

Computational Sensitivity Analysis of Knee-Joint Tissue Properties on Pulsed Electromagnetic Field Distribution Using COMSOL Multiphysics 6.0

Iliya Ladan Filibus¹

¹Department of Electrical and Electronic Engineering, Abubakar Tafawa Balewa University, Bauchi, Nigeria
E-mail: iliya221221@gmail.com

ARTICLE INFO

Keywords:

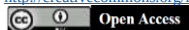
Bioelectromagnetics,
COMSOL Multiphysics,
finite element method,
knee osteoarthritis,
magnetic flux density,
pulsed electromagnetic field (PEMF),
sensitivity analysis, tissue dielectric properties.

Vol.01, No.05, 2026

<https://doi.org/10.5281/zenodo.23100986>

Copyright © 2026 by author(s) and Scientific Research Publishing Inc. This work is licensed under the Creative Commons Attribution International License (CCBY4.0).

<http://creativecommons.org/licenses/by/4.0/>



ABSTRACT

Pulsed electromagnetic field (PEMF) therapy is increasingly investigated as a non-invasive adjunct for knee osteoarthritis, yet clinical outcomes remain heterogeneous, partly because the field actually delivered to intra-articular tissues is rarely quantified. This study presents a finite-element sensitivity analysis of how the electromagnetic and geometric properties of knee-joint tissues affect PEMF distribution. A multilayer knee model comprising bone, articular cartilage, muscle, subcutaneous fat, and skin, surrounded by an excitation coil and an air domain, was implemented in the Magnetic Fields interface of COMSOL Multiphysics 6.0 and solved in the time domain. Using a one-at-a-time design, the electrical conductivity (σ), relative permittivity (ϵ_r), relative magnetic permeability (μ_r), and thickness (d) of each tissue were varied around literature-based baseline values, and the response of the volume-averaged magnetic flux density was quantified through percentage change and normalized sensitivity coefficients $S \sim X$. Under baseline conditions, the average flux density ranged from 3.80 mT (skin) to 6.39 mT (cartilage), with peak values between 5.5 and 9.4 mT; the corresponding peak induced electric field and current density were approximately 0.20 V/m and 0.10 A/m². Relative permeability produced the highest normalized sensitivity ($|S| = 0.92\text{--}0.95$), followed by tissue thickness ($|S| = 0.11\text{--}0.20$, inverse relationship) and conductivity ($|S| = 0.10\text{--}0.19$), whereas permittivity had the weakest effect ($|S| = 0.06\text{--}0.17$). Among tissues, muscle and the superficial fat and skin layers were the most sensitive to conductivity and thickness, while cartilage exposure was governed chiefly by geometry. A mesh-convergence study confirmed numerical stability ($<0.2\%$ change beyond 1.29×10^5 elements). The results identify the tissue parameters that most strongly control predicted PEMF exposure and provide a quantitative basis for parameter prioritization, uncertainty assessment, and future patient-specific optimization of PEMF devices for knee pain management.

Introduction

Knee osteoarthritis (OA) is one of the leading causes of chronic musculoskeletal pain and disability worldwide. It is characterized by progressive degeneration of articular cartilage, subchondral bone remodeling, synovial inflammation, and the resulting pain, stiffness, and loss of physical function [1]. Pharmacological management, physiotherapy, and, in advanced cases, joint replacement remain the mainstays of treatment; however, adverse effects, cost, and invasiveness have motivated continued interest in non-invasive physical therapies.

Pulsed electromagnetic field (PEMF) therapy exposes tissue to time-varying magnetic fields generated by an external coil. By Faraday induction, the changing magnetic field induces electric fields and eddy currents in conductive tissues, which have been associated with modulation of ion transport, adenosine-receptor signaling, inflammatory pathways, and chondrocyte and osteoblast activity [2]–[4]. Meta-analytic evidence indicates that PEMF can reduce pain and stiffness and improve physical function in OA [5], and recent systematic reviews report promising clinical outcomes, while emphasizing

considerable heterogeneity in field amplitude, frequency, waveform, device type, and treatment duration among published protocols [6], [7]. This heterogeneity complicates comparisons between trials and makes it difficult to determine whether differences in outcome arise from the biological response itself or from differences in the electromagnetic dose that actually reaches the target tissue.

A central obstacle is that the field experienced by intra-articular structures such as cartilage cannot be measured directly in vivo without invasive instrumentation. The field generated by an external coil must traverse skin, subcutaneous fat, and muscle before reaching cartilage and bone, each of which has distinct electrical conductivity, permittivity, and geometry [8]–[11]. Computational bioelectromagnetics offers a practical alternative: numerical models allow anatomical geometry, tissue properties, and source configuration to be specified and the resulting field quantities to be calculated throughout the anatomy [12], [13]. The finite element method (FEM) is particularly suitable because it accommodates complex multi-domain geometries and heterogeneous material properties [13], [14].

However, the reliability of any such model depends on the tissue properties assigned to it. Published dielectric data vary considerably with measurement method, frequency, tissue condition, and inter-subject variability; reported uncertainty in tissue properties can exceed that in geometry by an order of magnitude [8], [15]. Parametric and surrogate-model approaches have therefore been proposed to explore exposure variability efficiently at low frequencies [16]. Sensitivity analysis, i.e., the systematic quantification of how variations in model inputs propagate to model outputs [17], is the established tool for identifying which inputs require accurate characterization and which can be approximated without materially affecting predictions.

Despite the growing number of clinical PEMF studies on knee OA, systematic sensitivity analyses of how the electromagnetic and geometric properties of *individual* knee tissues influence the predicted PEMF distribution remain scarce. Most previous work has focused either on clinical efficacy or on the field produced by a particular device, typically assuming a single fixed set of tissue properties. Consequently, it is unclear which tissue parameters dominate the predicted exposure of cartilage and surrounding tissues, and how much uncertainty is introduced when literature-derived property values are used.

To address this gap, the present study develops a multilayer FEM model of the knee in COMSOL Multiphysics 6.0 and performs a one-at-a-time sensitivity analysis of four tissue parameters—electrical conductivity (σ), relative permittivity (ϵ_r), relative magnetic permeability (μ_r), and tissue thickness (d)—for five tissues: bone, cartilage, muscle, fat, and skin. The specific contributions are as follows:

- 1) a baseline characterization of magnetic flux density, induced electric field, and current density in each tissue domain under a defined PEMF excitation;
- 2) quantification of the percentage change and normalized sensitivity coefficient of the tissue-averaged magnetic flux density with respect to each parameter and tissue;
- 3) a ranking of tissue parameters by influence, identifying those that should be prioritized in model development, uncertainty assessment, and future patient-specific PEMF optimization.

The remainder of the paper is organized as follows. Section II describes the model, governing equations, numerical implementation, and sensitivity procedure. Section III presents the results. Section IV discusses their physical interpretation, implications, and limitations, and Section V concludes the paper.

Materials and Methods

A. Study Design

A computational, simulation-based design was adopted; no human or animal subjects were involved. A baseline knee model with reference tissue properties was first established. Each tissue

parameter was then perturbed one at a time (OAT) around its baseline value while the geometry of all other tissues, coil configuration, excitation, boundary conditions, mesh strategy, and solver settings were held constant, so that changes in the output could be attributed to the parameter under investigation [17].

B. Model Geometry and Tissue Domains

The knee was represented as a simplified, axially layered multi-domain geometry consisting of five tissue domains arranged from the core outward: bone, articular cartilage, muscle, subcutaneous fat, and skin (Fig. 1). Each tissue was modeled as a separate domain with its own material definition and identifiable interfaces, enabling independent modification of its properties and thickness. Geometry was checked for overlapping domains, gaps, and invalid boundaries before meshing. A simplified rather than image-based anatomy was deliberately chosen to isolate the effect of tissue parameters from anatomical irregularities and to keep the parametric study computationally tractable.

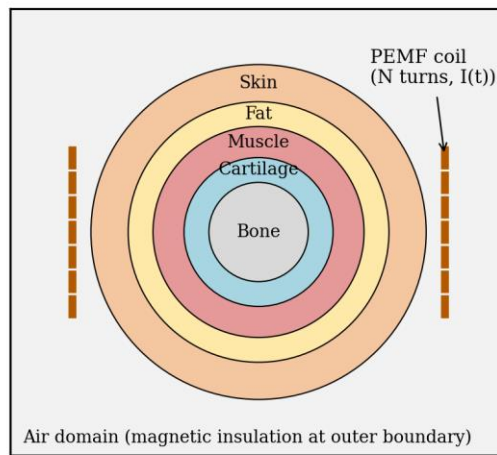


Fig. 1. Schematic cross-section of the multilayer knee-joint model showing the five tissue domains, the external PEMF excitation coil, and the surrounding air domain (not to scale).

The PEMF source was a multi-turn circular coil positioned coaxially around the knee at a fixed distance, which was kept constant in all simulations. The knee and coil were enclosed in an air domain ($\sigma = 0$, $\epsilon_r = \mu_r = 1$) sufficiently large relative to the model that the truncation boundary had negligible influence on the computed fields.

C. Tissue Properties

Baseline electromagnetic properties were selected from the low-frequency ranges reported in the dielectric literature [8]–[11] and are summarized in Table I together with the baseline layer thicknesses and the variation ranges applied in the sensitivity analysis. Because biological tissues are essentially non-magnetic (volume susceptibility of order 10^{-5} [18]), $\mu_r = 1$ was used as the baseline for all tissues; a $\pm 2\%$ range was applied as a numerical probe of the model's magnetic sensitivity.

Table 1. Baseline Tissue Properties and Sensitivity Ranges

Tissue	σ (S/m)	ϵ_r	μ_r	d (mm)
Bone	0.04 (0.02–0.06)	10 (≤ 15)	1 (0.98–1.02)	30 (20–40)
Cartilage	0.20 (0.10–0.30)	70 (50–90)	1 (0.98–1.02)	35 (25–45)
Muscle	0.50 (0.30–0.70)	65 (45–85)	1 (0.98–1.02)	40 (30–50)
Fat	0.04 (0.02–0.06)	9 (5–13)	1 (0.98–1.02)	45 (35–55)
Skin	0.20 (0.10–0.30)	55 (35–75)	1 (0.98–1.02)	50 (40–60)

Values outside parentheses are baseline values; ranges in parentheses were used in the OAT sensitivity analysis. d denotes the radial layer dimension in the simplified model.

D. PEMF Excitation

The coil was excited by a time-dependent current $I(t)$ applied through the Coil feature of the Magnetic Fields interface, with $[N]$ turns, peak current $[I_0 \text{ (A)}]$, and repetition frequency $[f \text{ Hz}]$ [waveform: e.g., sinusoidal/pulsed, pulse width]. For a sinusoidal approximation the excitation is

$$I(t) = I_0 \sin(2\pi ft) \quad (1)$$

and for a pulsed waveform a piecewise or analytic function reproducing the pulse train was used. The excitation parameters were identical across all sensitivity simulations.

E. Governing Equations

The electromagnetic problem was formulated in terms of the magnetic vector potential $\mathbf{A}$, with $\mathbf{B} = \nabla \times \mathbf{A}$. Combining Ampère–Maxwell's law with the constitutive relations gives

$$\sigma \partial \mathbf{A} / \partial t + \epsilon_0 \epsilon_r \partial^2 \mathbf{A} / \partial t^2 + \nabla \times (\mu_0^{-1} \mu_r^{-1} \nabla \times \mathbf{A}) = \mathbf{J}_e \quad (2)$$

where $\mathbf{J}_e$ is the externally applied coil current density. The induced electric field follows from Faraday's law,

$$\nabla \times \mathbf{E} = -\partial \mathbf{B} / \partial t, \quad \mathbf{E} = -\partial \mathbf{A} / \partial t \quad (3)$$

and the induced (conduction) current density in tissue is

$$\mathbf{J} = \sigma \mathbf{E} \quad (4)$$

with $\mathbf{B} = \mu_0 \mu_r \mathbf{H}$. At the low frequencies typical of PEMF therapy, the quasi-static approximation holds because tissue dimensions are much smaller than the wavelength and the skin depth [19]; the displacement-current term is retained in (2) for completeness so that the effect of ϵ_r can be evaluated.

F. Boundary Conditions, Mesh, and Solver

Magnetic insulation ($\mathbf{n} \times \mathbf{A} = 0$) was imposed on the outer boundary of the air domain. Continuity of the tangential field components was enforced automatically at interior tissue interfaces. The initial condition was $\mathbf{A} = 0$ at $t = 0$.

The geometry was discretized with free tetrahedral elements, refined around the coil windings and at tissue interfaces where field gradients are largest. A mesh-convergence study was performed with five mesh densities (32 450 to 286 710 elements). The time-dependent study used the default implicit backward-differentiation-formula (BDF) time stepping with a direct linear solver; convergence was verified from the solver log and the stability of the computed field quantities.

G. Output Quantities

The principal output was the magnitude of the magnetic flux density, $B = |\mathbf{B}|$, evaluated as the volume average over each tissue domain V :

$$\bar{B} = (1/V) \int_V |\mathbf{B}| dV \quad (5)$$

Maximum and minimum values within each domain were also extracted to characterize spatial uniformity. The induced electric field $|\mathbf{E}|$ and current density $|\mathbf{J}|$ were evaluated from (3) and (4).

H. Sensitivity Analysis

For each parameter $X \in \{\sigma, \epsilon_r, \mu_r, d\}$ and each tissue, five levels (baseline and two levels on either side, Table I) were simulated, giving $4 \times 5 \times 4 = 80$ perturbed simulations in addition to the baseline. The percentage change relative to baseline was computed as

$$\% \Delta B = (B_i - B_0) / B_0 \times 100 \quad (6)$$

where B_0 and B_i denote the baseline and perturbed tissue-averaged flux densities. Because the parameters were varied over different relative ranges, a normalized (elasticity-type) sensitivity coefficient was used for comparison [17]:

$$S_X = (\Delta B / B_0) / (\Delta X / X_0) \quad (7)$$

S_X was evaluated by central differences across the full investigated range (forward difference where only one-sided data were available). Parameters were ranked by $|S_X|$; $|S_X| = 1$ indicates that a 1%

change in the parameter produces a 1% change in $\bar{B}$. Post-processing was performed with the COMSOL Results module and exported for tabulation and plotting.

Results

A. Mesh Convergence

Table II and Fig. 2 show that the average and maximum flux density in the bone domain changed by 2.32% from the coarse to the normal mesh, 1.05% from normal to fine, and only 0.17% from fine to finer, with no further change thereafter. The fine mesh (128 650 elements) was therefore adopted for all subsequent simulations as a compromise between accuracy and computational cost.

Table 2. Mesh Convergence (Bone Domain)

Mesh	Elements	$\bar{B}$ (mT)	$B_{\max}$ (mT)	Δ (%)
Coarse	32 450	5.61	7.94	—
Normal	67 820	5.74	8.11	2.32
Fine*	128 650	5.80	8.20	1.05
Finer	198 430	5.81	8.22	0.17
Very fine	286 710	5.81	8.22	0.00

Adopted mesh. Δ is the change in $\bar{B}^*$ relative to the previous mesh level.

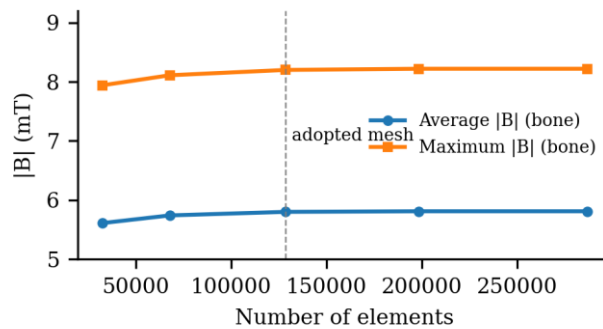


Fig. 2. Convergence of the average and maximum magnetic flux density in the bone domain with mesh refinement.

B. Baseline PEMF Distribution

Under baseline conditions, the PEMF penetrated all tissue layers (Table III, Fig. 3). The highest volume-averaged flux density occurred in cartilage (6.39 mT), followed by bone (5.80 mT), muscle (4.90 mT), fat (4.30 mT), and skin (3.80 mT). Peak values ranged from 5.5 mT (skin) to 9.4 mT (cartilage). The ratio of maximum to minimum flux density was approximately 2.0–2.2 in all tissues, indicating moderate spatial non-uniformity within each domain. The peak induced electric field was approximately 0.20 V/m, and the peak induced current density was approximately 0.10 A/m², the latter occurring in the comparatively conductive muscle domain, consistent with (4).

Table 3. Baseline Magnetic Flux Density in Each Tissue

Tissue	$\bar{B}$ (mT)	$B_{\max}$ (mT)	$B_{\min}$ (mT)	$B_{\max}/B_{\min}$
Bone	5.80	8.2	3.9	2.10
Cartilage	6.39	9.4	4.8	1.96
Muscle	4.90	7.1	3.3	2.15
Fat	4.30	6.3	2.9	2.17
Skin	3.80	5.5	2.6	2.12

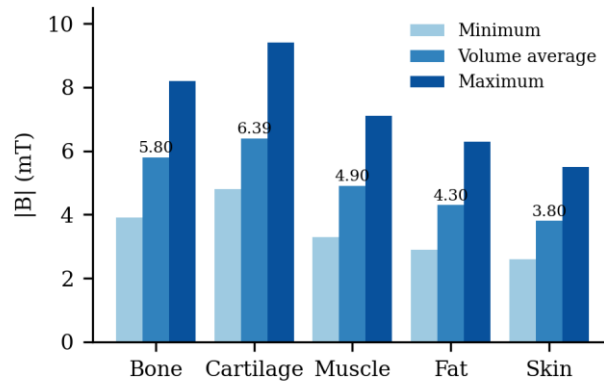


Fig. 3. Baseline minimum, volume-averaged, and maximum magnetic flux density in each knee-tissue domain.

C. Sensitivity to Electrical Conductivity

Increasing σ produced a monotonic increase in $\bar{B}$ in every tissue (Fig. 4(a)). Over the investigated ranges, the largest upward changes were found for fat (+7.91% at $\sigma = 0.06$ S/m) and muscle (+7.76% at $\sigma = 0.70$ S/m), followed by skin (+6.05%), bone (+4.83%), and cartilage (+4.54%). Normalized coefficients (Table IV) were highest for muscle ($S_{\sigma} = 0.194$) and fat (0.158) and lowest for cartilage (0.103), i.e., a 10% uncertainty in muscle conductivity translates into an approximately 2% uncertainty in its predicted flux density.

D. Sensitivity to Relative Permittivity

Relative permittivity also showed a positive, approximately linear relationship with $\bar{B}$ (Fig. 4(b)). Maximum changes were +5.26% (skin), +5.12% (fat), +5.10% (muscle), +3.44% (cartilage), and +2.76% (bone). After normalization, muscle ($S_{\epsilon} = 0.166$) and skin (0.145) were the most sensitive, whereas bone was clearly the least sensitive (0.055).

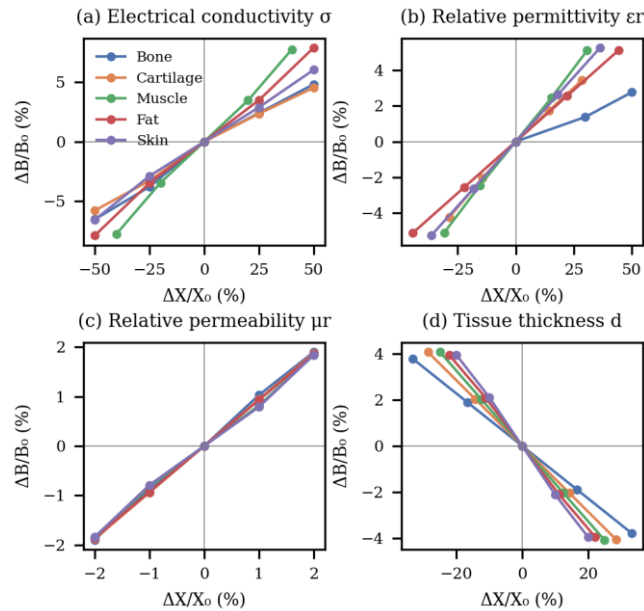


Fig. 4. Percentage change in tissue-averaged magnetic flux density versus percentage change in (a) electrical conductivity, (b) relative permittivity, (c) relative magnetic permeability, and (d) tissue thickness.

E. Sensitivity to Relative Magnetic Permeability

Varying μ_r by only $\pm 2\%$ changed $\bar{B}$ by $\pm 1.84\%$ to $\pm 1.90\%$ in all five tissues (Fig. 4(c)). The response was almost perfectly linear and nearly identical across tissues, giving the largest normalized

coefficients of the study ($S_\mu = 0.92\text{--}0.95$). This near-unity elasticity is consistent with the direct proportionality $\mathbf{B} = \mu_0\mu_r\mathbf{H}$.

F. Sensitivity to Tissue Thickness

Tissue thickness was the only parameter with an inverse effect (Fig. 4(d)): increasing the thickness of any layer reduced $\bar{B}$ in that layer, and decreasing it increased $\bar{B}$. A 10 mm change produced changes of approximately $\pm 3.8\%$ to $\pm 4.1\%$ in all tissues. Because superficial layers had larger baseline thicknesses and therefore smaller relative perturbations, their normalized coefficients were larger: skin ($|S_d| = 0.197$), fat (0.178), muscle (0.163), cartilage (0.142), and bone (0.114).

G. Comparative Sensitivity and Ranking

Table 4 and Fig. 5 summarize all normalized coefficients. Ranked by mean $|S_x|$ across tissues, the parameters were ordered as $\mu_r (0.931) \gg d (0.159) > \sigma (0.139) > \epsilon_r (0.123)$. By tissue, muscle was the most sensitive to the electrical parameters (σ and ϵ_r), skin and fat were the most sensitive to thickness, and bone was the least sensitive overall to the electrical parameters. For cartilage—the principal therapeutic target—thickness ($|S_d| = 0.142$) and permittivity (0.134) exerted greater influence than its own conductivity (0.103).

Table 4. Normalized Sensitivity Coefficients of Tissue-Averaged B

Tissue	σ	ϵ_r	μ_r	d
Bone	0.114	0.055	0.948	0.114
Cartilage	0.103	0.134	0.939	0.142
Muscle	0.194	0.166	0.918	0.163
Fat	0.158	0.115	0.930	0.178
Skin	0.126	0.145	0.921	0.197
Mean	0.139	0.123	0.931	0.159

S_d is negative for all tissues (inverse relationship); absolute values are shown. Bold denotes the largest value in each column among tissues.

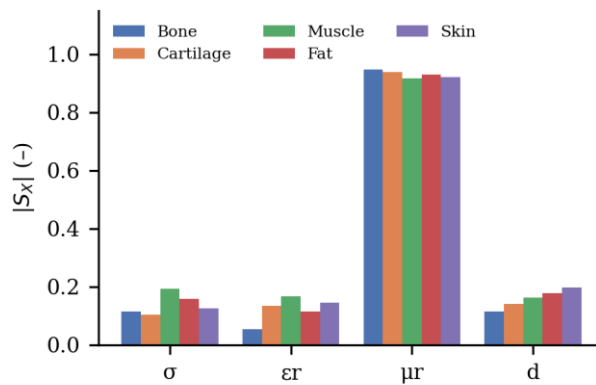


Fig. 5. Normalized sensitivity coefficients $|S_x|$ of the tissue-averaged magnetic flux density for each parameter and tissue.

Discussion

A. Baseline Distribution

The baseline results demonstrate that a PEMF applied externally does not produce a uniform exposure throughout the knee. The tissue-averaged flux density differed by a factor of about 1.7 between cartilage and skin, and within each tissue the peak value was roughly twice the minimum. The relatively high flux density in the central cartilage and bone domains is consistent with the coaxial coil arrangement, in which the field from the encircling windings superposes in the interior of the coil [14].

From a therapeutic perspective, this is encouraging, as it suggests that a suitably positioned circumferential coil can deliver a field to the cartilage that is comparable to or larger than that at the surface. The computed peak induced electric field (≈ 0.20 V/m) and current density (≈ 0.10 A/m²) are in the low range typically associated with therapeutic PEMF and are of the same order as the reference levels used in exposure guidelines for the low-frequency range [20]; they should nevertheless be compared directly with the applicable basic restrictions for the specific excitation frequency used.

B. Interpretation of Parameter Sensitivities

The very high normalized sensitivity to μ_r must be interpreted with care. It reflects the direct proportionality between $\mathbf{B}$ and μ_r and therefore indicates how strongly the *model* responds to this parameter, not how much uncertainty it introduces in practice. Real tissues are weakly diamagnetic or paramagnetic, with $|\chi| \approx 10^{-5}$ [18]; realistic deviations of μ_r from unity are thus about three orders of magnitude smaller than the $\pm 2\%$ range tested, and their practical contribution to uncertainty in $\bar{B}$ is negligible. Assigning $\mu_r = 1$ to all tissues, as is common practice, is therefore justified.

In contrast, conductivity and permittivity are known to vary widely between subjects, measurement techniques, and tissue states, with reported ranges often spanning a factor of two or more [8]–[11], [15]. Although their normalized coefficients are modest (0.06–0.19), their realistic uncertainty ranges are large, so they can produce the largest *practical* variation in predicted exposure. For example, a factor-of-two uncertainty in muscle conductivity corresponds, through $S_\sigma \approx 0.19$, to an uncertainty of up to $\approx 19\%$ in its predicted flux density. This distinction between sensitivity and practical uncertainty contribution is essential when prioritizing which tissue properties require experimental characterization.

Conductivity also governs the induced current through (4): even where the effect of σ on $\mathbf{B}$ is small, $\mathbf{J}$ scales directly with σ for a given $\mathbf{E}$. Hence, uncertainty in conductivity has a first-order effect on the predicted current density, which is the quantity most often hypothesized to mediate biological effects [2], [4]. The comparatively high sensitivity of muscle to both σ and ϵ_r is consistent with its high water and electrolyte content and its intermediate position between the coil and the joint.

The inverse relationship with thickness shows that geometry matters for every tissue. Thicker superficial layers increase the separation between the windings and the deeper structures, while thicker internal layers distribute the flux over a larger volume, both reducing the domain average. This finding is clinically relevant because subcutaneous fat thickness varies substantially with body mass index; patients with thicker soft-tissue envelopes may therefore receive lower intra-articular exposure from the same device setting. For cartilage, thickness was the dominant tissue-specific parameter, which suggests that cartilage thinning in OA may itself alter local exposure—an aspect that warrants patient-specific investigation.

C. Implications for PEMF Modeling and Therapy

The present results provide practical guidance for PEMF model development. First, accurate representation of anatomical dimensions, particularly the thickness of fat and skin and of the cartilage itself, should be prioritized. Second, conductivity values for muscle and fat should be selected carefully and, where possible, accompanied by an uncertainty range. Third, permittivity has comparatively limited influence at low frequencies and, together with μ_r , can be taken from standard databases without materially affecting predicted $\bar{B}$. These findings support the view that part of the heterogeneity in clinical PEMF outcomes [6], [7] may stem from differences in the dose actually delivered to the joint, which depend on both device parameters and patient anatomy. Dosimetry-informed, patient-specific planning could therefore improve the reproducibility of future clinical trials.

D. Limitations

This study has several limitations. 1) The knee geometry was a simplified layered representation rather than an image-based anatomical model; menisci, ligaments, synovial fluid, and the patella were not represented. 2) Tissue properties were treated as linear, isotropic, homogeneous, and frequency-independent within each domain, whereas real tissues exhibit dispersion and anisotropy [10]. 3) The OAT design does not capture interactions between parameters; variance-based global sensitivity analysis [17] would provide a more complete picture. 4) Only magnetic flux density was used for sensitivity ranking; a similar analysis of $\mathbf{E}$ and $\mathbf{J}$ would be valuable because these are expected to be more sensitive to σ . 5) The model has not been validated against experimental measurements in phantoms or cadaveric specimens, and no biological or clinical endpoint was evaluated. The results should therefore be regarded as a computational basis for further investigation and not as evidence of therapeutic efficacy.

Conclusion

A multilayer finite-element model of the knee was developed in COMSOL Multiphysics 6.0 to quantify how the electromagnetic and geometric properties of bone, cartilage, muscle, fat, and skin affect the distribution of an externally applied PEMF. The baseline simulation showed non-uniform exposure, with tissue-averaged flux densities of 3.80–6.39 mT and the highest exposure in cartilage. One-at-a-time sensitivity analysis revealed that relative magnetic permeability has the highest normalized sensitivity ($|S| \approx 0.93$) but negligible practical relevance because of the non-magnetic nature of tissue; tissue thickness ($|S| = 0.11$ – 0.20 , inverse) and electrical conductivity ($|S| = 0.10$ – 0.19) are the most influential parameters in practice, while permittivity has the weakest influence. Muscle, fat, and skin were the most sensitive tissues, and cartilage exposure was governed mainly by geometry. These findings identify which parameters must be characterized most carefully in PEMF dosimetry and provide a basis for future work involving image-based patient-specific knee models, frequency-dependent and anisotropic tissue properties, global sensitivity analysis including the induced electric field and current density, experimental validation, and optimization of coil design and excitation parameters for knee pain management.

Acknowledgment

The author thanks the Department of Electrical and Electronic Engineering, Abubakar Tafawa Balewa University, Bauchi, for providing computational facilities..

References

- [1] D. J. Hunter and S. Bierma-Zeinstra, “Osteoarthritis,” *Lancet*, vol. 393, no. 10182, pp. 1745–1759, Apr. 2019.
- [2] M. S. Markov, “Expanding use of pulsed electromagnetic field therapies,” *Electromagn. Biol. Med.*, vol. 26, no. 3, pp. 257–274, 2007.
- [3] L. Massari *et al.*, “Biophysical stimulation of bone and cartilage: State of the art and future perspectives,” *Int. Orthop.*, vol. 43, no. 3, pp. 539–551, 2019.
- [4] K. Varani *et al.*, “Characterization of adenosine receptors in bovine chondrocytes and fibroblast-like synoviocytes exposed to low frequency low energy pulsed electromagnetic fields,” *Osteoarthritis Cartilage*, vol. 16, no. 3, pp. 292–304, 2008.
- [5] J. Tong *et al.*, “The efficacy of pulsed electromagnetic fields on pain, stiffness, and physical function in osteoarthritis: A systematic review and meta-analysis,” *Evid.-Based Complement. Alternat. Med.*, vol. 2022, Art. no. 9939891, 2022.
- [6] L. Cianni *et al.*, “Current evidence using pulsed electromagnetic fields in osteoarthritis: A systematic review,” *J. Clin. Med.*, vol. 13, no. 7, Art. no. 1959, 2024.

- [7] Y.-S. Chang, C.-Y. Lin, and W.-C. Huang, “Pulsed electromagnetic field therapy in people with knee osteoarthritis: A systematic review and meta-analysis,” *Medicina*, vol. 62, no. 4, Art. no. 677, 2026.
- [8] C. Gabriel, S. Gabriel, and E. Corthout, “The dielectric properties of biological tissues: I. Literature survey,” *Phys. Med. Biol.*, vol. 41, no. 11, pp. 2231–2249, 1996.
- [9] S. Gabriel, R. W. Lau, and C. Gabriel, “The dielectric properties of biological tissues: II. Measurements in the frequency range 10 Hz to 20 GHz,” *Phys. Med. Biol.*, vol. 41, no. 11, pp. 2251–2269, 1996.
- [10] S. Gabriel, R. W. Lau, and C. Gabriel, “The dielectric properties of biological tissues: III. Parametric models for the dielectric spectrum of tissues,” *Phys. Med. Biol.*, vol. 41, no. 11, pp. 2271–2293, 1996.
- [11] P. A. Hasgall *et al.*, “IT’IS database for thermal and electromagnetic parameters of biological tissues,” Version 4.1, IT’IS Foundation, Zurich, Switzerland, 2022, doi: 10.13099/VIP21000-04-1.
- [12] Y. Diao *et al.*, “Recent advances and future perspective in computational bioelectromagnetics for exposure assessments,” *Bioelectromagnetics*, 2025, doi: 10.1002/bem.70002.
- [13] J.-M. Jin, *The Finite Element Method in Electromagnetics*, 3rd ed. Hoboken, NJ, USA: Wiley-IEEE Press, 2014.
- [14] COMSOL AB, *AC/DC Module User’s Guide, COMSOL Multiphysics v. 6.0*. Stockholm, Sweden, 2021.
- [15] A. Krupa and I. Antoniuk, “Digital representation of tissues in high compute bioelectromagnetics,” *J. Comput. Appl. Math.*, vol. 397, Art. no. 113643, 2021, doi: 10.1016/j.cam.2021.113643.
- [17] A. Saltelli *et al.*, *Global Sensitivity Analysis: The Primer*. Chichester, U.K.: Wiley, 2008.
- [18] J. F. Schenck, “The role of magnetic susceptibility in magnetic resonance imaging: MRI magnetic compatibility of the first and second kinds,” *Med. Phys.*, vol. 23, no. 6, pp. 815–850, 1996.
- [19] R. Plonsey and D. B. Heppner, “Considerations of quasi-stationarity in electrophysiological systems,” *Bull. Math. Biophys.*, vol. 29, no. 4, pp. 657–664, 1967.
- [20] International Commission on Non-Ionizing Radiation Protection, “Guidelines for limiting exposure to time-varying electric and magnetic fields (1 Hz to 100 kHz),” *Health Phys.*, vol. 99, no. 6, pp. 818–836, 2010.