

DEVELOPMENT OF MINIATURIZED, LOW-FREQUENCY MAGNETOELECTRIC ANTENNA FOR DATA AND POWER TRANSFER TO BIOMEDICAL IMPLANTS

DIBYAJYOTI MUKHERJEE



**DEPARTMENT OF ELECTRICAL ENGINEERING
INDIAN INSTITUTE OF TECHNOLOGY DELHI**

July 2026

©Indian Institute of Technology Delhi (IITD), New Delhi, 2026

DEVELOPMENT OF MINIATURIZED, LOW-FREQUENCY MAGNETOELECTRIC ANTENNA FOR DATA AND POWER TRANSFER TO BIOMEDICAL IMPLANTS

by

DIBYAJYOTI MUKHERJEE

DEPARTMENT OF ELECTRICAL ENGINEERING

Submitted

In fulfilment of the requirements of the degree of Doctor of Philosophy

to the



INDIAN INSTITUTE OF TECHNOLOGY DELHI

July 2026

THESIS CERTIFICATE

This is to certify that the thesis titled **Development of Miniaturized, Low-Frequency Magnetoelectric Antenna for Data and Power Transfer to Biomedical Implants**, submitted by **Dibyajyoti Mukherjee (2019EEZ8346)**, to the Indian Institute of Technology, Delhi, for the award of the degree of **Doctor of Philosophy(Ph.D.)**, is a bonafide record of the research work done by him under our supervision. The contents of this thesis, in full or in parts, have not been submitted to any other Institute or University for the award of any degree or diploma.

Dhiman Mallick

Prof. Dhiman Mallick

Department of Electrical Engineering,
Indian Institute of Technology Delhi.

Place: New Delhi

Date: July 9, 2026

ACKNOWLEDGEMENTS

The long journey of the PhD has been full of ups and downs. However, certain pillars of my life provided unwavering support, and without their presence, reaching this stage would not have been possible. Firstly, I would like to express my sincere and profound gratitude to my PhD supervisor, Prof. Dhiman Mallick, for his unwavering support and guidance. His mentorship extended beyond technical discussions and research improvements, as he also guided me whenever I approached him for suggestions regarding my career objectives and future goals. He has consistently inspired me to strive for excellence and has provided both moral and financial support to help me carry out high-quality research work. I am deeply inspired by your passion and dedication towards the development of microscale devices, wireless power transfer systems, and magnetoelectrics. I sincerely thank the Student Research Committee (SRC) members, Prof. Bhaskar Mitra, Prof. Ankesh Jain, and Prof. Samaresh Das for their invaluable suggestions, technical advice, and insightful comments during my Comprehensive Viva and SRC presentations. I sincerely thank Prof. Neetu Singh for her support and guidance and for granting permission to carry out in vitro experiments in her lab. I am also grateful to Prof. Ankur Goswami and Prof. Soutik Betal for their continuous support and guidance. I would like to thank IIT Delhi for awarding me the Research Scholar Travel Award (RSTA) grant to present my research at the IEEE Sensors 2023 conference in Vienna, Austria, and the Research Scholar Travel Excellence Award (RETA) grant to attend IEEE BioSensors 2024 in Cambridge, UK. Finally, I would like to thank the Ministry of Education (MoE), Government of India, for providing financial support throughout my PhD.

I am very much thankful to my friends and colleagues at the Interdisciplinary Microsystems Lab (IML), IIT Delhi, Dr. Nadeem, Dr. Pankaj, Mr. Vinit, and Mr. Faizan, Mr. Sujoy and Palak for making a mesmerizing and friendly environment in the lab. I am thankful to my friends Dr. Shruti, Dr. Sailendra, Dr. Nandan, and Sudipto for their constant support throughout the journey of my PhD. Heartfelt thanks to Nairrita for the constant encouragement, kindness, motivation, and belief that helped me immensely during the final phase of the journey. Finally, I am deeply grateful to all my family members for their unwavering love and support. My heartfelt gratitude goes to my parents, Mr. Prabhat Kumar Mukherjee and Mrs. Kakali Mukherjee, and to my Gurudev, Mr. Bijon Sarkar, and Gurumata, Mrs. Nupur Sarkar, who have been a constant source of strength, unconditional love, and blessings, helping me overcome every challenge along the way. Without their blessings and support, this would not have been possible.

I am also thankful to my elder uncle, Mr. Pranab Kumar Mukherjee, whose motivation, inspiration, and consistent support have always guided me. I am truly grateful to my uncle, Mr. Joydip Banerjee, for his valuable suggestions and constant encouragement throughout this journey. I express my heartfelt gratitude to God, the Almighty for his blessings and for granting

me the strength and perseverance to complete my Ph.D. journey.

Dibyajyoti Mukherjee

Dibyajyoti Mukherjee

Department of Electrical Engineering,
Indian Institute of Technology Delhi.

Place: New Delhi

Date: July 9, 2026

ABSTRACT

KEYWORDS: Magnetolectric Antenna, Wireless Power Transfer, Near Field Data Communication, Self-Biased Magnetolectric Devices, Magnetically Graded Magnetolectric Heterostructure, Power Management Circuit, Implantable Medical Devices

With the rapid proliferation of low-power electronics and smart sensor devices, the demand for miniaturized antennas has garnered significant attention for potential applications in implantable medical devices (IMDs), including therapeutic medication, neural stimulation, deep body communications, and energy harvesting applications. The IMDs are earlier powered or communicated through an external power supply via transcutaneous wire, which connects the implants to the extrinsic environment and increases the risk of infection. It also hinders the free movement of the implanted device in relation to the targeted muscles, reducing the performance or quality of data communication with the IMDs. On the other hand, battery-powered IMDs can replace wired devices by providing complete isolation from the extrinsic environment, thereby reducing the risk of infection to a certain extent and allowing for free muscular movements. However, battery-powered IMDs still fail to act as a safer alternative as they need periodic replacement through surgical invasion. Wireless Power Transfer (WPT) systems have emerged as a safer alternative to existing powering techniques for IMDs, addressing the aforementioned challenges. There are several successful WPTs incorporated in IMD applications that could solve the existing problems, one of them is radio frequency antennas that operate in the radio frequency (RF) range, which are used extensively in portable applications due to their smaller electrical wavelength, which miniaturizes the overall size of the antenna[1], [2]. However, they are denied for IMD applications due to the higher signal loss resulting from electromagnetic (EM) wave absorption in biological tissues. On the contrary, low-frequency (LF) EM antennas, which are widely used in meteorological broadcasting, radio communications, RFID technology, and underwater communication, can be used in IMDs due to the lower EM wave absorption. The LF EM antennas typically operate in the frequency range of 30 kHz to 300 kHz. The associated wavelength(λ), being proportional to the antenna aperture, ranges from 1 to 10 km, nullifying the feasibility of IMD applications. Using the Inductively coupled power transfer (ICPT) technique, the problem of power absorption in the tissue can be addressed. In the ICPT system, the power is coupled from an external transmitter (Tx) coil to the implanted receiver (Rx) coil through EM coupling. Maximum efficiency can be achieved by optimizing the number of turns of the Tx-Rx coils. Despite the ICPT system's high power conversion efficiency (PCE), it suffers from angular misorientation between the Tx and Rx coils, which incurs higher power loss and degrades the PCE. Acoustic Power Transfer (APT), which operates on the principle of ultrasonic (> 20

kHz-10 MHz) wave propagation, can overcome the problems associated with ICPT. As the speed of ultrasonic signal propagation is significantly lower than that of EM wave propagation, the associated wavelength of the APT device is orders of magnitude smaller than that of EM-based WPTs. However, higher acoustic loss in the bones and tissues ≥ 22 dB/cm.MHz [2] degrades the device performance. Additionally, APTs are very sensitive to angular misorientation and must be placed in close contact with the tissues. Therefore, a slight movement of Tx or Rx results in the degradation of the PCE. Lately, the antennas that combine the effect of magnetoelectric (ME) coupling and electromechanical resonance (EMR) have been explored by researchers for very high frequency (VHF) and ultra-high frequency (UHF) band applications using standard silicon integration that overcomes the significant drawbacks of the conventional EM antenna in terms of efficiency and size miniaturization. However, the VHF and UHF ME antennas suffer from higher losses due to the absorption of high-frequency magnetic fields in biological tissues. The LF ME antenna, which works on the same principle as its HF counterparts and operates at the EMR, overcomes the drawbacks of conventional LF antennas. In contrast to conventional EM antennas, the LF ME antenna resonates at its natural mechanical frequency, which dramatically reduces its overall size, and also significantly lowers the LF magnetic field absorption in human tissue. Most of the reported works on magnetoelectric transducers are carried out using Metglas/PZT, FeGaB/AlN, Terfenol-D/PZT, Metglas/PMN-PT heterostructure and incorporating bulk-sized NdFeB or Rb magnets for providing biasing magnetic field[1], [3]. The state-of-the-art reported works either suffer from large size (due to bulk biasing magnets), contain toxic substances such as lead (Pb), or operate at higher frequencies, making them unviable for IMD applications. Therefore, replacing them with bio-compatible alternatives and restricting the appropriate operating frequency for deep-body applications is essential. Moreover, the majority of the reported works do not provide any experimental demonstration with human body ambience, raising concerns over the performance of the ME antennas in such an environment. Additionally, no complete system-level demonstration of a biocompatible ME antenna for near-field data transfer with a low-cost power management interfacing circuit for a potential magnetically chargeable WPT device has been reported so far.

In this thesis, a self-biased, lead-free, LF ME heterostructure-based receiving (Rx) antenna is proposed that provides advantages over the existing ME antennas with higher operating frequency or DC biasing magnets, in terms of improved performance due to lower signal attenuation and miniaturized size, suitable for IMD applications. This work demonstrates the operation of a magnetic bias-independent, both epoxy-based laminate-based and thermocompression-bonded, magnetically graded magnetoelectric (ME) antenna for low-frequency data communication in short-range biomedical applications. The proposed antenna exhibits enhanced noise immunity and supports high data transfer rates of up to 5 kHz. In addition, the ME device is integrated with a micrometer-scale power management circuit (PMC). The ME WPT system generates a regulated DC output voltage of 3.5 V from an input voltage range of 0.7-3.3 V and remains operational up to a distance of 20 cm from the Tx coil. The antenna is characterized using a planar spiral coil with 50 turns (N), which is designed and

fabricated on a 2-layer printed circuit board (PCB). The *in vitro* cytotoxicity analysis in the human hepatic cell line WRL-68 confirmed the biocompatibility of the Parylene-C encapsulated ME device for up to 7 days, suggesting its potential use in implantable electronic devices for biomedical and clinical applications. Finally, the operational capability of the ME antenna is tested under three different conditions: in air, in a tissue-mimicking phantom, and in animal tissue. Further, the optimized ME antenna is explored for regulated tissue heating applications by controlling the ON/OFF time of the AC magnetic field.

सार

कीवर्ड: मैग्नेटोइलेक्ट्रिक एंटीना, वायरलेस शक्ति स्थानांतरण, निकट-क्षेत्र डेटा संचार, स्व-पक्षपाती मैग्नेटोइलेक्ट्रिक उपकरण, चुंबकीय-ग्रेडेड मैग्नेटोइलेक्ट्रिक हेटरोस्ट्रक्चर, शक्ति प्रबंधन परिपथ, प्रत्यारोपणीय चिकित्सीय उपकरण।

कम-शक्ति इलेक्ट्रॉनिक प्रणालियों तथा स्मार्ट सेंसर उपकरणों के तीव्र विकास के साथ, सूक्ष्म आकार के एंटीनाओं की आवश्यकता ने विशेष महत्व प्राप्त किया है। ऐसे एंटीनाओं के प्रत्यारोपणीय चिकित्सीय उपकरणों (IMDs) में चिकित्सीय औषधि वितरण, तंत्रिका उद्दीपन, गहन-शारीरिक संचार तथा ऊर्जा संचयन जैसे संभावित अनुप्रयोग हैं। प्रारम्भिक IMDs को शक्ति अथवा संचार बाहरी विद्युत स्रोत से त्वचा के आर-पार जाने वाले तार के माध्यम से प्रदान किया जाता था, जो प्रत्यारोपित उपकरण को बाह्य वातावरण से जोड़ता है तथा संक्रमण के जोखिम को बढ़ाता है। इसके अतिरिक्त, यह लक्षित मांसपेशियों के सापेक्ष प्रत्यारोपित उपकरण की स्वतंत्र गति में भी बाधा उत्पन्न करता है, जिससे IMDs के प्रदर्शन अथवा डेटा संचार की गुणवत्ता प्रभावित होती है। दूसरी ओर, बैटरी-आधारित IMDs तारयुक्त उपकरणों के स्थान पर बाह्य वातावरण से पूर्ण पृथक्करण उपलब्ध कराते हैं, जिससे संक्रमण का जोखिम कुछ सीमा तक कम हो जाता है तथा मांसपेशियों की प्राकृतिक गति के अनुरूप उपकरण स्वतंत्र रूप से कार्य कर सकता है। तथापि, बैटरी-आधारित IMDs भी पूर्णतः सुरक्षित विकल्प सिद्ध नहीं होते, क्योंकि बैटरी के आवधिक प्रतिस्थापन के लिए पुनः शल्य-चिकित्सा की आवश्यकता होती है। इन चुनौतियों के समाधान के रूप में वायरलेस शक्ति स्थानांतरण (WPT) प्रणाली IMDs को शक्ति प्रदान करने की एक सुरक्षित एवं प्रभावी तकनीक के रूप में विकसित हुई है। वर्तमान में IMDs के लिए अनेक WPT तकनीकों का सफलतापूर्वक उपयोग किया गया है, जो उपर्युक्त समस्याओं के समाधान में सहायक हैं। इनमें से एक प्रमुख तकनीक रेडियो आवृत्ति (RF) एंटीना है, जो रेडियो आवृत्ति परास में कार्य करता है। छोटे विद्युत तरंगदैर्घ्य के कारण इन एंटीनाओं का समग्र आकार अत्यंत छोटा बनाया जा सकता है, जिसके कारण इनका व्यापक उपयोग पोर्टेबल अनुप्रयोगों में किया जाता है। किन्तु जैविक ऊतकों द्वारा विद्युतचुंबकीय (EM) तरंगों के अधिक अवशोषण से उत्पन्न उच्च संकेत क्षीणन के कारण ये IMDs के लिए उपयुक्त नहीं हैं। इसके विपरीत, निम्न-आवृत्ति (LF) विद्युतचुंबकीय एंटीना, जिनका व्यापक उपयोग मौसम विज्ञान प्रसारण, रेडियो संचार, RFID प्रौद्योगिकी तथा जलमग्न संचार में किया जाता है, जैविक ऊतकों में EM तरंगों के अपेक्षाकृत कम अवशोषण के कारण IMDs में उपयोग किए जा सकते हैं। सामान्यतः LF विद्युतचुंबकीय एंटीना 30 kHz से 300 kHz की आवृत्ति सीमा में कार्य करते हैं। इनका संबद्ध तरंगदैर्घ्य (λ), जो एंटीना के आकार के समानुपाती होता है, लगभग 1 से 10 km तक होता है, जिसके कारण ऐसे एंटीनाओं का प्रत्यारोपणीय चिकित्सीय उपकरणों में उपयोग व्यावहारिक रूप से संभव नहीं रह जाता। इंडक्टिवली कपल्ड पावर ट्रांसफर (ICPT) तकनीक का उपयोग करके जैविक ऊतकों में शक्ति अवशोषण की समस्या का समाधान किया जा सकता है। ICPT प्रणाली में शक्ति का स्थानांतरण बाहरी ट्रांसमीटर (Tx) कॉइल से प्रत्यारोपित रिसीवर (Rx) कॉइल तक विद्युतचुंबकीय युग्मन के माध्यम से किया जाता है। Tx तथा Rx कॉइलों के टर्न की संख्या का उपयुक्त अनुकूलन करके अधिकतम शक्ति रूपांतरण दक्षता (PCE) प्राप्त की जा सकती है। यद्यपि ICPT प्रणाली उच्च PCE प्रदान करती है, तथापि Tx तथा Rx कॉइलों के मध्य कोणीय असंरेखण इसकी प्रमुख सीमाओं में से एक है। परिणामस्वरूप, शक्ति हानि बढ़ जाती है तथा प्रणाली की PCE में कमी आती है। अल्ट्रासोनिक (20 kHz–10 MHz) तरंगों के प्रसार के सिद्धांत पर आधारित ध्वनिक शक्ति स्थानांतरण (APT) प्रणाली, ICPT से संबंधित समस्याओं का समाधान प्रदान कर सकती है। चूंकि अल्ट्रासोनिक संकेतों का प्रसार वेग विद्युतचुंबकीय तरंगों की तुलना में काफी कम होता है, इसलिए APT उपकरण का संबद्ध तरंगदैर्घ्य EM-आधारित वायरलेस शक्ति स्थानांतरण प्रणालियों की अपेक्षा कई गुणा छोटा होता है। तथापि, अस्थियों तथा जैविक ऊतकों में होने वाली उच्च ध्वनिक हानि (≥ 22 dB/cm·MHz) उपकरण के प्रदर्शन को प्रभावित करती है। इसके अतिरिक्त, APT प्रणाली कोणीय असंरेखण के प्रति अत्यधिक संवेदनशील होती है तथा इसके प्रभावी संचालन के लिए इसे ऊतकों के निकट संपर्क में स्थापित करना आवश्यक होता है। परिणामस्वरूप, ट्रांसमीटर (Tx) अथवा रिसीवर (Rx) की स्थिति में होने वाला थोड़ा-सा परिवर्तन भी शक्ति रूपांतरण दक्षता (PCE) में कमी का कारण बनता है। हाल के वर्षों में, मैग्नेटोइलेक्ट्रिक युग्मन तथा विद्युत-यांत्रिक

अनुनाद (EMR) के संयुक्त प्रभाव पर आधारित एंटीनाओं ने शोधकर्ताओं का विशेष ध्यान आकर्षित किया है। इन एंटीनाओं का विकास मानक सिलिकॉन एकीकरण तकनीक का उपयोग करते हुए बहुत उच्च आवृत्ति (VHF) तथा अति-उच्च आवृत्ति (UHF) बैंड अनुप्रयोगों के लिए किया गया है। पारंपरिक विद्युतचुंबकीय एंटीनाओं की तुलना में ये एंटीना उच्च दक्षता तथा उल्लेखनीय आकार-लघुकरण प्रदान करते हैं। तथापि, VHF तथा UHF मैग्नेटोइलेक्ट्रिक एंटीनाओं में जैविक ऊतकों द्वारा उच्च-आवृत्ति चुंबकीय क्षेत्र के अधिक अवशोषण के कारण शक्ति हानि बढ़ जाती है। इसके विपरीत, निम्न-आवृत्ति मैग्नेटोइलेक्ट्रिक एंटीना, जो अपने उच्च-आवृत्ति समकक्षों के समान सिद्धांत पर कार्य करता है तथा विद्युत-यांत्रिक अनुनाद (EMR) पर अनुनादित होता है, पारंपरिक निम्न-आवृत्ति एंटीनाओं की सीमाओं को दूर करता है। पारंपरिक विद्युतचुंबकीय एंटीनाओं के विपरीत, निम्न-आवृत्ति मैग्नेटोइलेक्ट्रिक एंटीना अपनी प्राकृतिक यांत्रिक अनुनाद आवृत्ति पर कार्य करता है, जिससे इसका समग्र आकार उल्लेखनीय रूप से कम हो जाता है तथा मानव ऊतकों में निम्न-आवृत्ति चुंबकीय क्षेत्र का अवशोषण भी काफी कम हो जाता है। अब तक प्रकाशित अधिकांश शोध कार्यों में मैग्नेटोइलेक्ट्रिक रूपांतरण उपकरणों का विकास Metglas/PZT, FeGaB/AlN, Terfenol-D/PZT तथा Metglas/PMN-PT हेटरोस्ट्रक्चर का उपयोग करके किया गया है तथा आवश्यक बायस चुंबकीय क्षेत्र उपलब्ध कराने के लिए बड़े आकार के NdFeB अथवा Rb चुंबकों का प्रयोग किया गया है। वर्तमान अत्याधुनिक शोध कार्यों में प्रयुक्त प्रणालियाँ या तो बड़े आकार के बायस चुंबकों के कारण आकार में बड़ी होती हैं, अथवा इनमें सीसा (Pb) जैसे विषैले पदार्थों का उपयोग किया जाता है, या फिर ये उच्च आवृत्तियों पर कार्य करती हैं, जिसके कारण ये प्रत्यारोपणीय चिकित्सीय उपकरणों के लिए उपयुक्त नहीं हैं। इसलिए, इन प्रणालियों के स्थान पर जैव-संगत विकल्पों का उपयोग करना तथा गहन-शारीरिक अनुप्रयोगों के लिए उपयुक्त कार्यकारी आवृत्ति का चयन करना अत्यंत आवश्यक है। इसके अतिरिक्त, उपलब्ध अधिकांश शोध कार्यों में मानव-शरीर जैसे वास्तविक परिवेश में कोई प्रयोगात्मक प्रदर्शन प्रस्तुत नहीं किया गया है, जिससे ऐसे वातावरण में मैग्नेटोइलेक्ट्रिक एंटीनाओं के प्रदर्शन को लेकर संदेह बना रहता है। साथ ही, जैव-संगत मैग्नेटोइलेक्ट्रिक एंटीना का उपयोग करते हुए, कम लागत वाले शक्ति प्रबंधन परिपथ के साथ एकीकृत निकट-क्षेत्र डेटा संचार तथा चुंबकीय रूप से चार्ज होने वाले वायरलेस शक्ति स्थानांतरण उपकरण के लिए संपूर्ण प्रणाली-स्तरीय प्रदर्शन भी अब तक प्रकाशित नहीं किया गया है।

इस शोधकार्य में प्रत्यारोपणीय चिकित्सीय उपकरणों के लिए एक सेल्फ-बायस्ड, सीसा-मुक्त, निम्न-आवृत्ति मैग्नेटोइलेक्ट्रिक हेटरोस्ट्रक्चर आधारित अभिग्राही (Rx) एंटीना प्रस्तावित किया गया है। प्रस्तावित एंटीना, उच्च कार्यकारी आवृत्ति अथवा DC बायस चुंबकों पर आधारित वर्तमान मैग्नेटोइलेक्ट्रिक एंटीनाओं की तुलना में कम संकेत क्षीणन तथा सूक्ष्म आकार के कारण बेहतर प्रदर्शन प्रदान करता है, जिससे यह प्रत्यारोपणीय चिकित्सीय अनुप्रयोगों के लिए अधिक उपयुक्त बनता है। इस शोधकार्य में चुंबकीय बायस से स्वतंत्र, एपॉक्सी-आधारित लैमिनेट तथा थर्मोकम्प्रेसन-बॉन्डेड, चुंबकीय-ग्रेडेड मैग्नेटोइलेक्ट्रिक एंटीनाओं के संचालन का प्रदर्शन निम्न-आवृत्ति निकट-क्षेत्र डेटा संचार के लिए अल्प-दूरी जैव-चिकित्सीय अनुप्रयोगों में किया गया है। प्रस्तावित एंटीना बेहतर शोर प्रतिरोधक क्षमता प्रदर्शित करता है तथा 5 kHz तक की उच्च डेटा स्थानांतरण दर का समर्थन करता है। इसके अतिरिक्त, विकसित मैग्नेटोइलेक्ट्रिक उपकरण को माइक्रोमीटर-स्तरीय शक्ति प्रबंधन परिपथ के साथ एकीकृत किया गया है। विकसित मैग्नेटोइलेक्ट्रिक वायरलेस शक्ति स्थानांतरण प्रणाली 0.7–3.3 V के इनपुट वोल्टेज से 3.5 V का विनियमित DC आउटपुट वोल्टेज उत्पन्न करती है तथा ट्रांसमीटर (Tx) कॉइल से 20 cm तक की दूरी पर भी प्रभावी रूप से कार्य करती है। एंटीना का अभिलक्षणन 50 फेरे (N) वाले प्लेनर स्पाइरल कॉइल का उपयोग करके किया गया है, जिसे 2-लेयर प्रिंटेड सर्किट बोर्ड (PCB) पर अभिकल्पित एवं निर्मित किया गया है। पैरिलीन-सी से आवृत मैग्नेटोइलेक्ट्रिक उपकरण का मानव यकृत कोशिका रेखा WRL-68 पर इन-विट्रो साइटोटॉक्सिसिटी विश्लेषण किया गया, जिससे 7 दिनों तक इसकी जैव-संगतता की पुष्टि हुई। यह परिणाम प्रत्यारोपणीय इलेक्ट्रॉनिक उपकरणों के जैव-चिकित्सीय तथा नैदानिक अनुप्रयोगों में इसके संभावित उपयोग को दर्शाता है। अंततः, विकसित मैग्नेटोइलेक्ट्रिक एंटीना की कार्यक्षमता का परीक्षण तीन विभिन्न परिस्थितियों वायु, ऊतक-अनुरूप) फैंटम तथा पशु ऊतक में किया गया। इसके अतिरिक्त, अनुकूलित मैग्नेटोइलेक्ट्रिक एंटीना का उपयोग प्रत्यावर्ती (AC) चुंबकीय क्षेत्र के ON/OFF समय को नियंत्रित करके विनियमित ऊतक तापन अनुप्रयोगों के लिए भी प्रदर्शित किया गया।

TABLE OF CONTENTS

ACKNOWLEDGEMENTS	i
ABSTRACT	iii
LIST OF TABLES	xi
LIST OF FIGURES	xvii
ABBREVIATIONS	xviii
NOTATIONS	xx
1 Introduction	1
1.1 Motivation	1
1.2 Based on the above-mentioned research aim, the research objectives are identified as follows:	3
1.3 Organization of Thesis Chapters	3
2 Literature Review	5
2.1 Introduction	5
2.1.1 Inductively Coupled Power Transfer (ICPT)	7
2.1.2 Acoustic Wave based Power Transfer (APT)	7
2.1.3 RF WPTs	8
2.1.4 Magnetoelectric Antennas	8
2.2 Summary	15
3 Design Considerations of Low-frequency Magnetoelectric Antennas for Biomedical Implants	18

3.1	Introduction	18
3.1.1	Absorption of AC magnetic field in biological tissues	19
3.1.2	Safety Measures with maximum SAR	20
3.1.3	Design methodology of the ME transducer	20
3.1.4	Theoretical Modeling of the ME transducer	23
3.1.5	FEM analysis of the ME device	26
3.2	Summary	27
4	Development and Characterization of Miniaturized, Low-frequency Magnetolectric Antenna	28
4.1	Introduction	28
4.1.1	Structural Design & Fabrication	30
4.1.2	Opimization of Aspect ratio of ME Laminate	31
4.1.3	ME WPT Device Characterization	31
4.1.4	ME Antenna Characterization for Near Field Communication	36
4.1.5	Experiment On Radiation Mechanism And Data Transfer	38
4.1.6	Demonstration of Near Field Communication	39
4.1.7	Localized Heating Using ME Antenna	41
4.2	Summary	43
5	Development of Low-temperature Bonded Magnetolectric Heterostructure based Antenna	45
5.1	Introduction	45
5.2	Development of ME Antenna	46
5.2.1	Structural Analysis	46
5.2.2	Development of Low-temperature Bonded Magnetolectric Heterostructure based Antenna	49
5.2.3	Characterization of Material Properties	51
5.2.4	FESEM Analysis	54
5.3	Result & Discussion	54

5.3.1	Electrical Characterization of ME Antenna	54
5.3.2	Data Transfer Through ME Antenna	57
5.4	Summary	61
6	System-level Demonstration of Magnetoelectric Antenna for Data and Power Transfer Applications	62
6.1	Introduction	62
6.2	Characterization of ME antenna	64
6.2.1	Human Tissue Mimicking Phantom Preparation and Characterization of ME antenna	64
6.2.2	Electrical Characterization of the ME antenna	66
6.3	Experimental Results of PMC	67
6.3.1	A Off-the-shelf component based interface circuit for ME voltage regulation	72
6.3.2	In vitro Cytotoxicity studies of the ME Device	74
6.4	Summary	75
7	Conclusion and Future Scope	76
7.1	Overall Conclusion of the Thesis	76
7.2	Future Works	77
	BIBLIOGRAPHY	77
	LIST OF PUBLICATIONS	91
	BIO-DATA	94

LIST OF TABLES

2.1	A comprehensive comparison of the ME device with state-of-the-art works on WPTs	10
2.2	List of different magnetostrictive materials with properties usable for ME device development.	16
2.3	List of different commonly used piezoelectric materials for magnetoelectric applications.	17
3.1	Properties of 6-layered tissue phantom	20
3.2	Design parameters of ME transducer	24
4.1	A qualitative comparison of the ME antenna with state-of-the-art works	43
5.1	A qualitative comparison of the ME antenna with state-of-the-art works	59
6.1	Properties of 3-layered tissue phantom	66
6.2	Comparison of the ME antenna system with the State-of-the-art works	72

LIST OF FIGURES

1.1	Market size growth and future prediction for implantable medical devices from the year 2018-2030 [7].	2
2.1	A roadmap to the innovations in implantable bioelectronics and powering techniques with the device prototype images.	6
2.2	Wireless power transfer topologies: (a) ICPT topology schematic and (b-d) Tx coil and miniaturized Rx coil inside rat body[49]. (e) APT topology schematic, (f-g) photographs of the device before and after implantation[50]. (h) RF WPT topology schematic, (i-j) photograph of implantable capsule RF antenna in minced pork[51].	9
2.3	(a) ME effect using a phase diagram. (b) Mechanism of ME device under AC magnetic field	12
2.4	(a) A NEMS scale microfabricated ME antenna operating at 2.5 GHz [25]. (b) A microfabricated circular disk shape SMR ME antenna that is capable of operating at 1.75 GHz and shows high gain [71]. (c) A $600 \times 500 \mu m^2$ MEA with IDT structure and array or magnetostrictive micromagnet which operates ≈ 1 GHz[70].	13
2.5	(a) A mm scale ME antenna operating at ≈ 24 kHz has the capability of long-range data communication up to a few Kms with a 100 Hz data rate[72]. (b) An mm-scale ME implantable bioelectronic device with a size of $\approx 95 mm^3$ for endovascular nerve stimulation applications that operates at 350 kHz [62]. (c) A magnetically powered ME stimulator that has an integrated CMOS PMC and neural stimulator circuit operating at 250 kHz and generates ME voltage up to 1 V [91]. (d) A near-field communication system using ME antenna and backscatter communication for implantable applications [93].	14
3.1	(a) FEM simulation showing the propagation of B_{AC} at 50 kHz through the six layers of brain tissues (scalp, skull bone, dura mater, CSF, and brain (comprising of the grey matter, white matter, etc.)) (b) Shows the variation of AC magnetic field when the distance between the coil and the scalp layer (D) increases from 0 to 30 mm. At D=30 mm and a tissue depth of 15 mm, the maximum B_{AC} available close to the brain interface is ≈ 0.45 mT.	19

3.2	Optimization of the device dimension and SAR with the variation in the operating frequency of the ME device. The left axis shows the variation of SAR (dB) with the increase in frequency (f). In the right axis, variation in the length of the ME device (L_{ME}) with frequency is illustrated. Inset shows the variation of SAR inside the human brain at the radiation frequency $f = 50$ kHz, 250 kHz, 1 MHz and 13.54 MHz.	21
3.3	(a) Schematic representation of the ME transducer showing the structural details. (b) Fabricated prototype of the ME transducer. (c) Direction of polarization of piezoelectric material (P) and direction of magnetization of magnetostrictive material (M).	22
3.4	(a-c) FEM simulation results of the ME antenna resonating at the first three resonant modes. (d) The variation of mode frequencies for the different resonant modes are compared using FEM simulation and analytical equation.	23
3.5	The equivalent circuit model of (a) the coil, (b) ME transducer. (c) The equivalent circuit simulation result of ME transducer, which indicates the variation of V_{ME} and output power with increasing H_{AC} at the f_{res} of 50 kHz. Inset shows the resonant mode of the ME device and the FEM simulation results. (d) Change in induced magnetic field strength due to angle of misalignment between Tx-Rx ($\varphi = 0^\circ - 90^\circ$).	25
4.1	(a) Fabrication flow sequence of the ME antenna. (b) Prototype image of the encapsulated ME device. (c) Cross sectional FESEM image of the ME heterostructure.	29
4.2	The AC Magnetic Flux Distribution considering longitudinal-transverse mode of ME samples with different aspect ratios (length/width):(a) AR=0.55, (b) AR=1, (c) AR=1.45, (d) AR=2. (e) Variation in Magnetic Field and ME Voltage with different ARs. (f) Output voltage response of the ME laminate at for different ARs.	32
4.3	Experimental setup for characterization of ME device, consisting of Lock-in Amplifier, DC current supply, DC electromagnet, Gauss meter, AC Amplifier, pair of Helmholtz coil, and the ME transducer.	33
4.4	(a) Measured ME voltage response of the ME transducer with varying H_{DC} without applying H_{AC} . (b) Represents the zoomed view of the M-H loop of the unannealed Metglas sample having dimension 3×4 mm ² . Measured ME voltage co-efficient (α_{ME}) of the ME transducer with varying H_{DC} and under the influence of $H_{AC} = 0.1$ Oe: (c) At $f = 50$ kHz, (d) At $f = 25$ kHz, (e) At $f = 10$ kHz, and (f) At $f = 1$ kHz.	34

4.5	(a) Demagnetization factor (N_D) variation as a function of variation in the length/width(L_M/W_M) ratio of the sample. For a L_M/W_M of 1.5, the operating frequency of the device is $\approx 50\text{kHz}$. (b-c) The magnetic force microscope (MFM) image of a Metglas sample shows the height and phase response, indicating the presence of magnetic domains (observable from the color contrasts).	35
4.6	(a) Measured ME voltage co-efficient, (b) Measured output voltage response of the ME transducer with different H_{AC} , 0.1 Oe, 0.2 Oe, 0.4 Oe, obtained from Oscilloscope. (c) Measured output power as a function of V_{ME} obtained from the ME transducer at 50kHz for a range of resistances $10\ \Omega$ - $20\ \text{M}\Omega$	35
4.7	(a) Fabricated prototype of planner coil with a number of turns of 50 and resonant frequency of $49.9\ \text{kHz}$. (b) Prototype of the Magnetoelectric antenna (c) The experimental setup for the magnetically tuned LF ME antenna includes a transmitting planner coil and a receiving ME antenna.	36
4.8	(a) Experimental measurements of the impedance and phase response of the antenna as a function of frequency. (b) Measured output RMS voltage PSD response of the device with various $H_{AC}= 0.1\ \text{Oe}$, $0.2\ \text{Oe}$, and $0.4\ \text{Oe}$, respectively, (c) The measured frequency spectrum of the ME antenna with varying distance (D) between Tx and Rx is obtained from the spectrum analyzer. (d) The limit of detection (LoD) of the magnetic field for the device.	37
4.9	(a) Schematic illustration of the near field radiation pattern measurement of the ME antenna at a $D=100\ \text{cm}$ in various configurations: (b) X-Y rotation; (c) Y-Z rotation.	39
4.10	Data transmission using digital modulation techniques: (a) A periodic frequency shift keying modulation at a planner orientation of 0° and 90° , (b) Similarly, an arbitrary amplitude shift keying modulation at the same conditions. (c) Signal-to-noise ratio (SNR) and bit error rate (BER) with increasing gap between Tx and Rx. (d) Measured spectrum of received signal and noise.	40
4.11	Temperature of the ME antenna after 90 mins. (a) Off State. (b) At a distance of 15 cm from the coil. (c) At a distance of 40 cm from the coil. (d) At a distance of 80 cm from the coil. (e) Temperature variation in ME antenna with time for different gap distance between coil and antenna ($D= 15\text{-}80\ \text{cm}$).	41
4.12	Temperature of the ME antenna at (a) $t=0$ mins, (b) $t=45$ mins and (c) 90 mins. Received data with ASK modulation rate of $2\ \text{kHz}$ after: (d) $t=45$ mins, (e) $t=90$ mins. Received data with ASK modulation rate of $5\ \text{kHz}$ after: (f) $t=45$ mins, $t=90$ mins.	42

5.1	(a) Schematic representation of Metglas/Ni/PVDF/Ni/Metglas thermocompression bonded ME antenna. (b) Fabricated prototype image. (c) Cross sectional FESEM of Metglas/Ni/PVDF. (d) EDX image of the cross section.	47
5.2	The eigen modes of mode of the M/N/P/N/M ME antenna (a) $f=13.25$ kHz, (b) $f=25.5$, (c) 48.15 kHz, (d) 65.7 kHz. (e) Magnetic field distribution in the ME antenna at resonant mode. (f) Maximum von mises stress developed at 48.15 kHz. (g) S_{11} analysis from the Butterworth-van Dyke (BVD) model as a function of frequency for different Q values. Inset shows the ME voltage from FEM simulation.	48
5.3	Fabrication sequence of the ME antenna with details of thermocompression bonding.	49
5.4	Fabrication sequence of the ME antenna with details of thermocompression bonding.	50
5.5	(a) Thermogravimetric analysis (TGA) of PVDF sample. (b) Differential scanning calorimetry (DSC) is conducted to analyze the thermal behavior of PVDF. (c) Fourier Transform Infrared Spectroscopy (FTIR), (d) X-ray Diffraction (XRD) analysis of PVDF samples	51
5.6	(a) Bond strength measurement of the thermocompression bonded ME antenna using Instron Instrument. (b) Crosssectional image of the ME device (c) Plot of bond strength for different samples annealed at different temperatures, (d) Maximum sheer bond strength for different samples.	52
5.7	(i) Cross-sectional FESEM images of the Metglas/PVDF/Metglas ME sample: (a) FESEM image (b-g) Elemental mapping of the different elements (EDX) image of cross-section of thermocompression bonded ME antenna, (h) EDX spectrum of the sample. (ii) Cross-sectional FESEM images of the optimized Metglas/Nickel/PVDF sample: (a) Cross-sectional EDX of image, (b-g) EDX image of cross-section, (h) Intensity of the mapping.	53
5.8	(a) ME voltage measurement for MPM device prepared with different annealing temperatures. (b) DC biasing field and α_{ME} measurement with different Ni layer thicknesses. (c) Characterization of magnetoelectric voltage coefficient of MPM and MNPNM devices. (d) Output voltage response of the ME antennas as a function of frequency for $H_{AC}=0.4$ Oe. (e) Limit of detection of the ME devices	55
5.9	Radiation pattern of the ME antenna: (a) Phi plane (out of the plane) condition (b) Theta plane (in-plane) condition.	56

5.10	(a) Characterization setup of thermocompression bonded ME antenna with in vitro chicken breast. (b) Received power spectrum with variation in gap between transmitter and ME antenna.(d-e) ASK modulation data at two different misaligned positions 0° and 90°	58
5.11	Thermal images of ME antenna at different conditions: (a) OFF State, (b) At D=40 cm and after 90 min, (c) At D= 10 cm and after 90 min, (d) Change in temperature in ME surface as a function of time. (e) ASK data received after 45 mins of continuous data transfer through the ME antenna, (f) Received ASK bits after 90 mins.	60
6.1	A conceptual illustration of the mechanism of the magnetically powered ME WPT system for futuristic implantable neural stimulation applications.	63
6.2	(a) Human tissue mimicking phantom characterization using VNA. (b-c) Measured relative permeability and conductivity of the phantom samples as a function of frequency.	65
6.3	(a) Schematic representation of the experimental setup for characterization of ME device, consisting of a lock-in amplifier, DC current supply, DC electromagnet, Gauss meter, AC amplifier, pair of Helmholtz coils, and the ME transducer. (b) ME device under testing with Helmholtz coil and electromagnets, and (c) shows the measurement setup including Tx planar coil and ME device. The inset shows the Parylene-encapsulated ME device and the optimized human tissue-mimicking phantoms.	67
6.4	The <i>in vitro</i> experimental result of ME device (a) ME voltage generated at resonance frequency of 50 kHz in air and 3-layered tissue mimicking phantom medium with different B_{AC} of 10 μ T, 20 μ T, and 40 μ T. (b) Output power variation with increase in ME voltage across the load resistance (10 Ω - 20 M Ω) in air and tissue-mimicking phantom medium.	68
6.5	(a) Near Field data transfer setup consisting of ME antenna Tx coil and tissue mimicking phantom. (b) Experimental measurement of power received by the ME antenna (MEA) and same-sized loop antenna (LA) in air and optimized muscle phantom as a function of variation in distance between the transmitting coil and Rx antenna (D). Received power spectrum in air and phantom at D=50 cm:c) ME antenna (MEA), (d) Loop antenna (LA).	69
6.6	(a) Block diagram of the proposed power management circuit for ME transducer-based WPT system design. (b) Prototype of the proposed magnetoelectric (ME) WPT system consisting of an ME device and power management circuit fabricated on a 2-layer PCB. The overall dimension of the ME WPT system is $12 \times 12 \text{ mm}^2$	69

6.7	Experimental results of the ME WPT system under applied magnetic field. (a) The output voltage of the PMC is recorded with different AC magnetic excitation [B_{AC} = 0.09 mT, 0.27 mT, and 0.5 mT], where the corresponding V_{ME} are 500 mV, 700 mV, and 1V, respectively. (b) Output voltage of the PMC is tested with increasing distance (D) between the customized transmitting coil and ME device from D=0 to 30 cm. (c) The settling time (T_s) , V_{DCP} , and V_{SYS} are measured while increasing the ME voltage.	70
6.8	Experimental result of load resistance variation analysis of the ME WPT system: (a) Output voltage across the SYS pin (V_{SYS}). (b) Output voltage of DCP (V_{DCP}).	71
6.9	(a) Schematic representation of ME WPT system consisting of ME device, Dickson Charge Pump (DCP), interface circuit, peak detector and switching circuit. (b) Top and bottom view of the fabricated 2-layer PCB prototype of the proposed PMC.	72
6.10	Experimental results of the ME WPT system (ME device and PMC) at different H_{AC} with corresponding V_{ME} : (a) 0.5 V, (b) 0.7 V.	73
6.11	Efficiency plot of the self-powered PMC prototype at 50 kHz frequency. The plot shows that when the V_{ME} increases from 2-5 V the conversion efficiency increases from 40% to 90%)	73
6.12	<i>In vitro</i> cytotoxicity of Parylene-C and ME device performed on WRL-68 cell line up to 7 days. (a) Alamar blue assay. (b) Fluorescence microscopy images (Scale bar: 100 μ m) for live-dead assay using dyes Calcein-AM and Propidium Iodide (PI).	74

ABBREVIATIONS

AC	Alternating Current
AFM	Atomic Force Microscope
AlN	Aluminum Nitride
APT	Acoustic Power Transfer
AR	Aspect Ratio
ASK	Amplitude Shift Keying
BER	Bit Error Rate
BVD	Butterworth van Dyke
CMOS	Complementary Metal Oxide Semiconductor
CNC	Computerized Numerical Control
CSF	Cerebrospinal Fluid
Cu	Copper
DCP	Dickson Charge Pump
DMF	Dimethylformamide
DMSO	Dimethyl Sulfoxide
DSC	Differential Scanning Calorimetry
DSO	Digital Storage Oscilloscope
EDX	Energy Dispersive X-ray Spectroscopy
EH	Energy Harvesting
EM	Electromagnetic
EMR	Electromechanical Resonance
FDTD	Finite Difference Time Domain
FEGaB	Iron Gallium Boron
FEM	Finite Element Method
FESEM	Field Emission Scanning Electron Microscopy
FSK	Frequency Shift Keying
FTIR	Fourier Transform Infrared Spectroscopy
Au	Gold
ICNIRP	International Commission on Non-Ionizing Radiation Protection
ICPT	Inductively Coupled Power Transfer
IEEE	Institute of Electrical and Electronics Engineers
IIoMT	Industrial Internet of Medical Things
IMD	Implantable Medical Devices
IoT	Internet of Things

ISM	Industrial, Scientific and Medical
JCPDS	Joint Committee on Powder Diffraction Standards
LF	Low Frequency
LoD	Limit-of-Detection
M/N/P/N/M	Metglas/Nickel/PVDF/Nickel/Metglas
MEA	Magnetoelectric Antenna
ME	Magnetoelectric
MEMS	Micro Electromechanical Systems
MPE	Maximum Permissible Exposure
NdFeB	Neodymium Iron Boron
NEMS	Nanoelectromechanical Systems
PCB	Printed Circuit Board
PCE	Power Conversion Efficiency
PMIC	Power Management Integrated Circuit
PMMA	Polymethyl Methacrylate
PMN-PT	Lead Magnesium Niobate-Lead Titanate
PMC	Power Management Circuits
PMU	Power Management Unit
PVDF	Polyvinylidene Fluoride
PZT	Lead Zirconate Titanate
Q-Factor	Quality Factor
RF	Radio Frequency
RPM	Revolutions Per Minute
SAR	Specific Absorption Rate
SEM	Scanning Electron Microscopy
Ag	Silver
SNR	Signal-to-Noise Ratio
TGA	Thermogravimetric Analysis
UHF	Ultra High Frequency
UTM	Ultrasonic Thickness Measurement
VHF	Very High Frequency
VNA	Vector Network Analyzer
VSD	Voltage Spectral Density
WPT	Wireless Power Transfer
XRD	X-ray Diffraction

NOTATIONS

V_{ME}	Magnetoelectric Voltage
H_{AC}	AC Magnetic Field (Oe)
H_{DC}	DC Bias Field (Oe)
B_{AC}	AC Magnetic Field (T)
I_{AC}	Alternating Current
σ_0	Tissue Conductivity
ρ_0	Tissue Density
E_0	Induced Electric Field Vector
f	Frequency
L_{ME}	Length of the ME Laminate
W_{ME}	Width of the ME Laminate
f_{res}	Resonant Frequency
c_{11}^H	Magnetoelastic Stiffness
$e_{11,M}$	Piezomagnetic Constant
t_M	Thickness of Metglas Layer
t_P	Thickness of PVDF Layer
k_e^2	Electromechanical coupling factor
$e_{31,P}$	Piezoelectric constant
c_{11}^E	Piezoelastic stiffness
C_0	Piezoelectric capacitance
$T_{1,M}$	longitudinal stress
$B_{1,M}$	magnetic flux density
$S_{11,M}$	Longitudinal Strain
$\mu_{11,S}$	permeability at steady strain
$T_{1,P}$	stress developed in the piezoelectric layer
$S_{11,P}$	strain developed in the piezoelectric layer
E_3	electric field in the 3-direction
D_3	dielectric displacement in the 3-direction
$e_{31,P}$	electric field constant in (3) direction
$\epsilon_{33,S}$	permittivity of the dielectric under constant strain
H_1	Magnetic field in 1 direction
R_L	Load Resistance
i	Integer Value
m_0	inertial mass
c_0	mechanical damping co-efficient
k_0	mechanical stiffness
x	displacement
F_M	Equivalent Voltage Generated by Induction of B_{AC}
R_M	Electrical equivalent of damping co-efficient
C_M	electrical equivalent mechanical stiffness

L_M	electrical equivalent of inertial mass
C_0	electrical equivalent of piezoelectric capacitance
$\phi_i(x)$	Spatial displacements for the i -th mode of resonance
$\delta\phi_i(x)$	Difference in $\phi_i(x)$ at the boundary condition
H_{DCOpt}	Optimum DC Magnetic Field
λ_s	Magnetostriction
α_{ME}	Magnetoelectric Voltage Coefficient
P_r	Received Power
S_{EMR}	Magnetic field sensitivity
N_{SD}	Power spectral density (PSD) of the noise
G_c	Gain of the LF ME antenna
Q_{Rd}	Radiation quality factor
Q_{Chu}	Chu's Limit of Quality factor
k	Wave Number
b	Radius of encircling sphere
ΔT	Temperature change
FOM	Figure of Merit
ϵ_r'	relative permittivity
A_{Rx}	Area of the receiving ME antenna
$D_1 - D_n$	Diode Names
$C_1 - C_n$	Capacitor Names
C_{SYS}	Capacitance across SYS Pin of ADP5092
$L_1 - L_n$	Inductor Names
P_L	Load Power
V_{BAT}	Voltage stored across BAT pin
C_p	Piezoelectric Capacitance
D	Distance between Tx coil and Rx ME antenna
V_{DCP}	Voltage Dickson Charge Pump
C_{BAT}	Capacitor across BAT pin
T_s	Settling time
$M_1 - M_n$	MOSFET names
V_{RECT}	Rectified Voltage
V_{REG}	Regulated Voltage
μ_r	Relative permeability
φ	Azimuthal angle
θ	Elevation angle
T_x	Transmitter
R_x	Receiver
S_{11}	Return Loss
α	Alpha phase of PVDF
β	Beta phase of PVDF
γ	Gamma phase of PVDF