Multi-Resonator Wireless Inductive Power Link for Wearables on the 2D Surface and Implants in 3D Space of the Human Body

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Abstract

This paper presents a novel resonance-based, adaptable, and flexible inductive wireless power transmission (WPT) link for powering implantable and wearable devices throughout the human body. The proposed design provides a comprehensive solution for wirelessly delivering power, sub-micro to hundreds of milliwatts, to deep-tissue implantable devices (3D space of human body) and surface-level wearable devices (2D surface of human skin) safely and seamlessly. The link comprises a belt-fitted transmitter (Belt-Tx) coil equipped with a power amplifier (PA) and a data demodulator unit, two resonator clusters (to cover upper-body and lower-body), and a receiver (Rx) unit that consists of receiver load and resonator coils, rectifier, microcontroller, and data modulator units for implementing a closed-loop power control (CLPC) mechanism. All coils are tuned at 13.56 MHz, FCC-approved ISM band. Novel customizable configurations of resonators in the clusters, parallel for implantable devices and cross-parallel for wearable devices and vertically oriented implants, ensure uniform power delivered to the load, PDL, enabling natural Tx power localization toward the Rx unit. The proposed design is modeled, simulated, and optimized using ANSYS HFSS software. The Specific Absorption Rate (SAR) is calculated under 1.5 W/kg, indicating the design's safety for the human body. The proposed link is implemented, and its performance is characterized. For both the parallel cluster (implant) and cross-parallel cluster (wearable) scenarios, the measured results indicate: 1) an upper-body PDL exceeding 350 mW with a Power Transfer Efficiency (PTE) reaching 25%, and 2) a lower-body PDL surpassing 360 mW with a PTE of up to 20%, while covering up to 92% of the human body.
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Abstract—This paper presents a novel resonance-based, adaptable, and flexible inductive wireless power transmission (WPT) link for powering implantable and wearable devices throughout the human body. The proposed design provides a comprehensive solution for wirelessly delivering power, sub-micro to hundreds of milliwatts, to deep-tissue implantable devices (3D space of human body) and surface-level wearable devices (2D surface of human skin) safely and seamlessly. The link comprises a belt-fitted transmitter (Belt-Tx) coil equipped with a power amplifier (PA) and a data demodulator unit, two resonator clusters (to cover upper-body and lower-body), and a receiver (Rx) unit that consists of receiver load and resonant coils, rectifier, microcontroller, and data modulator units for implementing a closed-loop power control (CLPC) mechanism. All coils are tuned at 13.56 MHz, FCC-approved ISM band. Novel customizable configurations of resonators in the clusters, parallel for implantable devices and cross-parallel for wearable devices and vertically oriented implants, ensure uniform power delivered to the load, PDL, enabling natural Tx power localization toward the Rx unit. The proposed design is modeled, simulated, and optimized using ANSYS HFSS software. The Specific Absorption Rate (SAR) is calculated under 1.5 W/kg, indicating the design's safety for the human body. The proposed link is implemented, and its performance is characterized. For both the parallel cluster (implant) and cross-parallel cluster (wearable) scenarios, the measured results indicate: 1) an upper-body PDL exceeding 350 mW with a Power Transfer Efficiency (PTE) reaching 25%, and 2) a lower-body PDL surpassing 360 mW with a PTE of up to 20%, while covering up to 92% of the human body.

Index Terms—Wireless Power Transmission, Multi-Coil Inductive Link, Parallel and Cross-parallel Resonators, Implant Medical Devices (IMDs), Wearable Medical Devices (WMDs).

I. INTRODUCTION

The growing adoption of implantable medical devices (IMDs) and wearable devices has transformed the healthcare sector from hospital-centric to individual-centric systems. These devices play a crucial role in precision medicine by enabling targeted and personalized medical treatments, ultimately leading to improved patient outcomes and a higher quality of life [1]-[4]. As these devices continue to shrink in size and increase in functionality, they facilitate continuous real-time monitoring of various physiological functions. This ability allows for patient-specific treatments, early detection of medical conditions, proactive decision-making for medical interventions, and precise drug therapies with reduced side effects [5]-[12].

The implantable and wearable medical devices encompass a wide range of applications, which can be categorized as follows: 1) brain-related diseases and neurodegenerative diseases (NDDs); brain-machine interfaces (BMIs), deep brain stimulation (DBS) electrodes and implants, neuro-stimulator, neuro-prosthetics, optogenetic implants, and brain-wearables for monitoring NDD-related motor impairments [1], [13], [14], cochlear implant, retinal implant [1], [3], 2) cardiovascular diseases: pacemakers, defibrillators, ventricular assist devices (VADs) [1], [15], stents, cardiac loop recorders [16], 3) tissue repair and regeneration: photo-bio-modulation (PBM) and low-level light/laser therapy (LLLT) [17]-[19], 4) gastrointestinal diseases and urology: wireless capsule endoscopy [5], [6], insulin pump, implanted bladder stimulator [20], 5) monitoring multiple physiological parameters in real-time: implantable and wearable biosensors [1], smart electronic contact lens [21], fitness tracker, smart textiles with integration of wearable medical devices (WMDs) [22], and 6) others: orthopedic implants, foot drop implants, bone growth stimulator [1], [20], floating implants capable of electromyography (EMG) sensing [23], portable pulse monitoring [24], doppler ultrasound blood flowmeter used for a smart graft [25] and more.

An emerging application of WPT involves its use in implantable and wearable devices, facilitating the utilization of (1) light for neural stimulation, such as optogenetics, and (2) light for its therapeutic effects. The optogenetics approach combines genetics and optics to control cell activity, offering insight into studying NDDs and neuropsychiatric disorders [26]-[30]. Light can activate or inhibit light-responsive proteins such as opsins [26], allowing researchers to control the activity of specific cells in a living organism. Optogenetics, employed in studying neural activity, muscle contraction, and gene expression, also utilizes implantable devices to stimulate specific nerves, like the vagus nerve, for appetite regulation [27]-[28]. WPT technology-assisted optogenetic devices help avoid passing fibers and wires through tissue layers [29]-[30].

Other therapeutic light application includes Low-Level Light Therapy (LLLT), Photo-biomodulation (PBM),
Intravascular Photo-biomodulation (iPBM), or Intravascular Laser Irradiation of Blood Therapy (ILIB, using laser light through optical fibers inserted into veins). These are all proven low/non-invasive treatment methods with potential benefits such as wound healing, pain relief, anti-inflammatory responses, anti-edema effects, nerve regrowth, tissue repair, and regeneration [31]-[33]. They enhance blood circulation and optimize blood cell functionality, ensuring beneficial effects in reducing inflammation, enhancing antioxidants, and improving hemodynamic parameters. Efficient and precise light delivery is crucial for achieving expected clinical outcomes in PBM devices [31], [19]. While delivering light deep into the body is challenging [34], WPT links can efficiently deliver energy to implants, producing sufficient light density for internal organs and expanding therapeutic applications for humans. This capability opens new possibilities by enabling the assessment of invasive ILIB for various applications, including cardiovascular disease, ensuring precise and efficient dose delivery, and offering promising prospects for innovative treatment approaches in future healthcare technologies.

However, medical devices relying on built-in batteries present drawbacks, necessitating periodic surgeries for replacements, with associated risks and financial burdens [1]-[11]. To address these challenges, various wirelessly powered technologies, including inductively coupled WPT, capacitively coupled short-ranged WPT, and newer options like ultrasonic, mid-field, and far-field coupled WPT, are widely used for implants and wearables, although challenges persist [1]-[2], [13]-[15], [20]-[23], [35]. Inductive power links face issues like suboptimal performance during misalignments, requiring innovative design solutions [1]-[2], [35]. Near-field capacitive coupling encounters obstacles, including heightened sensitivity to Tx-Rx separation and increased power fluctuations [23], [35]. Ultrasonic energy transfer faces challenges due to varying organ densities, soft tissue attenuation, and sensitivity to distance, potentially impacting overall system performance [1], [21], [35]. Mid-field WPT, with lower delivered power levels, demands a comprehensive safety analysis [1], [35]. Far-field EM Coupling, extensively studied for long-range power transmission, has limited research in biomedical implant applications, constrained by minimal received power and reduced efficiency [1], [35].

Magnetic resonance-based WPT (MR-WPT) technology, incorporating multi-coil (resonator) with various configurations, has emerged as an efficient and widely adopted solution for powering implants and wearable devices. Recent studies have aimed to expand the transfer distance, extend the effective 3D coverage area, adhere to safety standards, and enhance adaptability [36]-[42]. Different multi-coil Tx array configurations, namely switching approaches, floating resonators, parallel resonators, overlapping designs, two sandwiched structures, and so on, have been proposed to increase the effective active area where that implant or wearable devices can receive sufficient power [6], [36]-[42]. In floating, switching, and overlapping states, a disconnected array may lead to multiple resonance frequency peaks [38]-[40], [41], [43], [44]. The parallel array configuration resolves this by confining energy in a closed channel, ensuring continuous propagation, delivering a consistent voltage to each array element from the PA, generating higher current at the array's coil element adjacent to the receiver, enabling automatic positioning, and optimizing power consumption [38]-[40], [41].

The parallel-connected multi-resonator Tx array designs have shown excellent performance in wirelessly powering (1) wearable headstages in animal caged (for long-term in vivo experiments with freely moving small animals), (2) IMDs, and (3) multiple smartphones simultaneously, using a single power source [40]-[43], [45]. Such configuration provides great uniformity of power transfer efficiency (PTE) and power delivered to the load (PDL), as well as enabling natural power localization property, only activating the resonator near the Rx unit with no additional circuitry [39]-[41]. However, the parallel approach with horizontally oriented Tx resonators does not efficiently deliver power to the vertically oriented implants or wearables.

This paper presents a smart, efficient, range-adaptive, and flexible human-specific cluster-based WPT inductive link. This novel WPT link offers reliable, uniform, distance-insensitive, and transcutaneous wireless power delivery solutions for a wide range of implantable and wearable medical devices. Configurable resonator cluster setups, i.e., parallel connections for implantable devices and cross-parallel connections for wearables, facilitate seamless and effective power supply to both types of medical devices. Section II discusses the design review, circuit model analysis, and simulation results for developing a comprehensive design rule and optimization process for the proposed resonance-based inductive link. Section III incorporates the experimental results, followed by Section IV’s discussion and Section V’s conclusion.

II. DESIGN REVIEW

Fig. 1a presents the conceptual design of the proposed customizable resonator-based WPT inductive link, which is tailored to the anatomical dimensions of the human body. It enables wireless power transfer from the belt-fitted transmitter unit to the implant or wearable device, where the Rx unit can be located anywhere inside or at the surface of the human body. Fig. 1b represents the equivalent circuit of the proposed flexible
and wearable transmitter unit, including Class E PA and control unit circuitry, Belt-Tx coil, and two clusters of Tx resonators, (1) upper body cluster, C1, consisting of \( L_{C1,1} \) to \( L_{C1,8} \), and (2) lower body cluster, C2, comprising of \( L_{C2,9} \) to \( L_{C2,13} \). Fig. 1c illustrates the conceptual parallel and cross-parallel configurations of the resonators in the clusters, controlling the orientation of the generated electromagnetic fields. The parallel configuration is effective for implants placed inside a human body, but weak coupling occurs for vertically positioned Rx units (such as wearable devices) due to zero flux conditions. To address this issue, a cross-parallel configuration is proposed to achieve optimal PTE and PDL for wearable device/s by shaping the electromagnetic field toward the body’s surface, offering better efficacy than the parallel configuration.

Fig. 2a shows the entire system’s equivalent circuit block diagram, indicating the closed-loop power control (CLPC) mechanism, in which the load shift keying (LSK) back telemetry (BT) mechanism is implemented by switching the Rx resonator On and Off. CLPC mechanism is crucial for WPT links for preventing a substantial variation of the power delivered to the load, enabling adaptation to the changes in load impedance, the gap between Rx and Tx coils, and angular and lateral misalignments between Rx and Tx coils [7]. The Rx unit is comprised of a Rx resonator \( (L_{R2}) \), a Rx coil \( (L_L) \), a rectifier and regulator units, and a microcontroller and modulator for sensing the received power level and generating BT data for the CLPC mechanism. The BT data is received at the Tx external unit via sensing the Tx coil’s current and demodulating the BT signals to adjust the power level by processing the data in the utilized microcontroller (nRF52840).

The entire system (resonators) is tuned to carry the energy wirelessly at the frequency of 13.56 MHz, safe for biological tissues according to FCC-approved ISM (industrial, scientific, and medical) band and safety standards, with negligible power absorption [36]-[41]. The number of resonators, \( n \), within a single cluster is determined by the coverage requirements of the intended application area and the separation between the Rx unit and the Tx unit.

A. Equivalent Circuit Model Analysis

To analyze and optimize the proposed system, cluster’s equivalent inductance is calculated as the first step using the equivalent circuit model of C1 presented in Fig. 2b considering \( n \) resonators in parallel. In this model, the individual coils’ self-inductances are denoted as \( L_{C1,1} \), \( L_{C1,2} \), ..., \( L_{C1,n} \) and mutual inductance between adjacent coil elements are represented as \( M_{C1-1,C1-2} \), \( M_{C1-2,C1-3} \), ..., \( M_{C1-n-1,C1-n} \). For the sake of simplicity, mutual inductance values are assumed to be equal, \( M_{C1-1,C1-2} = M_{C1-2,C1-3} = ... = M_{C1-n-1,C1-n} = M \). Although coil’s parasitic series resistance, \( R \), is deliberately excluded from the circuit model for simplification, the \( R \) is considered in simulation models and numerical analyses. Based on this model, we calculate the voltage, \( V \), and total current, \( I \), of the paralleled resonators (for the cross parallel resonators, the polarity of mutual inductances can be reversed) to find the equivalent inductance of the cluster, resonance frequency, optimizing the resonators and the entire system’s efficiency. The total current, \( I \), in the cluster is the sum of all the resonators’ currents, expressed by (1).

\[
I = I_1 + I_2 + ... + I_n = \sum_{i=1}^{n} I_i \tag{1}
\]

We use mesh circuit analysis technique to find currents of the resonators, developed in (2) [44], [46]-[47].

\[
\begin{bmatrix}
   sL_{C1,1} & -sM_{12} & 0 & 0 & 0 \n
   -sM_{21} & sL_{C1,2} & -sM_{23} & 0 & 0 \n
   0 & 0 & 0 & ... & 0 \n
   0 & 0 & 0 & ... & 0 \\
\end{bmatrix}
\begin{bmatrix}
   I_1 \\
   I_2 \\
   ... \\
   I_{n-1} \\
   I_n \\
\end{bmatrix} =
\begin{bmatrix}
   V \\
   V \\
   ... \\
   V \\
\end{bmatrix}
\tag{2}
\]

where, \( s = j\omega \) and \( s[M]_{n \times n} \times [I]_{n \times 1} = [V]_{n \times 1} \)

\[
[I] = \frac{1}{s} \times [M]^{-1} \times [V] \tag{3}
\]

Considering (2), the current of any branch \( I_i \) \((i = 1, 2, ... , n)\) can be calculated by (4).

\[
I_i = \frac{1}{s} \times V \sum_{j=1}^{n} M_{ij}^{-1} \tag{4}
\]

Therefore, the total equivalent inductance, \( L_{eq} \), for \( n \) number of mutually coupled inductors in parallel for a cluster, C1, can be calculated by (5).

\[
L_{eq} = \frac{1}{s} \times \sum_{j=1}^{n} M_{ij}^{-1} \tag{5}
\]

Using (4) and (5), we can write,

\[
L_{eq} = \frac{1}{\sum_{i=1}^{n} \sum_{j=1}^{n} M_{ij}^{-1}} \tag{6}
\]
Finally, the resonance frequency, \( f_{r,C1} \), of the cluster, C1, is calculated by (7).

\[
f_{r,C1} = \frac{1}{2\pi \sqrt{(L_{C1eq} + C_{C1eq} \cdot sL_{C1eq})}}
\]  

(7)

where, \( C_{C1eq} \) is the equivalent capacitance value for a cluster, \( n \times C \). A variable capacitor is added in parallel with \( C_{C1eq} \) to fine-tune the system’s resonance frequency, enabling precise adjustment of the clusters’ resonance frequency to 13.56 MHz. To validate the calculation of \( L_{C1eq} \) for clusters with two, three, four, and five mutually coupled identical resonators in parallel, we have used MATLAB to find \( L_{C1eq} \) values using (6) and compared them with the results found in LTSpice. The matched results presented in Table I indicate the accuracy of (6) in calculating \( L_{C1eq} \) for different numbers of resonators in a cluster. Therefore, we can model down the proposed multi-resonator system to a simple four-coil inductive link to analyze and optimize it. Using Kirchhoff’s Voltage Law, Mesh analysis, the inductive link of Fig. 2a is modeled by it in (8), considering a voltage source, \( V_s \) (representing the PA model), source resistance \( R_s \), and the entire load resistance model, \( R_L \), [48]. The currents of the loop (\( I_n, n \in \{1,2,3,4\} \)) are calculated by (8) using MATLAB to find the four-coil link’s \( PTE_{4-coil} \) [48],[49].

\[
\begin{bmatrix}
Z_{11} & Z_{12} & 0 & 0 \\
Z_{12} & Z_{22} & Z_{23} & 0 \\
0 & Z_{23} & Z_{33} & Z_{34} \\
0 & 0 & Z_{34} & Z_{44}
\end{bmatrix}
\begin{bmatrix}
I_1 \\
I_2 \\
I_3 \\
I_4
\end{bmatrix}
=
\begin{bmatrix}
V_s \\
0 \\
0 \\
0
\end{bmatrix}
\]  

(8)

where, \( Z_{11} = sL_T + \frac{1}{sC_T} + R_S \), \( Z_{22} = sL_{C1eq} + \frac{1}{sC_{C1eq}} \), \( Z_{33} = sL_{RL} + \frac{1}{sC_{RL}} \), \( Z_{44} = sL_L \), \( \frac{1}{sC_{C1eq}} \), \( \frac{1}{sC_{RL}} \), \( R_L, Z_{12} = -sM_{Lp,L_{C1eq}}, Z_{23} = -sM_{Lp,L_{RL}}, Z_{34} = -sM_{Lp,L_{RL}} \), \( M_{Lp,L_{C1eq}} \), \( M_{Lp,L_{RL}} \).

The PTE of the proposed cluster-based 4-coil link can be calculated by finding the ratio of the delivered power, \( P_L \), to the source power, \( P_S \), also defined as the squared of transmission coefficient, \( S_{21} \), using (9) and (10).

\[
S_{21}^2 = \frac{P_L}{P_S} = \frac{|I_{4}|^2 |R_K|}{|I_{4}|^2 |R_S|}
\]  

(9)

\[
PTE_{4-coil} = S_{21}^2 \times 100 \%
\]  

(10)

B. Inductive Link Optimization by ANSYS HFSS Modeling and Simulation

The design of the Tx unit can be independent of the Rx unit design when the Rx unit’s dimensions are much smaller than the Tx coil or Tx resonator(s) [41]. We have used the same approach to design and optimize the proposed inductive link. The proposed comprehensive optimization guideline is presented in Fig. 3, targeting to 1) maximize the PTE, 2) achieve uniform PTE/PDL for both the 2D surface and 3D space of the body, 3) minimize the number of resonators in the cluster (ensuring optimum PTE and PDL), and 4) minimize the Specific Absorption Rate, SAR. The design flowchart of Fig. 3 indicates a very general input parameter, design requirements for covering the entire body, while the design approach is adaptive to any other types of requirements such as targeting different parts of a human body (e.g., stomach area, legs, head, etc.) or different size of human bodies.

Our initial optimization criteria encompass PDL > 50 mW, PTE > 10%, frequency (\( f \)) = 13.56 MHz, SAR < 1.6 W/kg, Tx to Rx gap coverage of 0-100cm (uniformity of PTE & PDL), Back Telemetry (BT) for power adjustment, Rx dimension of 2cm x 6cm (diameter). Customizability of cluster design (adjustable/adaptable design), human/user height and waist size.
whether it covers the upper or lower-body devices. To find the optimal distance between the Tx coil and the nearest Tx resonator and therefore the optimum coupling between them \((k_{T,R_1,1} \text{ or } k_{T,R_2,9})\), we calculate the Tx coil’s reflection coefficients, \(S_{11}\), to minimize it. The \(S_{11}\) can be calculated by (11),

\[
S_{11} = \frac{Z_{11} - R_S}{Z_{11} + R_S},
\]

where \(Z_{11}\) and \(R_S\) are Tx coil’s impedance and source resistance, respectively, while the \(Z_{11}\) is calculated when the equivalent inductance of C1’s resonators, \(L_{C1eq}\), from (6) is coupled with the Tx coil, \(L_T\). To find the optimum gap and \(k_{T,C1,1}\), (minimum \(S_{11}\)), the gap between the \(L_T\) and \(L_{C1,1}\) (closest resonator to the Tx coil when C1 is active) is swept according to the optimization flowchart. The diameters (and shapes) of the flexible Tx coil and the resonators are determined by the application requirements and target body area where the devices (implants/ wearables) are located. For different numbers of resonators in a cluster, the optimum gap between the Tx and the first resonator might be different, and the proposed optimization flowchart must be followed to determine the optimum location of the resonator for achieving the best possible PTE and most uniform PDL. The gaps between the resonators (and, therefore, the number of resonators) in parallel or cross-parallel configurations are determined accordingly to achieve the best possible PTE and the most uniform PDL in the targeted volume and surface of the body. The same optimization procedure (optimizing \(S_{22}\) for Rx coil and Rx resonator) is used to optimize the Rx unit’s coils by determining the optimum coils’ turns, dimensions, and the gap between them.

The proposed link is modeled by ANSYS HFSS software (Cecil Township, PA) for supporting the optimization flowchart of Fig. 3 by using a standard realistic human body model and modeling the Tx unit’s coils (Tx coil and Tx resonators in parallel/cross-parallel configurations) and Rx coils. Thin wires are used to implement the connection between the Tx resonators. A flat copper foil with a width of 13 mm and a thickness of 0.1 mm is designed in HFSS for the resonators.

Figs. 4a, 4b, and 4c present the HFSS simulation results of the proposed inductive link, indicating the coupling coefficients between the Rx coil and two paralleled resonators (at upper body, arms, and legs, respectively) for different gaps between the resonators while sweeping the location of the Rx coil.
between the resonators. These simulations are conducted to find the optimum gap between the resonators. The inset pictures of these figures show the close-up of the developed HFSS model of the inductive link and the human body. Optimal resonator separations for the torso, forearm, and leg areas are identified as 200 mm, 100 mm, and 100 mm, respectively.

Fig. 5a shows the difference between the parallel and cross-parallel resonators’ coupling coefficients with a vertically oriented Rx coil while sweeping the location of the Rx coil from the lower resonator to the upper resonator. When the wearable Rx unit is placed vertically, the coupling coefficient approaches zero in the parallel configuration. This phenomenon arises from electromagnetic field cancellation at the midpoint between the resonators, leading to weak mutual coupling (k) between the vertically oriented Rx and the two paralleled resonators. An innovative cross-parallel configuration is proposed to address this weak coupling to redirect the electromagnetic fields toward the Rx. The cross-parallel connection demonstrates superior performance in wirelessly powering vertically oriented wearables and implants.

The proposed design offers flexible choices between two clusters: simultaneous operation of C1 and C2 or alternate operation to power the Rx unit at an application-specific target location. The active cluster activates nearby resonators of the target Rx location using the same L_T. The clusters can establish connections in three ways without causing any disruption to the wireless system: 1) parallel, 2) cross-parallel, or 3) one cluster as parallel and the other as cross-parallel. These adaptable and interchangeable designs enable efficient and uniform power delivery to devices targeting diverse areas and applications across the entire human body. This capability relies on the automatic, convenient, and reliable tracking of the Rx unit at any location using the same L_T via a single or multiple resonators within an active cluster.

C. Power Transfer Efficiency, PTE, Calculation

Fig. 6 shows simulated S_{21}’s as a function of frequency, utilizing parallel-configured clusters for horizontally oriented Rx unit (implant) and cross-parallel-configured clusters for vertically oriented Rx unit (wearable), positioned on the subject’s chest and leg areas. The peak magnitudes for S_{21}’s are approximately -6.42dB at the working frequency of 13.56 MHz. PTEs calculation in HFSS, with input and output port resistances = 50 Ω, allows us to evaluate the functionality and effectiveness of the active upper body or lower body clusters while transferring efficient power to implants or wearables, specifically catering to a designated application using both configurations.

D. Specific Absorption Rate, SAR

Fig. 7 illustrates the average SAR simulation results of the proposed design. The design includes a Tx-Belt coil, 13 resonators, an Rx unit with two coils, and the human body HFSS model. The simulated SAR is calculated under 1.5 W/kg at both the upper body (Fig. 7a) and the lower body (Fig. 7b), below the standard limit of 1.6 W/kg. In this simulation, the system’s input power has been set to 1 W, and the average SAR is calculated at the power carrier resonance frequency when the Rx unit is located at the center of the body (worst-case scenario). In addition, the SAR plot shows the natural power localization mechanism of the proposed design, in which the only nearest Tx resonators to the Rx unit mainly contribute to delivering power to the Rx unit.

III. Implementation and Experimental Results

A. Prototype Implementation and Specifications of the Coils

A mannequin male transparent human body model (with a height of 190 cm and chest circumference of 84 cm, made of
transparent plastic) is utilized to assemble the proposed flexible cluster-based WPT prototype on the mannequin. The implemented prototype, shown in Fig. 8a, comprises cluster C1 with eight resonators and cluster C2 with five resonators to provide distance-insensitive powering of implants/wearyable devices on the upper and lower human body sides, respectively. Fig. 8a inset indicates the designed Class E PA to generate the 13.56 MHz power signal to energize the \( L_T \). The Rx unit, depicted in Fig. 8b inset, consists of \( L_3 \) and \( L_4 \) and can function as either an implanted-Rx or wearable-Rx, connected to a 216 \( \Omega \) resistor as the load to measure PDL and also to a red LED for visualizing the delivered power to the Rx unit. The specifications of the inductors \( (L_1, L_{2,1}, L_3, \text{and } L_4) \) and the resonance capacitors of the implemented prototype of Fig. 8 are presented in Table II. One-turn coils are formed using a half-way-folded 3M-RF-EMI shielding conductive and flexible copper tape/foil for the Belt-Tx coil and cluster resonators, excluding the hand resonators. The hand resonators are made with the same foil but with two-turn coils while maintaining a minimum gap of 0.5 mm between each turn. The optimum numbers of the turns are determined by following the optimization process of Fig. 3. The copper foil has a thickness of 0.1 mm. It is covered with Kapton adhesive tape on both sides to 1) provide excellent electrical isolation, 2) protect the copper foil against moisture, and 3) make it more robust against mechanical stress while it is completely flexible.

### Table II. Specifications of the Implemented Coils at 13.56 MHz

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Belt-Tx Coil ( (L_3) )</th>
<th>Cluster 1, C1: Res. Coils ( (L_{C1}, L_{C1}) )</th>
<th>Cluster 2, C2: Res. Coils ( (L_{C2}, L_{C2}, L_{C2}) )</th>
<th>Rx Res. Coil ( (L_3) ) / Rx Load Coil ( (L_3) )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Inductance, ( L ) (( \mu )H)</td>
<td>0.89</td>
<td>0.79/0.9/0.99</td>
<td>0.97/0.54/0.94</td>
<td>1.76/1.54</td>
</tr>
<tr>
<td>Quality Factor, ( Q )</td>
<td>66</td>
<td>66/88/71/79/90/90/92/95</td>
<td>65/67/67/90</td>
<td>125/90</td>
</tr>
<tr>
<td>Outer Dia., do (cm) (Oval shape)</td>
<td>28x18</td>
<td>36x18</td>
<td>31x21.3/34x21</td>
<td>31x26</td>
</tr>
<tr>
<td>Inner Dia., di (mm)</td>
<td>do-0.1</td>
<td>do-0.1</td>
<td>do-0.1</td>
<td>do-0.4</td>
</tr>
<tr>
<td>Copper Thickness ( 900\mu m )</td>
<td>Foil</td>
<td>Foil</td>
<td>Foil</td>
<td>Foil</td>
</tr>
<tr>
<td>Turns (N)</td>
<td>1</td>
<td>1/1/1/1</td>
<td>1</td>
<td>1/1/1/1/1</td>
</tr>
<tr>
<td>In-between Optimum Distance, OD (cm)</td>
<td>D1, (L3-L3,1)=8</td>
<td>D1, (L3-L3,1)=15</td>
<td>D1, (L3-L3,1)=15</td>
<td>D1, (L3-L3,1)=15</td>
</tr>
<tr>
<td></td>
<td>D2, (L3-L3,2)=8</td>
<td>D2, (L3-L3,2)=13</td>
<td>D2, (L3-L3,2)=13</td>
<td>D2, (L3-L3,2)=13</td>
</tr>
<tr>
<td></td>
<td></td>
<td>=14, D2</td>
<td></td>
<td>=14, D2</td>
</tr>
<tr>
<td>Coil length (cm)</td>
<td>( L_T = 85 )</td>
<td>( L_{C1} = 79 )</td>
<td>( L_{C2} = 39 )</td>
<td>( L_{C2} = 39 )</td>
</tr>
<tr>
<td>Capacitors’ Values (pF)</td>
<td>-</td>
<td>( C_{C1,4}=355/18 )</td>
<td>( C_{C2,3}=240 )</td>
<td>( C_{C2,1}=310 )</td>
</tr>
</tbody>
</table>

Fig. 9. The measured PDLs and PTEs for Rx unit’s different locations along the Z-axis involving varying numbers of parallel configured resonators: five, four, and thirteen for the lower body, torso, and the forearm areas, respectively.

VNA ports were 50 \( \Omega \). The PTE is calculated using (10) by measuring the \( S_{21} \) where the VNA’s two ports are connected to the Tx and Rx load coils. The PDL is calculated using (12) by measuring the pick-to-pick voltage (\( V_{p-p} \)) across the Rx load, \( R_L = 216 \Omega \), using an Oscilloscope.

\[
PDL = \left( \frac{V_{p-p}}{2} \right)^2 / (2 \times R_L)
\]  

(12)

Fig. 9 illustrates the PTEs and PDLs as functions of the Rx unit’s location, where the Rx location is swept along the Z-axis for different scenarios involving 1) a lower body cluster of five resonators in parallel to cover lower body area, 2) an upper body cluster of four resonators in parallel to cover the torso, and 3) an upper body cluster of eight resonators in parallel to cover the entire upper body areas, including the forearm area. The reference location to move the Rx along the Z-axis is the center of the \( L_T \) coil (position zero of the coordinate system), in which negative Z indicates the location in the lower body. For this measurement, the Rx unit position is swept along the \( +Z/-Z \) axis at the center of resonators, where it is shifted along X axis to sweep the Rx unit location along Z in the forearm resonators. Fig. 9 indicates the location of the resonators, too, where the forearm resonators are away from the Belt-Tx coil by up to 90 cm and can even be located further out on the body without any PTE reductions (covering the entire human body).

In Fig. 9, the measured PDLs vary from 187 mW to 550 mW around the lower body, 161 mW to 523 mW around the torso.
and 202 mW to 339 mW around the forearm areas, with average measured PTEs at around 20%, 20%, and 12%, respectively, within the enclosed 3D space. The covered transmission distances for these scenarios are 35 cm, 50 cm, and 90 cm along the Z-axis.

To compare the performances of the parallel (P) and cross-parallel (CP) resonator configurations for horizontally (H) and vertically (V) oriented Rx unit cases, 1) Fig. 10a shows the three scenarios of CP-V, P-V, and P-H, and 2) Fig. 10b presents the PDLs of these scenarios while varying the Rx position along the Z axis within lower body cluster area. The measured results of Fig. 10b confirm the effectiveness of the cross-parallel configuration of resonators in powering a vertically oriented Rx unit compared to parallel configuration, while the parallel design is efficient in powering a horizontally oriented Rx unit. Together, the proposed parallel and cross-parallel resonators wirelessly power wearable and implantable devices in the 2D surface and 3D space of the human body. Although the presented prototype has 13 resonators and covers a large percentage of the body, it can be optimized to focus on any specific body area (depending on the target application), involving fewer resonators.

Fig. 11 illustrates the uniformity of measured PDLs for various positions of the Rx unit. Fig. 11a shows the Rx unit’s horizontal sweeping area in the upper body area (X-Y plane). Fig. 11b presents the measured PDLs of the proposed design while sweeping the Rx unit’s location in the area highlighted in Fig. 11a, passing the forearms and torso (at Z=16 cm). These results indicate the uniformity of the PDLs in a 3D plot for resonators in parallel configuration while the Rx unit is oriented horizontally. For a vertically oriented Rx unit, Fig. 11c indicates the measured PDLs in a 3D plot for the proposed cross-parallel design while sweeping the location of the Rx unit vertically over the surface of the body and torso area. Similarly, Fig. 11d shows a cross-section (X-Y plane) in the lower body (legs) area where the Rx unit location is swept across for measuring PDL. The measured PDL results are presented in Fig. 11e for the parallel configuration while the Rx unit is oriented and moved horizontally. Fig. 11f indicates the...
measured PDL in a 3D plot at the lower body area with resonators in cross-parallel configuration while sweeping the location of the Rx unit vertically over the surface of the leg.

To conduct an in vitro experiment, ground beef is used to fill the space between hand resonators. The in vitro test setup and the prototype are shown in Fig. 12a. Fig. 12b shows the close-up view of the hand with resonators and meat, wherein the Rx unit is positioned within the meat. The PDLs are measured at 339 mW for parallel configuration and 266 mW for cross-parallel configuration, exhibiting no notable distinctions compared to the setup devoid of meat.

C. Back Telemetry and Closed-Loop Power Control (CLPC) Mechanism

The location and orientation of an implantable or wearable Rx unit can vary depending on the Tx resonators’ positions. In addition, the Tx resonators and Rx unit might experience some movement due to the subject’s body movement, leading to coupling variations between the Belt-Tx, cluster resonators, and Rx coil unit. Furthermore, the power consumption of the Rx unit tends to vary based on the specific power requirements of the target application. Therefore, it is necessary to continuously adjust the delivered power to the load by developing a closed-loop power control (CLPC) mechanism. Such a mechanism needs feedback information from the load (PDL level). A back telemetry mechanism is needed to transfer data from the Rx unit to the Tx unit. We have proposed using the power link and load-shift-key (LSK) to implement the back telemetry mechanism.

In the proposed design, unlike the conventional techniques of modulating the load coil (with a switching transistor), we have modulated the Rx resonator (as shown in Fig. 2a) using a transistor to influence the current of the Belt-Tx coil efficiently. Modulating the Rx resonator instead of the Rx load coil provides a more reliable data link (due to a stronger impedance reflection toward the Tx unit), especially when the coupling between the Rx unit and the Tx unit is very weak. In this design, when the Rx unit receives more power than it needs, the data will be sent to the primary Tx unit to reduce the Tx power level, and when the received power level is not at the required level, it does not send data. When the Tx unit does not receive feedback data, it starts increasing the Tx power level till it receives feedback data to avoid increasing the Tx power further.

The proposed CLPC mechanism is implemented by 1) the Rx unit’s microcontroller (with embedded analog to digital converter, ADC) that monitors the rectifier’s output voltage and generate pulses (back telemetry data) for the Rx modulator unit, 2) modulator unit that can drive the switch/transistor connected to the Rx resonator, 3) current sensor in Tx unit to measure the current of Belt-Tx coil to sense the back telemetry data, 4) demodulator unit to amplify the back telemetry signal and generate digital data for the microcontroller unit, and 5) microcontroller unit that controls the level of the Tx power output via its driver (Fig. 2a). The CLPC mechanism ensures that the Rx unit consistently receives a constant power level, irrespective of any changes in the gaps and alignments between Tx and Rx units.

Using the implemented inductive link, the back telemetry data transmission mechanism is verified by sending pulses from the Rx unit to the Tx unit through the weak coupling between them. Fig. 13 shows the measured back telemetry data, including the input data at the Rx unit, the signal across the Rx resonator, the signal across the Belt-Tx coil, and recovered data.

<table>
<thead>
<tr>
<th>Ref./Year</th>
<th>WPT Approach</th>
<th>Application</th>
<th>Uniformity (%)</th>
<th>Rx Volume (cm³)</th>
<th>Frequency (MHz)</th>
<th>Tx-Rx Gap (mm)</th>
<th>Coverage Area (%)</th>
<th>PTE (%)</th>
<th>PDL (mW)</th>
<th>FoM</th>
</tr>
</thead>
<tbody>
<tr>
<td>[6] 2018</td>
<td>Magnetic Resonance (MR)</td>
<td>Capsule Endoscope</td>
<td>94% in H-Field</td>
<td>1.7280</td>
<td>0.25</td>
<td>±130</td>
<td>30%, Y/Z=±14/±8.85 cm</td>
<td>5.4</td>
<td>570</td>
<td>0.09</td>
</tr>
<tr>
<td>[9] 2021</td>
<td>MR, Segmented Helmholtz Coil</td>
<td>Doppler System</td>
<td>~94% in M-Field</td>
<td>58.90</td>
<td>6.78</td>
<td>60–140</td>
<td>20%</td>
<td>81–86</td>
<td>~5000</td>
<td>0.29</td>
</tr>
<tr>
<td>[25] 2014</td>
<td>MR, Segmented Helmholtz Coil</td>
<td>Cardiac Pacemaker</td>
<td>~70% in M-Field</td>
<td>0.7696</td>
<td>-</td>
<td>-</td>
<td>25%</td>
<td>-</td>
<td>60 mA</td>
<td>-</td>
</tr>
<tr>
<td>[45] 2019</td>
<td>MR, Sandwiched</td>
<td>Cardiac Pacemaker</td>
<td>~70% in M-Field</td>
<td>9.8174</td>
<td>0.16</td>
<td>0–100</td>
<td>25%</td>
<td>88</td>
<td>~5000</td>
<td>1.13</td>
</tr>
<tr>
<td>[50] 2015</td>
<td>MR, Circular Array Coil</td>
<td>Implantable Hearing Aid</td>
<td>-</td>
<td>0.6157</td>
<td>0.1342</td>
<td>40</td>
<td>15%</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>[51] 2018</td>
<td>Magnetolectric (ME)</td>
<td>Neural Implant</td>
<td>-</td>
<td>0.82</td>
<td>0.25</td>
<td>30</td>
<td>Spot Powering</td>
<td>0.064</td>
<td>0.024</td>
<td>-</td>
</tr>
<tr>
<td>[42] 2023</td>
<td>MR, Textile Coil Array</td>
<td>Wearable Multi-sensors</td>
<td>-</td>
<td>1.2566</td>
<td>13.56</td>
<td>300</td>
<td>50%</td>
<td>11–29</td>
<td>32–122</td>
<td>1.27</td>
</tr>
<tr>
<td>[52] 2010</td>
<td>MR, Helmholtz Coil</td>
<td>Capsule Endoscope</td>
<td>~60% in M-Field</td>
<td>0.48</td>
<td>1</td>
<td>~150</td>
<td>30%</td>
<td>-</td>
<td>~330</td>
<td>-</td>
</tr>
<tr>
<td>[54] 2017</td>
<td>MR, Segmented Tx Coil</td>
<td>Heart Pump</td>
<td>~75% in H-Field</td>
<td>23.8267</td>
<td>6.78</td>
<td>100</td>
<td>25%</td>
<td>54</td>
<td>19700</td>
<td>1.12</td>
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<tr>
<td>[55] 2019</td>
<td>Microwave, Rectenna</td>
<td>Cardiac Pacemaker</td>
<td>-</td>
<td>80.352</td>
<td>954</td>
<td>110</td>
<td>Spot Powering</td>
<td>65</td>
<td>10</td>
<td>-</td>
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<tr>
<td>[56] 2023</td>
<td>Ultrasound, External Phased Array</td>
<td>Gastric Applications</td>
<td>-</td>
<td>0.0175</td>
<td>0.94</td>
<td>30 at 90°</td>
<td>40%</td>
<td>-</td>
<td>0.53</td>
<td>-</td>
</tr>
<tr>
<td>This Work</td>
<td>MR, Parallel 8 Res.</td>
<td>Implants</td>
<td>88% in PDL</td>
<td>4.9480</td>
<td>13.56</td>
<td>900</td>
<td>90%</td>
<td>12</td>
<td>339</td>
<td>6.12</td>
</tr>
<tr>
<td></td>
<td>MR, Cross-parallel 5 Res.</td>
<td>Wearables</td>
<td>85% in PDL</td>
<td>4.9480</td>
<td>13.56</td>
<td>350</td>
<td>&gt;92%</td>
<td>18</td>
<td>350</td>
<td>1.32</td>
</tr>
</tbody>
</table>

\[ \text{Uniformity} = 1 - \frac{\text{Max} - \text{Min}}{\text{Max}} \times 100\% , \text{Coverage Area} = \frac{\text{Total Referred Human Body Volume}}{\text{WPT Covered Area}} \times 100\% , \text{FoM} = \frac{\text{Coverage Area} \times \text{PTE} \times \text{PDL} (\text{mW}) \times \text{Max. Range} (\text{Tx-Rx Gap})^2 (\text{cm})}{\text{Rx Volume} (\text{cm}^3)} \]
from the signal recorded at the Tx unit. The measured results, the data stream transmitted from the Rx unit, confirm the ability to stabilize the power delivered to the load using feedback data. Note that all measured PDL data are presented in Figs. 9, 10, and 11 were obtained without activating the CLPC mechanism to evaluate the pure PDL ability of the proposed design.

IV. DISCUSSION

A. Comparison with Other Works

Table III presents a concise overview and comparison of essential attributes between the proposed cluster-based WPT link design and prior state-of-the-art studies for powering implants and wearables across a large portion of the human body. The proposed approach outperforms in terms of extending coverage area and Figure of Merit, FoM. Moreover, the proposed approach demonstrates the following key advantages: 1) adaptability to various target locations within and outside the human body’s skin surface, 2) robustness against receiver misalignment and body movement, 3) provision of uniform PDL in 2D/3D target area, 4) natural power localization near the receiver, minimizing the need for complex circuitry to detect target locations, 5) consistent PTE, and 6) capability to simultaneously power multiple implants or wearables in different locations.

B. Customization of the Proposed Adaptive Design for Different Applications

In the proposed design, the resonators grouped under two clusters have both standby and direct mode capability to power up medical devices, whether located in the outer body surface or precisely injected deep inside. All resonators are flexible to conform to the curvature of the body and can be expanded for ease of use (flexibility and enlarging), relaxed wearing, and smart fit according to different body shapes and sizes.

Considering the adaptability nature of the proposed design, the number of resonators and coverage area of the system can be modified based on the optimization flowchart of Fig. 3 for different applications. For example, it can be optimized for only powering a bladder implant with only one Tx resonator or a wearable LLLT bandage on foot. The proposed cluster-based WPT link design can also be adapted for powering brain implants by incorporating an additional head-sized free-floating resonator coupled with Cluster C1. The proposed technique can simultaneously power multiple brain implants and other implants or wearables (or artificial organs) in other parts of the human body and facilitate data exchange between them.

V. CONCLUSION

A comprehensive and range-adaptive multi-resonator WPT link is proposed to wirelessly power implantable devices in 3D space and wearable devices over the 2D surface of the human body. The proposed link consists of a belt transmitter (Belt-Tx) coil at the waist connected with a power amplifier and a control unit, two upper and lower body clusters of resonators in parallel (for implants) and cross-parallel (for wearables) configurations, and a receiver (Rx) unit. The power link is equipped with a closed-loop power control (CLPC) mechanism using a load modulation back telemetry technique. The proposed link ensures uniform and range-insensitive PDL and PTE without the need for precise alignment of the Rx unit, enabling natural Tx power localization toward the Rx unit. We have modeled the proposed link on the human body model with HFSS software to simulate and optimize it, considering the safety study, where the SAR is calculated at 1.5 W/kg. The prototype of the design is implemented and characterized. In both the parallel cluster (implant) and the cross-parallel cluster (wearable) scenarios, the measured data suggests: 1) an upper-body PDL of more than 350 mW and a PTE of up to 25%, and 2) a lower-body PDL over 360 mW with a PTE as high as 20%, covering as much as 92% of the human body’s volume and surface.

REFERENCES


