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Review

Electrospun Polymeric Composites for Dental Tissue Engineering: Materials, Fabrication Strategies, Biological Performance, and Clinical Translation

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Abstract

Oral diseases impact approximately 3.5 billion individuals globally, often leading to the loss or damage of periodontal, alveolar bone, dentin-pulp, or oral soft tissues. Consequently, dental tissue engineering necessitates scaffolds that can simultaneously offer a three-dimensional extracellular matrix-like microenvironment conducive to cell adhesion, migration, and differentiation. These scaffolds must also maintain spatial and mechanical integrity during the healing process and degrade at a rate that aligns with new tissue formation. Conventional biomaterials frequently fail to meet all these criteria. Natural polymers, such as collagen, provide cell-recognition cues and support cellular responses but may lose mechanical integrity before regeneration is complete. In contrast, many synthetic polymer barriers maintain structural stability and offer adjustable degradation rates but exhibit limited intrinsic bioactivity. Electrospinning offers a method to integrate these complementary characteristics into fibrous scaffolds with high surface areas that replicate essential structural aspects of the native extracellular matrix. This structure provides cells with a greater surface area for adhesion, migration, and remodeling than cast films or solid scaffolds. This review brings together the electrospinning literature related to dental applications. It begins with the basics of the process, including Taylor cone formation, jet whipping, and the solution, instrument, and environmental factors that influence fiber diameter and morphology, as well as variations such as coaxial, emulsion, melt electrowriting, and needleless electrospinning. It then explores the natural, synthetic, and composite polymer systems used to produce dental nanofibers and their applications in periodontal regeneration, alveolar bone repair, dentin-pulp regeneration, antibacterial and drug delivery functions, and oral wound healing. The obstacles, such as low production throughput, inconsistent sterilization and testing protocols, and the scarcity of large-animal and human trials, that still hinder the transition of laboratory-scale electrospun scaffolds to regular use in dental practice are discussed, and the research directions most likely to bridge this gap are highlighted.

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1. Introduction

Oral diseases are one of the largest unmet health issues worldwide. According to the World Health Organization's (WHO) Global Oral Health Status Report, approximately 3.5 billion people are affected by some form of oral disease. Among these, untreated caries, severe periodontitis, edentulism, and oral cancer are the leading causes [1]. Periodontitis is usually cited as the leading cause of tooth loss, affecting approximately 15% of adults globally, and the incidence of periodontal disease is climbing roughly 76% to approximately 89.6 million cases, with no sign of leveling off through 2035 [2,3]. This pattern continues to be reflected elsewhere in the mouth. Pulp and dentin injury from caries or trauma and oral mucosal ulcers affect an estimated 5–20% of people at some point, while peri-implant diseases, such as mucositis and peri-implantitis, affect approximately 43% and 22% of people, respectively [4,5]. The major impact of these oral diseases is tissue loss, which is a major concern.

The materials currently used in dentistry to address these problems often have well-known limitations. For example, collagen membranes are widely used for guided tissue and bone regeneration because they support cell growth and healing processes. However, the same natural properties that make collagen useful also cause it to break down quickly. Native collagen has low mechanical strength and usually loses its structure within approximately six weeks in the body [6] which is often before the underlying periodontal tissue has fully healed. Cross-linking can slow this degradation, but it may also cause foreign body reactions and reduce proper tissue integration [7], creating a different set of challenges. A recent review highlighted that current Guided bone regeneration (GBR)/Guided tissue regeneration (GTR) membranes still fall short in several important areas, including bone regeneration, antibacterial performance, controlled degradation, mechanical strength, and compatibility with the immune system [8]. Non-resorbable Polytetrafluoroethylene (PTFE) membranes avoid rapid degradation, but they usually require a second surgery for removal. Similar challenges exist in the field of endodontics. Common disinfectants, such as sodium hypochlorite and triple antibiotic paste, are effective in eliminating bacteria; however, at the concentrations required for disinfection, they can also damage the dental pulp stem cells that clinicians aim to preserve [9]. This pattern is observed across many areas of dentistry, including dental implants, restorative materials, and oral wound healing. In most cases, materials are either durable but biologically inactive or biologically active but short-lived, making it difficult to achieve both properties in a single material.

In the advancement of biomaterials, a nanofabrication technique called electrospinning has been gaining momentum owing to its simple and cost-effective process. The setup consists of an electric field (10–30 kV) that causes the polymer solution to be pulled off from the needle tip. Under the influence of this electric field, the polymer jet undergoes whipping and thins on its way to the grounded collector, resulting in solidified fibers ranging from nanometers to micrometers in diameter [10]. The small fiber size is not the main advantage. Instead, the fiber network created during electrospinning closely resembles the natural extracellular matrix (ECM), providing an ideal environment for tissue regeneration. These nanofiber mats have a high surface area-to-volume ratio and a highly porous structure, providing cells with many places to attach, grow, and communicate [11]. The use of periodontal scaffolds has been consistently demonstrated to be superior in promoting cell adhesion, migration, growth, and differentiation compared to cast films or

solid materials in previous studies [12]. Cells grow in a flexible fibrous scaffold, can move through it, and remodel it, mimicking the way they grow in the native ECM.

The electrospinning process has a wide range of material processing capabilities. Natural polymer materials such as collagen, gelatin, chitosan, and silk fibroin, and synthetic polymer materials such as polyvinyl alcohol (PVA), poly L-lactic acid (PLLA), and polycaprolactone (PCL) have been used to successfully form dental nanofibers. In addition, a large number of nanocomposite scaffolds have been developed with hydroxyapatite and functional additives such as silver, titanium dioxide, zinc oxide, chlorhexidine, and ampicillin [13]. Furthermore, processing parameters, such as polymer concentration, solution viscosity, molecular weight, applied voltage, tip-to-collector separation, and solvent type, can be varied to influence the properties of electrospun fibers [14]. This enables researchers to manipulate the diameter, morphology, and performance of scaffolds for various applications. Further development of the electrospinning technique has led to more advanced techniques, such as coaxial electrospinning, which offers even more flexibility. The technique involves placing drugs or growth factors in the fiber core and controlling the slow release of the contents using the shell. For instance, a scaffold consisting of a polycaprolactone shell and a polyethylene glycol core containing enamel matrix derivative (EMD) has been found to induce osteogenic differentiation in periodontal ligament stem cells (PDLSCs) [15]. Other attempts have been made to control the delivery of growth factors, deliver antimicrobial agents during endodontic therapy, preserve the viability of pulp stem cells, and generate aligned nanofibers to guide tissue growth in specific directions rather than being a passive scaffold [16,17].

However, the use of electrospun nanofibers in dentistry has increased, and although there are many laboratory studies, they have not yet been used routinely in clinical practice. This review aims to synthesize all available research to obtain a better understanding of the field. It describes the main design factors that influence the biological performance and their significance in electrospinning: fiber diameter, alignment, and porosity. It also assesses the mechanical properties and degradation behavior of various electrospun materials for specific dental applications, including periodontics, endodontics, oral and maxillofacial surgery, implantology, and restorative dentistry. Moreover, this review appraises the strategies for incorporating antimicrobial agents, growth factors, and bioceramics into these scaffolds and critically assesses their effectiveness in laboratory experiments. Finally, it highlights the significant challenges that still hinder clinical translation, such as sterility, large-scale production, product quality, and the absence of long-term animal and human studies. These problems must be solved before electrospun nanofiber scaffolds can be used as common clinical tools in dentistry. Thus, the novelty of the present review is to provide an updated cross-indication synthesis that links polymer/composite selection and electrospinning architecture with biological performance, evidence level, and translational readiness. This approach seeks to identify the most advanced design strategies for particular dental uses and highlight areas where evidence is lacking.

2. Fundamentals of Electrospinning for Dental Biomaterials

2.1. Principles of Electrospinning

Electrospinning is an electrohydrodynamic technique in which an electric field is used to attenuate a polymer solution or melt into fibers with diameters ranging from tens of nanometers to a few micrometers in size. A typical setup consists of four core components: a high-voltage power supply (10–30 kV), a syringe pump that delivers the polymer solution/melt at a controlled flow rate, a spinneret (metallic needle or nozzle) connected to the power supply, and a grounded collector set at a defined distance from

the spinneret tip [18,19]. The polymer solution is held at the needle tip by surface tension before voltage is applied, resulting in a pendant hemispherical droplet.

Under the applied voltage, the polymeric droplet becomes electrically charged, and the resulting repulsive charges at the droplet surface act against the surface tension. As the voltage increases, the droplet elongates into a conical shape known as the Taylor cone [18]. Once the electrostatic force exceeds a critical threshold that the surface tension can no longer balance, a thin, charged liquid jet is ejected from the apex of the cone toward the grounded collector [20]. In the initial segment of its trajectory, the jet travels in a relatively straight line, but as solvent evaporation increases the surface charge density, the mode of current flow shifts from an ohmic to a convective regime, and the jet undergoes a rapid, electrically driven bending (whipping) instability [18]. This whipping motion subjects the jet to an enormous degree of stretching and thinning over a very short timeframe, which is the principal mechanism by which micrometer-scale jets are reduced to nanometer-scale fiber diameters [19]. Concurrent with this elongation, the solvent (or, in melt electrospinning, the thermal energy) is progressively lost to the surrounding atmosphere, so that the jet solidifies into a continuous solid fiber before it reaches the collector, where the fibers accumulate as a randomly oriented or aligned nonwoven mat, depending on the collector design [19,21].

The foundational research in modern polymer-electrospinning literature laid the groundwork for understanding the process and the physics of the jet involved. In 1995, Doshi and Reneker detailed the electrospinning technique, its operational parameters, and the resulting fiber structures [22]. Following this, Reneker and Chun showcased the creation of polymer fibers with nanometer-scale diameters through electrospinning [23]. Later, Reneker and his team elaborated on a mechanistic explanation of the electrically induced bending and whipping instability, which is crucial for significant jet elongation and the formation of nanoscale fibers [24]. These pivotal studies underpin the sequence involving the Taylor cone, straight jet, and whipping instability. Hence, this underlies the production of high-surface-area and extracellular-matrix-mimicking fibrous architectures that make electrospun nanofibers attractive for dental biomaterials. Figure 1a–c illustrate the basic working principle of the electrospinning technique, representing the Taylor cone with a whipping instability jet region.

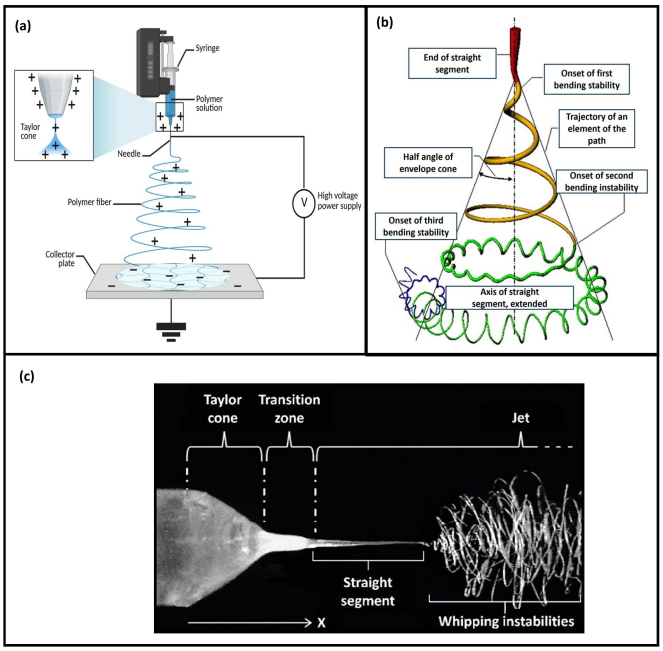


Figure 1. Principles of the electrospinning technique. (**a**) Basic electrospinning setup. Created in BioRender. Chandrashekar, C. (2026) https://BioRender.com/xdxrxem. (**b**) Trajectory representation

of the electrospinning jet. Reproduced with permission from [25] under CC-BY-4.0. (**c**) High-speed photographic image of Taylor cone formation, representing straight and whipping jet regions. Reproduced with permission from [26] under CC-BY-4.0.

Process Parameters

The morphology, diameter, and uniformity of electrospun nanofibers are governed by an interdependent set of variables that are conventionally grouped into three categories: solution parameters, process (instrument) parameters, and ambient parameters. [21,27]. Because these parameters are interdependent, optimization for a given dental application typically requires a systematic and iterative approach rather than tuning a single variable independently.

Among the solution parameters, the polymer concentration and the resulting solution viscosity are widely regarded as the most influential. Sufficient chain entanglement is required for the jet to be stretched into a continuous fiber rather than breaking into droplets. Below a polymer-specific critical concentration, electrospraying or beaded fibers result instead of smooth fibers, whereas excessively high concentrations can impede pumpability and clog the spinneret [25,27]. The molecular weight of a polymer, in conjunction with the concentration of its solution, dictates the extent of chain overlap and entanglement necessary for the formation of stable fibers. As a result, polymers with higher molecular weights can frequently achieve the critical entanglement threshold at lower concentrations compared to those with lower molecular weights [27]. The solvent volatility and electrical conductivity of the solution further influence fiber formation: more volatile solvents accelerate the transition from liquid jet to solid fiber, while higher conductivity increases the net charge carried by the jet and tends to promote thinner fibers through greater electrostatic stretching [20,25].

Process parameters such as the applied voltage, solution flow rate, needle-to-collector distance, and needle gauge are directly controlled at the instrument level. Increasing the voltage generally narrows the fiber diameter by intensifying the electrostatic stretching force on the jet, although excessively high voltages can destabilize the Taylor cone and introduce beading or branching defects [28]. The flow rate must be balanced against the voltage: higher flow rates deliver more material to the jet and tend to thicken the fibers unless compensated for by a proportional increase in the voltage. The needle-to-collector distance determines the flight time available for solvent evaporation and jet elongation before deposition. Distances that are too short can leave residual solvent in the deposited fiber, causing fiber fusion or ribbon-like morphologies, while excessive distances can reduce the electric field strength experienced by the jet [25]. Several of these parameter–morphology relationships have been confirmed directly in dental-relevant polymer systems; for instance, in poly(lactic-co-glycolic acid) (PLGA) fibrous membranes intended for guided tissue and bone regeneration, systematic variation in the applied voltage, flow rate, solution concentration, and spinneret-to-collector distance produced statistically significant differences in the mean fiber diameter [27].

Ambient parameters, such as temperature and relative humidity, are comparatively less controllable in many laboratory settings but can meaningfully alter outcomes. Elevated temperature lowers solution viscosity and accelerates solvent evaporation, generally narrowing fiber diameter, whereas humidity effects are more complex and solvent-dependent. In some hygroscopic solvent systems, ambient water vapor can condense directly into the Taylor cone rather than only onto the jet in flight, altering local solvent quality and promoting fiber fusion or surface pore formation [20,27]. A systematic review of biopolymer nanofibers similarly found that the fiber diameter of cellulose increased with increases in

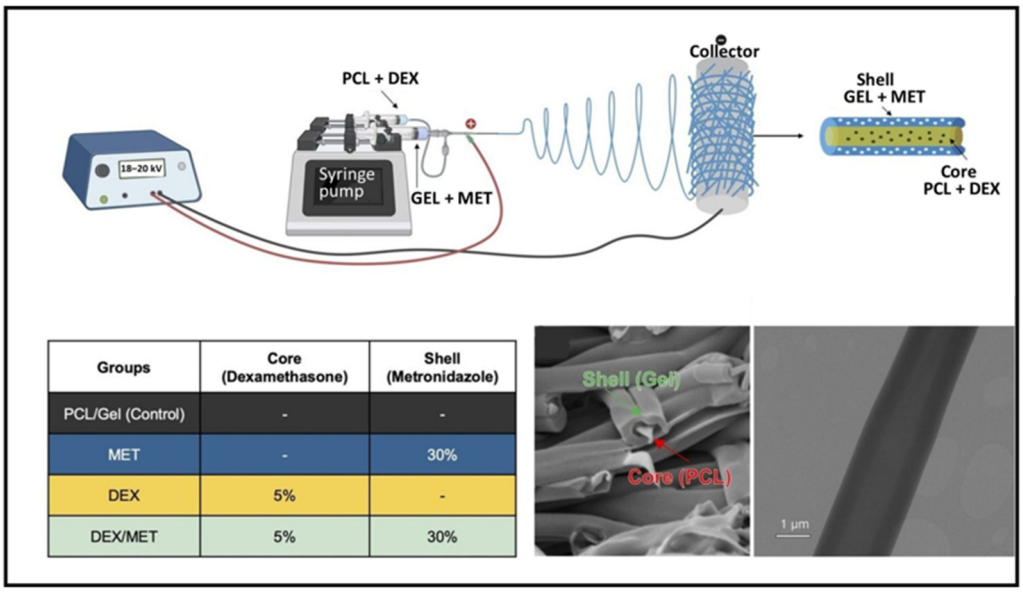
concentration, voltage, orifice diameter, temperature, and humidity, whereas the tip-to-collector distance had an inverse relationship, underscoring that parameter–morphology relationships are not universal but are polymer- and solvent-system specific [27,28].

2.2. Variants of Electrospinning Relevant to Dental Materials

Although single-fluid, single-needle electrospinning is most commonly utilized, multiple variations in the process have been designed to satisfy the specific functional needs of dental tissue engineering and drug delivery, such as spatially controlled drug release, encapsulation of hydrophilic bioactives, control of scaffold architecture, and volume suitable for clinical translation.

2.2.1. Coaxial/Core–Shell Electrospinning

In coaxial electrospinning, two different fluids are concurrently extruded from a coaxial dual-capillary spinneret, forming fibers with a core material surrounded by a shell. The shell material and thickness can control the release rate of the drug or bioactive agent from the shell into the external environment, and the sensitive bioactive agent or drug is protected from immediate exposure to the external environment in this architecture [29]. This has been applied in a dental application to control periodontal disease, and coaxial scaffolds (Figure 2) composed of poly(ε-caprolactone) and gelatin with a separate compartment containing an antimicrobial agent (metronidazole) and an anti-inflammatory corticosteroid (dexamethasone) have been designed to deliver the two agents in a sustained and independently tunable manner, avoiding a simultaneous burst release by single fluid blended fibers [30]. Genes have also been successfully delivered using a coaxial electrospinning approach for periodontal regeneration, which employs a PLGA shell to protect a DNA plasmid encoding bone morphogenetic protein-2 (BMP-2) loaded into the fiber core. The core–shell configuration exhibited gene expression in human periodontal ligament stem cells for over 28 days, markedly longer than that of single-axial fibers fabricated from the same materials [31].



Groups	Core (Dexamethasone)	Shell (Metronidazole)
PCL/Gel (Control)	-	-
MET	-	30%
DEX	5%	-
DEX/MET	5%	30%

Figure 2. Schematic illustration of the electrospun core–shell structure with an integrated drug and corresponding scanning electron microscopy and transmission electron microscopy micrographs of the nanofibers, respectively. Reproduced with permission from [30] under CC-BY-4.0.

2.2.2. Emulsion Electrospinning

Emulsion electrospinning (Figure 3) creates a core–shell-like fiber structure from a single spinneret by first forming a continuous hydrophobic polymer solution and then emulsifying a hydrophilic bioactive agent (usually dispersed in an aqueous phase) in this polymer solution using a surfactant before the mixture is electrospun as a single fluid. The tendency for the dispersed aqueous droplets to migrate to and coalesce around the fiber core is similar to that of the coaxial fiber and is obtained without the need for a dual-capillary spinneret [32]. This approach is also applicable to dental hard-tissue regeneration, where a mixture of hydrophilic hydroxyapatite nanoparticles and hydrophobic polymer fiber matrices was applied to enhance the homogeneous dispersion of the bioceramic phase and its known role in inducing osteoconduction and osteointegration in alveolar bone and periodontal scaffolds [33].

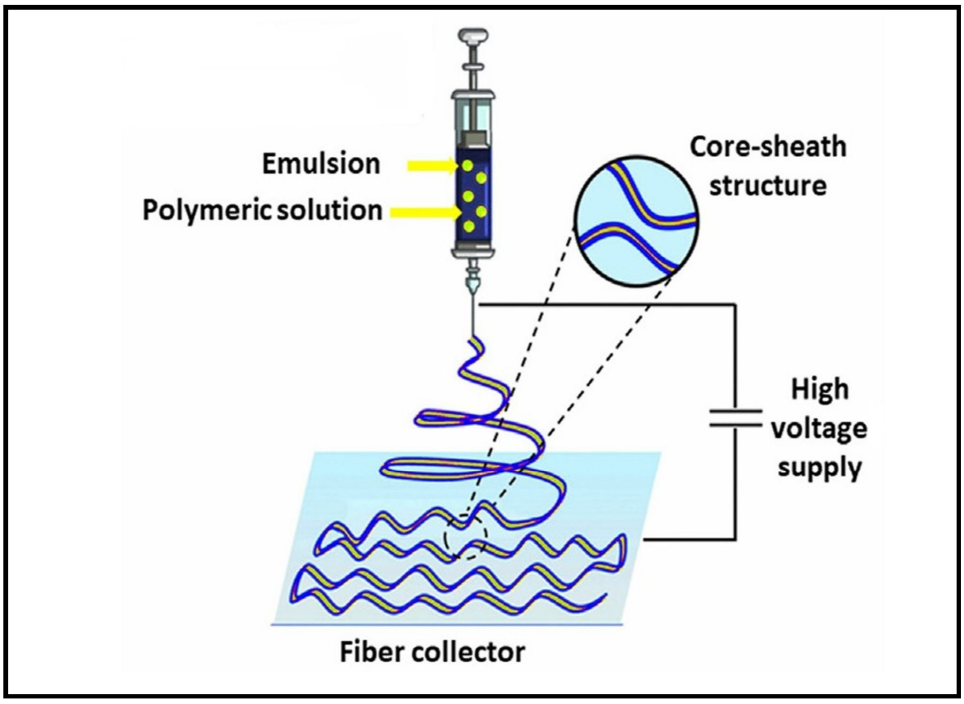


Figure 3. Schematic illustration of the emulsion electrospinning working principle. Reproduced with permission from [32] under CC-BY-4.0.

2.2.3. Melt Electrowriting

Melt electrowriting (MEW) emerged from early demonstrations of direct writing through melt electrospinning [34]. The principle of melt electrowriting (MEW) is similar to that of solution electrospinning with a direct-write computer-controlled deposition technique, except that the instability associated with the chaotic whipping of the polymer solution in solution electrospinning is eliminated and replaced by a near-field path through which the microfiber jet is deposited layer-by-layer onto a moving collector under a comparatively low and stable voltage [35–37]. This creates a highly structured and repeatable scaffold structure (such as a 0–90° crosshatch matrix, honeycomb pattern, or graded strand spacing) instead of the random fiber orientations found in traditional electrospinning. This level of architectural control has made MEW particularly attractive for periodontal tissue engineering, where strand spacing and fiber alignment can be tuned to direct cell behavior. Scaffolds with 500 μm strand spacing have been shown to upregulate osteogenic marker expression in human periodontal ligament stem cells, while differently structured regions of the same composite scaffold (Figure 4) can be tailored to support either bone

or periodontal ligament regeneration in a single, spatially graded construct [38]. MEW scaffolds with engineered fiber-guiding microchannel architectures have been shown, in a rodent periodontal defect model, to significantly increase oblique periodontal ligament fiber attachment to the tooth root surface relative to non-guiding scaffold designs, directly illustrating how MEW-enabled architecture can be leveraged to direct the functionally oriented tissue attachment required for periodontal regeneration [39].

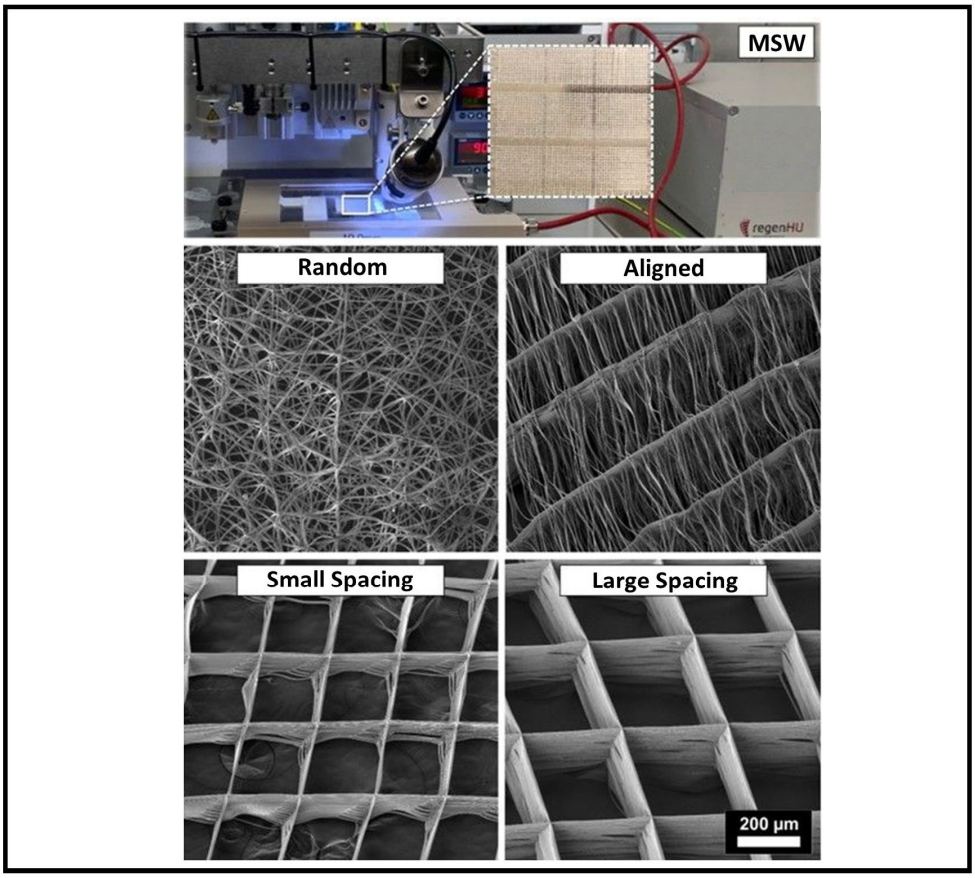


Figure 4. Schematic illustration of the melt electrowriting (MEW) electrospinning variant for producing nanofibrous scaffolds. Reproduced with permission from [38] under CC BY-NC-ND.

2.2.4. Multi-Jet/Needleless Electrospinning

A persistent barrier to the clinical translation of electrospun dental biomaterials is the very low production rate of conventional single-needle electrospinning, typically on the order of 0.01–1 g/h, which is insufficient for manufacturing membranes and scaffolds at clinically or commercially relevant scales [40]. Multi-needle electrospinning is achieved by placing several spinnerets in parallel to achieve a higher production rate. However, the electric field distribution is not uniform in neighboring charged jets, and the fact that the fiber is charged can result in an uneven distribution of the electric field if no auxiliary electrodes or optimized needle spacing are used [41]. In contrast, needleless electrospinning (Figure 5) forms multiple Taylor cones directly from the free surface of a polymer solution or melt, such as a rotating disk, wire, or roller, minimizing the risk of spinneret clogging and significantly enhancing the production rate [42]. Reported production rates for some configurations are hundreds of grams per hour, but usually with a broader fiber diameter distribution than that attained with a single needle [40,43]. The transition from laboratory research to clinical application, especially in the field of dental surgery, such as with guided bone regeneration (GBR) membranes, remains a significant challenge. Numerous studies have created electrospun barrier membranes for

oral bone regeneration on a laboratory scale, but few have moved beyond this point. A thorough search for reports on their use did not reveal any clinical applications, and only a few animal studies were found. While these technologies are still under development, this underscores the need for scalable manufacturing methods to enable the transition of electrospun dental biomaterials from research laboratories to routine clinical use, such as multi-jet and needleless electrospinning [44].

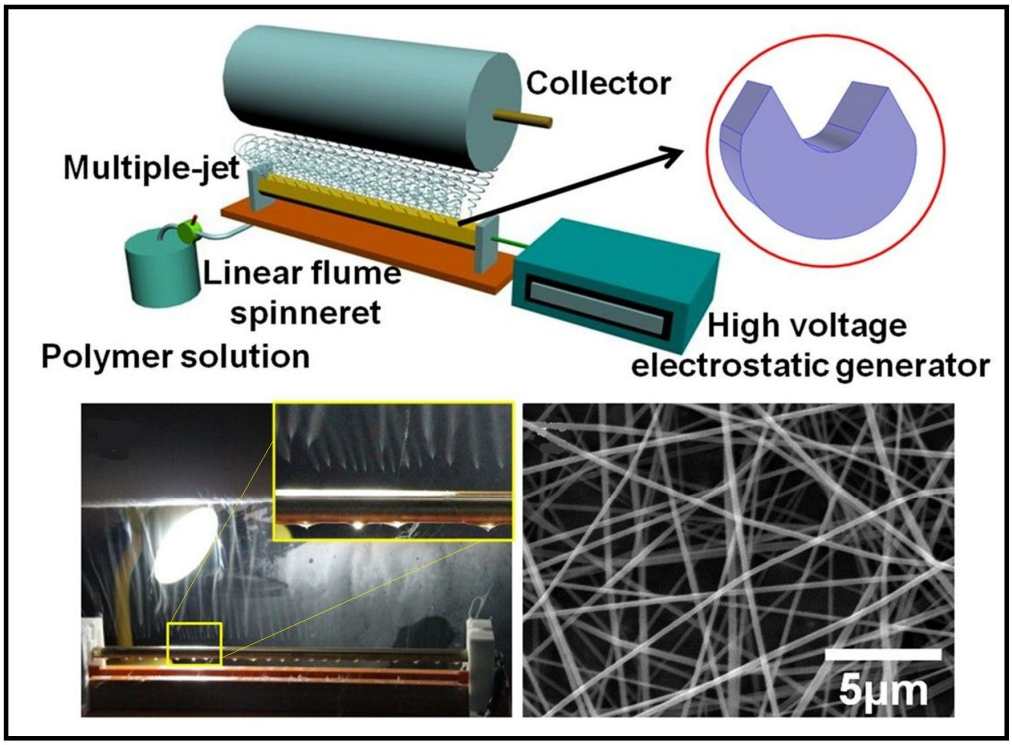


Figure 5. Schematic illustration of the multiple-jet needleless electrospinning variant for producing nanofibrous scaffolds, with the yellow square box indicating the enlarged image of multi-jet formation. Reproduced with permission from [43] under CC-BY-4.0.

3. Polymer and Composite Systems Used in Dental Nanofibers

The choice of polymer system used in the production of electrospun dental scaffolds is crucial in determining their biological and mechanical properties. Natural polymers offer biochemical signals that closely resemble those of the extracellular matrix (ECM), promoting cell attachment. In contrast, synthetic polymers provide consistent and adjustable mechanical properties as well as controlled degradation rates. Since neither class alone fully meets the requirements for an effective dental scaffold, the field is increasingly moving toward combinations of natural and synthetic polymers, as well as inorganic and organic materials containing osteo/odontogenic mineral fillers.

3.1. Natural Polymers

In dental nanofiber design, the biomaterials used are natural polymers, which have a structure similar to the components of natural dental and periodontal tissues, and are highly biocompatible and non-immunogenic, with intrinsic bioactivity that facilitates cell adhesion, proliferation, and differentiation [45–47]. The major inherent disadvantage of the polymers in this class is their mechanical properties; most of them are natural polymers, which, when electrospun into fibers, have relatively low tensile strength and are prone to enzymatic degradation at a relatively fast rate. They are also batch-to-batch variable because they are of natural origin (animal, plant, or microbial) [45,48].

Collagen, the principal structural protein of the periodontal ligament, dentin, and pulp ECM, is one of the most extensively studied natural polymers for dental scaffolds. This is because of its close compositional match to native tissue and its capacity to support cell attachment through integrin-binding motifs [49]. Gelatin, a denatured derivative of collagen obtained by partial hydrolysis, retains many of collagen's cell-recognition Arginyl-glycyl-aspartic acid (RGD) sequences while being more economical and easier to process by electrospinning. Hence, gelatin has become a widely used surrogate in dental tissue engineering, including hydrogel, electrospun, and hybrid composite scaffold formats for endodontic and periodontal regeneration [47]. However, both proteins are mechanically weak in their pure electrospun form and degrade rapidly and unpredictably in aqueous or enzymatic environments. Hence, they are either chemically crosslinked or combined with a synthetic partner polymer rather than being used alone [46,50,51]. Electrospun gelatin and gelatin/polycaprolactone (PCL) nanofibers with fiber diameters approximating those of native collagen fibrils have demonstrated favorable biocompatibility and have been investigated for periodontal and pulp-capping applications [46,50].

Chitosan, a cationic polysaccharide derived from chitin, is valued in dentistry for its biocompatibility and biodegradability. In addition to its intrinsic antibacterial and antifungal activities against common oral pathogens, few other scaffold polymers possess this property natively [2,12]. This antimicrobial property makes chitosan particularly attractive for periodontal barrier membranes, where infection control and guided tissue regeneration are design priorities [2,52]. Chitosan/PCL nanofibrous scaffolds incorporating decellularized periodontal ligament-derived extracellular matrix have been shown to support the attachment and proliferation of periodontal ligament stem cells, illustrating the polymer's suitability as a bioactive component of composite periodontal scaffolds [52]. However, the electrospinnability of chitosan is limited by its high crystallinity and poor solubility in mild solvents. Hence, it necessitates blending with a synthetic carrier polymer such as polyethylene oxide (PEO) or PCL [12].

Silk fibroin, a structural protein obtained from Bombyx mori silkworm cocoons, is one of several electrospinnable natural proteins explored for dental and craniofacial applications. It offers a desirable combination of mechanical robustness (relative to other natural proteins), slow and controllable biodegradation, and low immunogenicity [53,54]. In dental pulp tissue engineering, silk fibroin-based fiber composites loaded with bioactive agents have been reported to promote osteogenic differentiation and attenuate the inflammatory response associated with pulp injury [55]. Silk fibroin has also been electrospun in combination with synthetic carriers such as poly(lactic acid) (PLA) to overcome PLA's poor cell adhesion and slow in vitro degradation of PLA [54]. Despite these advantages, raw silk fibroin solutions are difficult to electrospin without the addition of carrier polymers or surfactants, and, similar to other animal-derived proteins, fibroin preparations show variability depending on the extraction protocol and silkworm source [53].

Cellulose-derived polymers, especially cellulose acetate, are promising for dental drug delivery because of their hydrophilicity, electrospinnability, and chemical compatibility with antimicrobial agents for restorative and endodontic therapy [56,57]. Nanofiber mats of cellulose acetate loaded with chlorhexidine for controlled release have been developed for use in root canal spaces and carious voids. These mats can absorb water and adapt to irregular shapes while providing antibacterial activity against Streptococcus mutans and Enterococcus faecalis [57,58]. Coaxial cellulose acetate/polyvinylpyrrolidone scaffolds have also been explored as substrates for regenerative endodontic procedures, exploiting a core–shell architecture in which the hydrophobic cellulose acetate forms a structurally stable core, while the hydrophilic shell promotes early cell attachment [59]. The main limitation of cellulose acetate for tissue engineering is its slow biodegradation and absence

of bioactive cell-recognition motifs, restricting its use to a carrier/barrier material rather than a cell-instructive scaffold [56].

Alginate, a biocompatible anionic polysaccharide derived from brown algae, rapidly gels with Ca^{2+}. These properties are utilized in oral and periodontal local drug-delivery films, such as lidocaine-releasing alginate/carboxymethylcellulose/PEO barriers for post-surgical dental applications [60]. The principal limitation of alginate for nanofiber fabrication is its poor inherent electrospinnability. Its polyelectrolyte nature and high solution viscosity at spinnable concentrations make pure alginate difficult to process into continuous defect-free fibers; therefore, it is almost always electrospun as a blend with a carrier polymer such as PEO or polyvinyl alcohol (PVA) [61,62]. Once fibers are formed, post-spinning ionic crosslinking can improve mechanical stability; however, batch variability in molecular weight and guluronate/mannuronate ratio (which depend on algal source) remains a recognized limitation for reproducible scaffold performance [63].

3.2. Synthetic Polymers

Synthetic polymers are the structural backbone of most of the electrospun dental scaffolds. Their molecular weights, crystallinities, and copolymer compositions can be controlled during synthesis. This yields fiber mats with reproducible mechanical properties and tunable, predictable degradation profiles, in contrast to the batch variability characteristics of naturally derived polymers [64]. Several of the most widely used synthetic polyesters have a long regulatory history. They were used in FDA (Food and Drug Administration) and CE (European Conformity) approved resorbable sutures, drug-delivery systems, and orthopaedic devices well before their adoption in dental tissue engineering [65,66].

Polycaprolactone (PCL) is the most extensively used synthetic polymer in dental applications. This is possible owing to its excellent processability, mechanical durability, slow and predictable hydrolytic/enzymatic degradation, and long-standing FDA approval for biomedical and drug delivery devices [67,68]. PCL's main limitation is its hydrophobicity, which leads to poor cell attachment and limited bioactivity when used alone. This is why it is typically used as a mechanical scaffold in blends or composites rather than as a standalone material [67,69]. PCL nanofiber membranes loaded with nano-hydroxyapatite and gentamicin have been studied for guided bone regeneration around dental implants. Additionally, PCL membranes coated with MTA/hydroxyapatite exhibit good biocompatibility for pulp-capping applications [70].

Polylactic acid (PLA) and its copolymer, poly(lactic-co-glycolic acid) (PLGA), are FDA-approved aliphatic polyesters. Their adjustable lactide: glycolide ratio allows the tuning of degradation rates from weeks to over a year, making them suitable for matched resorption and tissue regeneration applications [71,72]. PLGA electrospun nonwovens have been evaluated for oral mucosa regeneration, where their degradation behavior in simulated body fluid and artificial saliva was found to be dependent on the copolymer ratio and sterilization method [66]. One of the drawbacks of PLA and PLGA is that they generate acidic degradation products, which may lead to an inflammatory response if hydrolysis is too fast or if there is insufficient buffering. Additionally, PLA is more brittle and degrades more slowly than pure PCL [73,74].

Polyvinyl alcohol (PVA) is a water-soluble synthetic polymer that is frequently mixed with cellulose acetate and/or chitosan to improve wettability and processability. It has also been investigated as a filler for Polymethyl methacrylate (PMMA)-based dental composite resins [75–77]. Polyurethane (PU) is used in membranes that have a mechanical requirement for oral wound healing and mucosal repair, as they can be used as strong, elastic, and biodegradation-resistant membranes [4,78,79]. Polyethylene oxide (PEO) has three main applications: as a spinning aid, structural polymer, and electrolyte. It pos-

sesses excellent flexibility and can stabilize the electrospinning jet of difficult-to-electrospun polyelectrolytes, such as alginate and chitosan. Furthermore, PEO has been incorporated to enhance the mechanical properties and hydrophilicity of PLGA barrier membranes surrounding dental implants for GBR [75,80,81].

3.3. Polymer Blends and Composite Nanofibers

Natural polymers provide bioactivity but lack mechanical strength, whereas synthetic polymers provide mechanical strength but lack bioactivity. Therefore, a natural-synthetic blend is the most popular design approach used in dental electrospinning, which is designed to incorporate the desirable properties of both materials into one fiber [69].

Blends of PCL/gelatin and PCL/chitosan are among the most thoroughly reported systems. The PCL/gelatin nanofibers exhibited good biocompatibility, and their diameters were comparable to those of native collagen fibrils. Gelatin counteracts the mechanical strength and hydrophobicity of PCL, promoting cellular attachment [46,67]. The rationale behind the use of PCL/chitosan blends for periodontal regeneration is the same, with the former providing mechanical properties appropriate for the use of the blend as a guided tissue regeneration membrane and the latter providing bioactivity and antibacterial properties [52,82].

PLGA/gelatin blends extend this rationale to faster-degrading synthetic backbones. Aligned PLGA/gelatin electrospun sheets were combined with the native dental pulp extracellular matrix and treated dentin matrix to simulate the periodontium and dentin-pulp microenvironments. This facilitates the odontogenic induction of dental stem cells and illustrates how blended nanofiber sheets can serve as a structural substrate for more complex ECM-augmented constructs in tooth-root regeneration [69]. PLGA was blended with gum tragacanth and tetracycline hydrochloride for periodontal regeneration. Additionally, triple-modified PLGA/PCL fiber scaffolds with silver impregnation and collagen coating enhanced biocompatibility, antimicrobial activity, and osteogenic performance, demonstrating that multiple components can be combined in a single composite scaffold [83,84].

PLA/PCL and other synthetic blends are primarily used to balance the mechanical and degradation properties. PLA's brittleness and slow degradation can be countered by PCL's flexibility and faster erosion. Blending PLA/PCL with nano-hydroxyapatite and zeolite enhances the proliferation and adhesion of human dental pulp stem cells, indicating that dental nanofiber design can effectively use synthetic-synthetic combinations [85].

3.4. Bioceramic/Mineral-Incorporated Nanofibers

Mineralized dental and periodontal tissues (dentin, cementum, and alveolar bone) are composite materials with an organic collagen matrix and inorganic calcium phosphate. Incorporating osteogenic or odontogenic mineral fillers into electrospun polymer fibers is an effective strategy for creating scaffolds that mimic the native tissue composition. Additionally, it promotes mineralized tissue regeneration beyond providing structural support [18,86,87].

Hydroxyapatite (HA), the principal mineral phase of bone, dentin, and enamel, is the most commonly used filler in dental and periodontal nanofiber composites. Nano-HA-doped PCL/gelatin electrospun scaffolds have been shown to enhance the proliferation and odontogenic differentiation of dental pulp stem cells (DPSCs) [88]. Furthermore, HA-coated PLLA/PCL nanofibrous scaffolds have been reported to direct DPSC differentiation toward osteoblast-like cells [89]. Aligned PCL-based guided bone regeneration barriers incorporating xenograft-derived nano-HA have also been shown to support the viability and osteogenic differentiation of dental pulp-derived mesenchymal stem cells. This illustrates HA's role of HA in barrier membranes placed around dental implants and alveolar bone

defects [90]. Higher HA loading can enhance bioactivity but may reduce fiber porosity and pore diameter, potentially hindering cell infiltration. This suggests the need to optimize the balance between mineral content and structural accessibility rather than maximizing either [91,92].

Bioactive glass (BAG), developed by Hench as an alternative to inert implants, forms a hydroxyapatite-like layer in physiological fluids [93]. It enhances odontogenic gene expression (DSPP, DMP1, ALP, and osteopontin) and boosts the proliferation of DPSCs when used in combination with calcium phosphate cements or polymer scaffolds [94,95]. Bioactive glass nanoparticles doped into PCL/gelatin nanofibers at a 70:30 polymer ratio have been reported to accelerate apatite-forming bioactivity relative to undoped fibers [96]. Furthermore, BAG-functionalized scaffolds have been shown to support endothelial migration and tubular network formation. This suggests a role in promoting the vascularization needed for pulp-dentin complex regeneration, in addition to mineralization [94]. Several BAG-based formulations have FDA-cleared clinical precedents for bone and periodontal repair. This lends regulatory familiarity to their incorporation into newer electrospun composite designs [97].

β-Tricalcium phosphate (TCP) and biphasic HA/TCP ceramics are resorbable calcium phosphates. Unlike HA, these materials undergo active dissolution and remodeling in vivo, thereby improving DPSC adhesion, proliferation, and odontogenic differentiation in TCP-based scaffolds [98]. Resorbable calcium phosphates are preferred over pure HA in applications where scaffold resorption must align with new tissue formation. This is particularly relevant for the repair of alveolar bone and periodontal defects, as these materials actively participate in the turnover of mineralized tissue rather than remaining as inert fillers [99]. Mesoporous bioactive glass nanoparticles and other engineered mineral carriers have been used as platforms for the sequential or sustained delivery of osteogenic growth factors from hollow or core–shell electrospun fiber composites. This extends the role of mineral fillers beyond passive bioactivity toward active, time-controlled biochemical signaling [100,101]. Table 1 summarizes the various polymer/composite-based scaffolds used in dental applications.

Table 1. Summary of polymer and composite systems used in electrospun dental nanofibrous scaffolds, their key properties, and associated dental applications.

Polymer/Composite	Type	Key Properties	Dental Applications	Refs.
Collagen/Gelatin	Natural	ECM mimicry, cell-binding motifs	Pulp capping, periodontal scaffolds	[46,50]
Chitosan	Natural	Intrinsic antibacterial activity	Periodontal barrier membranes	[2,52]
Silk fibroin	Natural	Mechanical robustness, slow degradation	Dental pulp/craniofacial scaffolds	[53,55]
Cellulose acetate	Natural	Hydrophilicity, drug-carrier capacity	Root-canal antiseptic delivery (chlorhexidine)	[57,58]
PCL	Synthetic	Slow, tunable degradation	Guided bone regeneration, pulp capping	[69,70]
PLA/PLGA	Synthetic	Tunable degradation	Oral mucosa regeneration, GBR membranes	[71,72,81]

Table 1. *Cont.*

Polymer/Composite	Type	Key Properties	Dental Applications	Refs.
PVA/PU/PEO	Synthetic	Processing aid; mechanical durability	Composite reinforcement, mucosal repair	[77–79,81]
PCL/Gelatin	Synthetic/natural composite	Balanced bioactivity	Periodontal regeneration, pulp capping	[46,67,70]
PLGA/Gelatin, PLGA/Collagen	Synthetic/natural composite	ECM mimicry with tunable degradation	Tooth-root/dentin-pulp regeneration	[69,83,84]
PCL/HA, PLGA/HA	Synthetic/bioceramic composite	Osteo/odontogenic bioactivity	Alveolar bone, dentin-pulp regeneration	[88–90]
PCL/Gelatin/Bioactive glass	Synthetic/natural/ bioceramic composite	Apatite-forming bioactivity, angiogenesis	Pulp-dentin complex regeneration	[94,96]

4. Applications in Dentistry

Electrospun nanofibers have transitioned from proof-of-concept polymer mats to application-specific platforms aimed at addressing distinct challenges in dentistry, including periodontal regeneration, alveolar bone repair, pulp revitalization, and antimicrobial wound management in the oral cavity. A bibliometric analysis of the literature showed a gradual increase in dental electrospinning research publications, with the majority of studies focusing on periodontal and bone regeneration [13]. The key physical property identified by reviews of the field as playing a major role in the adoption is that the diameter of the fibers is close to the size of the native extracellular matrix collagen, thus enabling cell attachment and infiltration, which flat films do not [11]. More recent reviews have discussed this developing body of evidence in terms of three translation themes: fabrication control, functional loading of active agents, and movement towards clinical products [102]. The new generation of electrospun mats for oral disease management features a combination of mechanical scaffolding and drug delivery, which simultaneously treats infection, inflammation, and loss of tissues at the same time [81].

4.1. Periodontal Tissue Regeneration

Guided tissue regeneration (GTR) employs a barrier membrane to prevent the incursion of rapidly proliferating gingival epithelium and undesired connective-tissue cells into a periodontal defect. This approach preserves space for the slower regeneration of periodontal ligament, cementum, and alveolar bone tissues. Current research is exploring electrospun membranes as advanced GTR barriers, as their fibrous architecture offers a combination of cell-occlusive properties with adjustable bioactivity, degradation rates, and localized therapeutic delivery [44]. Budai-Szűcs et al. designed a periodontal wound-healing scaffold with controlled-release functionality to maintain the membrane's activity throughout the healing process, not just as a physical barrier [103]. Direct-writing electrospinning is a technique that deposits fibers with controlled and digital precision rather than collecting the fibers in the form of a random mat. This technique has been used to create a PD scaffold (Figure 6) with tunable pore geometry and has been validated in vitro [104]. The membranes reviewed by Epicoco et al. were functionalized with loaded agents, which were specifically designed to regenerate the periodontal complex, not only the bone [105]. A recent study dedicated to periodontal applications lists this growing arsenal of drug-eluting and bioactive electrospun membranes, ranging from antibiotic-loaded fibers to

platelet-derived factor carriers [12]. A related study outlined the use of decellularized extracellular matrix as a component of the nanofibrous periodontal scaffolding to restore a more native microenvironment [106], and another review provided an overview of the developments regarding electrospun nanofibers for periodontal applications [107].

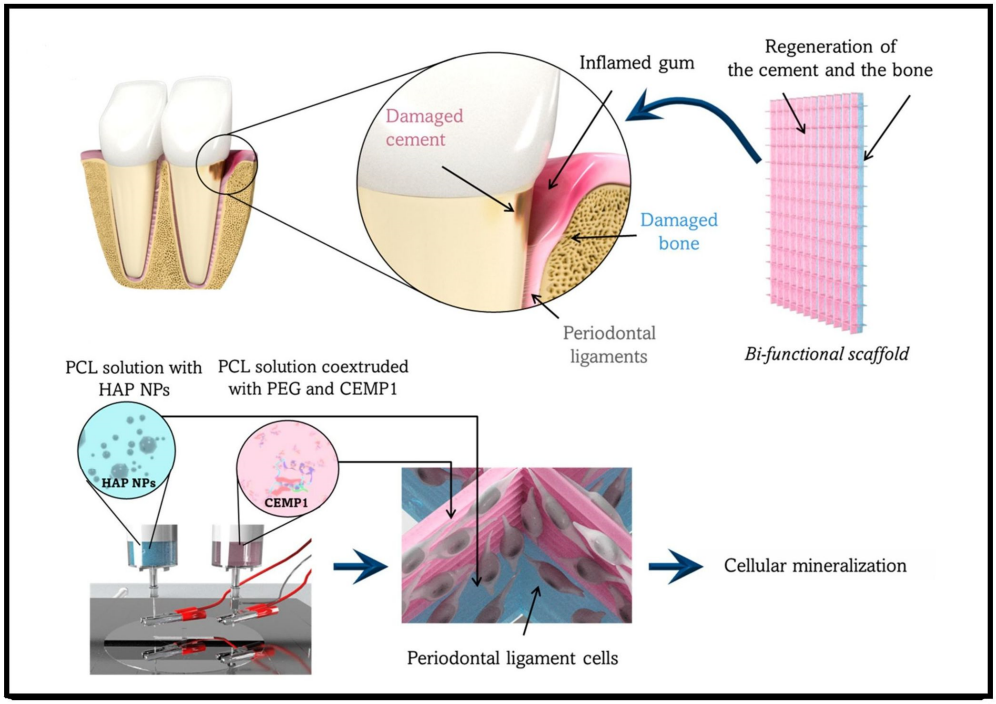


Figure 6. Schematic representation of the development of electrospun scaffolds for periodontal tissue regeneration. Reproduced with permission from [104] under CC-BY-4.0.

Antibacterial activity has become a recurring design target in periodontal membranes because periodontitis is fundamentally an infection-driven disease. Zhuang, Lin, and Yu reviewed nano-composite electrospun fibers built for periodontal regeneration, noting a shift toward mats that incorporate inorganic nanoparticles for both mechanical reinforcement and antimicrobial effects [2]. Liu et al. designed a biodegradable multifunctional nanofibrous membrane that combined periodontal tissue regeneration with additional functional layers in a single device [108], and Abdelaziz et al. embedded biodegradable nanoparticles within nanofiber membranes (Figure 7) to boost antibacterial performance alongside guided periodontal and bone regeneration [109]. Drug-loaded periodontal membranes have also been used to target bone metabolism directly. Azami et al. constructed a dual-function nanofiber membrane that released alendronate sodium, a bisphosphonate that slows bone resorption, for combined periodontal tissue and bone regeneration [110]. Inorganic and antibiotic loading has been explored from several angles, including magnesium oxide nanoparticles incorporated into coaxial PCL/gelatin-derived nanocellulose membranes for periodontal tissue regeneration [111], moxifloxacin-loaded mesoporous silica carriers embedded in electrospun PLGA membranes [112], and cerium oxide nanoparticles loaded into nanofibrous membranes to support bone regeneration in a periodontal context [113]. More structurally complex fiber architectures have also been developed, such as a uniaxial Janus nanofiber that releases zeolitic imidazolate framework-8 and the immunosuppressant FK506 in two separate phases for periodontal tissue regeneration [114], a 3D-oriented polycaprolactone/gelatin scaffold designed to guide periodontal tissue regrowth along a specific direction [115], and a trilayer eccentric Janus nanofiber built for the combined treatment of periodontitis that layers distinct functions across the fiber cross-section [116]. Coaxial fiber designs have also been used to manage the inflammatory side of periodon-

titis directly, as in a coaxial nanofiber loaded with agents to control inflammation while promoting alveolar bone regeneration in periodontitis-related defect models [117].

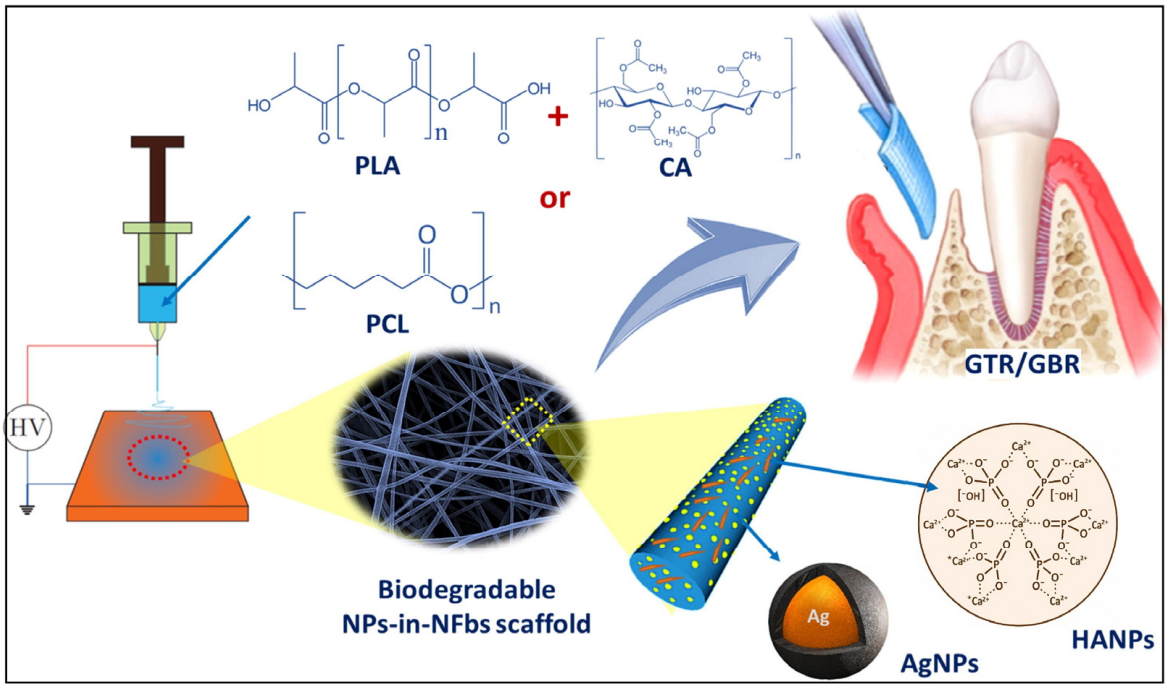


Figure 7. Schematic representation of the development of electrospun scaffolds for enhanced antibacterial and bone regeneration. Reproduced with permission from [109] under CCBY-NC-ND.

4.2. Alveolar Bone Regeneration

Beyond the periodontal complex, electrospun scaffolds are widely used for guided bone regeneration (GBR) of alveolar bone lost to periodontitis, trauma, or tooth extraction, where the same barrier membrane principle applies, but the emphasis shifts toward osteoconductive mineral content. Boda et al. developed mineralized nanofiber segments coupled with calcium-binding BMP-2 peptides to promote alveolar bone regeneration, an approach that pairs a mineral phase with a growth factor-mimicking peptide rather than the growth factor itself [118]. Composite membranes (Figure 8) built around β-tricalcium phosphate embedded in a GelMA/PCL matrix have been evaluated for GBR at the nanoscale [119], and a bilayered membrane combining polyamide-6/chitosan nanofibers with a nano-hydroxyapatite/polyamide-6 composite layer has been designed to reinforce bone regeneration while providing the device with sufficient strength to hold its shape under a bone defect [120].

Some groups have combined nanofiber scaffolding with other biomaterial classes rather than relying on fibers alone, as in a multifunctional hydrogel/platelet-rich fibrin/nanofiber scaffold built to provide both a cell barrier and an osteogenic signal for combined GTR/GBR [121]. Growth factor delivery for alveolar healing has moved toward extraction-socket applications. Huang et al. loaded concentrated growth factors and nano-hydroxyapatite into electrospun fibers (Figure 9) to accelerate alveolar bone healing after tooth extraction [122]. This bone-focused study sits within a broader body of electrospun scaffold research for bone tissue engineering that is not dentistry-specific but underlies much of the dental bone literature, including biomimetic nanofibrous scaffold designs reviewed for regenerative medicine [123] and adipose-derived mesenchymal stem cell-based scaffolds developed for oral and maxillofacial reconstruction [124].

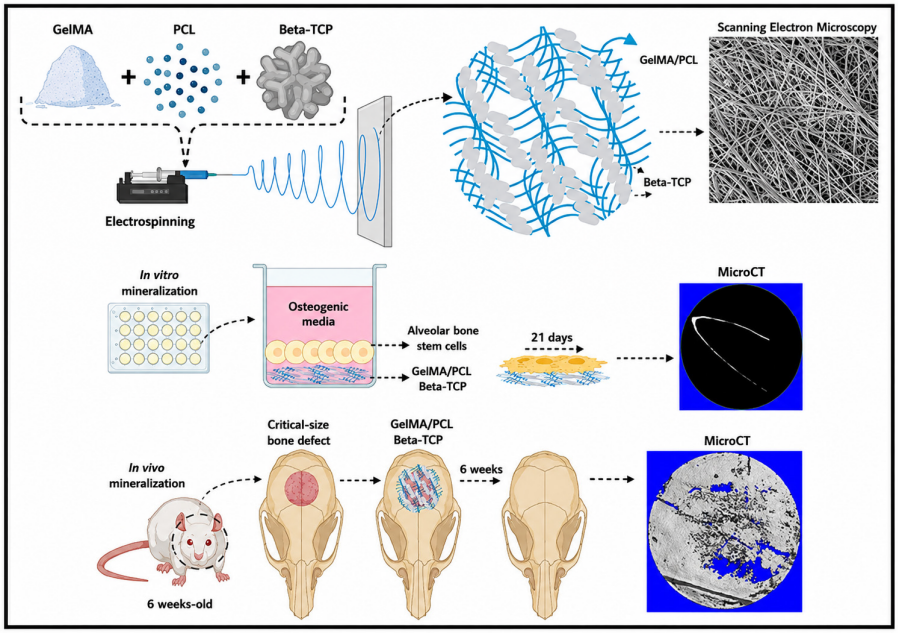


Figure 8. Schematic illustration of the development of electrospun GelMA/PCL-TCP composite scaffolds and their potential application in guided bone regeneration (GBR). Reproduced with permission from [119]. Copyright © 2023 American Chemical Society.

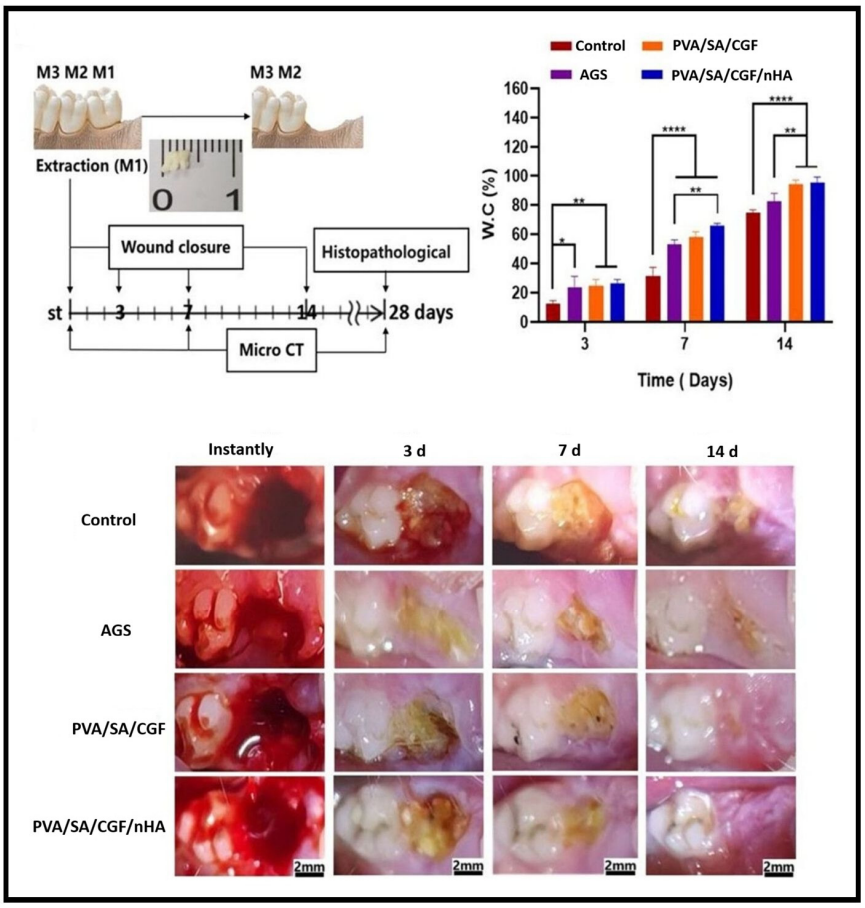


Figure 9. Schematic demonstrating the application and healing process of an electrospun PVA/SA/CGF/nHA-based membrane on a wound model of an extracted rat tooth. Statistical significances were shown as * $p < 0.05$, ** $p < 0.01$, and **** $p < 0.0001$. Reproduced with permission from [122]. Copyright © 2025 American Chemical Society.

4.3. Dentin-Pulp Regeneration and Regenerative Endodontics

Pulp regeneration requires more scaffolding than bone or periodontal repair because the fiber network must support a densely vascularized, innervated tissue inside a confined root canal space. Ballikaya et al. incorporated a proanthocyanidin/β-cyclodextrin inclusion complex into electrospun polylactic acid nanofibers (Figure 10) and observed measurable effects on dental pulp stem cell behavior, positioning the fiber as both a delivery vehicle and a scaffold for pulp cells [125]. A more recent bioengineering approach loaded polycaprolactone nanofibers with an RGD cell-adhesion peptide alongside asiatic acid, a compound associated with tissue repair, specifically for dentin-pulp regeneration [126]. Growth factor delivery has also been adapted for pulp applications. An electrospun heparin-modified nanofiber membrane was designed to carry multiple growth factors for dental pulp regeneration, using heparin to bind and slow-release the factors rather than losing them in a fast burst [16]. Outside of regeneration itself, electrospun fibers have been proposed as an alternative to the antibiotic pastes conventionally used in regenerative endodontic procedures, with a direct comparison reporting differences in how the two delivery formats control canal disinfection [17].

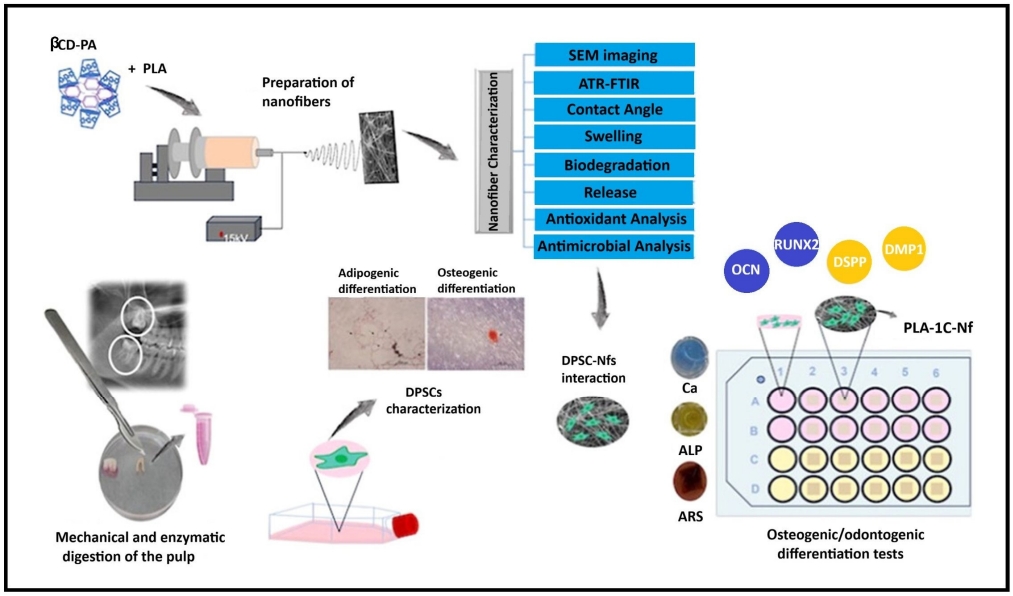


Figure 10. Schematic illustrating the development of Proanthocyanidin (PA)/β-Cyclodextrin (β-CD)/PLA-based nanofibers and their evaluation as a scaffold for dental pulp stem cell viability and odontogenic differentiation. Reproduced with permission from [125] under CC-BY-4.0.

4.4. Antibacterial and Drug-Delivery Functionalities

Nearly every dental application described above depends on an antimicrobial or drug delivery function layered onto the fiber, which is why this has become its own design discipline within the field rather than an afterthought. A review of antibacterial nanomaterials integrated into electrospun membranes (Figure 11) surveyed the fabrication routes used to impart antimicrobial properties to fibers, from metal nanoparticle loading to antibiotic encapsulation, across biomedical applications [127]. Ye et al. reviewed how electrospinning process parameters govern drug loading efficiency and release kinetics in tissue engineering scaffolds, which is a set of principles that dental-specific studies routinely draw on [128]. Two further reviews cover the broader polymer toolbox behind these functions: biopolymer-based electrospun nanofibers and their biomedical applications [129], and collagen- and chitosan-based electrospun fibers for tissue engineering and drug delivery [130], both of which underlie the material choices seen in the periodontal and pulp studies discussed above.

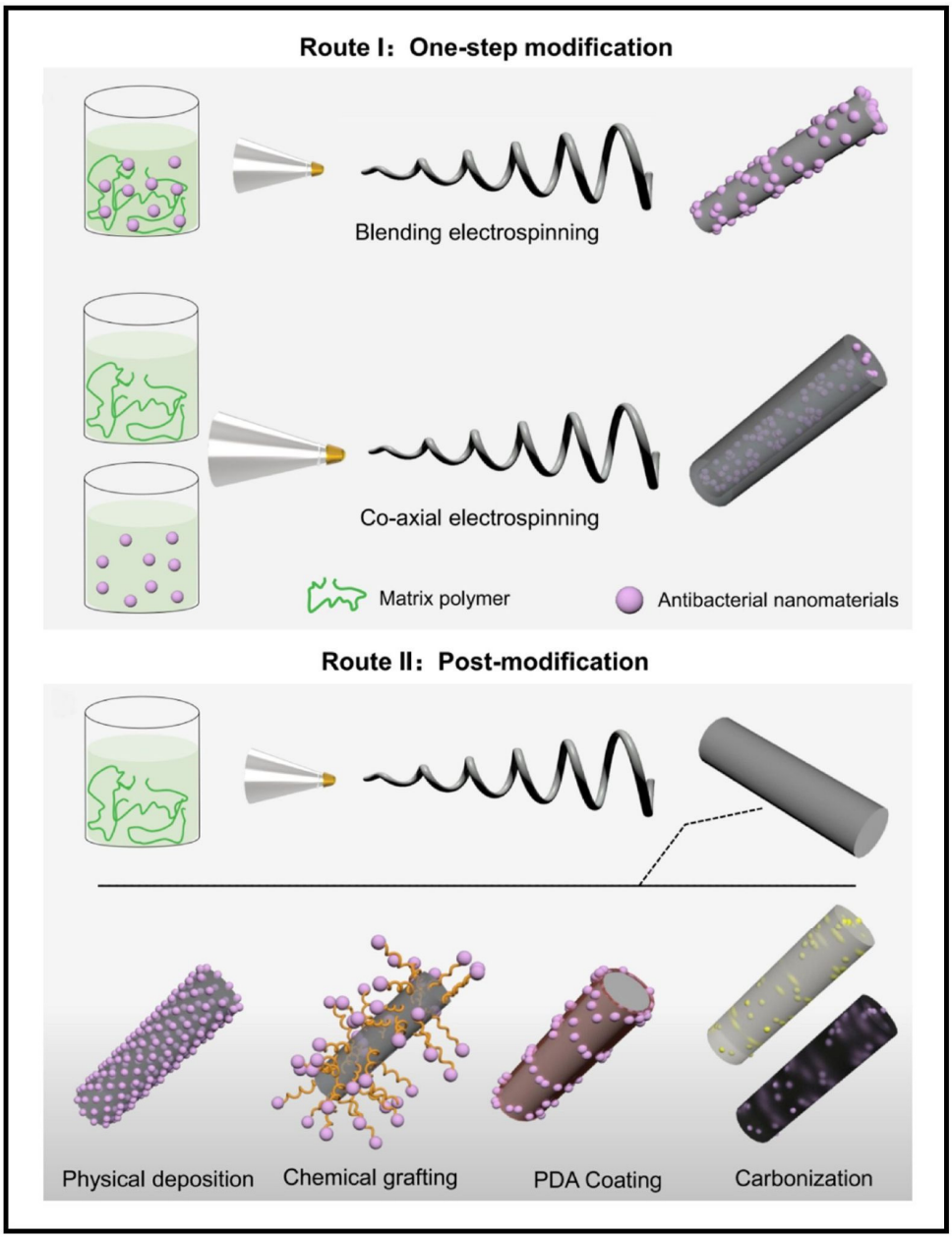


Figure 11. Schematic illustration of strategies for fabricating antibacterial nanofiber-based materials, comparing Route I, one-step modification via blending and co-axial electrospinning method, and Route II, post-modification of plain electrospun fibers through physical deposition, chemical grafting, polydopamine (PDA) coating, or carbonization to incorporate antibacterial functionality. Reproduced with permission from [127] under CC BY-NC-ND.

4.5. Oral Wound Healing and Soft-Tissue Repair

Oral soft-tissue healing, whether after periodontal surgery, extraction, or mucosal injury, draws on wound-healing literature that is largely generic to the skin and mucosa rather than exclusive to the mouth; however, the underlying mechanisms translate directly. Gao et al. laid out the wound-healing theories relevant to nanofiber dressings, covering how fiber porosity and surface chemistry affect the phases of healing from hemostasis to remodeling [131]. Reviews of gelatin-based nanofiber dressings [132] and medicated nanofiber dressings [133] describe design choices that map onto oral applications, where a scaffold must manage a moist, bacteria-rich environment, similar to an oral wound. Further reviews have tracked recent progress in nanofiber wound dressings [134], polysaccharide-based electrospun dressings [135], and composite nanofiber dressings moving toward

clinical use [136], along with organic/inorganic hybrid fiber designs aimed at accelerating healing [137]. Two further reviews focused on the structural and functional design choices behind wound-healing nanofibers [138] and the fabrication-to-application pipeline for functional electrospun mats in wound care [139], and another survey on bioactive compound-loaded electrospun fibers for wound healing across natural and synthetic polymer platforms [140]. Metal–organic framework composites represent one of the newer additions to the wound-care toolbox, with MOF-loaded electrospun fibers (Figure 12) reviewed for their combined antimicrobial and regenerative effects [141]. None of these studies specifically targeted the oral cavity; however, periodontal and post-extraction soft-tissue applications borrowed directly from these general wound-healing design principles.

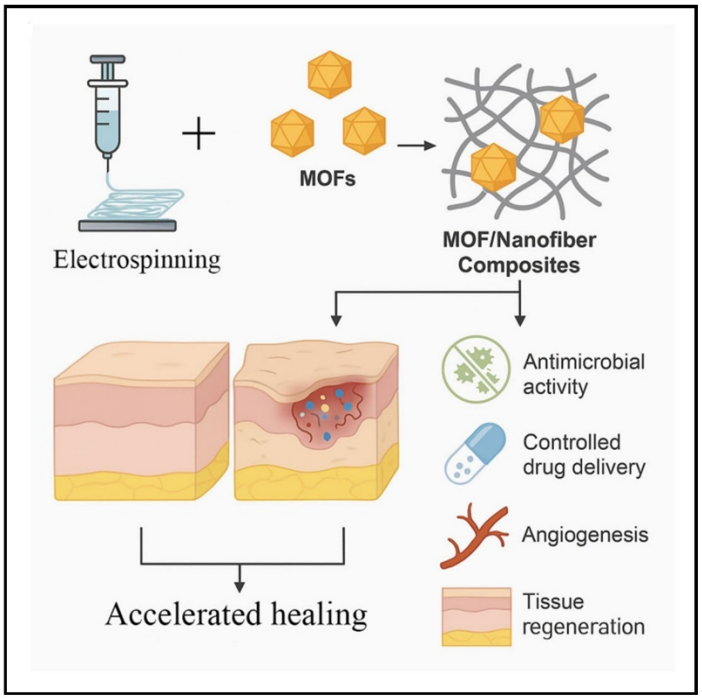


Figure 12. Schematic representation of Metal–Organic Framework (MOF)/nanofiber composites as a promising advancement in the field of wound healing, providing a versatile platform for the controlled delivery of therapeutic agents and the promotion of rapid healing. Reproduced with permission from [141] under CC-BY-4.0.

4.6. Comparative Performance and Evidence Strength

Currently, no single electrospun formulation can be deemed universally optimal for dental regeneration. Scaffolds primarily composed of PCL are known for their consistent mechanical durability and ease of electrospinning. Yet, their hydrophobic nature and limited inherent bioactivity often necessitate surface alterations or blending with other materials. Fibers that include natural polymers, such as gelatin/collagen or chitosan, tend to enhance cell adhesion and biological signaling, though they typically require reinforcement or crosslinking to maintain their barrier properties. Adding HA, TCP, or bioactive glass can improve mineralization outcomes in hard-tissue applications, but excessive amounts of inorganic materials might negatively affect fiber structure, pore accessibility, and mechanical uniformity. Advanced designs like coaxial, multilayer, and Janus architectures can offer better-controlled release and compartmentalized functions, yet they are less developed for practical use due to increased process complexity, stringent quality control, and higher regulatory demands when drugs or biologics are involved. Further, Table 2 illustrates the evidence-based potential of electrospun nanofibers in dental applications, indicating that electrospun nanofibrous membranes are primarily at the

in vitro and animal experiment stages and highlighting the necessity for additional animal studies before human trials.

Table 2. Illustrates the evidence-based potential of electrospun nanofibers in dental applications.

Material	Dental Indication	Study Type	Major Outcome	Translation Stage	Ref.
PCL/gelatin/ nHA	Dentin/pulp regeneration	In vivo and in vitro	Scaffolds supported DPSC adhesion/proliferation and odontoblastic differentiation.	Small-animal preclinical	[88]
PCL/equine bone-derived nHA	Guided bone regeneration/alveolar bone regeneration	In vitro	Enhancement in cell viability and calcium deposition, and the expression of osteogenic markers.	In vitro proof of concept	[90]
PCL/gelatin	Periodontal regeneration	In vivo and in vitro	Scaffold supported bone regeneration.	Small-animal preclinical	[115]
PLGA/PCL/Ag/ collagen	Periodontal/alveolar bone regeneration with antibacterial activity	In vivo and in vitro	Improved cell adhesion/proliferation and osteogenic markers.	Small-animal preclinical	[84]
PLGA/collagen/ gelatin/calcium-binding bone morphogenetic protein 2	Post-extraction/ maxillary alveolar bone regeneration	In vivo	Greater new bone volume and bone mineral density than unfilled defects.	Small-animal preclinical	[118]
Chitosan/PCL/ PEO/ornidazole	Direct pulp capping/local antimicrobial delivery	In vitro	Minimal cytotoxicity and antimicrobial inhibition zones, supporting feasibility as a pulp-capping coating.	In vitro proof of concept	[142]
PVA/ sodium alginate/ concentrated growth factor/ nHA	Post-extraction alveolar bone healing	In vitro and in vivo	Scaffold supported enhanced osteogenic differentiation and promoted bone-healing outcomes.	Small-animal preclinical	[122]

5. Clinical Translation and Challenges

Although extensive preclinical research has been conducted, the integration of electrospun scaffolds into routine dental applications remains limited, with obstacles falling into several common categories.

Manufacturing is a significant challenge. Single-needle electrospinning produces approximately 0.01–1 g/h, which is adequate for small-scale experiments but insufficient for commercial membrane production [40]. While needleless and multi-jet systems can increase output to hundreds of grams per hour, they also result in a broader and less controllable fiber-diameter distribution. This poses a problem for barrier membranes, where pore size is crucial for cell permeability. The use of natural polymers introduces further variability: materials like collagen, gelatin, and alginate can differ between batches due to variations in extraction and biological sources, meaning a scaffold that works well

with one batch might not perform similarly with another. Combining electrospinning with melt electrowriting or 3D printing could be beneficial, as it allows for the addition of macro-scale structures, channels, and patient-specific geometries based on CT data, complementing the ECM-like fiber texture [143]. Machine learning is already reducing the trial-and-error process in adjusting parameters like voltage, flow rate, and concentration in other electrospinning applications [144], and applying this to dental polymer systems could help manage batch-to-batch variability.

Sterilization presents another challenge. The standard methods for sterilizing resorbable devices, gamma irradiation and ethylene oxide, can impact degradation rate, mechanical strength, and the stability of any incorporated antibiotics or growth factors. This effect was observed in PLGA nonwovens used for oral mucosa repair, where the sterilization method influenced degradation as much as the copolymer ratio [66,145]. However, most studies do not specify the sterilization method used or whether it affected the payload before implantation, complicating cross-study comparisons.

Regulatory challenges also exist. Scaffolds that release a drug or growth factor face a more stringent regulatory process than passive membranes, as they are typically assessed as drug-device combinations rather than standalone devices [146]. This increases both time and cost, likely explaining why few designs loaded with antibiotics or growth factors have advanced beyond the laboratory stage.

Preclinical validation is also inconsistent. Testing conditions vary significantly between studies, including differences in wet versus dry mechanical testing, strain rate, gauge length, degradation medium (PBS, simulated body fluid, artificial saliva), and the timing of release-kinetics measurements. None of these factors are standardized or consistently linked to the clinical site intended for the scaffold. A scaffold that appears mechanically superior in one laboratory's setup may not perform well in another's, and there is currently no standardized protocol for reporting these parameters. Stimuli-responsive fiber chemistries that release payloads based on local pH, enzyme activity, or bacterial presence, rather than a fixed diffusion schedule, offer a way to align drug release with the actual progression of infection or healing instead of a lab-derived timetable [147].

Clinical evidence is another area of concern. The majority of studies discussed in this review are limited to cell culture or small-rodent defect models. There is a lack of data from large-animal studies, and controlled human trials are nearly absent, except for a few products that have already been commercialized. Chaudhari et al. found that the range of nanofiber products used in clinical settings is notably more limited in dental applications than in areas such as wound care and drug delivery [148]. This highlights the main issue: the transition from lab-scale production to a scalable, regulator-approved product is a major barrier for these applications. Bridging this gap will require the development of standardized testing protocols, sterilization studies that clearly report their impact on scaffold performance, and large-animal or preliminary human trials to evaluate the effectiveness of these membranes in oral applications compared to laboratory settings.

6. Conclusions

This review examines the journey of electrospun dental biomaterials from their foundational fabrication principles to their potential clinical applications. At the fabrication stage, the creation of fibers is influenced by the interplay of solution characteristics, operational settings, and environmental factors. Advanced techniques like coaxial and emulsion electrospinning, melt electrowriting, and needleless electrospinning enhance control over the localization of payloads, scaffold structure, and production efficiency. Regarding materials, natural polymers offer cell-recognition signals and beneficial biological interactions but often fall short in mechanical strength. In contrast, synthetic polymers provide consistent

mechanical properties and adjustable degradation rates, though they have limited inherent bioactivity. Consequently, blends and composites filled with bioceramics or minerals are the preferred approach to achieve structural resilience alongside osteogenic, odontogenic, or antimicrobial properties.

In dental applications, these design principles have led to the development of increasingly specialized scaffolds for periodontal regeneration, guided alveolar-bone regeneration, dentin-pulp engineering, localized antimicrobial/drug delivery, and oral wound healing. The most advanced systems surpass passive barriers by managing spatial architecture, offering bioactive signals, or ensuring sustained and localized therapeutic delivery. However, the evidence remains largely preclinical, and the promising results observed in laboratories have yet to become standard in dental practice.

The main challenges are now as much about translation as they are about materials: inconsistent and low manufacturing throughput, variability in natural polymer batches, inconsistent mechanical/degradation and release-testing conditions, changes in scaffold or payload performance due to sterilization, and a lack of large-animal and controlled human studies. Advancing toward clinical use will necessitate scalable manufacturing with precise control over fiber and pore sizes, validated sterilization methods, standardized testing relevant to applications, and phased in vivo assessments. Hybrid manufacturing techniques that integrate electrospinning with MEW or 3D printing, along with stimuli-responsive delivery and data-driven process optimization, could further enhance the design of scaffolds tailored to specific patients and defects. Future research should focus on reproducibility, regulatory-ready characterization, and clinically significant comparative studies rather than merely additional proof-of-concept formulations.

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Abbreviations

The following abbreviations are used in this manuscript:

GBR	Guided bone regeneration
GTR	Guided tissue regeneration
EMD	Enamel matrix derivative
PTFE	Polytetrafluoroethylene
ECM	Extracellular matrix
PVA	Polyvinyl alcohol
PLLA	Poly-L-lactic acid
PCL	Polycaprolactone
PDLSCs	Periodontal ligament stem cells
PLGA	Poly(lactic-co-glycolic acid)

BMP-2	Bone morphogenetic protein-2
MEW	Melt electrowriting
RGD	Arginyl-glycyl-aspartic acid
PEO	Polyethylene oxide
PMMA	Polymethyl methacrylate
PU	Polyurethane
HA	Hydroxyapatite
DPSCs	Dental pulp stem cells
BAG	Bioactive glass
TCP	Tricalcium phosphate
SA	Sodium alginate
CGF	Concentrated growth factor
ALP	Alkaline phosphatase
PA	Proanthocyanidin
CD	Cyclodextrin
PDA	Polydopamine
MOF	Metal–organic framework
PBS	Phosphate-buffered saline

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