



OPEN In situ electric field dosimetry analysis for powerline frequency peripheral nerve magnetic stimulation

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Humans are exposed to environmental 60 Hz magnetic fields (MFs), inducing in our body electric fields (EFs) and currents, potentially stimulating the peripheral nervous system (PNS). Uncertainties exist regarding the 60 Hz MF PNS stimulation threshold. The spatially extended nonlinear node model (SENN) is used to help define international MF exposure guidelines and standards protecting workers and the general public. However, other models exist, particularly the McIntyre–Richardson–Grill (MRG) model, the new gold standard for electrostimulation. This study aims (1) to model a new extremely low frequency MF exposure system for the human leg and (2) to investigate the in situ EFs generated by the system at 60 Hz at the skin level and in the nerves of the leg using a realistic human body model with both the SENN and the MRG models. A Helmholtz like-coil system was designed to generate in situ EFs sufficient for nerve stimulation, modeled using Biot–Savart and Faraday laws. Sim4Life simulations assessed the induced EFs at skin and nerve levels using a detailed human body model and two nerve excitation frameworks: the SENN and MRG models. High EF intensities were observed in four sensory and sensory-motor nerves, with MRG-derived thresholds lower than SENN-derived thresholds. Results also highlight the significance of nerve orientation in EF induction. This study emphasizes the critical role of comprehensive modeling for the design and validation of MF exposure systems and underscores the need for experimental data to refine models, standards, and guidelines.

Keywords Dosimetry, Extremely low-frequency magnetic fields, MRG, Powerline frequency, SENN

Throughout the twentieth century, human exposure to electromagnetic fields (EMF) steadily increased due to the growing prevalence of man-made electromagnetic sources in the environment. The advancement of technology and the rising electricity demand resulted in the widespread presence of artificial EMFs in both homes and workplaces. Extremely low frequency (ELF < 300 Hz) magnetic fields (MFs), are especially common due to the generation, distribution, and use of alternating current at power lines operating either at 50 or 60 Hz, depending on geographical locations¹. International organizations have established guidelines and standards, such as those provided by the International Commission for Non-Ionizing Radiation Protection (ICNIRP)^{2–4} and the International Committee on Electromagnetic Safety from the Institute of Electrical and Electronics Engineers (IEEE-ICES)^{5,6} to protect workers and the general public against potential adverse health effects from EMFs. Indeed, electrostimulation is considered the mechanism for biological and health effects at ELF. The guidelines and standards are based on scientific data and address various areas of the body, including the central nervous system (CNS) and the peripheral nervous system (PNS), to avoid adverse effects.

While there has been considerable research on the CNS stimulation by ELF-MFs, particularly with regards to the acute perception of magnetophosphenes^{7–12}, the acute effects of continuous powerline sinusoidal ELF-MFs on PNS stimulation have not been thoroughly investigated to date.

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The occurrence of electrostimulation (nerve stimulation) is governed by the strength and distribution of the internal (in situ) electric field (EF) inside the human tissue. Therefore, the field strength of the in situ EF (V/m) is the appropriate physical quantity that is used in both standards and guidelines to set limits (i.e., the thresholds). IEEE-ICES and ICNIRP define two sets of metrics: ICNIRP uses the basic restriction (BR) and the reference level (RL)³, while IEEE-ICES defines the dosimetric reference limit (DRL) and the exposure reference level (ERL)⁶. BR and DRL correspond to the in situ EF, which directly relates to electrostimulation within the body. In contrast, RL and ERL pertain to the MF strength, expressed as magnetic flux density (B) in Tesla (T). These two sets of limits are interconnected since the time-varying MF induces an in situ EF within the body. The MF levels (RL/ERL) are calculated or modeled to ensure that the internal EF remains within safe thresholds set by BR/DRL, with the consideration of reduction factors.

Electrostimulation models are essential tools for predicting neural responses to induced EF. Among them, the Spatially Extended Nonlinear Node (SENN) model^{13,14} was developed to capture neural excitation under spatially diffuse and varying fields. Unlike traditional models, as McNeal¹⁵ or Rattay¹⁶ models, that rely on focal stimulation and strong field gradients, SENN accounts for action potential (AP) initiation at the neural terminus, making it particularly suited for evaluating exposure limits in human safety standards^{17,18}. Its application has been foundational in defining regulatory thresholds for MF exposure in clinical and public settings⁵.

In this paper, all reported EF and MF values are presented as peak values. Instances where root mean square (RMS) values are used will be explicitly indicated. The experimental groundwork for threshold values was laid by studies such as^{19,20}. The 2019 IEEE-ICES C95.1 standards⁶ utilize Reilly's ellipsoidal model to relate the MF to the internal EF²¹. The “*presumed median for nerve excitation threshold*” has been derived using a nerve fiber model known as the SENN model^{13,21,22}, where the nerve fibers have a diameter of 20 μm . Thus, IEEE-ICES is based on SENN model calculations and establishes the PNS threshold at 6.15 V/m. Using a heterogeneous human model and from computed minimum EF amplitudes ranging from 3.6 to 5.8 V/m on volunteer exposure to the switched gradient fields of magnetic resonance, ICNIRP calculated a minimum EF threshold of about 4 V/m_{RMS} for PNS stimulation²³. High-resolution mathematical modeling of in situ EFs with voxel sizes of approximately $2 \times 2 \times 2 \text{ mm}^3$ ²⁴ is used to calculate the RL from the BR. However, there are ambiguities as it is not specified if the threshold reflects a population median or percentile.

It is important to note that most of these thresholds, both in IEEE-ICES and ICNIRP, are based on experimental data's interpolations of pulsed stimulations with various waveform shapes, such as rectangular or damped sinusoidal pulses, rather than continuous sinusoidal powerline frequency exposures.

Currently, ELF-MF PNS thresholds are estimated based on the SENN model, which represents a nerve fiber as an extended structure with alternating nodes and internodes (myelinated segments)¹³. The model uses a 20 μm diameter axon with internodal spacing set to 2 mm²¹. The nodes follow dynamics derived from the Frankenhaeuser-Huxley model for myelinated neurons²⁴ while internodes contribute resistive coupling between nodes. This spatially extended structure is governed by a cable equation, which describes electrical activity along the axon in the presence of extracellular fields¹⁵. However, this model does not account for the complex anatomical and physiological characteristics of nerve fibers and may not accurately capture the detailed behavior of nerves in realistic scenarios. Both ICNIRP and IEEE-ICES mention a need to improve the excitation modeling to improve accuracy of restrictions^{6,25}.

The current gold standard for myelinated nerve fiber electrostimulation modeling is the McIntyre-Richardson-Grill (MRG) model²⁶. This model provides a more anatomically and physiologically detailed representation of peripheral nerve fibers by using a nonlinear framework that incorporates the paranodal ultrastructures^{27,28}. This region, located near the nodes of Ranvier where the axon meets the myelin sheath, plays a critical role in signal transmission and influences nerve excitability. The MRG model uses a multi-compartmental approach, representing distinct physiological sections of the nerve fiber—such as the axon, myelin sheath, and nodes of Ranvier—each with their own properties. This enables a more realistic simulation of nerve activity than simpler models like SENN. The MRG model includes a detailed description of ion channels and their dynamics, enabling a more accurate simulation of nerve excitation. This model has been used and validated for PNS stimulation thresholds for electromagnetic coils such as a MRI gradient coil²⁹. It has been applied to describe perceptual thresholds observed in experiments on human arms and legs exposed to MF²⁹ and has demonstrated the ability to accurately predict experimental PNS thresholds in male and female body models³⁰. Furthermore, the MRG model was refined to be adapted to the type of nerve studied, i.e. “MRG-Sensory” and “MRG-Motor”³¹, predicting a lower threshold for sensory axons than motor axons of the same diameter.

Given the uncertainties regarding the ELF-MF PNS stimulation threshold, we propose to design a new ELF-MF system for human limbs exposure. Before obtaining experimental data, it is essential to develop a theoretical model to predict whether the desired effects can be theoretically induced. This is necessary due to the significant effort required in data collection and equipment creation. This study aims (1) to model a new ELF-MF exposure system for human limbs, and (2) to characterize the capability of the coil to generate PNS stimulation by simulation modelling. In this study, we present a theoretical and numerical investigation of the generation of in situ EFs by ELF-MF leg exposure system for PNS stimulation. Specifically, we estimate the maximum in situ EFs generated by the system at 60 Hz at the skin level and in the nerves of the leg using a realistic human body model with both the SENN and the MRG model.

Methods

Coils design

We propose to design a set of coils generating at least the 6.15 V/m in situ EF threshold predicted by the SENN model. As found in the North America environmental sources the fields will be sinusoidal waves at 60 Hz.

Based on experience and results from our previous works^{11,32} and the material available (interested reader may refer to the supplementary section II-A), a Helmholtz-like coil configuration was chosen for its ability to

generate an uniform MF at the center of the coils^{11,33}. To estimate the *in situ* EF induced, Maxwell–Faraday’s law of electromagnetic induction was applied. According to this law, the induced EF is proportional to the rate of change of magnetic flux density (dB/dt) through a loop. For a sinusoidal MF at all frequencies f , the magnetic flux density can be expressed as:

$$B(t) = B_0 \sin(2\pi ft). \quad (1)$$

From this, the time derivative of B is given by:

$$\frac{dB}{dt} = 2\pi f B_0 \times \cos(2\pi ft). \quad (2)$$

Given that $\max(\cos(x)) = 1$, we can therefore write that the peak dB/dt value follows:

$$\max\left(\frac{dB}{dt}\right) = 2\pi f B_0. \quad (3)$$

The peak value of the induced EF (E_{peak}) was calculated using the equation:

$$E_{\text{Peak}} = \frac{r}{2} \max\left(\frac{dB}{dt}\right) \quad (4)$$

where r is the radius of the limb (in m), f is the MF frequency (in Hz), B_0 is the magnetic flux density (in T), and dB/dt is the time derivative for B (in T/s). According to (4), the *in situ* EF is proportional to the limb’s radius. The torso was not considered due to concerns about cardiac electrical activity and its unintended potential stimulation. The lower limb was selected as the target location, as it has a large radius.

The MF at the center of the coil system was estimated using the Biot-Savart law (4), which relates the MF generated by a current-carrying conductor

$$B(x) = \frac{\mu_0 n I R^2}{(R^2 + x^2)^{\frac{3}{2}}} \quad (5)$$

where $B(x)$ is the MF at a distance x from the center, μ_0 is the permeability constant ($1.257 \times 10^{-6} \text{ T m A}^{-1}$), n is the number of turns in each coil, I is the coil current (in A), and R is the radius of the coils (in m).

Following these equations, with an average leg radius of 7.2 cm³⁴, and considering the amplifier system delivering up to 180 A, the choice of the coil dimensions, number of turns, and wire diameter were based on the desired *in situ* EF strength and the technical requirements for generating the targeted *in situ* EF. The theoretical dimensions are given by Fig. 1 (more details in the supplementary materials II-A).

Human body model

For the purpose of this study, the 3D neuro-functionalized human anatomical model JEDUK cV4.0 (male, 33 y, 1.62 m, 64.5 kg; IT’IS Foundation, Zurich, Switzerland) from the Virtual Population family, a set of high-resolution anatomical models generated based on MRI data from healthy volunteers, was used³⁵ (Fig. 2A). It is a highly detailed, computable anatomical human model including a set of 1186 tissues at high resolution. Anatomically correct nerve trajectories have been extracted from the segmentation and functionalized as splines for use with the NEURON solver³⁶ allowing a detailed analysis of the separate nerves in the leg. Simulations were

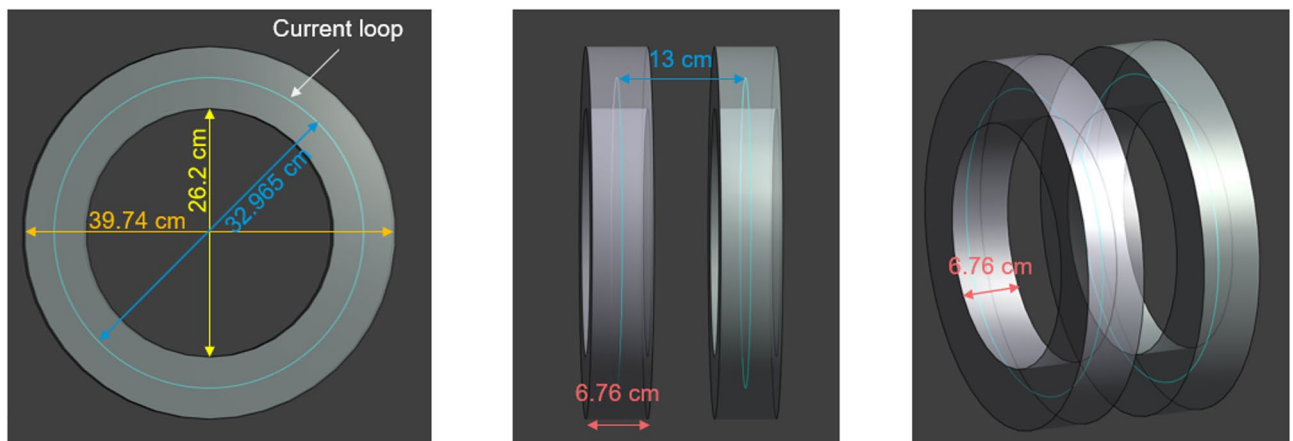


Fig. 1. Dimensions and configuration of the theoretical coils assembled in a Helmholtz-like configuration. The average current loop is represented by the blue circles.

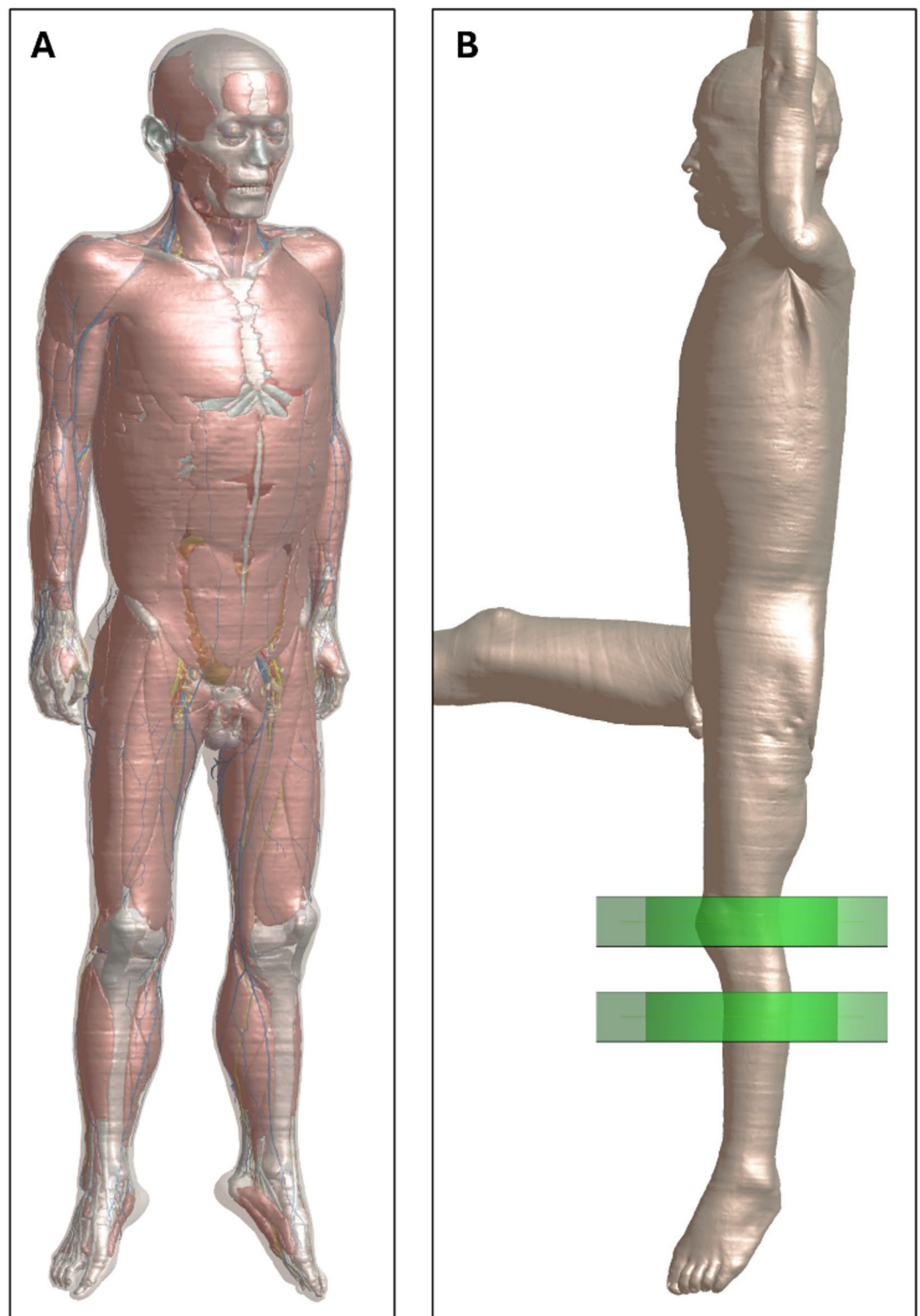


Fig. 2. Surface-based whole-body model used for the simulation of the MF. **(A)** JEDUK cV4.0 human model from IT'IS Foundation (Zurich, Switzerland) and the Virtual Population family (10.13099/ViP11017-04-0)—*Reproduced with permission from the IT'IS Foundation under a CC BY license*; **(B)** Helmholtz-like coil system positioned at the knee/popliteal fossa level. The figure was created using Sim4Life version 6.2 (Zurich MedTech AG, Zurich, Switzerland, <https://zmt.swiss/>).

run on the leg. Figure 2B illustrates one of these positions, with the coils at the knee level. Results of the other positions are reported in the supplementary material (Fig. S1).

Single nerves consist of a bundle of nerve fibers (axons) which can transmit electrical stimuli, so-called action potentials (AP). We identified ten nerves of the PNS in a leg of the JEDUK model, which consist of a total of 161 nerve fibers. The resolution of the JEDUK model was redefined to 0.6 mm in the area of the coils.

Software and numerical accuracy

The dosimetry simulations were carried out with the simulation platform Sim4Life Version 6.2 (ZurichMedTech AG, Zurich, Switzerland) to study the induced EF in the tissues and nerve fibers of the leg for exposure to the MF created with the coils. The Sim4Life software is a well-established tool and is being widely used for simulating neurostimulation. The EF simulations were performed with a sinusoidal current at 60 Hz and with a peak amplitude of $20 \times 19.75 \times 180 \text{ A} = 71.1 \text{ kA}$ flowing through each current loop (Fig. 1).

The dielectric properties of Jeduk's virtual model were assigned according to the LF IT'IS database (in supplementary materials II-B).

The nerve fibers were parametrized to have a diameter of 20 μm . The calculations were performed at 37 °C core body temperature. The membrane depolarization threshold was set at 80 mV. The duration of the simulation was at least 83.3 ms (5 full 60 Hz wave cycles), with a timestep of 0.001 ms. The internodal distance was set to 2 mm in the SENN model and approximately 1.7 mm in the MRG model, reflecting their different structural resolutions. As a result, the spatial location of a given node number differs between the two models.

The T-neuro module³⁷ uses a titration procedure to estimate the EF strength which is necessary to generate an AP along the nerve fiber. The result is not directly the EF, but a titration factor which represents a multiplication factor from which the necessary EF strength can be calculated. To this end, one has to look up the local EF strength at a certain location of the nerve fiber and multiply it with the titration factor. Hence, a factor smaller or equal to one indicates that given the simulated current an AP would be generated, whereas a factor larger than one indicates that a higher current would be needed to elicit an AP.

Results

EFs calculations

Figure 3 shows the in situ EFs maps at the skin level of the stimulated leg and MF vectors during a stimulation of 442 mT (the theoretical maximal MF strength generated by the system) at 60 Hz. Across all views, a centralized region of high EF intensity ($\geq 6.15 \text{ V/m}$) is evident. This indicates that the area surrounding the coils is where the EF peaks, likely aligning with the coil setup. And therefore suggesting that this leg's zone may be subjected to greater stimulation, potentially facilitating effective PNS stimulation.

The MF vectors, represented by arrows transitioning from dark to light, exhibit a pattern consistent with Helmholtz-like coil fields, showing circular gradients around the coil regions. Importantly, the orientation of the MF vectors with respect to the leg's anatomy could influence which nerves are more likely to be stimulated. The MF vectors demonstrate a more parallel alignment to the leg's surface in the high-EF region.

Figure 4 illustrates the in situ EF distribution along principal nerves of the leg during stimulation. The color gradient reflects varying EF intensities, with white denoting areas where EF values meet or exceed 6.15 V/m, and black representing areas with 0 V/m. This visualization highlights specific nerve segments that experience higher EF levels, indicating their potential for targeted stimulation. The branches of the femoral nerve—the posterior cutaneous and medial cutaneous—both primarily sensory, show local maxima of 6.01 V/m and 6.07 V/m, respectively, and an ICNIRP average³ of 5.6 V/m and 5.5 V/m. The ICNIRP average is implemented in Sim4Life and corresponds to the 99th percentile value found over a region of $2 \times 2 \times 2 \text{ mm}^3$. The common fibular nerve, responsible for the motor innervation of the muscles in the anterior and lateral compartments of the leg, and has several branches that provide sensory innervation, exhibits the highest local maximum EF at 7.55 V/m, and an ICNIRP average of 6.46 V/m. The tibial nerve, which has both sensory and motor functions, has a local maximum EF of 7.31 V/m, and an ICNIRP average of 4.86 V/m. This elevated EF along these nerves indicates the system's capability to stimulate sensory-motor pathways, potentially triggering both sensory and motor responses in the target region.

Titration study

Tables 1 and 2 summarize the titration results comparing two models, the SENN and MRG models, as well as the differentiation between sensory and motor components within the MRG model.

In Table 1, the range of titration factors across the nerve sets shows considerable variability between the SENN and MRG global models. For instance, the SENN model exhibits a broader titration range in nerves sets compared to the MRG global model and, the SENN model consistently displays higher upper titration values. The variability in titration factors across different nerve sets highlights the differences in model sensitivity and the potential impact of nerve structure and model parameters on EF thresholds.

Table 2 presents the titration results within the MRG model, distinguishing between sensory and motor components. Here, the titration factor ranges indicate that motor nerves require significantly higher EF strength to generate an AP than sensory nerves across all nerve sets. The nerve set 3 is the more likely set where AP can occur (titration factor close to 1). This disparity underscores the higher threshold requirements for activating motor fibers compared to sensory fibers, which may reflect the inherent physiological differences in excitability between these two types of nerves.

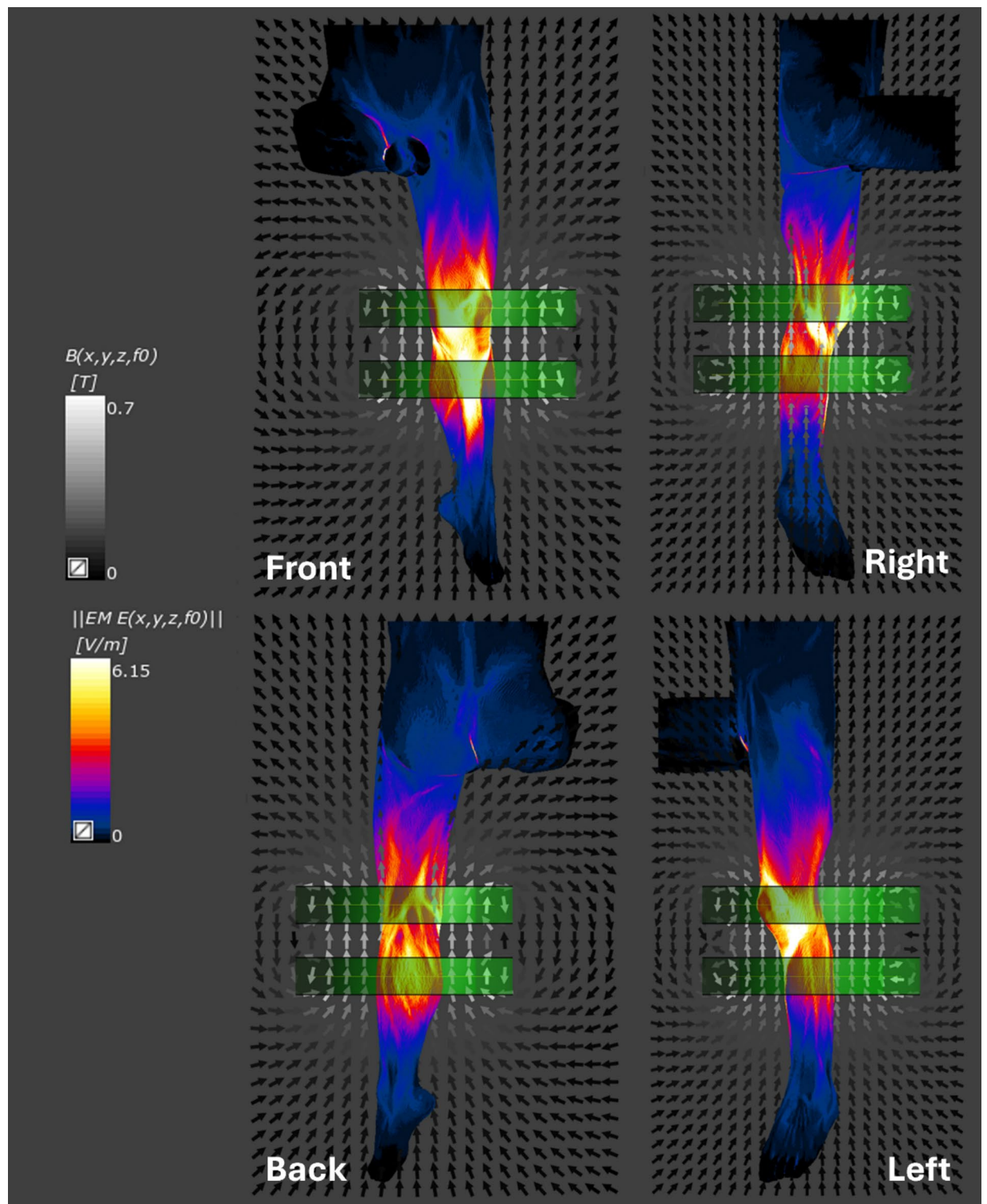


Fig. 3. EF maps for four views (front, back, left and right) at the skin level on the left leg during a MF stimulation by the Helmholtz-like coil system. The colors represent the level of induced EF in the leg, with white indicating an EF equal to or greater than 6.15 V/m and black equal to 0 V/m. The MF vectors are represented by arrows, the dark ones corresponding to 0 T and the lighter ones to 0.7 T. The figure was created using Sim4Life version 6.2 (Zurich MedTech AG, Zurich, Switzerland, <https://zmt.swiss/>).

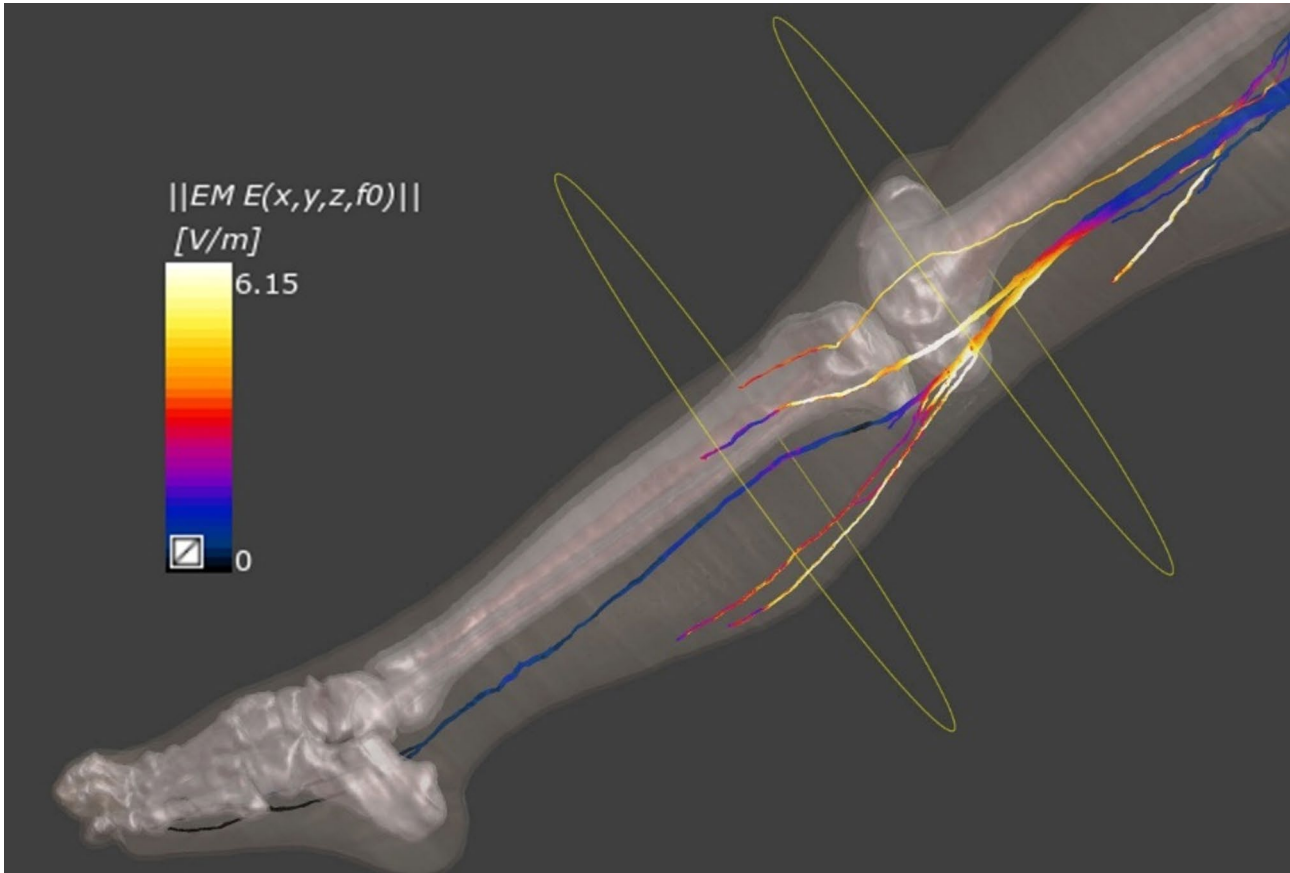


Fig. 4. Simulated principal leg’s nerve segments, with highest in situ EF. The colors represent the level of induced EF in the leg, with white indicating an EF equal to or greater than 6.15 V/m and black equal to 0 V/m. The two yellow circles represent the middle of each coil. The figure was created using Sim4Life version 6.2 (Zurich MedTech AG, Zurich, Switzerland, <https://zmt.swiss/>).

Nerve set	Range of titration factor	
	SENN	MRG global
1	21.75–99.5	4.156–20.5
2	11.9375–27.25	3.1875–8.4375
3	4.531–20.75	4.5625–17.5
4	7–22.625	7.3125–17.625

Table 1. Comparison of titration factors across SENN and MRG models for different nerve sets.

Nerve set	Range of titration factor	
	MRG-sensory	MRG-motor
1	4.9375–15.125	43.25–47.75
2	2.39–10.625	10.875–11.625
3	1.75–1.828	1.906–12.125
4	2.672–2.734	4.031–12.062

Table 2. Sensory versus Motor titration factors in the MRG model across different nerve sets.

Orientations

Figure 5 shows the orientations of the EFs and their strengths along one fiber nerve. The EFs are predominantly close to orthogonal relative to the nerve fiber along much of its trajectory. However, the figure also demonstrates that the nerve fiber is not linear; its pathway exhibits significant curvature and variability.

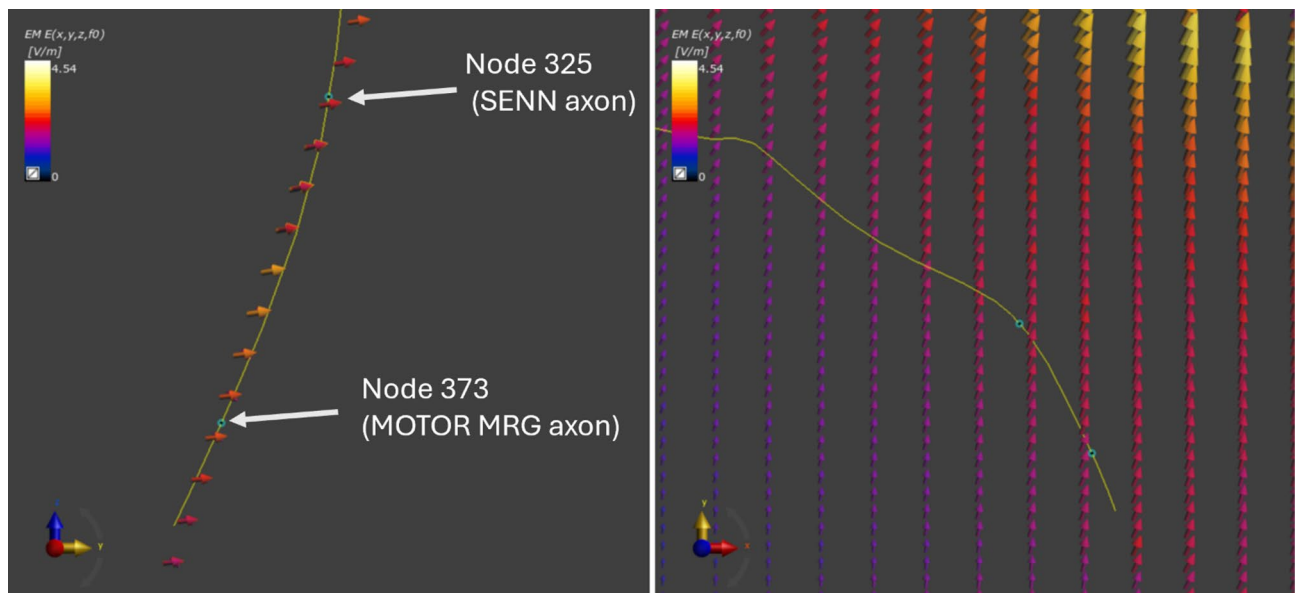


Fig. 5. EF orientation in relation to nerve trajectory. The figure illustrates the spatial distribution and orientation of the electric field ($E(x,y,z,f_0)$) in V/m, with field intensity indicated by color and arrow size. The color scale ranges from 0 to 4.54 V/m. The yellow line represents the nerve pathway through the field, with markers showing specific points along its trajectory.

In both the SENN and MRG models, nodes are indexed numerically along the nerve trajectory, with lower indices corresponding to more proximal segments. However, due to differences in spatial discretization between the models (i.e., internodal distances of 2 mm in SENN and ~1.7 mm in MRG), the same node number does not correspond to the same physical location. As a result, the identified nodes of AP initiation differ: node 325 in the SENN model and node 373 in the MRG model, which are located 4.158 mm apart.

Discussion

In this study, we conducted a dosimetry analysis to estimate the in situ EFs in the tissues and nerves of the human leg, aiming to assess the potential for PNS stimulation under sinusoidal MFs at powerline frequencies of 60 Hz. We first designed a Helmholtz-like coil setup to achieve an in situ EF target of 6.15 V/m, then modeled the effects of this MF exposure within a high-fidelity anatomical model (JEDUK) from the Virtual Population.

Our results indicate that four nerves in the leg—the common fibular, femoral posterior, femoral medial and tibial nerves—are found within high enough in situ EFs, exceeding the current 6.15 V/m SENN model threshold. And also the threshold for the MRG-Sensory model which is significantly lower by about a factor of 10^{38} . This indicates that these nerves could potentially be modulated. The two femoral branch nerves are sensory, while the tibial and common fibular nerves have both sensory and motor functions. Havel et al.²⁰ tested rectangular and damped sinusoidal pulses on 14 forearms using solenoidal coils. This study reported in situ EF thresholds of 5.87 ± 1.31 V/m. The Purdue University Study¹⁹, based on 84 healthy subjects in MR gradient coils (torso exposition), found values of 2.07 V/m for the y-coil and 2.30 V/m for the z-coil. Based on these data, computations and extrapolations using anatomical models report thresholds between 3.6 and 4.6 V/m at ELF-MF^{39,40}. Moreover, Kavet⁴¹ calculated a “nominal” 0.5-to-99.5 percentile population range of 3.5 to 7.0 V/m for in situ EFs as the threshold for PNS stimulation, based on previous studies^{14,39,42}. Based on these large ranges, the relatively high EF values along the nerves (< 6 V/m) in our study could induce PNS responses, potentially triggering sensory perception and, in the case of motor nerves, even motor contractions.

The titration study results highlight distinct differences between the SENN and MRG models and within the sensory and motor divisions of the MRG model. The broader range of titration factors in the SENN model across all nerve sets suggests that it may have a higher activation threshold compared to the MRG model, possibly due to differences in model structure or assumptions in sensitivity to EF induction. The more concentrated titration range of the MRG model may imply a more efficient alignment with physiological responses in nerve activation, making it potentially more representative of actual nerve behavior under MF stimulation. The disparity between sensory and motor thresholds in the MRG model, as shown in Table 2, is in line with Gaines et al.³¹. Our results are in line with Reilly¹⁷ and Soyka et al.³⁸, which found significant differences between thresholds depending on the model and parameters used.

Furthermore, the variability observed across different nerve sets (see supplementary materials Table S1) may suggest that nerve location, structural differences, and model-specific assumptions can influence EF titration thresholds. This insight could be valuable in designing more precise and customizable stimulation protocols, particularly when aiming to target specific nerve types or when attempting to fine-tune the balance between sensory and motor activation. Additionally, the differences between models may warrant further investigation

to understand the implications of each model's assumptions on EF, potentially guiding the choice of model based on desired outcomes or specific nerve targets.

Limitations of our study pertain. First, our calculation of the PNS stimulation was performed for a single male body model and does not yet include estimation of the uncertainty due to the normal biological, dielectric, and anatomical variability of people. Indeed, So et al.⁴⁰ have demonstrated that the size, shape, and accuracy of the body model significantly affect the generation of internal EFs. Furthermore, as we can see, depending on the location of the coil (supplementary materials Fig. S1, Table S2), the in situ EFs are not the same along the nerves (from 0 to 7.33 V/m for example), with large variations in amplitude and titration factors (Tables 1, 2). Moreover, another source of uncertainty in electromagnetic dosimetry lies in the dielectric properties of biological tissues, particularly conductivity, which govern how MF interact with the body and influence the induced in situ EF. For example, Schmid et al.⁴³ reported that skin conductivity could be substantially higher than the canonical values established by Gabriel et al.⁴⁴, potentially leading to inaccurate surface-level EF estimates. Santis et al.⁴⁵ also showed that dermal conductivity plays a dominant role in determining EF magnitude under LF-MF exposure. These studies underscore the importance of accounting for variability in skin properties, especially in peripheral tissues. Although our model used average tissue values from the IT'IS database, which is state-of-the-art⁴⁶, individual differences^{47–49} can still lead to substantial EF variability, particularly at boundaries between tissues with contrasting conductivities. These uncertainties may significantly affect EF predictions and the estimation of PNS stimulation thresholds. To address this, future studies will incorporate experimental data in humans to validate simulations and better capture inter-individual variability.

The positioning of the coil relative to the body model is also crucial, as highlighted in studies^{40,50}. Indeed, if the EF is orthogonal to the nerve fiber (i.e., MFs tend to run parallel to the nerves), stimulation may not occur, regardless of field strength. This underscores the importance of field orientation^{51,52}, not just intensity, in PNS modeling and experimental design. This critical dependence on orientation arises from the fundamental biophysics of nerve excitation. For an AP to be initiated, the induced EF must generate a suprathreshold transmembrane potential difference along the neuronal axon^{53,54}. When the EF is collinear to the nerve fiber, it creates a potential gradient along the membrane, leading to charge accumulation and depolarization at specific nodes of Ranvier or along unmyelinated sections. Conversely, an EF applied orthogonal to the nerve fiber generates no potential difference along the fiber's axis, thus failing to reach the depolarizing threshold and initiate an AP, even at high field strengths. As depicted in Fig. 5, the EF orientations tend to be orthogonal which could account for the wide variation in titration values observed in both models. However, it is important to note that nerves are not perfectly linear, and vary between individuals due to differences in body anatomy and nerve trajectories, tissue irregularities, and non-planar geometries^{55,56}.

This nonlinearity in the nerve's structure affects how the EF interacts with it, as the angle between the EF vector and the nerve axis changes along the trajectory. Such variability can influence the likelihood of AP initiation, as EF orientation is a critical determinant for effective nerve stimulation. This discrepancy likely reflects differences in the underlying assumptions and sensitivity to EF orientation within each model. The identification of two different nodes of Ranvier activated in the SENN and MRG models (Fig. 5) reflects their underlying structural and functional differences. The MRG model includes a more anatomically detailed double-cable representation of myelinated axons, with explicit modeling of internodal segments and nonlinear ion channel dynamics at each node. In contrast, the SENN model is simpler and less anatomically resolved includes fewer spatial compartments. Notably, the spatial discretization differs between the models: the internodal space is 2 mm in the SENN model and 1.7 mm in the MRG models. As a result, the number and spatial locations of nodes do not align exactly across the two models, leading to differences in where AP initiation is predicted. In our simulations, AP initiation occurred at node 325 in the SENN model and node 373 in the MRG model, corresponding to distal nodes and positions approximately 4.16 mm apart. While this spatial offset is modest, it underscores the influence of model resolution and structural assumptions on neural activation predictions. This discrepancy further illustrates the importance of model selection in predicting realistic neural activation and highlights the added value of high-fidelity models in dosimetry assessments. These findings emphasize the importance of both EF magnitude and orientation, as well as the structural complexity of nerves, in determining stimulation outcomes.

While the present study focused on the lower limb, it is important to acknowledge that PNS stimulation effects may differ in other regions. As previously discussed, the likelihood of AP initiation may vary across different regions of the body due to factors such as limb radius and nerve fiber orientation. For instance, the arms have a smaller radius than the legs, which leads to lower induced in situ EF under the same exposure. In contrast, the torso, despite its larger size, includes vital organs and the spinal cord, raising ethical and modeling limitations that make it unsuitable for isolated PNS investigations. Moreover, as shown in Fig. 3 (and Fig. S1 in the supplementary materials), the induced EF distribution is geometry and tissue dependant. While titration factors identified here are specific to the leg, and reflect its unique geometry and tissue structure, the modeling approach and key principles are applicable to other peripheral regions.

In addition to nerve fibers, the PNS contains specialized sensory receptors, such as Meissner and Pacinian corpuscles, which play a crucial role in sensory perception. These receptors differ from nerves not only in their orientation but also in their structural properties. For instance, Pacinian corpuscles are larger in diameter than typical nerve fibers, and Meissner corpuscles often have orientations that are perpendicular to the nerves^{57,58}, which may make them susceptible to different patterns of EF and MF stimulation. Notably, both Meissner and Pacinian corpuscles are tuned to frequencies relevant to our study: Pacinian corpuscles respond best to mechanical high-frequency vibrations (20–1000 Hz), while Meissner corpuscles are more sensitive to lower frequencies (10–50 Hz)^{57,58}, aligning closely with the 50 and 60 Hz frequencies of the powerline frequencies MF. These receptors operate through graded potentials (GPs) rather than the all-or-nothing APs seen in nerve fibers, meaning they can respond to subtle changes in stimulus intensity. This graded response characteristic is similar

to the potential mechanisms behind magnetophosphenes¹¹, where low-frequency MFs stimulate retinal cells (primarily rods) that also generate GPs. This parallel suggests that Meissner and Pacinian corpuscles could be similarly responsive to low-frequency MF, contributing to sensory effects beyond nerve activation alone.

The presence of these receptors and their varying orientations⁵⁸ underscore that in situ EF strength alone may not fully predict PNS activation. Field orientation relative to these structures is equally important, as a perpendicular EF might stimulate a receptor more effectively than it would a nerve fiber. Understanding the roles and activation thresholds of these specialized receptors could improve our dosimetry models by accounting for non-nerve elements within the PNS that may respond to MFs. Future work should consider these structures when assessing the thresholds and effects of low-frequency MF exposure, as they may play a significant role in sensory perceptions under such conditions.

There is still a critical gap in our understanding regarding the precise thresholds at which PNS stimulation by MF at powerline frequencies occurs. Establishing these thresholds is essential for accurately assessing the potential risks associated with exposure to EMF within this frequency range. In 2019 and 2020, IEEE⁶ and ICNIRP²⁵ respectively recommended the need for experimental data for threshold assessments in humans to enhance the restrictions' accuracy. Our work directly responds to this call, providing a crucial message for the scientific community: high-fidelity computational models, particularly those with detailed anatomical and physiological representations like the MRG model, are indispensable tools for bridging this gap and refining our understanding of PNS responses to ELF-MF exposure. By investigating perception thresholds in human subjects, we can further refine dosimetry models and develop more accurate safety guidelines for mitigating potential health risks. This future research endeavor holds promise for enhancing our understanding of the effects of EMF on the PNS and ensuring the safety and well-being of individuals exposed to EMF.

Given the results and the limitations of the simpler SENN model, the MRG model—due to its anatomical detail and precision—could be a valuable tool for developing standards and guidelines for ELF-MF exposure. Its comprehensive representation of nerve fiber properties allows for more accurate dosimetry assessments, potentially leading to more effective and protective exposure standards.

By offering a dosimetry analysis and identifying specific nerve stimulations, this study opens doors for both fundamental and applied research in peripheral neuromodulation. Understanding these precise stimulation thresholds, as well as the effects of threshold exposures, is not only vital for establishing robust safety guidelines and standards but also offers the knowledge needed to intentionally modulate the PNS. This has profound implications for understanding underlying neuromechanisms and their integration with the CNS. Furthermore, these insights are directly applicable to the development and refinement of medical devices and functional evaluations that utilize peripheral stimulation^{59–61}.

Conclusion

The evaluation and model comparisons have been performed for PNS stimulation in human models exposed to uniform MF at 60 Hz. The quantitative results found in this work are specific to the investigated simulation setup, i.e., a leg in Helmholtz-like coil configuration. Our results demonstrate the critical importance of thorough modeling to guide the design and validation of experimental setups for ELF-MF exposure systems.

This study highlights the necessity of developing a robust theoretical foundation before embarking on extensive data collection and equipment creation. By accurately predicting the in situ EFs and their potential to induce PNS stimulation, we can ensure that the experimental setup will achieve the desired outcomes. The use of both the SENN and MRG models provided valuable insights into the stimulation thresholds and the effectiveness of the designed coil system.

Our findings underscore the importance of validating the coil through experimental data to confirm the simulation results and refine the exposure system and localisation. This step is crucial for translating theoretical predictions into practical applications and ensuring the reliability of the designed system.

Future work should focus on conducting experimental studies to validate the coil system and exploring other anatomical regions and exposure conditions. This will further enhance the precision and applicability of the dosimetry assessments, ultimately advancing the field of bioelectromagnetic safety.

Data availability

The datasets generated during and/or analyzed during the current study are available from the corresponding author on request, to any qualified researcher.

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Author contributions

E.F., N.B. and A.L. developed the basic concepts of the framework, E.F. wrote the original draft. C.A. and F.S. implemented and ran the simulation and the computational methods. C.A. created all the figures. E.F., N.B., F.S., and C.A. contributed to the data interpretation. N.B., F.S., C.A., F.S.P and A.L. participated in the reviewing and editing of the manuscript. A.L. obtained the funding. All authors read and approved the submitted manuscript.

Declarations

Competing interests

The authors declare that they have no competing interests.

Ethics declarations

This study was approved by Western University's Ethics Board (protocol REB# 120854). This study is based entirely on a computational dosimetry model and does not involve any human participants, interventions, or biological tissue. Therefore, informed consent and following the declaration of Helsinki were not required for this specific work.

Additional information

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