Carbon Nanomaterials as Functional Fillers in Polymer Scaffolds for Bone and Joint Healing

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Abstract

This review examines recent developments in the use of carbon nanomaterials (carbon nanotubes (CNTs) and graphene-based materials) as functional fillers in polymer scaffolds for bone and joint healing. Organized by fabrication–structure-property-function linkages, the review synthesizes how dispersion methods and scaffold manufacturing techniques (e.g., three-dimensional (3D) printing and electrospinning) govern percolation behavior and, in turn, enable piezoresistive self-sensing and the delivery of low-amplitude electrical cues during mechanical loading. Compiled are representative device-level measurements reported in the literature (including conductivity versus filler fraction, piezoresistive gage factors under cyclic strain, impedance spectra, and elastic moduli), and evidence is summarized linking these properties to early osteogenic and chondrogenic responses. Emerging design guidelines are distilled (such as operating at filler loadings near the percolation threshold, controlling nanoscale dispersion and interfacial bonding, and balancing electrical functionality with biocompatibility), alongside translational considerations (dispersion stability, cytotoxicity thresholds, wear debris in articulating environments, and manufacturing scalability). Current challenges are identified (notably the lack of standardized testing protocols, uncertainties in long-term *in vivo* safety, and data fragmentation), and open research gaps are highlighted, including the need for comparative datasets across materials and architectures. In general, nanocarbon-filled scaffolds represent a promising convergence of nanotechnology and regenerative engineering, creating mechanically competent, electrically active, and biologically instructive implants for complex bone and joint repair.

Contents

1 Introduction

Large bone and osteochondral defects resulting from trauma, tumor resection, or degenerative diseases remain a significant clinical challenge. Autografts (patient-derived bone) and allografts (donor bone) are still considered the clinical gold standard for reconstruction but are limited by donor-site morbidity, finite supply, and variable integration [13]. Metal and ceramic implants can provide mechanical stability, yet are often too stiff and bioinert relative to native bone, resulting in stress shielding and long-term complications such as aseptic loosening [13, 19].

Tissue engineering seeks to bridge this gap by combining scaffolds, cells, and bioactive factors to regenerate bone and cartilage. Ideal scaffolds must be porous, mechanically competent, biocompatible, and gradually degradable, allowing native tissue to remodel and replace the scaffold over time. They should also support vascularization, facilitate nutrient diffusion, and integrate with host tissue [1, 2].

Recent work has shown that incorporating carbon nanomaterials, primarily CNTs and graphene-based derivatives, into polymer scaffolds offers a powerful pathway for engineering multifunctional constructs [12, 2, 14]. Carbon nanomaterials introduce three capabilities that are difficult to achieve simultaneously with conventional scaffold materials. First, they provide mechanical reinforcement: CNTs and graphene have exceptional stiffness and strength (Young's moduli on the order of 1 TPa), so even low filler fractions can significantly stiffen relatively soft polymers [12]. Second, they introduce electrical conductivity: when properly dispersed, nanocarbon fillers form percolating networks that convert insulating polymers into conductive or semiconductive scaffolds, allowing electrically active cell culture environments and electrotherapeutic strategies [17, 16]. Third, they create high-surface-area biointerfaces: the nanoscale roughness and large specific surface area of CNTs and graphene support protein adsorption and focal adhesion formation, which can promote cell attachment and osteogenic or chondrogenic differentiation [2, 15]. Figure 1 situates these clinical and materials challenges within the hierarchical structure of bone and a CT-based 3D-printing workflow, highlighting the multiscale context in which nanocarbon-reinforced scaffolds must operate.

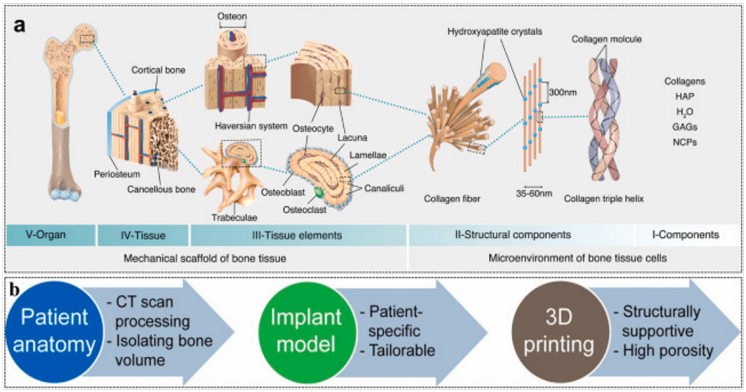

Figure 1: Schematic of bone's hierarchical structure (organ to nanoscale) and a CT-based 3D-printing workflow for patient-specific porous scaffolds.

2 Carbon Nanomaterials in Bone Scaffolds

2.1 Carbon nanotubes: structure and properties

Carbon nanotubes are cylindrical nanostructures formed by rolling one or more graphene sheets into a tube. Single-walled CNTs (SWCNTs) consist of a single graphene cylinder, whereas multi-walled CNTs (Multi-Wall Carbon Nanotubes) are nested concentric cylinders [12]. The electronic properties of a CNT depend on its atomic structure (chirality) and diameter; as a result, CNTs can be metallic or semiconducting [12].

Mechanically, CNTs exhibit extremely high Young's modulus (on the order of 1 TPa) and very high tensile strength, combined with low density. Their high aspect ratio (lengths from hundreds of nanometers up to many micrometers, with diameters of a few nanometers) and large surface area make them effective nanoreinforcements in polymer matrices [12, 19]. Electrically, CNT networks can become conductive at relatively low volume fractions, enabling nanocomposite scaffold conductivities in the range needed for electrical stimulation of cells [12, 17]. The way in which such CNT and graphene networks embed within polymer scaffolds and present electroactive interfaces to adherent cells is schematically summarized in Figure 2, which emphasizes the dual mechanical and electrical roles of these fillers in a bone defect setting.

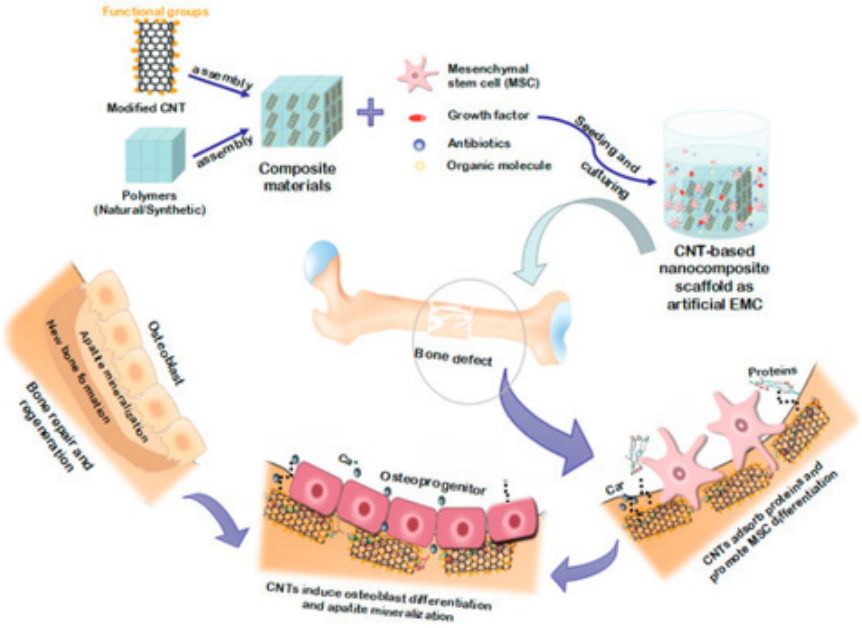

Figure 2: Conceptual illustration of CNTs and graphene as functional fillers in polymer scaffolds for bone defects: the nanocarbon network reinforces the scaffold and provides electroactive interfaces that guide stem cells toward osteogenic and chondrogenic lineages.

2.2 Graphene and graphene oxide

Graphene is a single atomic layer of sp^2-bonded carbon atoms arranged in a hexagonal lattice. It shares many extraordinary properties with CNTs, including an in-plane stiffness of approximately

1 TPa and excellent electrical and thermal conductivity [2]. Graphene oxide (GO) is an oxidized form of graphene decorated with oxygen-containing functional groups, such as hydroxyl, epoxy, and carboxyl moieties, which improve its dispersibility in water and polar solvents and enhance its interactions with polymers and biological molecules [2, 15]. GO, which is insulating or semiconducting, can be partially reduced to form reduced graphene oxide (rGO), which is more electrically conductive.

Graphene-based fillers, particularly GO and rGO, have been widely explored in bone scaffold applications because they can be processed from solution, easily incorporated into electrospun fibers or 3D-printed inks, and tuned between more insulating and more conductive states. At low doses, graphene derivatives have been reported to promote osteogenesis and exhibit favorable immunomodulatory and antimicrobial properties [14, 15].

2.3 Role as functional fillers in polymer matrices

When CNTs or graphene are embedded in biodegradable polymer matrices such as PLA, PLGA, or PCL, they act as nanoreinforcements and conductive pathways within the scaffold. By forming a nanoscale interpenetrating network, the fillers can simultaneously enhance the mechanical stiffness and introduce electrical conductivity into an otherwise non-conductive polymer scaffold [12, 16]. The nanocarbon network can be viewed as a nanoscale "rebar" that both shares mechanical load and carries electronic signals.

In addition, CNTs and graphene provide nanoroughness and chemically active surfaces that facilitate the adsorption of adhesion proteins, such as fibronectin and vitronectin, and growth factors. These interfaces can bias stem cell fate toward osteogenic and chondrogenic lineages [2, 15]. As conceptually illustrated in Figure 2, CNT bundles and graphene sheets are combined with natural or synthetic polymers to form composite materials that are processed into scaffolds filling bone defects.

2.4 Compatibility with common scaffold polymers

Incorporating CNTs and graphene into a broad range of scaffold-forming polymers has been successful. One primary class comprises synthetic polyesters such as poly(lactic acid) (PLA), poly(lactic-co-glycolic acid) (PLGA), and poly(caprolactone) (PCL), which are widely used scaffold materials due to their biodegradability and established safety profiles. CNT- or GO-reinforced PLGA and PCL electrospun fibers and 3D-printed lattices show improved stiffness and enhanced osteogenic responses compared to neat polymer controls [3, 8, 16]. A second class includes natural polymers such as collagen, gelatin, chitosan, and silk fibroin, which can be combined with CNTs or graphene (often in combination with hydroxyapatite) to create hybrid scaffolds that more closely mimic the native extracellular matrix [1, 15]. A third category is composite systems in which polymers are combined with hydroxyapatite or other calcium phosphates and further reinforced with CNTs or graphene, thereby providing both osteoconductive mineral phases and electroactive nanocarbon phases within a single construct [1, 14].

Surface functionalization of CNTs, for example, by carboxylation, and the use of GO instead of pristine graphene generally improve compatibility by enhancing dispersibility, reducing aggregation, and providing sites for covalent or noncovalent bonding with polymer chains [5, 2].

3 Fabrication Methods

3.1 Three-dimensional printing

Three-dimensional (3D) printing, including extrusion-based additive manufacturing and fused deposition modeling, enables precise control over scaffold geometry and architecture. Patient-specific scaffolds can be designed from CT scan data, with pore size, porosity, and strut geometry tuned to match mechanical and mass transport requirements for a given bone defect [9].

Nanocarbon–polymer composites can be formulated as printable inks or filaments for 3D printing. Seyedsalehi and co-workers [16], for example, used a PCL resin with 0.5–3 wt% rGO to 3D-print porous scaffolds, achieving uniform pore architectures and well-distributed graphene within the printed struts. Mechanical testing revealed substantial increases in compressive stiffness and strength at only 0.5 wt% rGO, with diminishing returns at higher loadings [16]. These results suggest that even a small amount of graphene can significantly reinforce the scaffold, but beyond a certain point, additional filler yields little benefit and may complicate processing. Representative CNT-functionalized 3D-printed lattices with varying morphology and associated patterns of cell coverage, as depicted in Figure 3, visually reinforce this link between composite formulation, print architecture, and device-level biological performance.

The translational interest in nanocarbon-enabled 3D printing is underscored by patents on graphene-based printing inks [4] and on CNT–polymer composite scaffolds for bone tissue engineering [6].

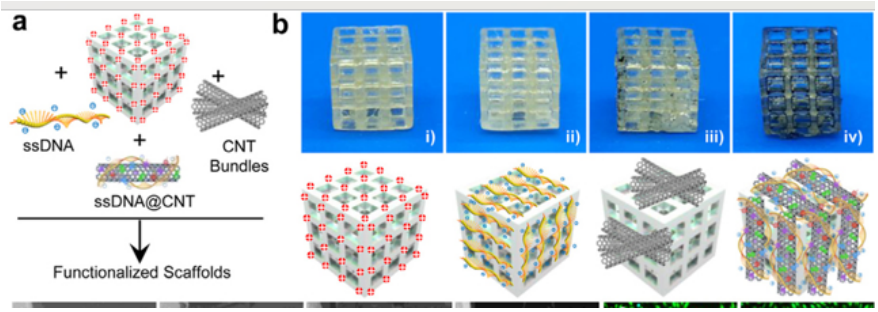

Figure 3: Examples of fabricating CNT/graphene scaffolds: CNT-functionalized 3D-printed PLA lattices with changing morphology and cell coverage.

3.2 Electrospinning

Electrospinning uses a high voltage to draw a polymer solution jet into ultrafine fibers, typically 100–1000 nm in diameter, that collect as a nonwoven mat. The resulting fibrous architecture closely resembles the native extracellular matrix of bone and cartilage [2].

CNTs or GO can be incorporated into the polymer solution prior to electrospinning or deposited onto fibers post-spinning. Fu et al. [3] electrospun PLGA/hydroxyapatite (HAp) nanofibers containing GO and observed enhanced osteoblast proliferation and mineralization compared to GO-free fibers. Luo et al. [8] showed that GO-incorporated PLGA nanofiber mats promoted mesenchymal stem cell proliferation and upregulated early osteogenic markers, including alka-

line phosphatase and Runx2, relative to GO-free mats. In both cases, the addition of graphene oxide improved both mechanical stiffness and cellular responses.

One limitation of pure electrospun mats is their relatively low bulk mechanical strength. While electrospun architectures are excellent for cell attachment and surface-driven phenomena, they may require reinforcement for load-bearing applications [1]. The diversity of fiber diameters, alignments, and array configurations achievable by electrospinning is illustrated in Figure 4, underscoring how closely this fabrication route can mimic the nano- to microscale fibrous morphology of native extracellular matrix.

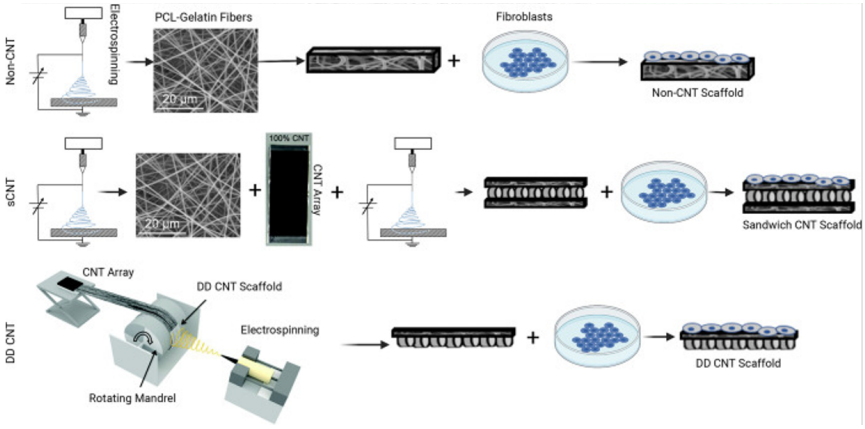

Figure 4: Examples of electrospun fiber scaffolds and arrays.

3.3 Hybrid fabrication strategies

To harness the advantages of both 3D printing and electrospinning, hybrid scaffolds have been developed that combine printed frameworks with electrospun fiber layers. For example, hybrid PCL scaffolds in which electrospun fibers are interleaved with 3D-printed layers exhibit improved hydrophilicity, enhanced preosteoblast adhesion, and elevated expression of osteogenic markers compared with scaffolds produced by either technique alone [9].

Hybrid architectures can be further augmented with CNTs or graphene. Three-dimensionally printed lattice structures can be coated with aligned CNT arrays or wrapped with GO-containing electrospun mats, thereby creating scaffolds with load-bearing cores and bioactive, conductive surfaces, conceptually linking the architectures shown in Figures 3 and 4 [17].

4 Dispersion and Percolation Behavior

4.1 Dispersion strategies

Achieving uniform dispersion of CNTs or graphene within a polymer matrix is critical for consistent mechanical and electrical properties and for minimizing cytotoxicity. Poorly dispersed nanocarbons tend to form agglomerates or bundles that act as stress concentrators and present large reactive surfaces to cells [5].

Standard dispersion techniques include ultrasonication (bath or probe), high-shear mixing, surfactant-assisted dispersion, and chemical functionalization. Kuroda et al. [5] compared conventional sonication to a high-intensity tip sonicator for dispersing tangled Multi-Wall Carbon Nanotubes. They found that the improved method produced much smaller CNT agglomerates, below 200 nm, virtually eliminating cytotoxic effects in vitro. In contrast, conventionally sonicated CNTs remained in larger bundles exceeding $1\,\mu\mathrm{m}$ and significantly reduced cell viability. This comparison demonstrates the importance of the dispersion method to both composite performance and biocompatibility.

Surface functionalization, such as introducing carboxyl groups, and noncovalent wrapping, for example with PEGylated pyrene, enhances nanocarbon wettability and affinity for polymer chains, stabilizing suspensions and reducing re-aggregation [2]. The resulting composites exhibit more homogeneous, reproducible properties.

4.2 Percolation thresholds and conductivity

As the nanocarbon filler fraction increases, the composite transitions from insulating to conductive when an interconnected filler network spans the material. This is the percolation threshold. Slight increases in filler fraction near this threshold can cause conductivity to increase by orders of magnitude [12]. This sharp transition, as well as its anisotropic expression in aligned fibrous composites, is captured in the conductance plots in Figure 5, where direction-dependent conductivity and the log-scale conductance versus CNT content together delineate the percolation regime.

In PCL–rGO scaffolds, Seyedsalehi et al. [16] observed a sharp rise in conductivity around 0.5–1 wt% rGO, corresponding to the formation of a continuous graphene network within printed struts. Similar behavior has been reported for CNT composites, with percolation thresholds often below 1 wt% when Multi-Wall Carbon Nanotubes are well dispersed [12, 17]. In some graphene–polymer systems, percolation at volume fractions as low as approximately 0.3 vol% has been reported [2].

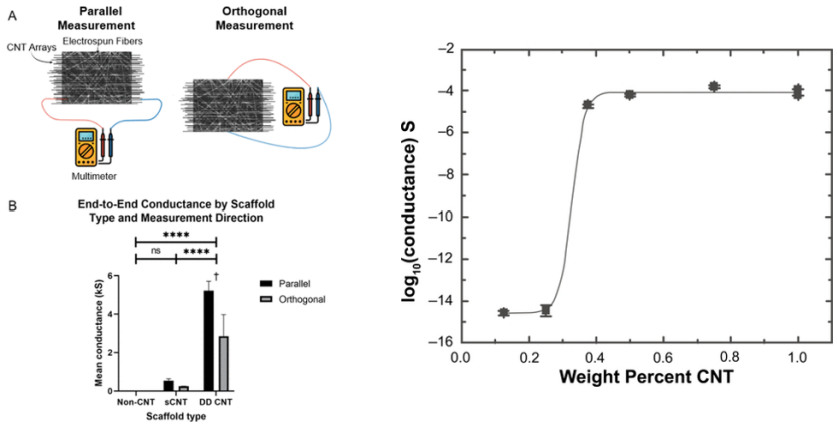

Figure 5: Representative percolation behavior in CNT scaffolds. Left: anisotropic conductance in aligned CNT/electrospun mats, with higher conductance along the fiber alignment direction. Right: log-scale conductance versus CNT weight fraction highlighting the percolation threshold.

4.3 Implications for mechanical integrity and sensing

Near the percolation threshold, small increases in nanocarbon content can dramatically alter a scaffold's electrical properties while keeping filler loading relatively low. This regime, just above percolation, is attractive because mechanical reinforcement remains substantial, piezoresistive behavior emerges, and biocompatibility is easier to maintain. Even at low loadings, well-dispersed CNTs or graphene can substantially increase compressive stiffness and strength [16]. At the same time, deformation of the conductive network under load alters tunneling and contact resistance, enabling strain sensing throughout the scaffold [17]. Designing for the minimum effective filler fraction also reduces the risk of nanocarbon-related toxicity by limiting the total amount of exposed nanomaterial in the construct [5].

In contrast, excessive filler beyond the optimal range can lead to agglomeration, brittleness, and potentially more cytotoxic debris. For this reason, many design strategies target filler fractions just above percolation to balance conductivity, mechanical performance, and safety.

5 Device-Level Functional Properties

5.1 Electrical conductivity

Once a percolating network is established, typical scaffold conductivities range from 10^{-6} to 10^{-2} S/cm, depending on filler type, loading, and processing [12, 17, 16]. These conductivities are sufficient to support direct current stimulation or alternating electric fields in the ranges used by clinical bone stimulators.

The ability to deliver electrical cues via the scaffold itself eliminates the need for separate metallic electrodes and enables more spatially distributed cell stimulation. Several studies report that conductive nanocomposite scaffolds combined with electrical stimulation elicit enhanced osteogenic responses compared to non-conductive controls subjected to the same stimulation protocols [12, 2].

5.2 Piezoresistive sensing

Because nanocarbon networks are sensitive to deformation, CNT/graphene scaffolds are inherently piezoresistive. Under compressive or tensile strain, distances and contact areas between neighboring CNTs or graphene sheets change, modulating tunneling and contact resistance throughout the network [17]. The resulting strain-dependent resistance change can be used to infer applied stress or strain.

Piezoresistive gauge factors, defined as the relative change in resistance per unit strain, for CNT-based composites can reach tens to hundreds in flexible sensor applications [17]. In load-bearing scaffolds, moderate gauge factors are sufficient to monitor changes in the mechanical environment as healing progresses. For example, a CNT/PCL scaffold could be wired to an external readout to provide real-time feedback on load sharing between the implant and regenerating bone, thereby functioning as a self-monitoring implant.

5.3 Mechanical properties

Mechanical performance is crucial for scaffolds intended for load-bearing applications, and CNT and graphene reinforcements significantly enhance mechanical properties when well dispersed. PCL–rGO scaffolds with 0.5 wt% rGO exhibit approximately 185% higher compressive strength and about 150% higher modulus than neat PCL, with little further gain at 1 and 3 wt% [16]. PLA/Multi-Wall Carbon Nanotubes scaffolds achieve moduli comparable to cancellous bone, far exceeding the modulus of PLA alone [12]. Graphene-reinforced 3D-printed lattices likewise show improved stiffness and toughness while retaining high porosity [9]. These improvements in mechanical competence are mirrored at the cellular level, where enhanced cell spreading and coverage on CNT-reinforced scaffolds relative to neat polymer controls, as visualized in Figure 6, indicate that mechanical optimization can proceed without sacrificing cytocompatibility. Matching the scaffold modulus to the target tissue, for example, cancellous versus cortical bone, is important to avoid stress shielding, because overly stiff scaffolds can reduce physiological loading on regenerating bone and thereby impair remodeling.

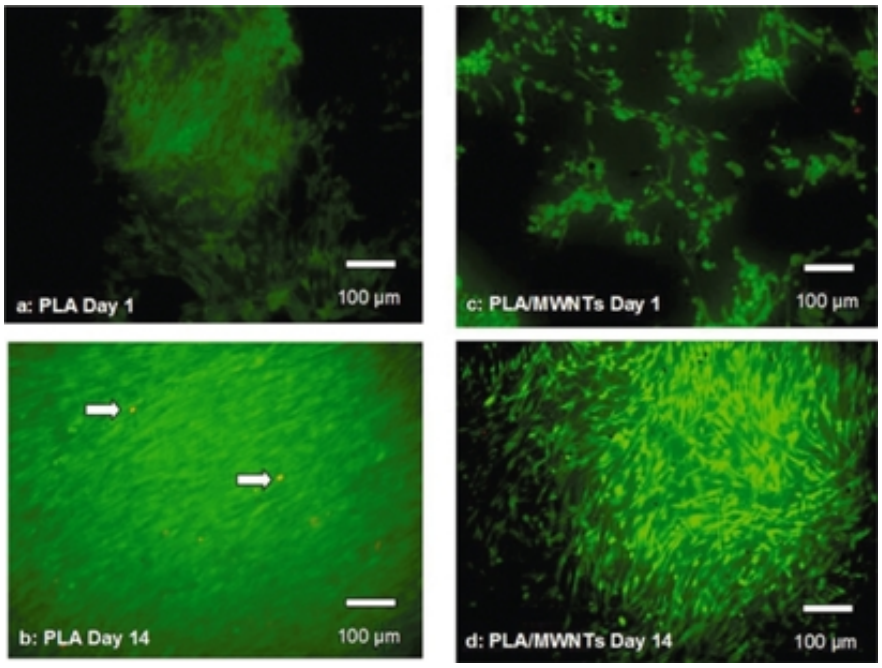

Figure 6: Representative device-level cell response to CNT-reinforced scaffolds: fluorescence images showing improved coverage and spreading of cells on PLA/CNT scaffolds compared to neat PLA over time.

6 Biological Responses

6.1 Cytocompatibility and safety

Polymer/CNT and polymer/graphene scaffolds can be cytocompatible under controlled conditions. Cytocompatibility tends to be maintained or improved when nanocarbons are well purified and free of catalyst residues, when filler fractions are moderate (often a few wt% or less), and when nanocarbons are well dispersed and immobilized mainly in the matrix [5, 2]. Under these conditions, mesenchymal stem cells, osteoblasts, and chondrocytes generally show equal or higher viability on nanocomposite scaffolds compared to neat polymer controls and often exhibit enhanced metabolic activity and spreading [3, 8, 7].

Many studies also report a viability window for nanocarbon loading. Within this optimal range, cell viability and function are preserved or enhanced, whereas higher loadings result in decreased proliferation, increased levels of inflammatory markers, or direct cytotoxicity [2]. This observation motivates the design of scaffolds that operate within a filler content window, delivering functional benefits while minimizing biological risk.

Dose and geometry also play important roles. Long, rigid CNTs can behave similarly to asbestos fibers in some contexts, provoking frustrated phagocytosis and chronic inflammation, whereas shorter, functionalized CNTs are more readily internalized and tolerated [12, 19, 2]. For graphene materials, sheet size and oxidation state influence uptake and immune response: smaller, highly oxidized GO is generally better tolerated than large, pristine graphene sheets [2, 15]. Long-term degradation and fate of nanocarbon debris remain open questions. GO may be partially degraded by oxidative enzymes over time, but pristine graphene and CNTs are largely persistent [2]. More long-term animal studies and careful immune characterization are needed before widespread clinical use.

6.2 Osteogenic differentiation and bone formation

CNT- and graphene-reinforced scaffolds have repeatedly been shown to promote osteogenic differentiation of mesenchymal stem cells and osteoprogenitor cells. Proposed mechanisms include increased stiffness and nano-roughness, which bias cells toward osteogenic lineages via mechanotransduction pathways; enhanced adsorption and presentation of osteogenic factors at the nanocarbon interface; and electrical conductivity, which enables electrostimulation and is known to support bone healing [12, 14]. GO- or rGO-containing PLGA and PCL nanofiber scaffolds elicit higher alkaline phosphatase activity, Runx2 expression, and calcium deposition than polymer-only controls [3, 8]. Multi-Wall Carbon Nanotube scaffolds have shown strong bone ingrowth and vascularization in mouse calvarial defect models, outperforming hydroxyapatite controls [18, 19]. Dense CNT coatings on scaffolds have also been shown to enhance osteogenic differentiation of mesenchymal stem cells and to promote new bone formation around implants [11].

6.3 Chondrogenic support

Graphene-based materials also show promise for cartilage regeneration. GO can bind and retain chondrogenic factors such as transforming growth factor beta (TGF-β), thereby creating local environments favorable for chondrogenic differentiation [1, 15]. GO-functionalized hydrogels and

scaffolds exhibit increased expression of cartilage markers, including type II collagen and aggrecan, and higher glycosaminoglycan content compared to controls [1]. A representative in vivo outcome is shown in Figure 7, where micro-computed tomography reveals more complete defect filling over 4–12 weeks for GO-containing constructs than for controls, illustrating how the osteogenic and chondrogenic trends discussed above translate into macroscopic defect repair. Osteochondral scaffolds integrating GO-rich regions for cartilage and CNT/hydroxyapatite-rich regions for subchondral bone highlight the versatility of nanocarbon fillers for designing multi-tissue interfaces [1].

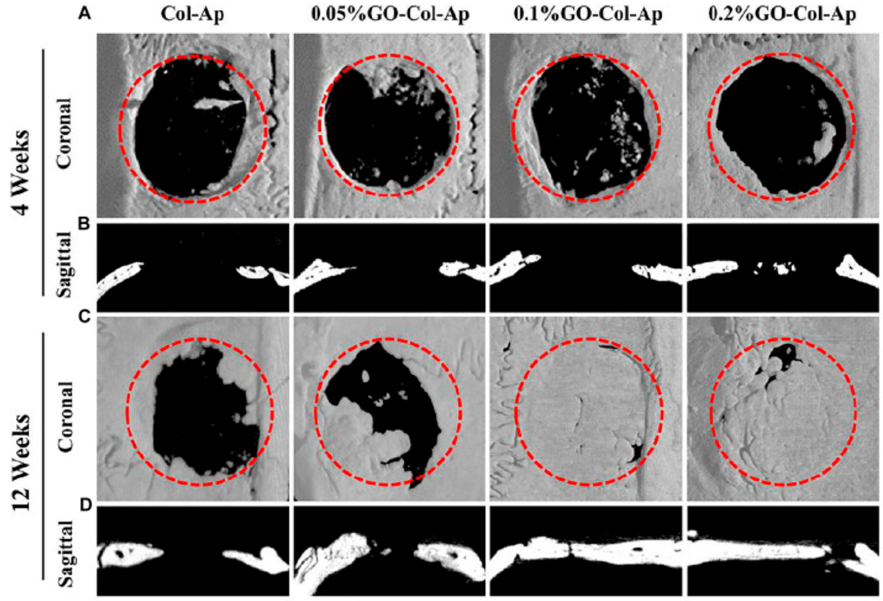

Figure 7: Osteogenic and chondrogenic responses to graphene-oxide–containing scaffolds. Micro-CT images showing improved bone defect filling with GO-containing constructs at 4 and 12 weeks.

7 Design Guidelines and Translational Considerations

7.1 Operating near percolation

A clear design rule emerging from the literature is to operate near the percolation threshold of the chosen nanocarbon/polymer system. This region offers the most significant marginal gain in conductivity and mechanical properties per unit of filler added. For many CNT and graphene composites, this design space corresponds to approximately 0.1–1 wt% filler [12, 16]. Higher loadings can lead to aggregation, processing difficulties, excessive stiffness, and increased cytotoxicity without significant additional benefit. Consequently, design should be driven by measured percolation behavior rather than by arbitrary filler targets.

7.2 Mechanical–electrical–biological trade-offs

Scaffold design must balance three interlinked domains. The first is the mechanical domain, in which stiffness and strength must be sufficient for the implant site without causing stress shield-

ing. The second is the electrical domain, in which conductivity must be adequate for the intended stimulation or sensing strategy. The third is the biological domain, in which both acute and chronic cytocompatibility and immunocompatibility are required. Maximizing CNT/graphene content might yield excellent conductivity and stiffness but risks brittleness and particle shedding. Conversely, overly low filler content may produce a safe but mechanically inadequate scaffold. Hybrid designs, such as conductive coatings on mechanically robust 3D-printed cores, can help decouple these trade-offs and achieve more favorable combinations of properties [9, 17].

7.3 Wear debris and long-term safety

In joint applications, cyclic loading and articulation can generate wear debris from any implant. For nanocomposite scaffolds, this debris may include polymer fragments and nanocarbon particles. Strategies to mitigate risk include using smooth outer surfaces or additional polymer/hydrogel layers to shield the nanocomposite core, choosing polymers and nanocarbon geometries with favorable wear behavior, and conducting detailed wear testing and particle characterization prior to clinical use. The long-term fate of CNTs and graphene *in vivo* remains an active area of research, and conservative designs that limit nanocarbon exposure and ensure strong matrix embedding are warranted until more data are available [2].

7.4 Manufacturability and scalability

Scaling production requires reproducible mixing, printing, or spinning, as well as post-processing under good manufacturing practice (GMP) conditions. Key challenges include maintaining consistent nanocarbon quality, including purity, aspect ratio, and surface chemistry; achieving batch-to-batch consistency in dispersion and rheology of printing inks or spinning solutions; and identifying sterilization methods that do not degrade nanocarbon or polymer properties. The growing availability of medical-grade CNTs and graphene derivatives, along with patents on graphene-based 3D-printing inks [4] and CNT composite scaffolds [6], suggests progress toward addressing these challenges.

8 Current Challenges and Open Research Gaps

8.1 Standardized testing protocols

The lack of standardized mechanical, electrical, and biological test protocols hampers direct comparison across studies. Differences in scaffold geometry, loading conditions, cell types, and assay methods make it difficult to determine whether observed differences are due to material design or experimental setup [10, 2]. Consensus protocols for mechanical testing, such as compression of cylindrical scaffolds with specified dimensions; for electrical testing, such as four-point probe measurements in hydrated conditions; and for biological testing, such as the use of common osteogenic markers at standardized time points, would significantly improve reproducibility and interpretability [2].

8.2 Cross-study comparison and data sharing

Variations in nanocarbon source, purification, and functionalization lead to heterogeneous materials even when nominally similar terms such as "Multi-Wall Carbon Nanotubes" or "GO" are used. Better reporting of nanomaterial characterization, including Raman spectra, impurity levels, size distributions, and surface area, is needed to enable cross-study comparison [2]. Shared databases capturing composition — processing — property — function relationships for nanocomposite scaffolds could be mined using machine learning to identify promising design spaces. However, such efforts require open data practices and consistent metrics across laboratories.

8.3 Long-term biocompatibility and regulatory pathways

Long-term *in vivo* studies in large-animal models are scarce. Outstanding questions include chronic immune responses, potential accumulation of nanocarbon debris, and interactions with systemic processes. Regulatory pathways for nanomaterial-containing implants are still evolving and will require extensive safety data [2, 10]. Addressing these gaps through standardized protocols, transparent data sharing, and multidisciplinary preclinical studies is essential before clinical translation.

8.4 Future directions: 4D printing and smart scaffolds

Several emerging directions are auspicious. Four-dimensional (4D) bioprinting extends 3D printing by introducing time- or stimulus-responsive behavior, enabling scaffolds that change shape, stiffness, or conductivity in response to temperature, load, or chemical cues; nanocarbons can provide responsive electroactive components within these constructs. Innovative, self-monitoring scaffolds that leverage piezoresistive behavior to sense their mechanical environment, relay data wirelessly, and potentially trigger on-demand electrostimulation or drug release represent another important direction [17]. Finally, AI-guided design using data-driven approaches to correlate composition and architecture with performance may allow algorithms to propose optimal scaffold designs for specific defect sites and patient populations [9].

9 Conclusion

Carbon nanomaterials, particularly CNTs and graphene-based derivatives, have emerged as powerful functional fillers for polymer scaffolds in bone and joint tissue engineering. When dispersed adequately at loadings near their percolation thresholds, these nanocarbons form networks that reinforce mechanical properties, introduce electrical conductivity and piezoresistive sensing, and provide high-surface-area interfaces that promote osteogenic and chondrogenic responses [12, 2]. In this sense, nanocarbon-reinforced scaffolds sit at the intersection of structural biomaterials, bioelectronics, and mechanobiology, transforming traditionally passive implants into active participants in the healing process. The fabrication–structure–property–function perspective developed in this review underscores that these advantages do not arise from the nanomaterials alone, but from their careful integration into architected polymer matrices with controlled dispersion, percolation, and interfacial chemistry.

Advances in 3D printing and electrospinning now allow nanocomposites to be organized into hierarchical architectures that mimic bone's multiscale structure and can be tailored to patient-specific defects [9, 16]. Hybrid constructs that combine mechanically competent printed cores with electrospun, nanocarbon-rich skins point toward a new generation of implants that simultaneously satisfy load-bearing requirements, provide biomimetic topography, and support electrical stimulation or self-sensing. In vitro and early *in vivo* studies demonstrate enhanced bone formation and promising cartilage support, with generally favorable biocompatibility when nanocarbon type, dose, and dispersion are carefully controlled [18, 3, 8]. These data, while still incomplete, suggest that appropriately engineered CNT/graphene composites can match or surpass current scaffold systems in both mechanical performance and biological function.

Significant challenges remain, including standardizing testing protocols, ensuring long-term safety, managing wear debris, and scaling manufacturing under GMP conditions. Addressing these gaps while harnessing emerging tools such as 4D bioprinting and AI-guided design will be critical to translating nanocarbon-enabled scaffolds from the laboratory to clinical practice. Long-term *in vivo* studies in relevant animal models, improved reporting of nanomaterial characteristics, and shared databases of composition–processing–property–function relationships will all be essential to de-risk clinical adoption and clarify regulatory pathways. Equally important will be systematic evaluations of immunological responses and degradation or persistence of nanocarbon debris over clinically relevant time scales.

Overall, CNT/graphene–polymer scaffolds represent a compelling convergence of nanotechnology and regenerative medicine, with the potential to deliver mechanically competent, self-sensing, and bioactive implants for complex bone and joint repairs. In the near term, the most impactful opportunities lie in rigorously mapping percolation-controlled design spaces, quantifying mechanobiological responses under physiologically relevant loading, and developing standardized, comparative datasets across materials and architectures. These efforts, many of which are accessible to undergraduate and early-stage researchers, can provide the foundational evidence base needed for future translational studies. If these scientific and engineering challenges are met, nanocarbon-filled scaffolds may help redefine how orthopedic implants are conceived, monitored, and integrated within the living skeleton.

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