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NUMERICAL ASSESSMENT OF ELECTROMAGNETIC EXPOSURE FROM MOTORCYCLE HELMET COMMUNICATION SYSTEMS

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Resume

In this study was numerically assessed electromagnetic exposure from a 2.4 GHz wireless communication system integrated into a motorcycle helmet. Using CST Studio Suite and an anatomically realistic human head model, the localized SAR were evaluated averaged over 10 g of tissue for five helmet shell materials at transmitter powers from 1 mW to 1 W. At 100 mW, polyimide produced the highest SAR (approximately 1.47 W/kg), whereas the carbon fibre produced the lowest (approximately 0.94 W/kg). All investigated configurations remained below the ICNIRP (International Commission on Non-Ionizing Radiation Protection) localized exposure limit of 2 W/kg for operating powers up to 100 mW, while higher-power scenarios exceeded this limit. The results demonstrate the importance of transmitter power, antenna placement, and helmet material properties for the electromagnetic safety of smart motorcycle helmets.

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1 Introduction

The rapid evolution of intelligent transportation systems (ITS) and the paradigm shift towards smart mobility have fundamentally transformed the design requirements of modern transport safety infrastructure. In the contemporary automotive and vehicular sectors, safety is no longer evaluated solely through the lens of passive mechanical crashworthiness, such as impact absorption and structural integrity. Instead, active safety and real-time connectivity have become core pillars of rider protection. This technological revolution is particularly prominent in the motorcycle domain, where riders remain among the most vulnerable road users (VRUs). To mitigate the high risks of severe injuries in dense or high-speed traffic environments, the integration of advanced wireless communication hardware into personal protective equipment (PPE), specifically smart motorcycle helmets, has experienced an unprecedented surge over the past decade [1-3].

Modern smart helmets are no longer mere passive shells lined with expanded polystyrene; they are complex, wearable technological hubs. These advanced protective systems standardly incorporate short-range Bluetooth transceivers, dynamic mesh communication (DMC) modules for inter-rider coordination, GPS tracking units, and ambient noise-cancelling systems. In the context of vehicle-to-everything (V2X) and vehicle-to-vehicle (V2V) communication networks, these integrated antennas allow motorcyclists to receive *real-time* telemetry, hazard warnings, and collision-avoidance data directly from the transport infrastructure. While these wireless systems undeniably enhance active traffic safety by increasing riders' situational awareness, they simultaneously introduce a new, often-overlooked engineering challenge: continuous exposure of the human body to non-ionising radiofrequency electromagnetic (EM) radiation [4-6].

From a bioelectromagnetic perspective, a wireless communicator operating inside or mounted onto a motorcycle helmet acts as an intentional EM transmitter

situated in the immediate near field of the human head. Unlike commercial cellular smartphones, which are typically placed against the ear intermittently during calls, helmet communication systems often transmit continuously for hours of riding, remaining in proximity to the highly sensitive occipital and temporal lobes of the brain. At microwave frequencies, particularly within the 2.4 GHz industrial, scientific, and medical (ISM) band used by Bluetooth and wi-fi protocols, biological tissues in the head absorb a fraction of the radiated energy. This energy absorption can lead to localised dielectric heating and potential thermal stress in the underlying cranial structures, raising vital questions regarding long-term user health and compliance with international dosimetric safety standards [1, 7-8].

Furthermore, the physical presence of the motorcycle helmet itself introduces severe EM boundary conditions that complicate standard free-space antenna assumptions. The outer shell of a modern helmet is fabricated from specialised structural materials: ranging from cost-effective high-impact thermoplastics like polycarbonate to high-performance advanced composites such as carbon fibre and E-glass fibre. These materials possess drastically different dielectric constants and electrical conductivities. When the embedded antenna radiates, the surrounding helmet shell behaves as a secondary complex reflector and dielectric loader. It can induce wave refractions, surface eddy currents, and localised field focusing, which can significantly alter the near-field radiation pattern and potentially amplify energy deposition (hotspots) within the rider's biological tissues [9-10].

Consequently, evaluating the specific absorption rate (SAR) within an anthropomorphic head model under realistic helmet-loaded conditions is an essential requirement for modern transport safety engineering. In this study was addressed this critical intersection of wireless communication performance, material science, and public health compliance within transportation systems. By deploying the high-resolution numerical modelling using the CST studio suite software (Dassault Systemes SIMULIA, Velizy-Villacoublay, France), combined with the anatomical "HUGO" voxel phantom, in this paper was systematically investigated the localized SAR distribution triggered by a 2.4 GHz dipole communicator.

The primary objective of this article is to quantify how different structural helmet shell materials influence energy deposition across various standard input power levels. Ultimately, this research aims to provide vehicle designers and transport safety engineers with rigorous, predictive data to optimize the integration of wearable wireless hardware, guaranteeing that next-generation smart protective gear protects the rider not only from mechanical impacts but also from electromagnetic hazards.

2 Theoretical background

In this section are presented the foundational EM principles and a review of the existing scientific literature concerning radiofrequency exposure in transport environments. To rigorously evaluate the SAR inside the human head, it is essential to analyze the underlying physics of dielectric heating and frequency-dependent tissue interactions. Furthermore, in this part is provided a comprehensive overview of the biomedical classification of electromagnetic field (EMF) effects, along with the evolution of international regulatory frameworks governing public safety. By bridging fundamental bioelectromagnetic with modern computational dosimetry, this theoretical framework creates the necessary baseline for interpreting the numerical results presented later in this study.

Since the experimental measurement of SAR inside a living human head requires highly invasive procedures (inserting micro-probes into the cranial cavity), physical validation is ethically prohibited and technically restrictive. Consequently, modern dosimetric research relies almost exclusively on computational electromagnetics (CEM) combined with high-fidelity anatomical phantoms. The two most prominent numerical methods deployed for human EMF simulations are the finite-difference time-domain (FDTD) method and the finite integration technique (FIT), the latter being the core solver framework of CST studio suite [10-12]. The accuracy of these numerical models has evolved significantly. Early studies in the late 1990s used homogeneous spherical or elliptical models to approximate the human head, severely underestimating the localized field concentrations arising from the complex internal boundaries of bone and fluid. The development of voxel-based anthropomorphic datasets, such as the "virtual family" and the "HUGO" phantom, revolutionized the field by partitioning the human body into millions of individual cubic cells (voxels), each assigned distinct frequency-dependent permittivity and conductivity values [13-14]. Recent literature on smart protective gear indicates that when an antenna is operated within an enclosed structure, such as a motorcycle helmet, the helmet shell material acts as a secondary boundary condition, potentially causing reflections, dielectric loading, or partial shielding [9]. In the context of modern ITS, evaluating the EM compatibility of wearable rider accessories has transitioned from a secondary design consideration to a strict regulatory mandate. The spatial deformation of the antenna's radiation pattern inside a multi-layered helmet-head configuration underscores the insufficiency of analytical approximations. It justifies the use of advanced full-wave numerical solvers. Furthermore, the continuous structural variance between the standard high-permittivity thermoplastics and advanced conductive carbon composites introduces

complex boundary conditions that radically alter near-field interactions. Consequently, establishing a rigorous theoretical foundation that bridges the dielectric relaxation physics with international safety standards is essential for interpreting and evaluating the subsequent localized SAR behavior.

The interaction between high-frequency EMF and complex biological systems represents a fundamental topic in modern bioelectromagnetic and transport safety engineering. When an organism is subjected to non-ionizing radiation within the microwave spectrum, specifically the 2.4 GHz ISM frequency band utilized by commercial Bluetooth and DMC systems, the primary mechanism of physical interaction is dielectric heating. Biological tissues are fundamentally heterogeneous and contain a vast concentration of polar molecules, predominantly water. At microwave frequencies, the rapidly alternating electric field induces a torque on these dipoles, continuously forcing them to realign with the changing field. This high-frequency rotational movement and the subsequent molecular-level friction dissipate electromagnetic energy into thermal energy, causing a localized temperature to rise within the exposed anatomical structures [1, 15].

To quantitatively evaluate and standardize the amount of radiofrequency (RF) energy absorbed by biological entities, the scientific community universally deploys SAR. Mathematically, the local SAR value at any specific coordinate within a lossy dielectric medium is derived from Maxwell's equations and defined by the following expression:

$$SAR = \frac{\sigma |E|^2}{2\rho} = c_i \frac{dT}{dt}, \quad (1)$$

where σ [S/m] represents the electrical conductivity of the localized tissue layer, E is the peak amplitude of the internal electric field intensity [V/m], ρ [kg/m³] denotes the mass density of the specific biological tissue, c_i is the specific heat capacity of the tissue [J/kg·°C], and dT/dt represents the initial time derivative of the temperature rise. Since the near-field radiation pattern of an embedded helmet antenna is inherently non-uniform, the local electric field exhibits strong spatial gradients. Consequently, international dosimetric standards require that the raw spatial SAR distribution be averaged over a standardized mass of continuous tissue, typically 1 g or 10 g, to smooth out extreme localized anomalies and provide a biologically relevant metric for thermal stress assessment [8, 16-17].

Biological tissues cannot be modelled as simple, ideal dielectrics. From an electromagnetic perspective, they represent a highly dispersive medium whose electrical conductivity and relative permittivity fluctuate dynamically with operating frequency. This frequency-dependent behavior is standardly quantified using the complex relative permittivity formula:

$$\varepsilon^*(\omega) = \varepsilon'(\omega) - j\varepsilon''(\omega) = \varepsilon'(\omega) - j\frac{\sigma}{\omega\varepsilon_0}, \quad (2)$$

Where ε' represents the real permittivity (a measure of the tissue's ability to store electrical energy), ε'' is the imaginary permittivity or loss factor (characterizing the energy dissipation), ω is the angular frequency, and ε_0 is the vacuum permittivity. At 2.4 GHz, the dielectric response of tissues is strongly governed by the dispersion phase, which arises from the dipolar relaxation of free water molecules. This relaxation process is mathematically modelled using the multi-frequency Cole-Cole or Debye formulation:

$$\varepsilon^*(\omega) = \varepsilon_\infty + \frac{\varepsilon_s - \varepsilon_\infty}{1 + (j\omega\tau)^{1-\alpha}}, \quad (3)$$

where ε_∞ is the relative permittivity at the high-frequency limit, ε_s is the static relative permittivity, τ represents the characteristic relaxation time of the tissue, and α is an empirical parameter describing the distribution of relaxation times [8, 13].

The literature comprehensively categorizes the biological consequences of RF exposure into two distinct domains: thermal (established) effects and non-thermal (low-level) effects. Thermal effects occur when the localized rate of EM energy deposition exceeds the biological organism's natural thermoregulatory capabilities. In the human head, metabolic heat dissipation is primarily regulated by forced convection via blood microcirculation. However, certain anatomical regions within the cranial structure, such as the crystalline lens of the eye or specific poorly vascularized areas of the skull, exhibit reduced blood flow and are highly susceptible to localized thermal stress. Prolonged exposure leading to a local temperature increase of more than 1°C can trigger cellular protein denaturation, alter enzyme activity, and cause irreversible tissue damage [18-19]. Conversely, non-thermal effects refer to biological alterations that occur at exposure levels well below those required to produce measurable temperature changes. Over the past decades, a significant body of epidemiological and in vitro research has investigated whether chronic low-level RF exposure can induce neurological symptoms, alter cognitive functions, or provoke electromagnetic hypersensitivity (EHS) [2, 5]. Recent experimental exposure systems using real telecommunication signals further demonstrate the importance of accurately controlled RF exposure conditions in bioelectromagnetic studies [20]. Some laboratory studies have reported subtle cellular transformations, including changes in calcium ion efflux across cell membranes, increased production of reactive oxygen species (ROS), and potential DNA strand breaks under specific exposure conditions [3]. Despite these isolated findings, the large-scale systematic reviews coordinated by the world health organization (WHO) maintain that a definitive, causal link between low-power microwave exposure and long-term carcinogenic or neurological pathologies has not been established conclusively [16]. Nevertheless, because wireless communicators mounted inside motorcycle helmets operate continuously in the immediate near

Table 1 The public exposure SAR limits according to international guidelines [1, 9]

Frequency range	Whole-body average SAR [W/kg]	Localized SAR (head and torso) [W/kg]	Localized SAR (extremities) [W/kg]
100 kHz - 10 MHz	0.08	2.0	4.0
10 MHz - 10 GHz	0.08	2.0	4.0

field of the brain's occipital and temporal lobes, rigorous numerical verification remains a vital precautionary requirement to ensure user safety in the transport sector [6].

As the EM wave emitted by the 2.4 GHz dipole antenna travels from free space into the multi-layered structure of the helmet and human head, it undergoes severe attenuation. The rate of this attenuation determines how deep radiation penetrates the human brain. The penetration depth δ , defined as the distance at which the amplitude of the EMF decays to e^{-1} (where e denotes Euler's number, corresponding to approximately 37% of the initial amplitude) of its initial value at the boundary, is calculated using the following expression:

$$\delta = \frac{1}{\omega} \left[\frac{\mu_0 \epsilon_0 \epsilon'}{2} \left(\sqrt{1 + \left(\frac{\sigma}{\omega \epsilon_0 \epsilon'} \right)^2} - 1 \right) \right]^{-\frac{1}{2}}, \quad (4)$$

where μ_0 is the magnetic permeability of free space. At 2.4 GHz, the penetration depth in high-water-content tissues (such as muscle or grey matter) is approximately 15-20 mm. This implies that most of the RF energy from a helmet communicator is absorbed within the outermost layers of the head, namely the skin, subcutaneous fat, and the skull. This physical limitation underscores the absolute necessity of using high-fidelity voxel phantoms rather than simplified spheres; a simplified model cannot accurately track wave refraction through the thin layer of high-attenuation skin before encountering the low-attenuation skull bone.

To safeguard the public against the proven thermal hazards of RF radiation, specialized international organizations have developed comprehensive exposure limits based on decades of peer-reviewed dosimetric literature. The two primary bodies responsible for establishing these safety guidelines are the international commission on non-ionizing radiation protection (ICNIRP) and the institute of electrical and electronics engineers (IEEE) via the international committee on electromagnetic safety (ICES) [1, 7]. Historically, the exposure thresholds were derived by determining the lowest operational SAR level capable of producing adverse biological effects in animal models—identified as a whole-body average SAR of 4.0 W/kg. By applying a conservative safety factor of 50 for the general public, the global safety threshold for whole-body average exposure was established at 0.08 W/kg and for localized exposure (such as the head and trunk), where higher energy concentrations are permissible due to localized thermoregulation, a threshold of 2.0 W/kg was mandated by ICNIRP, averaged over a 10 g tissue mass. In contrast, early guidelines from the United

States federal communications commission (FCC), based on ANSI/IEEE standards, adopted a stricter local threshold of 1.6 W/kg but averaged over a smaller mass of 1 g of tissue, creating a historical discrepancy in international compliance testing [17-18, 21]. In 2020, ICNIRP published updated guidelines that refined the exposure metrics for high-frequency fields above 6 GHz but maintained the local 2.0 W/kg limit for the 100 kHz to 10 GHz range, reaffirming its adequacy for protecting against localized tissue heating in consumer electronics [1]. In the Slovak republic, these international thresholds are legally codified and enforced via the decree of the ministry of health No. 534/2007 coll., ensuring that any electronic equipment deployed in transport and everyday life satisfies identical safety criteria [7]. Table 1 summarizes the current public exposure limits across the critical frequency domains relevant to consumer communications.

3 Numerical model and numerical simulation set-up

In this section are described the computational methodology and numerical configuration used to analyze EM interactions between the wireless communicator and the human body. To achieve high dosimetric accuracy, a complex simulation environment was established within CST studio suite, combining an advanced anthropomorphic phantom with precise geometric models of protective transport gear. The primary objective is to replicate realistic operational conditions of motorcycle communication systems and evaluate how different structural materials influence energy absorption in biological tissues.

To accurately evaluate the spatial distribution of the electromagnetic field and the resulting local SAR, a high-resolution anatomical voxel model, "HUGO", was deployed within the CST environment. The human head model was extracted from the high-fidelity voxel dataset to represent realistic biological boundaries. The tissue parameters, including electrical conductivity, mass density, and relative permittivity, were configured to match the international dielectric standard databases for human tissues at the targeted microwave frequency band of 2.4 GHz, Table 2.

The solver accounts for the distinct boundaries between these layers, as the high-water content in tissues like the muscle or brain significantly accelerates energy absorption compared to the low-conductivity bone structures. This high spatial resolution ensures

Table 2 Dielectric properties of key human head tissues at $f = 2.4$ GHz [13]

Tissue	Relative Permittivity ϵ_r [-]	Conductivity σ [S/m]	Mass Density ρ [kg/m ³]
Skin	38.1	1.46	1100
Muscle	52.7	1.74	1040
Bone (Skull)	11.4	0.39	1850
Brain (Grey Matter)	48.9	1.81	1040
Brain (White Matter)	36.2	1.22	1040
CSF (Cerebrospinal Fluid)	66.2	3.46	1000

that the computational model can precisely determine the location of maximum energy deposition and calculate localized thermal stress without undergoing simplification errors.

The motorcycle helmet geometry was imported into the computational domain using a standardized stereolithography (STL) surface mesh format. To investigate the EM attenuation and potential shielding behavior of the outer protective gear, five frequently utilized structural materials for helmet shell manufacturing were assigned and compared, Table 3: polycarbonate, a widely used impact-resistant thermoplastic polymer; polyimide, a high-performance polymer known for its thermal stability and structural strength; teflon (PTFE, a synthetic fluoropolymer of tetrafluoroethylene), a low-permittivity fluoropolymer utilized as a comparative baseline; carbon fibre, a lightweight, highly rigid conductive composite matrix designed for premium impact absorption; e-glass fibre, an industrial glass-reinforced composite bound with epoxy resin.

The numerical helmet model was simplified to specifically investigate the influence of the helmet material on electromagnetic energy absorption in the human head. The helmet geometry was imported from an STL model and resized to fit the anatomical HUGO head model. The same helmet geometry was maintained across all simulations, while the helmet material was varied among polycarbonate, polyimide, PTFE, carbon fibre, and E-glass fibre. The numerical model did not explicitly include the expanded polystyrene (EPS) impact-absorbing liner as a separate dielectric layer. Consequently, the model represents a simplified single-material helmet configuration rather than the complete multilayer structure of a real motorcycle helmet. This simplification should be considered when the absolute SAR values are interpreted. However, because the geometry, antenna position, operating frequency, and excitation conditions were identical across all investigated configurations, the model enables a consistent comparative assessment of how the selected helmet materials affect the resulting SAR distribution. Future work will investigate the effect of a realistic multilayer helmet structure, including an EPS liner with defined thickness and dielectric properties.

The combined numerical model, consisting of the anthropomorphic head model, the motorcycle helmet

shell, and the embedded dipole communicator, was analyzed using the high-frequency time-domain solver based on the FIT, Figure 1. The boundary conditions were explicitly set to “open (space)” in all directions to establish an unrestricted free-space radiation environment and prevent artificial reflections back into the simulation volume.

3.1 Antenna configuration and computational settings

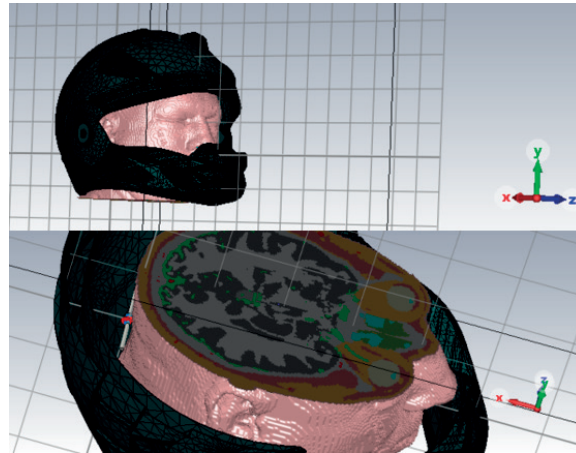
To ensure reproducible numerical exposure assessments, the geometrical configuration of the radiating source relative to the anatomical head model were kept identical for all investigated helmet materials. A half-wave dipole antenna operating at 2.4 GHz was positioned inside the helmet adjacent to the ear region of the anatomical head model. The antenna had a total dipole length of 55.76 mm, a diameter of 1.5 mm, and a 10 mm long central feeding section. The feed impedance was set to 73 Ω . The minimum distance between the radiating element and the head phantom surface was approximately 10 mm. Since the original simulation documentation did not record the antenna’s absolute Cartesian coordinates in the CST global coordinate system, its position was specified here by its anatomical location and separation from the head surface.

The centre of the dipole was set as the reference point, with the origin of the coordinate system $O = (0, 0, 0)$ at the centre of the head; x-axis: left-right, y-axis: posterior-anterior, z-axis: inferior-superior, and the antenna positioned near the ear, at a distance of 10 mm from the head surface. The dipole was oriented approximately in the superior-inferior direction, corresponding to the actual situation.

The geometrical position of the antenna, the head phantom, and the helmet was fixed throughout the simulation series; only the material assigned to the helmet shell was changed. Consequently, differences in the calculated SAR distributions can be attributed primarily to the electromagnetic properties of the investigated helmet materials rather than to changes in source geometry. The anatomical head model was derived from the HUGO model of the CST Voxel Family and included heterogeneous biological structures such as skin, bone, muscle, brain tissue, and blood.

Table 3 Physical and dielectric parameters of simulated motorcycle helmet shell materials [10]

Material	Relative Permittivity ϵ_r [-]	Conductivity σ [S/m]	Mass Density ρ [kg/m ³]
Polycarbonate	2.90	0.001	1200
Polyimide	3.40	0.002	1420
PTFE	2.10	0.000	2200
Carbon Fibre	5.50	10.500	1750
E-Glass Fibre	4.80	0.004	2580

**Figure 1** Numerical simulation model: anthropomorphic human head phantom model with helmet, CST Studio Suite

The dipole antenna was designed for operation at 2.4 GHz and evaluated over the frequency range of 2.3-2.5 GHz. Before the exposure simulations, the antenna model was verified separately by evaluating its reflection coefficient and radiation pattern. At the operating frequency of 2.4 GHz, the simulated reflection coefficient reached approximately -31.37 dB, indicating good impedance matching at the selected operating frequency. The antenna input power varied from 0.001 W to 1 W during the SAR calculations. The values of 0.001 W, 0.01 W, and 0.1 W represented the power levels considered for Bluetooth operation in the original simulation study. In contrast, we included higher power levels to investigate SAR dependence on increasing antenna power.

The computational domain was terminated using Open boundary conditions. The simulation included monitors for the far field, electric field (E-field), magnetic field (H-field), and power loss density. The specific absorption rate was evaluated using the CST SAR Calculation function with spatial averaging over 10 g of tissue.

To make the field interactions and calculate the SAR, localized field monitors were activated for the electric field, magnetic field, and power loss density. The internal SAR calculation routine was configured to average the power absorption over a standardized 10 g mass of continuous tissue, ensuring direct compatibility with regional public safety guidelines. The simulations were executed sequentially by sweeping the input power level across standard Bluetooth operational ranges.

3.2 Mesh and numerical settings

Electromagnetic simulations were performed at an operating frequency of 2.4 GHz. The computational domain used open (space) boundary conditions, allowing electromagnetic waves to propagate out of the domain with minimal artificial reflections. The numerical model included the anatomical HUGO head model, the helmet geometry, and the dipole antenna, and the same geometrical configuration was maintained for all investigated helmet materials. The electromagnetic field evaluation included electric-field (E-field), magnetic-field (H-field), far-field, and power-loss-density monitors. The specific absorption rate was calculated using the CST SAR evaluation procedure with spatial averaging over 10 g of tissue. The antenna excitation frequency, boundary conditions, head and helmet geometry, antenna position, and SAR averaging procedure remained unchanged throughout the simulation series to ensure consistent comparisons between the investigated material configurations. For an accurate analysis, the following mesh settings were applied: local mesh refinement in areas with intricate geometry, a minimum mesh step size of 0.2 mm, adaptive mesh refinement enabled, and a limit of 200 million mesh cells, ensuring a balance between accuracy and computation time.

The CST utilizes integrated full-wave electromagnetic solvers (specifically the Time Domain Solver) based on the Finite Integration Technique (FIT). The software calculates the electromagnetic field distribution across the entire discretized mesh

grid and then derives the resulting parameters during the post-processing using built-in analytical routines. Manual calculation was therefore unnecessary and less suitable, as the software accounts for complex boundary conditions, material losses, and high-frequency effects that standard analytical equations cannot fully capture.

All the helmet configurations were evaluated using an identical computational setup, keeping the head geometry, antenna position, excitation conditions, computational domain, and SAR evaluation procedure unchanged. The open boundary conditions were applied and the local SAR were evaluated using 10-g mass averaging. We assessed numerical consistency across twelve input-power levels from 0.001 to 1 W. For all materials, SAR exhibited the expected linear scaling with power; for example, the maximum 10-g averaged SAR for polycarbonate increased from 0.0145407 W/kg at 0.001 W to 1.45407 W/kg at 0.1 W and 14.5407 W/kg at 1 W. Similar proportional behavior was observed for the other materials, providing an internal numerical-consistency check.

The numerical model also underwent internal verification and regulatory benchmarking. Before integrating it into the helmet-head model, the half-wave dipole antenna was evaluated separately in CST Studio Suite over 2.3–2.5 GHz. The antenna was 55.76 mm long and 1.5 mm in diameter, with a 10-mm feed section and 73 Ω feed impedance. A return loss of –31.37 dB at 2.4 GHz confirmed appropriate antenna tuning. The primary simulations employed 10-g averaged SAR and were compared with the 2 W/kg local-head exposure limit. Additional 1-g SAR calculations for polycarbonate and E-glass yielded 0.0234611 and 0.0210642 W/kg at 0.001 W, and 2.34611 and 2.10642 W/kg at 0.1 W, respectively, allowing comparison with the 1.6 W/kg limit considered for the USA, Canada, and Korea. These procedures provided internal verification and regulatory benchmarking but did not constitute experimental validation.

4 Simulation results and discussion

The numerical simulation results provided highly granular insights into how different structural boundaries modify the near-field coupling between the 2.4 GHz transceiver and the heterogeneous tissues of the “HUGO” anatomical head phantom. Each case was analyzed independently across the designated input power spectrum.

Case 1: Polycarbonate shell configuration

Polycarbonate is one of the most widely deployed thermoplastics in the motorcycle helmet industry due to its excellent impact resistance and economic manufacturability. However, from an EM perspective, its low relative permittivity ($\epsilon_r = 2.90$) and near-zero electrical conductivity ($\sigma = 0.001$ S/m) render it almost entirely transparent to microwave radiation.

During the simulations, the wavefront emitted by the half-wave dipole antenna passed through the 3.0 mm Polycarbonate boundary layer with negligible insertion loss or reflection. Since, the shell fails to attenuate the reactive near-field, a massive portion of the radiated energy couples directly into the high-permittivity layers of the human scalp. At a standard high-power transmission of 100 mW, this configuration yielded a peak localized SAR of 10 g of 1.4520 W/kg, Figure 2. The resulting EM hotspot was intensely concentrated in the skin and underlying muscle tissues directly adjacent to the antenna feed point, showing minimal spatial dispersion.

Case 2: Polyimide shell configuration

The polyimide simulation case represents a high-performance polymer matrix often utilized in premium specialized headgear. Chemically and structurally distinct from polycarbonate, its higher relative permittivity ($\epsilon_r = 3.40$) and slightly elevated conductivity ($\sigma = 0.002$ S/m) introduce a subtle shift in the wave propagation dynamics.

The numerical results indicate that polyimide

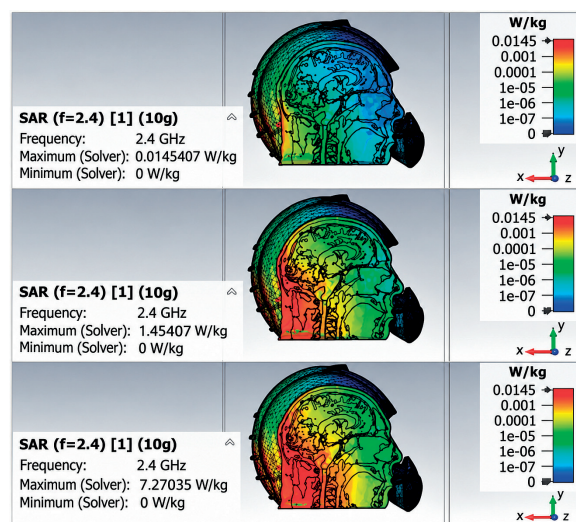


Figure 2 Numerical simulation results: polycarbonate helmet, antenna power of 0.001W, 0.1W, and 0.5W

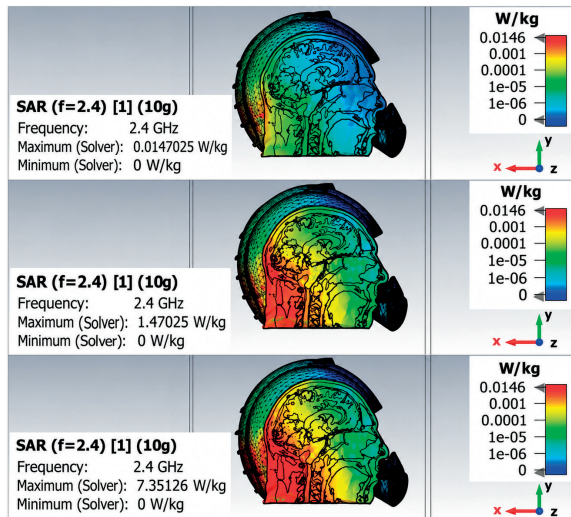


Figure 3 Numerical simulation results: polyimide helmet, antenna power of 0.001W, 0.1W, and 0.5W

presents the worst-case scenario for biological exposure among all five tested materials. At 100 mW input power, the peak localized SAR of 10g reached 1.4710 W/kg, the highest absolute value recorded in the medium-power tier, Figure 3. This phenomenon occurs because the specific permittivity of polyimide creates a subtle impedance matching effect between the free-space air domain and the high-permittivity biological phantom. Instead of reflecting the wave, the polyimide shell acts as an anti-reflective transition layer, effectively “pulling” more electromagnetic energy out of the antenna near-field and steering it straight into the occipital region of the head model.

Case 3: PTFE shell configuration

Teflon was introduced into the computational matrix as a low-loss, low-permittivity baseline calibration material ($\epsilon_r = 2.10$, $\sigma = 0.00$ S/m). Since the teflon possesses a permittivity extremely close to that of a free space, it introduces virtually no dielectric loading to the antenna structure. The simulations showed that the dipole maintained its ideal free-space resonance and S11 reflection coefficient. The wave passed through the shell without undergoing internal refraction or scattering. At 100 mW, the peak SAR was stabilized at 1.3240 W/kg, Figure 4. While this value is lower than that of Polycarbonate and Polyimide, the shape of the internal hotspot within the grey matter was found to be more deeply penetrated, as the lack of any material-induced scattering allowed the primary radiation beam to travel further into the cranial cavity before undergoing natural biological attenuation.

Case 4: Carbon fibre composite configuration

The carbon fibre simulation case introduced a radical shift in the EM environment due to the material's high macro-conductivity ($\sigma = 10.5$ S/m) and elevated permittivity ($\epsilon_r = 5.50$). Unlike the transparent thermoplastics, the conductive carbon fibre matrix behaves as a lossy, reflective EM shield.

When the 2.4 GHz field vectors encountered the

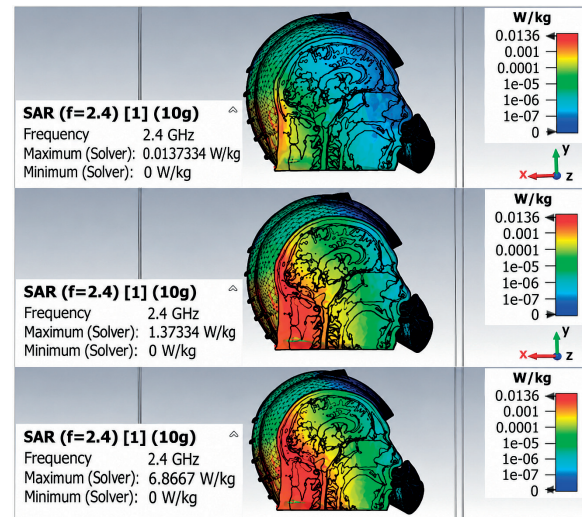


Figure 4 Numerical simulation results: PTFE helmet, antenna power of 0.001W, 0.1W, and 0.5W

Carbon Fiber shell, intense surface eddy currents were instantaneously induced within the shell mesh. These currents generated a powerful secondary magnetic field that actively opposed the incoming waves, triggering a massive back-reflection away from the human head. Consequently, at 100 mW, the peak localized SAR plummeted to just 0.9410 W/kg, an exceptional 35.3% reduction compared to polyimide, Figure 5. The visual hotspot analysis showed that the energy was successfully blocked from entering the deeper brain structures, with the remaining fraction being dissipated harmlessly as ohmic losses within the structural composite material itself. This pronounced EM shielding effectiveness highlights the role of skin-depth attenuation within the carbon fibre-reinforced polymer lattice. The high density of the interwoven conductive fibres creates an effective Faraday cage approximation, which severely restricts the reactive near-field components from capacitively coupling with the lossy skin phantom.

Case 5: E-glass fibre composite configuration

The E-glass fibre represents a standard non-conductive fiberglass composite structural material ($\epsilon_r = 4.80$, $\sigma = 0.004$ S/m). While it lacks the high electrical conductivity of carbon fibre, its remarkably high relative permittivity slows the electromagnetic wave significantly as it passes through the shell. This reduction in phase velocity causes severe wave refraction and localized scattering at the material-tissue interface. Instead of a single, sharp energy beam piercing the head phantom, the E-glass shell diffused the incoming wavefront over a significantly larger geometric surface area. This spatial defocusing successfully reduced the hotspot's peak intensity. At 100 mW, the peak localized SAR remained at a highly safe level of 1.0130 W/kg, demonstrating that high-permittivity dielectric scattering is an extremely effective passive method for reducing human dosimetric exposure without the expense of conductive shields, Figure 6.

The acquired results indicate a direct linear

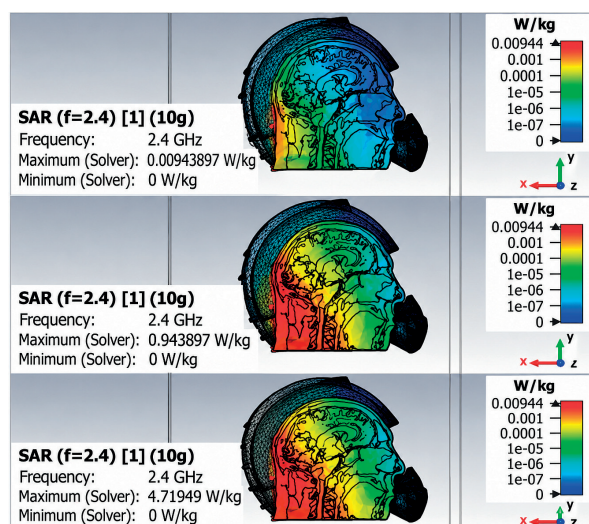


Figure 5 Numerical simulation results: carbon fibre, helmet, antenna power of 0.001W, 0.1W, and 0.5W

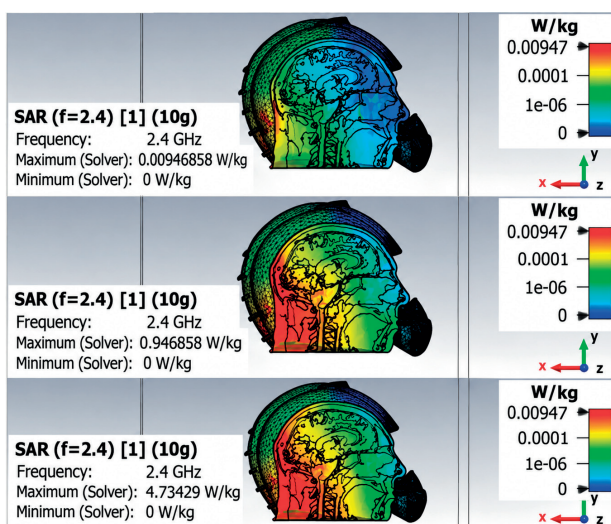


Figure 6 Numerical simulation results: E-glass fibre helmet, antenna power of 0.001W, 0.1W, and 0.5W

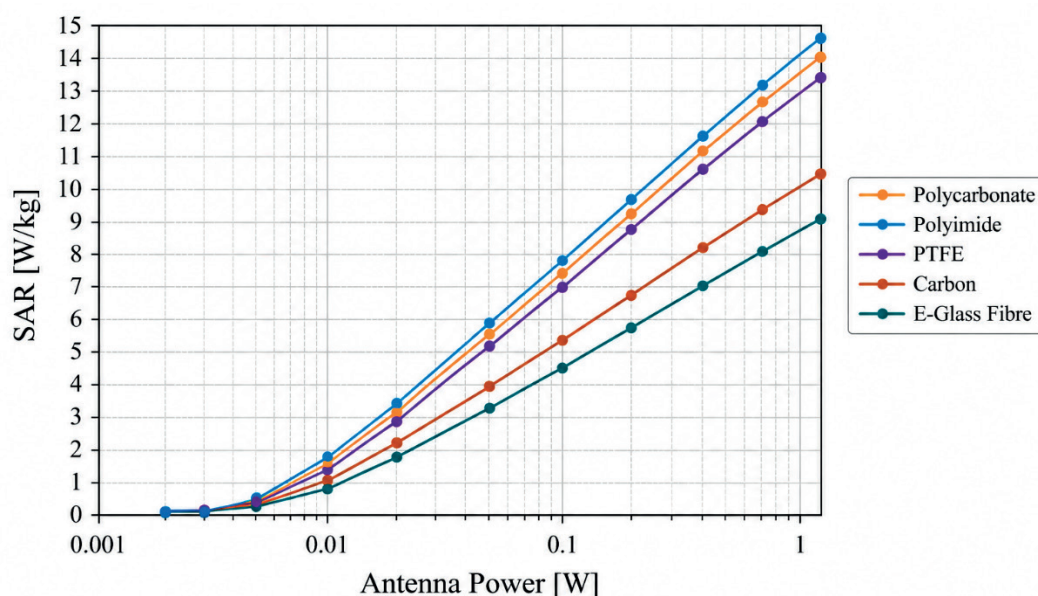


Figure 7 Comparison of the peak localized SAR as a function of antenna input power for the investigated helmet shell materials

correlation between the antenna input power and the peak SAR values across all investigated configurations. However, the dielectric and conductive properties of the helmet outer shell play a decisive role in modifying the electromagnetic near-field distribution. As shown in the data, carbon fibre and E-glass fibre exhibit the greatest EM attenuation. At a standard high-power transmission of 100 mW, the carbon fibre helmet structure reduced the peak SAR to 0.941 W/kg, approximately 35% lower than the polyimide shell (1.471 W/kg). This phenomenon is attributed to the relatively high electrical conductivity ($\sigma = 10.5$ S/m) of the carbon composite matrix, which acts as a partial EM shield, reflecting and absorbing a portion of the radiated energy within the shell structure itself before it penetrates the biological phantom.

The linear relationship between the antenna input power and the resulting tissue exposure highlights a critical

engineering constraint for transport communication hardware, Figure 7. The SAR progression showed the expected approximately linear dependence on antenna input power under unchanged geometrical and material conditions, providing an internal numerical consistency check of the simulation results. The divergence in the material becomes increasingly pronounced as the system transitions into higher power brackets. At the critical 500 mW tier, the shielding delta between the conductive carbon fibre mesh (4.705 W/kg) and the transparent polyimide polymer (7.355 W/kg) stretches to an absolute value of 2.650 W/kg. This massive gap emphasizes that under the extreme operating states, proper material selection alone can eliminate nearly a third of the thermal stress imposed on the user's cranial tissues, providing a critical baseline for the design of bio-compatible smart transport accessories.

5 Conclusion

In this study was numerically investigated the electromagnetic exposure of an anatomically realistic human head model caused by a 2.4 GHz wireless communication system integrated into a motorcycle helmet. Using the FIT-based time-domain solver in CST Studio Suite, we evaluated the influence of five helmet shell materials: polycarbonate, polyimide, PTFE, carbon fibre, and E-glass fibre on localized SAR averaged over 10 g of tissue under identical geometrical and excitation conditions.

The results demonstrate a strong dependence of SAR on both the transmitter input power and the electromagnetic properties of the helmet shell. At 100 mW, polyimide produced the highest peak of approximately 1.47 W/kg, whereas carbon fibre reduced the value to approximately 0.94 W/kg. Carbon fibre and E-glass fibre therefore provided the greatest reduction in the peak tissue exposure among the investigated materials. As expected for a linear electromagnetic model with unchanged geometry and material properties, SAR increased approximately proportionally with the applied antenna input power.

For the transmitter powers up to 100 mW, all investigated configurations remained below the ICNIRP

localized exposure limit of 2 W/kg averaged over 10 g of tissue. Conversely, the extended high-power scenarios demonstrated that passive attenuation by the helmet shell alone is insufficient to ensure compliance when substantially higher transmitter powers are used close to the head. Therefore, the electromagnetic safety of future smart helmets should be addressed through a combined optimization of transmitter power, antenna position and directivity, and helmet material composition. Future studies focus on multilayer helmet structures, experimental SAR benchmarking, and additional communication frequency bands.

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Conflicts of interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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