probable that using such a system could create “hot-spots” or areas of high B-field strength, thus violating the safety constraints.

Additionally, omnidirectional WPT coils that use a non-planar coil array may not be the ideal configuration for the WPTS transmit coil. The non-planar coil array construction means that at least one transmit coil will not be close to the skin but offset by a significant distance. This offset distance will mean that the B-field will be significantly weaker than a similar-sized transmit coil located closer to the skin, and hence the receiver.

Because of these limits, we investigate a design for creating a WPTS transmit coil that can adjust to implant misalignment or misorientation using a single coil that can change its position and orientation using motors, as seen in Figure 1. Using a single coil transmitter eliminates the creation of unintentional “hot-spots” of high B-field within the body. Furthermore, it has no limits to how close it can be placed by the body. This paper will explore two different types of dynamic transmit coils and present a comparison between the power transfer achieved using a normal static transmit coil versus a dynamic transmit coil for a given WPTS.

II. Methods

This paper will focus on WPTS with a ME transducer receiver designed for biomedical implants. ME transducers are an attractive option because they can achieve relatively high-power densities at small sizes using low radio frequency magnetic fields (kHz range). As the operating frequency increases, the magnetic field exposure limits become stricter, which makes operating at lower frequencies an appealing option. For a WPTS with an ME transducer, the maximum possible power, P_{avt}, that can be delivered to a load is proportional to the square of the magnetic field flux, B_0, given by the following equation

$$P_{avt} = \frac{1}{8}\frac{(\Gamma_m B_o)^2}{b\mu^2} \quad (1)$$

where Γ_m is the electrodynamic transduction factor, b is the mechanical damping coefficient, both of which are solely dependent on the material and mechanical properties of the receiver [7], and μ is the magnetic permeability of the operating space. For a given ME transducer WPTS receiver, maximizing power is equivalent to maximizing the magnetic field, B_0, at the ME transducer's location.

It is important to note that the B_0 in Eq. (1) is the B-field that is aligned in the same direction as the ME transducer. We have found when the B-field becomes misaligned, the power decreases as a function of the cosine of the misalignment angle, θ, or

$$P_\theta = P_0 cos^2(\theta) \quad (2)$$

where P_0 is the power delivered with no misalignment between the B-field and the receiver [1].

The most common method for producing the B-field at the ME receiver is supplying an AC current through a conductive transmit coil. The transmit coil is commonly a solenoid coil whose radius is much greater than its thickness. The B-field produced by such a solenoid is given by

$$B = \frac{\mu}{4\pi}\iiint_V \vec{j}\, d\tilde{v} \wedge \frac{\vec{r}-\vec{r'}}{\left|\vec{r}-\vec{r'}\right|^3} \quad (3)$$

where $\vec{j}$ is the current density, and $\vec{r}-\overrightarrow{r'}$ is a space vector between a point where the magnetic field is calculated and a location inside the solenoid conductor.

When comparing the power gain of WPTS using a dynamic transmit coil to that of a WPTS using a static transmit coil, we can define a figure of merit, Ψ, as

$$\Psi = \frac{P_{dyanmic}}{P_{static}} \quad (4)$$

where $P_{dynamic}$ is the power delivered to the receiver using the dynamic transmit coil and P_{static} is the powered delivered to the receiver using the static transmit coil defined in Eq. (1). However, since the ME transducer for both systems is equivalent, Γ_m, b, and μ are equal and Ψ can be reduced to

$$\Psi = \frac{B^2_{dynamic}}{B^2_{static}} \quad (5)$$

where $B_{dynamic}$ is the strength of the effective B-field (the strength of the B-field aligned with the ME transducer) for the dynamic transmit coil and B_{static} is the strength of the effective B-field for the static transmit coil.

When doing this analysis, we assumed that a person has a ME transducer receiver implanted. The WPTS transmit coil is located some distance away from the person and hence the implant. These assumptions reflect what we envision as a realistic operating environment, where the dynamic transmit coil will be in a fixed location (e.g. on a physician's table or a patient's desk) and therefore not directly touching the person's skin. The WPTS is aligned if the ME transducer receiver lies directly along the transmit coil's z-axis, as seen in Figure 1. The ME transducer is misaligned if the ME receiver is located off the transmit coils z-axis and/or rotationally misaligned with the transmitter by some angle θ.

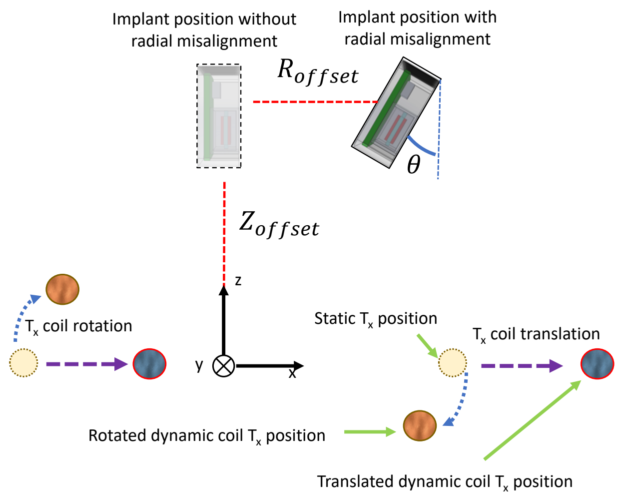

Figure 1 A planar view of the static and dynamic WPTS. The circles represent the cross-sectional area of the solenoid coils. The ME transducer WPTS receiver is located at some distance along the coil's z-axis, Z_{offset}, and some distance along the coil's radial axis (x-axis), R_{offset}. Additionally, the receiver can be misaligned with the static transmit coil by some angle θ. The position of the static transmit coil is shown, as well as a possible position for the dynamic transmit coil. The initial position of the dynamic transmit coil is concentric with the static coil. A rotational dynamic transmit coil (copper color) rotates about the y-axis to its optimal position. A translational dynamic transmit coil (blue color) moves along the x-axis to its optimal position.

For generality, all distances are non-dimensionalized by the transmit coil's radius. Additionally, we assumed the minimum distance between the transmit coil and the ME transducer receiver is three quarters (0.75) of the transmit coil's radius. This reflects what we envision as a likely operating scenario where the dynamic transmit coil is set on a surface with some non-trivial distance between it and the patient. Furthermore, it allows the dynamic transmit coil a rotational operating range of at least 50 degrees. This non-dimensionalization ensures that the coil comparisons are valid for any sized transmit coils operating with any given current.

When doing this analysis, we used a solenoid transmit coil for both the static and dynamic WPTS with a radius of 7.78 cm, made of 9 turns with 16 AWG (1.291 mm diameter) copper wire supplied with 1.67 amps of current. The ME transducer's location was varied along both the radial and transverse directions (along the xz-plane in Figure 1). At each point, the B-field for the static transmit coil, and the B-field for the dynamic transmit coil in its optimal alignment were computed and used to calculate Ψ for a given location. The optimal alignment was calculated by sweeping the dynamic transmit coil through its full range of motion, calculating $B_{dynamic}$ at each point, and then saving the largest $B_{dynamic}$ value.

The implant was also rotationally misaligned by either 30 or 45 degrees relative to its optimal orientation (i.e., in line with the coils z-axis). Figure 1 shows an implant that is both radially and rotationally misaligned and the relative position of both the static and dynamic transmit coils.

We examined two types of dynamic transmit coils; both are shown in Figure 1. The first is a rotational dynamic transmit coil that can rotate at its base in the xz-plane. The second is a translational dynamic transmit coil that can translate along

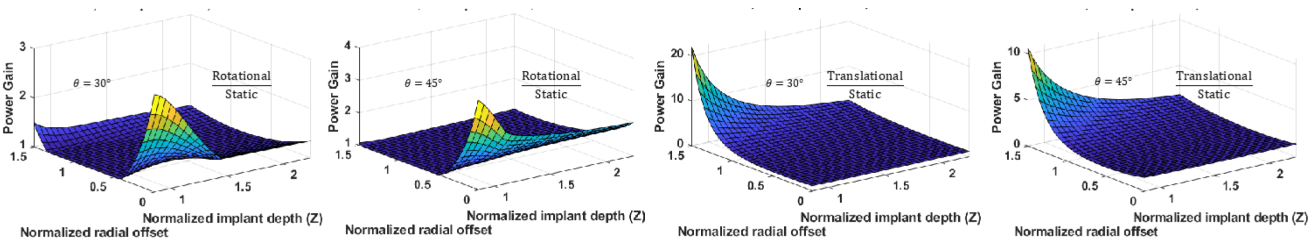

Figure 2 The power gain, Ψ, for a dynamic transmitter WPTS compared to a static transmit coil WPTS for an ME transducer receiver located at various distances with the receiver's angular misalignment, θ, equal to 30 or 45 degrees. The distances have been normalized by the radius of the transmit coil. The dynamic transmit coils equal or outperform the static coils in every case. Note, the scales are different for each graph and the normalized implant depth starts at 0.75.

the coils xy-plane. We examined how these two dynamic transmit coils compare against a static transmit coil and each other.

III. RESULTS

A. Modeling Results

Figure 2 shows the power gain from a dynamic transmit coil as a function of radial misalignment, rotational misalignment, and implant depth. As seen in Figure 2 the dynamic transmit coils equal or outperform the static transmit coils. Actuating the transmit coil leads to better alignment of the B-field and increases the strength of the B-field at the receiver, leading to an increase in available power for the system. While this result may seem obvious, the value of this work lies in modeling the significant power gains associated with adding a simple servo motor system to change a static wireless power transmit coil to a dynamic WPTS transmitter.

For the rotational dynamic transmit coil system, generally, the greater the angular misalignment of the receiver, the higher the power gain, Ψ, the dynamic transmit coil exhibited over the static transmit coil. For implants with low radial misalignment (less than one half of the coil radius) or extrememe misalignment the rotational transmitter outprefroms the static coil. There are some areas where the dynamic and static transmit coils preform equally well. It is interesting to note, though, the high Ψ values for implants with low radial offset and relatively close to the transmit coil. This spike is caused by the dynamic coil's rotation, which in addition to better aligning the B-field with the receiver, also increases the strength of the B-field by reducing the distance between the receiver and the inner plane of the transmit coil.

The optimal angle of rotation is not always equal to that of θ, the angle of the implant's misalignment. Instead, the optimal angle is determined by reducing the misalignment between the transmit coil's B-field and the receiver and by rotating to decrease the distance between the coil's center and the receiver. It is a combination of both factors that produces a rotation angle for the dynamic transmit coil that yields the highest Ψ.

The maximum power gain of the translational dynamic transmit coil, as seen in Figure 2, is greater than that of a rotational dynamic transmit coil. As the implant becomes more radially misaligned, the translational dynamic transmit coil system produces high Ψ values, up to nearly 10-20 times more power delivered when the receiver is located more than one coil radius away from the center of the static transmit coil.

Although the translational dynamic transmit coils produce higher power, they have a disadvantage of needing a larger footprint than static or rotational dynamic coil systems. This larger footprint is necessary for the supports or tracks that the dynamic coil will use to translate, similar to a 3-D printer. However, if translational misalignment is likely, it could be worth the operational cost for having a dynamic transmit coil with a large footprint.

Comparing the rotational dynamic transmit coil to the translational dynamic transmit coil, one does not consistently outperform the other. For implants with small rotational misalignments and/or large translational misalignments, the translational dynamic transmit coil outperforms the rotational dynamic coil. For implants with large rotational misalignments and/or small translational misalignments, the rotational dynamic transmit coil performs better. Either case, however, leads to an improvement over a static transmit coil.

B. Experimental Results

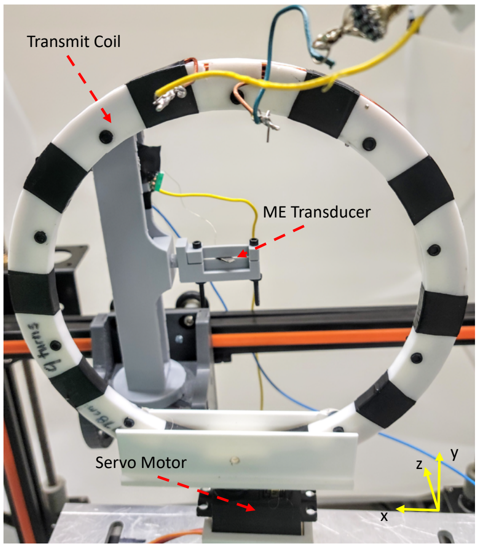

Figure 3 The experimental setup with a rotational dynamic transmit coil. A servo motor turns the coil to its optimal position. The self-biased ME transducer is placed at a misalignment angle of 45 degrees. A 3-D printer (composed of aluminum as to not interfere with the magnetic waves) moves the ME transducer along the radial axis (x-axis). At each ME transducer location, the power is measured across a 4.4 kΩ resistor.

A dynamic transmit coil was built by attaching a 7.78 cm radius, 9-turn (using 16-AWG copper wire) solenoid coil to a servo motor as seen in Figure 3. An Arduino Uno controls the servo motor. A self-biased ME transducer with dimensions of 13.5 mm x 5 mm x 1.5 mm was built from two layers of nickel, two layers of metglas, and a single layer of PZT. A self-biased ME transducer uses two different magnetostrictive materials (nickel and metglas) that have two different permeability and coercive fields, causing differences in saturation magnetizations. This difference in saturation magnetizations causes an internal magnetic field transverse to the layers mitigating the need for an external DC magnetic bias field.

The ME transducer was positioned 5.84 cm ($5.84 = 0.75 * 7.78$) away from the transmit coil along the z-axis with a rotational misalignment of 45 degrees. The radial misalignment of the ME transducer was varied. At each different radial misalignment, the dynamic transmit coil, in its optimal position, and the static transmit coil were supplied with 1.67 amps of current, and the power across a 4.4 kΩ load resistor (equal to the optimal resistance for the ME transducer receiver) was measured and plotted in Figure 4. The effective B-field (effective means the component of the B-field aligned with the ME transducer) at the implant with no radial misalignment for the static transmitter was 32 μT while the maximun effective B-field was 39 μT when the implant had a radial misalingment of $0.39 *$ Coil Radius. For the dynamic transmitter, the max effective B-field was 50 μT which occurred when the implant had zero radial misalignment.

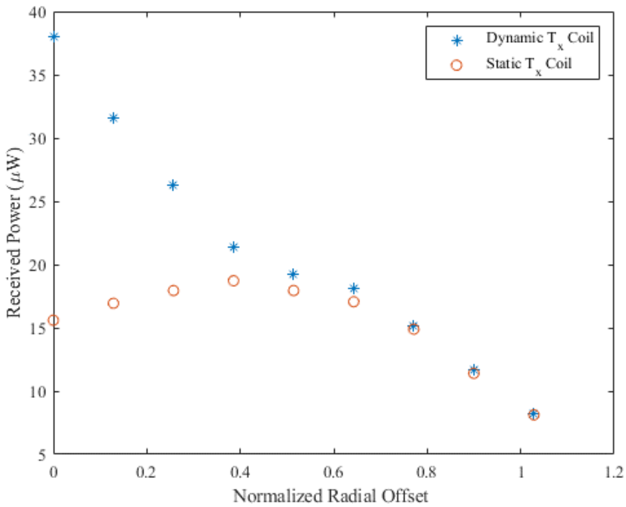

Figure 4 Experimental results of a dynamic transmit coil versus a static transmit coil for a ME transducer located at $Z_{offset} = 5.74$ cm with angular misalignment $\theta = 45$ degrees.

The experimental results show that the dynamic transmit coil provides up to 2.4 times the power of the static transmit coil for areas of shallow radial misalignment. As the radial misalignment approaches half of the coil radius, the dynamic and transmit coil deliver equivalent power. These results closely match the model's predicted results and confirm our hypothesis that dynamic transmit coils provide significantly more power than static transmit coils. The variation from the predicted values may be attributed to imperfections in the manufacturing of the ME transducers, the non-uniform B-field over the length of the transducer, and the friction of the clamp holding the ME transducer.

IV. CONCLUSION

This paper presents a unique dynamic transmit coil that can outperform static transmit coils and deliver higher power to systems with misaligned receivers. We modeled how rotational dynamic transmit coils can provide up to 3 times the power gain over static coils. Additionally, translational dynamic transmit coils can deliver power gains up to 10- 20x for systems with severe misalignment. A rotational dynamic transmit coil was built and shown that it can provide up to 2.4 times the amount of power to a misaligned receiver compared to a static transmit coil for some areas of misalignment but produces better or equal power gains for all ME transducer misalignment locations.

In this work, the dynamic transmit coil and the static transmit coil are equal in size. Future work will compare a larger static transmit coil to the performance of a smaller dynamic transmit coil. A larger transmit coil does not necessarily translate to a higher B-field at the implant. For a given current or power constraint, there is an optimal size for a transmit coil that maximizes the B-field for an implant at a given location [8]. Constructing a coil that is larger than the optimal size will lead to a lower B-field at the implant over a smaller coil that is closer to the optimal size, assuming each coil is supplied with an equivalent current.

ACKNOWLEDGMENT

This work is generously supported by the NSF Grant ECCS-1651438.

REFERENCES

[2] B. W. Flynn and K. Fotopoulou, “Rectifying loose coils: Wireless power transfer in loosely coupled inductive links with lateral and angular misalignment,” *IEEE Microw. Mag.*, vol. 14, no. 2, pp. 48–54, 2013.

[3] J. H. Kim, B. H. Choi, H. R. Kim, and C. T. Rim, “2-D synthesized magnetic field focusing technology with loop coils distributed in a rectangular formation,” *IEEE Trans. Ind. Electron.*, vol. 66, no. 7, pp. 5558–5566, Jul. 2019.

[4] C. Zhang, D. Lin, and S. Y. Hui, “Basic Control Principles of Omnidirectional Wireless Power Transfer,” *IEEE Trans. Power Electron.*, vol. 31, no. 7, pp. 5215–5227, Jul. 2016.

[5] “C95.1-2019 - IEEE Standard for Safety Levels with Respect to Human Exposure to Electric, Magnetic, and Electromagnetic Fields, 0 Hz to 300 GHz.,” 2019.

[6] “International Commission On Non-Ionizing Radiation Protection Icnirp Guidelines For Limiting Exposure To Electromagnetic Fields (100 kHz to 300 GHz),” 2020.

[7] B. D. Truong and S. Roundy, “Experimentally validated model and power optimization of a magnetoelectric wireless power transfer system in free-free configuration,” *Smart Mater. Struct.*, vol. 29, no. 8, p. 085053, Aug. 2020.

[8] E. Andersen, C. Casados, B. D. Truong, and S. Roundy, “Optimal Transmit Coil Design for Wirelessly Powered Biomedical Implants Considering Magnetic Field Safety Constraints,” *IEEE Trans Electromagn. Compat.*, 2021[accepted].