Chipscale Piezo-Magnetostrictive Interfaces – A new simplified and microminiaturized telemetry paradigm for Medical Device Packages

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***Abstract*— Miniaturized 3D package integration with piezo-magnetostrictive telemetry is analyzed towards advanced wireless medical systems. The paper is composed of two aspects. The first is to analyze telemetry using a multiferroic or piezo-magnetostrictive power interface. Parametric analysis was performed to estimate the impact of material, geometry and input magnetic fields. The second aspect is the co-integration of devices and the sensor-processing-telemetry interfaces in 3D embedded fan-out packages for smallest form-factors. The telemetric interface and sensor devices are fan-out packaged in a flexible substrate with 3D interconnections, leading to miniaturized solutions. The sensor-communication and the power source flex layers are separately fabricated and connected with via-fill layers that are flex-compatible. Flex integration allows the devices to be functional on curved surfaces. Coupled magnetostrictive and piezoelectric films were considered to power devices in neural recording, neurostimulation and biophotonic systems. Power analysis for these specific scenarios is shown to illustrate the geometric compatibility with emerging needs.**

Keywords—Power, Multiferroic, piezoelectric, magnetostrictive, medical device, package integration

I. Introduction

Recent developments in sensing and electrode interfaces, single-chip signal processing and modulation, wireless power and communication, and fabrication technologies have enabled medical health-diagnostic and therapeutic systems with low power, miniaturization with unparalleled interfacing to biological tissue and superior performance in complex physiological environment. More importantly, understanding vital human organs' complex functionalities, has improved considerably, leading to major advances in neuroscience, neurotechnologies and cardiovascular therapies. Three classes of such systems are of interest, as highlighted in Fig. 1, which illustrates neural recording, neurostimulation and biophotonic health-diagnostics in three rows. The simplified system architectures for these applications are also highlighted in the figure.

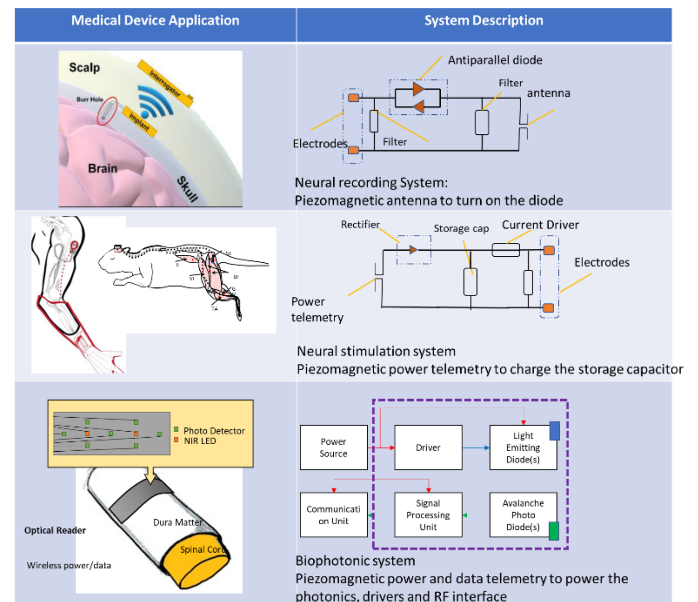

Fig. 1. Three classes of medical devices that require high-density power telemetry units: (a)Neural recording, (b)Neurostimulation, (c)Biophotonic sensors and communication

Currently, invasive implantable devices are implemented to record local field potentials (LFP), and action potentials. However, one of the significant problems in bioelectronics is providing power to miniaturized devices with adequate resolution. For instance, portable surface Electroencephalography (EEG) allows us to continuously monitor brain activity over a more extended period but suffers from low spatiotemporal resolution. fMRI (functional magnetic resonance imaging) has similar size and portability limitations as other imaging tools. Similar to EEG, they also

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face limitations from low spatiotemporal resolution. As a result, such devices are not convenient for neurotechnology advances that rely on monitoring single neuron activity. Wired or tethered invasive or implantable devices causes considerable damage to biological tissue and interruption in routine behavior of freely moving animals, thus compromising chronic in vivo experiments. It is a well-established fact that wire is a typical cause for failure in implantable devices. Percutaneous wires afflict infection and limit implantable devices to change position with the tissue. Eventually, it leads to foreign body response [1].

Development of wireless neurostimulation devices of mm-dimensions that function satisfactorily under scalp, skull, and tissue while animals involve in regular activities has been a perennial challenge in translating neurotechnologies to clinical practice. Implantable devices rely on the integration of various electrical components, such as, ICs (Integrated Circuit), off-chip components (capacitors, inductors, crystal oscillators), power telemetry (antenna, magnetic coils) with the required connectors and electrode arrays. Biophotonics with wireless telemetry will continue to dominate health-monitoring and diagnostics as they are widely established for applications ranging from oximetry, simple and complex biomarkers, along with spatiotemporal understanding of hemodynamics. All these medical systems need to meet the long-term safety and reliability constraints according to biological settings and safety concerns. Heterogeneous electronic components integration demands high interconnect density with low electrical parasitics for minimal signal losses. A system-level approach to electronics packaging is essential in order to accommodate multiple components and I/O interfacing. Another key requirement for such devices is miniaturization as it facilitates surgeons to implant and minimizes the possibility of inflammation or rejection by the tissues.

Packaging with 3D integration enables miniaturization, and also accommodates high-density circuits, flexibility, functionality, and eventually low cost. With die and package stacking, embedding dies is a major enabler technology in heterogeneous integration. Additionally, interconnect technology contributes to form-factor. Even though wire bonding technology is matured and cost-effective in the industry, it causes significant interconnection loss in RF/mm wave circuitry. Flip-chip offers small form-factor and high-density input/output connections with copper pillar interconnections. The key limitations arise from interconnection heights with high assembly temperatures for certain flexible substrates. Emerging fanout-embedded technology nullifies the need for wire bonding, and eliminates the high interconnection loss and decreases the size of footprint. Such fan-out technology is pursued with two manufacturing paths: 1) Fan-out wafer level packaging (FOWLP) 2) substrate- or panel-embedding. In this embedded packaging approach, interconnects from chip terminations to other system components are directly formed through substrate wiring processes rather than chip-last assembly technologies. Fan-out wafer level packaging has been in mobile and wireless basebands; however, now it is marching towards automotive and medical applications. Such in-package interconnection solutions between components result in low insertion loss, and reasonable return loss over frequency of interest, making them particularly suitable for RF sensors and passive neural recording.

As devices get miniaturized, particularly below 10 mm, providing adequate power becomes a hurdle. End-users typically utilize batteries to power up implanted devices; however, they limit miniaturization. Batteries demand replacement or suffer from charging issues, additionally increases the footprint of devices and weight. Two key technical challenges need to be addressed to realize power and data transfer in tiny <1 mm^3 nodes. Traditional approaches for wireless power are based on inductive link. With advances in wireless power transfer efficiency (PTE), inductively-coupled coils can potentially offer smaller footprints than batteries and less absorption compared to RF. However, as the receiver size is designed to mm-scale, the efficiency of this modality is constrained. This limits the power densities to less than 1 mW/mm^3.

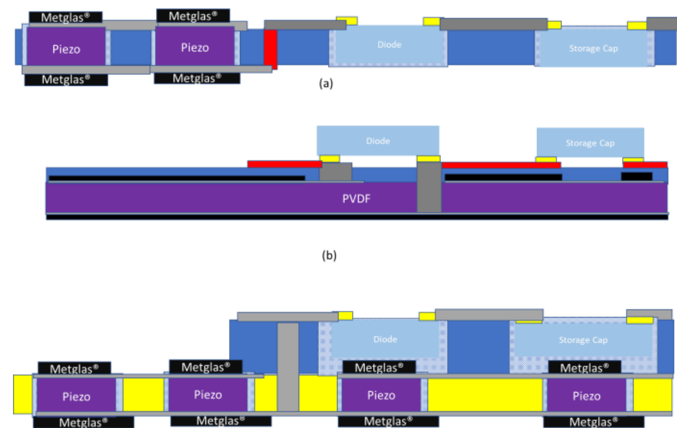


Fig. 2. a) 2- 2D Integration of piezomagnetic transducers with storage and sensing interfaces b) 3D Integration of piezomagnetic flexible films stacked under sensing and storage devices, (c) 3D Integration of piezomagnetic rigid tiles stacked under sensing and storage devices

RF (Radio Frequency) power transfer and harvesting is emerging as a key alternative to power medical devices [2]. This approach, on the other hand, requires an antenna with small features comparable to the electromagnetic wavelength. Therefore, mm-scale devices can transfer and communicate as the frequencies are in GHz bands, as reported in where a 2-mm size antenna resonating and operating at 34 GHz was designed and demonstrated. As a key milestone in this area, RF backscattering approach towards neural recording has been successfully demonstrated to realize zero-power health-monitoring systems [3, 4] by designing in ISM bands. However, EM (electromagnetic) loss in biological tissue is considerably high (100dB) at this frequency, and power transfer is inefficient [5]. Advanced RF designs also utilize the biological tissue as a part of the antenna structure, leading to size and efficiency benefits. In one such example, Ho et al., [6] have demonstrated that a mouse body can be considered as an electromagnetic resonant cavity in order to transfer energy efficiently to small implanted devices using RF power transfer. This method has also been shown to drive small LEDs for optogenetic stimulation [7]. In spite of such major

advances, RF power and data transfer are affected with power constraints or sensitivity issues towards the disturbance in the phase or distance between transmitter and receiver.

Wireless power transfer using ultrasound, as an alternative offers several orders of magnitude smaller power telemetry unit or date receiver (antenna) than EM antenna at specific frequencies because speed of the sound wave is much slower than that of EM waves. This technique, however, requires an intermediate electromagnetic transceiver implanted in the subdural region to communicate with external devices [8]. In addition, the issue of applying ultrasound for implantable devices is that loss in the skull alone (110dB) is significant. The maximum exposure limits of human tissue to ultrasonic power is inferior to magnetic fields, further limiting the amount of power that can be transmitted with this approach. In summary, all these approaches, face size, limitations in meeting the emerging needs in wireless sensing systems, compelling researchers to explore other power telemetry modes based on alternative multiphysics-based energy transduction modes.

Recently, a class of novel composite materials are being extensively studied. Broadly based on multiferroic transduction driven by multiphysics phenomena, these composites create unique opportunities to reduce the size of wireless transducers for power and data. The most prominent class is based on piezomagnetoelectric composite materials, leading to magnetoelectric (ME) transduction. ME is conversion of magnetic energy to electrical energy or vice versa [9]. Based on this concept, mechanical antennas, smaller than EM antennas, operating in kHz to GHz range were introduced in [10]. Microwave devices, such as inductors [11] and tunable filters [12] were engineered. This technology's advantage is that low-frequency antennas or energy harvesting can be realized in miniaturized form, virtually not vulnerable to misalignment issues, while also achieving high power density, and low tissue loss, making it stand out among other wireless technologies.

Multiferroic coupling with films of high piezoelectric and magnetostrictive coefficients that are coupled in acoustic resonance modes will create unique opportunities towards maximizing the received power or efficiency for various material stacks. Specifically, materials with such high piezoelectric and magnetostrictive coefficients need to be co-designed and processed to achieve the best multimode energy coupling. With optimal designs that utilize materials with best available properties, power densities of 1-20 mW/mm^3 were demonstrated while meeting the IEEE standards for Maximum Permissible Exposure [13]. Structural control at nanoscale is expected to further result in 4-5X further enhancement in properties. Referred to as giant piezoelectric or magnetostrictive materials, these are of major interest for transmitting power in mm^2 footprints. The power telemetry cell needs to be then integrated or co-packaged with wireless and sensing interfaces form compact 3D systems. Heterogeneous package integration with such new material architectures for superior performance is the key enabler for these systems.

Miniaturized power telemetry can be utilized in advanced packaging in multiple configurations, three of which are listed in Fig. 2. The process integration in 3D flex fan-out packages depends on the choice of the multiferroic layers in either flexible sheets (polyvinylidene fluoride, PVDF) or in rigid ceramic tiles (Lead zirconate titanate, PZT or Barium titanate). Appropriate via layers and stacking is illustrated to interconnect the telemetry layers to the power/ground layers. This paper analyzes advanced biomedical packages with multiferroic power telemetry. The first part of the paper discusses modeling and design of multiferroic transducers. Subsequent part discussed the fabrication approaches. The final part discusses the validation of transducers and initial package integration with new flex embedding and fan-out packaging technologies.

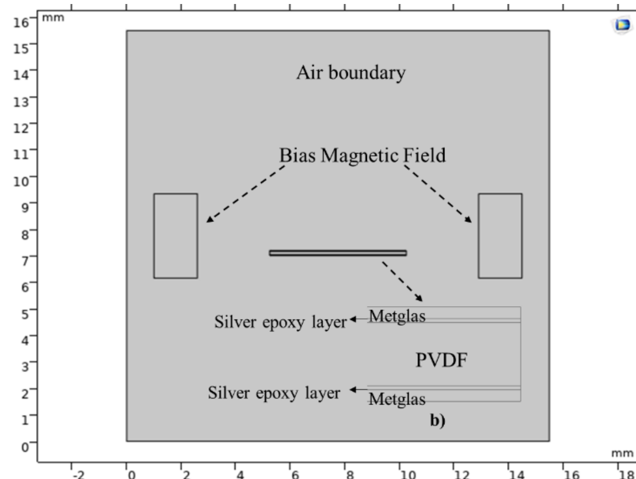


Fig. 3. a) 2-D design model of tri-layer multiferroic placed between two permanent magnets b) Close-up view of multiferroic tri-layer system

II. MODELING ANALYSIS

A. Multiferroic Model Set-Up

The primary goal of the multiphysics simulation of multiferroic transduction is to optimize geometric designs for a given set of properties to realize their effect on acoustic resonant frequency and maximum induced voltage. The analysis is also to perform parametric studies on the material combinations for highest power output for a given field and source-distance constraint. Piezoelectric and magnetostrictive multiphysics are coupled under the COMSOL^{TM} RF module in frequency domain to study the direct effect of the magnetoelectric phenomenon. The direct effect occurs when external magnetic fields influence the electrical polarization of the piezoelectric layer through induced strain. Maximum power transfer occurs at the acoustic resonance frequency harmonics.

Geometry of the piezo-magnetoelectric structure is optimized in 2-D COMSOL^{TM}, where the PVDF layer is sandwiched between two magnetostrictive Metglas® layers to resemble a tri-layer structure. In addition, the best outcome of the tri-layer is then compared to a bi-layer structure. The interfacial bodies are bonded by adhesive with thickness variation from 8 – 10 μm. Magnetostrictive and piezoelectric layers were assumed at 23 and 127 μm thickness, respectively.

Nonlinear isotropic relationship of the magnetostriction domain is chosen in order to define and sweep the magnetostriction coefficient across different values. Off-the-shelf Meglas® has magnetostriction coefficient of ~33 ppm. PVDF d_{31} coefficient is defined in strain-charge form, and chosen to be steable at – 33.8 pC/N (taken from the vendor data sheet). A 10 mm separation distance between the source and receiver is used as shown in Fig. 3. Air boundary is added to enclose the system to simulate magnetic fields under 100 mT magnetization. Electrical potential in the electric field domain is added to the interfacial layer of the top electrode and a ground layer is added to the bottom interfacial electrode layer. Frequency domain analysis is performed from 100 kHz to 400 kHz.

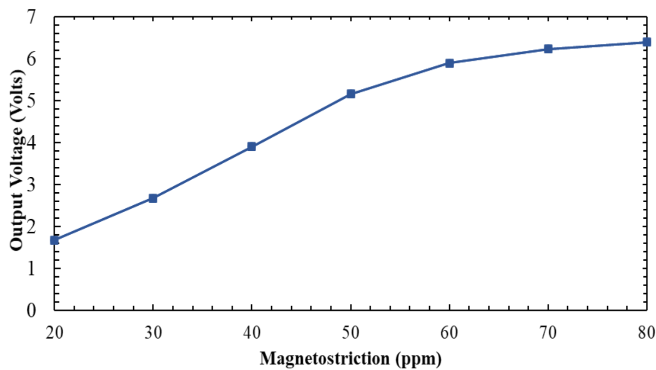
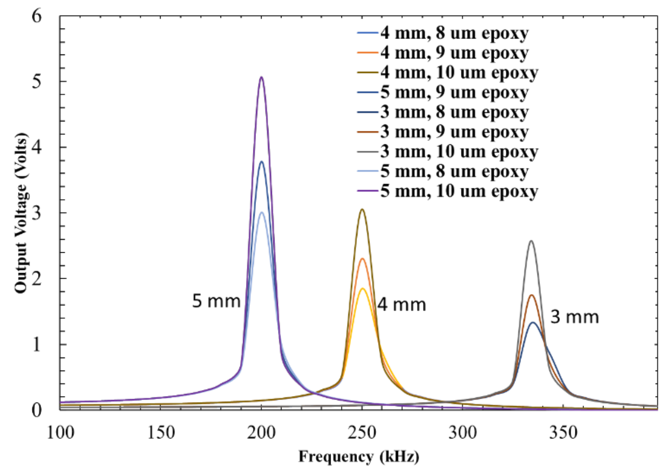


Fig. 4. Output voltage as function of magnetostriction coefficient sweep at the resonant frequency 200 kHz

B. Parametric Analysis with Properties, Geometry and Field-Strength

The induced voltage is a strong function of magnetostriction coefficient as higher magnetostriction results in higher output voltage, as shown in Fig. 4. Magnetostriction of 80 ppm in a 5 mm length tri-layer model shows an increase 282% when compared to 20 ppm magnetostrictive electrodes. The resonant frequency is an inverse function of the length. Larger length in the tri-layer model resonates at lower frequency with higher output voltage as displayed in Fig. 5. For a trilayer structure with 5 mm length, the resonant frequency is at 200 kHz. However, as length decreases, the resonant frequency increases. Layers with 4 mm and 3 mm showed corresponding resonant frequencies of 250 kHz and 334 kHz. The results also demonstrate the role of higher thickness of epoxy layer and larger stiffness in improving coupling to Metglas® and PVDF by increasing the strain mediation. The stress profile of the resonators was also noted at the resonant frequency of 200 kHz. The explicit compressive lateral strain represents the d31 response under applied magnetic field.

The bonding layer sitffness and thickness also affects the output voltage. A 5 mm length assembly with 10 µm epoxy interfacial bond results in a voltage of 5 V compared to 4V for an 8 µm thick of the same assembly, an increase of 27% in electrical potential. As a result, a 5 mm length tri-layer of Metglas®-PVDF-Metglas® with 10 µm thickness of adhesive bond shows the highest output voltage at lowest resonant frequency. A comparable bi-layer with only Metglas-PVDF is constructed to juxtapose the effects of resonant frequency and electrical potential with tri-layer Metglas®-PVDF-Metglas. A bi-layer with only one magnetostrictive electrode layer ensues a shift from the 200 kHz resonant frequency response with decrease of 20 kHz. However, the induced voltage in a bi-layer is limited to only 3.7V, which is approximately two times smaller than that of tri-layer since only one electrode strains the piezoelectric to influence its polarization.



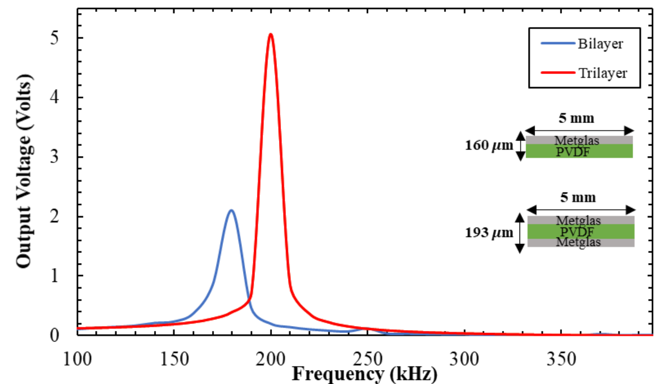


Fig. 5. Length parametric sweep from 3 – 5 mm and corresponding electrical potential. Comparative analysis of frequency responses between a tri-layer and a bi-layer multiferroic

Magnetic flux is a function of distance; if the magnetic source moves away, it decreases magnetic flux. Since piezo-magnetoelectric device is fed on magnetic field, so its induced voltage is also a function of distance. As the source moves away from our transducer, the induced voltage decreases. A 3-mm distance gives highest induced voltage of 20V; however, output voltage decreases as input magnetic flux decreases while distance increases, as shown in Fig. 6. Similarly, the bias magnetic field has direct effect on induced voltage, as observed in Fig. 7. As the magnetic field reduces, voltage reduces significantly. A voltage output of 5V is seen with bias magnetic field 100 mT.

The peak output power for a PVDF system with 5V peak is approximately estimated at a load that is equivalent to the transducer impedance of 80 kohm mm^2. This is roughly estimated to be 30 µW/mm^2 for a 127 µm thick PVDF film. For the multiferroic PZT/Galfenol system, detailed analysis was reported in [13], where the peak power is estimated as high as 21.6 mW/mm^3. These two systems will be considered in the subsequent power real estate analysis.

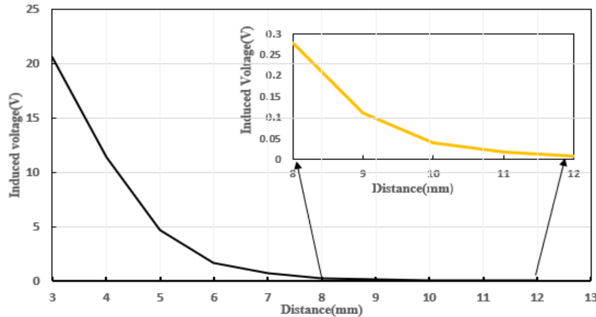


Fig. 6. Induced voltage vs Distance; millivolt range induced voltage

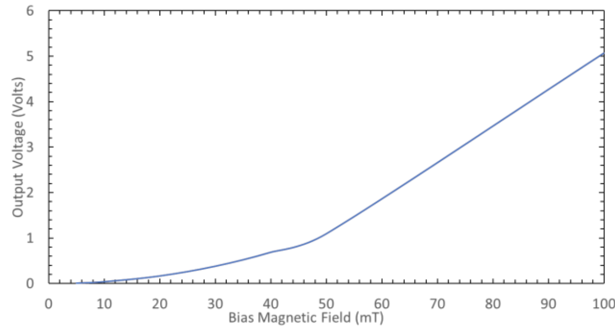


Fig. 7. Input bias magnetic field vs output(induced) voltage

III. PACKAGE INTEGRATION WITH PMD FILMS

Selection of magnetoelectric (ME) composites is based on the electrical and magnetic properties. Best ME response can be achieved with high magnetostriction coefficient, stiff interfacial bonding between the two phases (piezoelectric and magnetostrictive), high piezoelectric voltage constant, least piezoelectric loss, and low loss tangents [14]. The flex piezo-magnetostrictive system is fabricated as illustrated in the cross-section of Fig. 2b. More involved fabrication approaches are needed for the cavity-embedding of metal-ceramic rigid tiles in the flex carrier, as illustrated in Fig. 2a and 2c, and are not pursued here. In order to fabricate the structure in Fig. 2b, Metglas® (23 µm thick 2605SA1) and PVDF (127 µm thick, PolyK technologies, University Park, PA) materials were selected as magnetostrictive and piezoelectric layers. respectively. Both were diced into (5 mm x 5mm) size. Metglas® layer is bonded to both sides of the PVDF with epoxy (LOCTITE® EA M-21HP™) and hot-pressed to achieve minimum thickness of epoxy layer in between the multiferroic interface layers.

Flex packages are fabricated on thin and biocompatible LCP substrates. LCP layer is copper-patterned for high-density electrical circuitry. with low cost lithography. Laser vias were created with M-300 (Universal Laser system) laser at 5% Power and 20% Speed, to connect to the piezo-magnetostrictive layers. The drilled vias were filled with silver elastomer paste (CI 1036, Nagase Chemtex America Corporation, OH) to achieve low resistivity while also withstanding high curvature of strain without via cracking. Vacuum pressure gradients were applied to achieve the via fill using PTC VF-1000 Via Filler System (Pacific Trinetics Corporation, CA)The PMD power telemetry layer and the sensor signal processing layers were bonded together with adhesive gel (DOWSIL ™ 3-4207 Dielectric Tough Gel) and cured it at 120 °C for 5 minutes. With a pick-and-place tool, diode chip and storage capacitors are assembled onto the flexible LCP. The fabricated structures, illustrating the first 3D package integration of multiferroic telemetry layers with the sensing-processing-communication layer, are shown in Fig. 8.

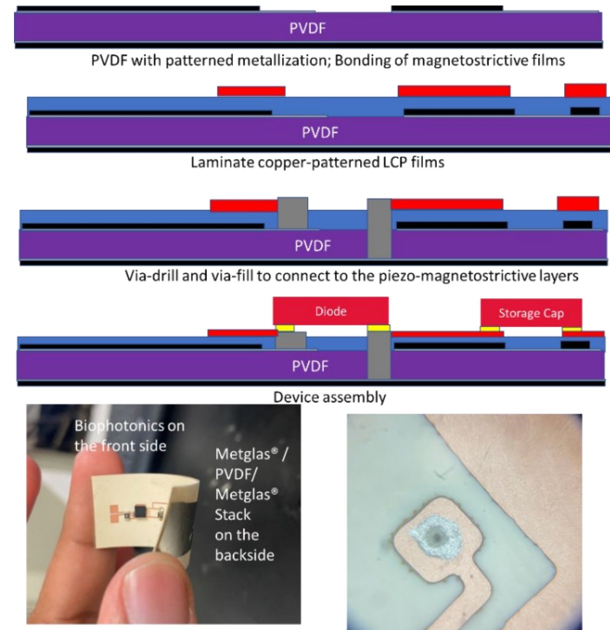


Fig. 8. a)Fabrication process flow for flex integration piezo-magnetostrictive telemetry and device layers, b) fabricated test-structure along with the zoomed view of a flex via-fill

IV. POWER ANALYSIS

In this section, we will discuss power requirement and PMD power supply analysis in neural recording, neurostimulation and biophotonic systems. These systems have been illustrated in Fig. 2. In the first configuration, ceramic piezoelectrics with magnetostrictive films as mm-scale tiles are embedded into flexible substrate cores and 2D fan-out connected to other system components. In the second configuration, the PVDF substrate sandwich with the magnetostrictive films can itself act as a flexible substrate. The functional sensor and communication module are then laminated onto this stack. In the third configuration, the piezomagnetostrictive tiles are embedded in the cavity of a flexible substrate, with the sensor-communication module laminated onto it to form the 3D interconnections.

The first application aims for passive neural recording and to retrieve data using multiferroic or piezo-magnetostrictive interfaces. RF backscattering technique is considered for this need as it can potentially achieve near-zero DC power [3]. However, the large antenna size has been a major impediment with RF telemetry. For passive neural recording system, an antiparallel diode is used as the sub-harmonic mixer for frequency modulation. Signal input is fed through the mixer to communicate to an external reader through backscattering. Two piezo-magnetic antennas are connected to either ends of a subharmonic mixer. The key power budget in this system arises from the power to turn on the diode. A state-of-the-art single diode or antiparallel diode requires -5 to -8 dBm to be turned on. In other words, a power of 300-500 μW will be adequate to operate the diode. Based on the projected power densities (30 μW/mm^2), PVDF/Metglas® piezo-magnetostrictive interfaces of 15-20 mm^2 are needed. The data telemetry units are within these dimensions, leading to 5x5 mm recording systems. With Galfenol/PZT tiles, the dimensions could be reduced to sub-mm.

Electrical stimulation therapies rely on current pulses that inject pulsed charges into the target neural interfaces. A neural stimulator typically requires biphasic waveforms at 1-5V, with peak currents of 5 ~5 mA at 10% duty cycle, assuming 100 Hz stimulation frequency at 1 millisecond pulse intervals. This mounts to 2.5 mW [15]. Target voltages are guided by the electrochemical limits at the electrode/tissue interface from the irreversible Faradaic reactions during the charge injection. The target voltage can be directly set by the transducer design parameters such as thickness, to be at the required 1-5 V, as noted in [15]. Assuming the flex PVDF/Metglas® system, leads to a size of 80 mm^2. For the more efficiency Galfenol/PZT system, a single tile of 2 mm^2 can address the power needs.

Biophotonic systems comprise of light sources such as LEDs, detectors that are typically in the form of photodiodes, drivers, and communication chips. The power consumption from light sources are estimated to be 10 mW, assuming a 20% duty cycle. The detector consumes relatively low power of less than 1 mW. The driver and transceiver consume a significant part of the power. With standard commercial transceivers, this power is estimated to be close to 50 mW, while emerging topologies can substantially reduce this power. In order to achieve such power levels with telemetry, the tile approach with PMD ceramic-magnetostrictove stacks is considered to be more suitable. Assuming a density of 0.2 mW/mm^2 for a 200 micron stack, a 3x4 array of tiles 5x5 mm (5 mW per tile) cavity-embedded tiles can adequately power such a system.

V. Conclusions

Wireless power transfer with piezo-magnetostrictive transduction was analyzed for implantable medical devices. Parametric simulations were performed to predict the output power with different transducer geometries and field input conditions. The high multiferroic coefficients lead to power densities superior to inductive link as the receiver dimensions approach mm dimensions. This advantage gives unique ability to power a range of future biomedical systems. Copackaging of the telemetry devices with functional chips on a flexible substrate where the substrate itself acts as both a power sources and also provides the mechanical support, is demonstrated for the first time. Initial power analysis shows that such a system can low-power neural recording systems and also certain neurostimulation applications. Cavity-embedded metal-ceramic tiles are more suitable for higher densities to power biophotonic systems with higher power needs.

Acknowledgment

Technical and process support from Advanced Materials Engineering Research Institute (AMERI) during the fabrication of test structures at FIU is duly acknowledged.

References

[1] R. Biran, D. C. Martin, and P. A. Tresco, "The brain tissue response to implanted silicon microelectrode arrays is increased when the device is tethered to the skull," *Journal of Biomedical Materials Research Part A,* vol. 82, pp. 169-178, 2007.

[2] K. Fotopoulou and B. W. Flynn, "Wireless power transfer in loosely coupled links: Coil misalignment model," *IEEE Transactions on magnetics,* vol. 47, pp. 416-430, 2010.

[3] A. Kiourti, C. W. Lee, J. Chae, and J. L. Volakis, "A wireless fully passive neural recording device for unobtrusive neuropotential monitoring," *IEEE Transactions on Biomedical Engineering,* vol. 63, pp. 131-137, 2015.

[4] S. Y. B. Sayeed, S. B. Venkatakrishnan, M. M. Monshi, A. Abdulhameed, J. L. Volakis, and P. Raj, "3D Heterogeneous and Flexible Package Integration for Zero-Power Wireless Neural Recording," in *2020 IEEE 70th Electronic Components and Technology Conference (ECTC)*, 2020, pp. 1003-1009.

[5] M. Zaeimbashi, H. Lin, Z. Wang, H. Chen, S. Emam, Y. Gao*, et al.*, "NanoNeuroRFID: A low loss brain implantable device based on magnetoelectric antenna," in *2018 IEEE International Microwave Biomedical Conference (IMBioC)*, 2018, pp. 205-207.

[6] J. S. Ho, Y. Tanabe, S. M. Iyer, A. J. Christensen, L. Grosenick, K. Deisseroth*, et al.*, "Self-tracking energy transfer for neural stimulation in untethered mice," *Physical Review Applied,* vol. 4, p. 024001, 2015.

[7] K. L. Montgomery, A. J. Yeh, J. S. Ho, V. Tsao, S. M. Iyer, L. Grosenick*, et al.*, "Wirelessly powered, fully internal optogenetics for brain, spinal and peripheral circuits in mice," *Nature methods,* vol. 12, pp. 969-974, 2015.

[8] D. Seo, J. M. Carmena, J. M. Rabaey, E. Alon, and M. M. Maharbiz, "Neural dust: An ultrasonic, low power solution for chronic brain-machine interfaces," *arXiv preprint arXiv:1307.2196,* 2013.

[9] M. Fiebig, "Revival of the magnetoelectric effect," *Journal of physics D: applied physics,* vol. 38, p. R123, 2005.

[10] T. Nan, H. Lin, Y. Gao, A. Matyushov, G. Yu, H. Chen*, et al.*, "Acoustically actuated ultra-compact NEMS magnetoelectric antennas," *Nature communications,* vol. 8, pp. 1-8, 2017.

[11] Y. Gao, S. Z. Zardareh, X. Yang, T. X. Nan, Z. Y. Zhou, M. Onabajo*, et al.*, "Significantly Enhanced Inductance and Quality Factor of GHz Integrated Magnetic Solenoid Inductors With FeGaB/${\rm Al} _ {2}{\rm O} _ {3} $ Multilayer Films," *IEEE Transactions on Electron Devices,* vol. 61, pp. 1470-1476, 2014.

[12] H. Lin, J. Wu, X. Yang, Z. Hu, T. Nan, S. Emori*, et al.*, "Integrated non-reciprocal dual H-and E-field tunable bandpass filter with ultra-wideband isolation," in *2015 IEEE MTT-S International Microwave Symposium*, 2015, pp. 1-4.

[13] T. Rupp, B. D. Truong, S. Williams, and S. Roundy, "Magnetoelectric transducer designs for use as wireless power receivers in wearable and implantable applications," *Materials,* vol. 12, p. 512, 2019.

[14] H. Palneedi, V. Annapureddy, S. Priya, and J. Ryu, "Status and perspectives of multiferroic magnetoelectric composite materials and applications," in *Actuators*, 2016, p. 9.

[15] A. Singer, S. Dutta, E. Lewis, Z. Chen, J. C. Chen, N. Verma*, et al.*, "Magnetoelectric materials for miniature, wireless neural stimulation at therapeutic frequencies," *Neuron,* vol. 107, pp. 631-643. e5, 2020.