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Advances in the Application of Polymeric Materials in Oral Care

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Abstract: *Polymeric materials serve as the core material foundation for the development of contemporary oral care technologies, playing an irreplaceable role in daily cleaning, disease prevention, tissue repair, and functional reconstruction. This paper systematically reviews the classification and characteristics, as well as the current research status, of polymeric materials used in oral care. It elaborates on their specific applications in oral hygiene products and oral therapeutic care products, and provides an in-depth discussion on the design strategies of functionalized polymers with antibacterial, repair-promoting, desensitizing, and controlled drug-release properties. Furthermore, the current status of biocompatibility and safety evaluations of these materials is analyzed. Finally, cutting-edge development directions such as eco-friendliness, intelligent responsiveness, and digital personalization are envisioned, aiming to provide theoretical references for scientific research and industrial translation in the field of oral care materials.*

Keywords: *Polymeric materials, oral care, functionalized design, application, biocompatibility*

1. Introduction

Oral health serves as a crucial window reflecting overall human health and is recognized by the World Health Organization (WHO) as one of the fundamental public health issues. According to WHO statistics, approximately 3.7 billion people worldwide suffer from chronic oral diseases such as dental caries, periodontal diseases, and oral mucosal diseases [1–4]. These conditions not only cause localized pain, impaired masticatory function, and compromised facial aesthetics but also exhibit close epidemiological associations with cardiovascular diseases, diabetes, adverse pregnancy outcomes, and digestive system disorders [5,6]. With the accelerating global aging process and the upgrading of public health consumption concepts, there is a growing societal demand for highly effective, safe, and comfortable oral care products and technologies, providing strong momentum for the interdisciplinary integration of oral medicine and materials science [7,8].

Throughout the extensive history of oral care technology, polymeric materials have undergone a profound transformation, evolving from auxiliary additives to functional core carriers [2,9]. Early oral cleaning and protection primarily relied on inorganic abrasives, surfactants, and simple natural colloids, which were limited in functionality and suboptimal in user experience. With the advancements in polymer chemistry, biomedical engineering, and nanotechnology, polymeric materials have gradually permeated various fields of oral care, owing to their tunable molecular chain structures, excellent biocompatibility, favorable mucoadhesive properties, and designable degradation behaviors [3,10,11]. As shown in Figure 1, To date, ranging from daily essentials such as toothpaste and toothbrushes to professional dental composite resins and clear aligners, and further to oral mucosal drug delivery systems with targeted release capabilities, polymeric materials have become an indispensable material foundation underpinning modern oral care technologies [9,12]. The oral environment possesses a unique complexity, characterized by continuous saliva secretion, dynamically changing temperatures and pH levels, highly diverse microbial communities, and sensitive mucosa susceptible to mechanical and

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chemical stimuli. These characteristics impose extremely stringent requirements on the stability, safety, and functionality of materials [3,13]. Most existing reviews on polymeric materials in the oral field focus largely on single application directions, such as drug delivery systems, oral and maxillofacial repair materials, or natural polymers. Few comprehensive reviews have been reported from the perspective of oral care that cover the full spectrum application scenarios of polymeric materials, including restoration, prevention, therapy and daily care products. This review aims to integrate advances in the application of polymers across the entire continuum of oral restoration, prevention, treatment, and daily care; critically evaluate key barriers and corresponding strategies for translation from laboratory investigations to clinical practice; and link material innovation to the global goal of oral health equity, thereby providing a forward looking guiding framework for the future research and development of polymeric materials. Therefore, systematically reviewing the mechanisms of action, current applications, and developmental trends of various polymeric materials within the oral environment holds significant theoretical and practical value for guiding the research and clinical translation of next-generation oral care materials [14]. Based on the above, this review intends to commence with material classification and fundamental properties, elaborate in detail on the application progress of polymeric materials in oral cleaning and professional treatments, deeply explore functionalized design strategies, and provide a comprehensive evaluation of their biosafety and future development directions.

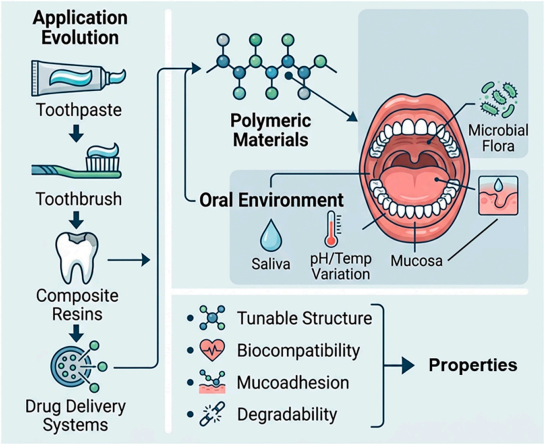

Figure 1. Applications of polymeric materials in oral care

2. Classification and fundamental properties of polymeric materials for oral care

Polymeric materials for oral care encompass a wide range of sources and diverse chemical structures. Based on their origin and properties, they can be broadly classified into two main categories: natural polymeric materials and synthetic polymeric materials. As shown in Table 1, the table includes information on the key properties and parameters, characteristics, advantages, and representative applications of main type of materials [2,4,9,14–20]. Due to their unique molecular structures and physicochemical properties, different categories of polymeric materials exhibit distinct advantages in the field of oral care [21]. Therefore, the selection of these materials requires a comprehensive consideration of the application scenarios, functional requirements, and biological constraints.

Table 1: Comparison of different types of polymeric material in oral care

Material	Key properties and parameters	Characteristics	Advantages	Representative applications
Sodium Carboxymethyl Cellulose (CMC)	Cellulose derivative with linear-chain structure; good water-solubility; broad molecular-weight range; viscosity: 200–12,000 mPa·s (in 1% or 2% aqueous solution).	Exhibits pseudoplastic flow; high thickening efficiency at low dosage; stabilizes abrasives and fluoride suspensions; film-forming capability.	Good affinity with oral mucosa and tooth surfaces; prolongs residence time of active ingredients; good tolerance to electrolytes and pH variations.	Thickener and binder for toothpaste; suspension stabilizer for mouthwash; matrix for oral gels.
Sodium Alginate	Derived from brown algae; block-copolymer composed of β-D-mannuronic acid and α-L-guluronic acid; rapidly gels with calcium ions; high viscoelasticity; extremely high water absorption; viscosity in aqueous solution: 15–850 mPa·s.	Forms viscous gel networks with excellent thixotropy; ionically cross-linked hydrogel; easy to mold and remove.	Biocompatible and biodegradable; strong mucoadhesiveness; enables controlled release of therapeutic ingredients; non-toxic; rapid gelation; reversible molding; low cost.	Gel skeleton for toothpaste; oral mucoadhesive preparations; periodontal sustained-release gels; traditional impression materials; temporary matrices and wound dressings; injectable hydrogels.
Xanthan Gum	Highly branched polysaccharide built from rigid pentasaccharide repeating units; molecular weight generally above 1×10^6 Da; solution viscosity: 1200–1800 mPa·s; for every 5°C increase in test temperature, the viscosity decreases by approximately 8%–12%.	Provides high viscosity at low concentrations; strong shear-thinning behaviour; excellent temperature-, pH- and salt-resistance.	Effectively prevents sedimentation and phase separation of solid components in toothpaste; spreads easily under oral shear force and rapidly recovers structural viscosity after rest.	Rheology modifier for high-end toothpastes; stabilizer for fluoride-free/fluoride-containing toothpastes; oral-care suspension systems.

(Continued)

Table 1 (continued)

Material	Key properties and parameters	Characteristics	Advantages	Representative applications
Chitosan	Degree of deacetylation and molecular weight determine solubility, antibacterial activity, and conformation; can be formed into gels, membranes, and nanofibers; viscosity of 1% acetic acid solution: 125–1500 mPa·s (25°C); MIC values: 0.025%–0.05% (*S. aureus, E. coli*).	Noticeable adhesive force; inhibitory rate >90% against oral pathogens such as *Escherichia coli* and *Staphylococcus aureus*; disrupts biofilm structures; the only cationic natural polysaccharide; dissolves by protonation in acidic media.	Intrinsic antibacterial property derived from cationic character; forms protective layers on enamel to reduce acid erosion and promote remineralization; broad-spectrum antibacterial (inhibits oral pathogens); hemostatic; biocompatible; non-toxic; film-forming, hydrogel-forming, and electrospinnable.	Functional antibacterial toothpaste; bioadhesive mouthwash; oral wound dressings and drug-delivery patches; antibacterial coatings; root canal irrigants; guided bone regeneration barrier membranes; periodontal sustained-release carriers; antibacterial modification of resin-based composites; bioinks; oral implant surface modifications.
Cellulose/ Nanocellulose (CNC/CNF)	Good film forming ability; high specific surface area; thermal decomposition commences at 220–300°C; the particles have a diameter of 5–20 nm, a length of 100–300 nm, and exhibit high crystallinity (>80%). Diameter: 10–100 nm (CNF).	Abundant source; rich surface hydroxyl groups available for functional modification.	Mechanical reinforcing filler; strong film-forming ability; tunable surface chemistry; capable of drug loading for controlled release; biodegradable.	Reinforcing fillers for restorative composite resins; periodontal barrier membranes; drug-controlled release carrier membranes; tissue scaffolds; bioink reinforcing components.
Hyaluronic Acid (HA)	High viscoelasticity; strong water retention; wide molecular weight range.	Natural component of extracellular matrix; natural anionic polysaccharide.	Excellent biocompatibility; promotes tissue healing; lubricating; can be hydrogelized.	Oral soft tissue regeneration; adjunctive treatment for peri-implantitis; drug delivery hydrogels; gingival aesthetic restorative materials.

(Continued)

Table 1 (continued)

Material	Key properties and parameters	Characteristics	Advantages	Representative applications
Gelatin/Collagen (Protein-based)	Temperature sensitive gelation; mechanical properties adjustable by cross linking; enzymatically degradable.	Natural proteins containing cell adhesion recognition sequences.	Promotes cell adhesion and proliferation; good biocompatibility; degradable.	Tissue regeneration scaffolds; electrospun fiber membranes; hydrogels; injectable cell carriers.
Polyacrylic Acid/Carbomer (PAA/Carbomer)	High molecular weight cross-linked polyacrylic acid forming 3 D microgel networks; Carbomer 940 exhibits short term rheological properties, reaching 63,000 mPa·s at 0.5% viscosity, making it suitable for high-viscosity products; whereas Carbomer 941 exhibits long term rheological properties, reaching 7500 mPa·s at 0.5% viscosity, making it suitable for low-viscosity products.	Forms high-viscosity, highly transparent gels at very low dosage (<1%); strong thixotropy and self-wetting property; electrolyte tolerance varies with grade.	Efficiently suspends abrasives and active drugs; excellent oral mucoadhesiveness prolonging drug contact time; easy to rinse with refreshing mouthfeel.	Transparent toothpaste gels; dental bonding primers; topical oral drug-delivery gels; thickening systems for mouthwash.

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Table 1 (continued)

Material	Key properties and parameters	Characteristics	Advantages	Representative applications
Polyethylene Glycol (PEG)	Flexible chain segments; miscible with water and many organic solvents; hydrophilic polymer with adjustable molecular weight; can form films and hydrogels; the viscosity of PEG varies significantly with its molecular weight; low molecular weight PEG (e.g., PEG 200–400) exhibits relatively low viscosity, ranging from approximately 20 to 90 mPa·s (25°C). Medium to high molecular weight PEG can exhibit higher viscosity, for instance, PEG 6000 has a viscosity of approximately 3000 mPa·s at 25°C.	Low-molecular-weight grades (e.g., PEG 400–600): mainly plasticizing and wetting functions; high molecular weight grades (e.g., PEG 4000): retard dissolution and enhance water retention; highly water-soluble; non-immunogenic.	Improves gel flexibility and sensory properties; lowers formulation surface tension and facilitates permeation of active agents; good biocompatibility; "stealth" properties; modifiable active groups; can be co-polymerized with other monomers to regulate hydrophilic/hydrophobic balance.	Humectant/plasticizer for toothpaste; oral fast-dissolving films; flexibility-regulating component for oral mucoadhesives; anti-protein adsorption coatings; drug delivery carriers; bioink base components.
Polyvinyl-pyrrolidone (PVP)	PVP exhibits excellent thermal stability; when heated in air below 100°C, it shows no significant changes. Based on the K-value (Fikentscher K-value), it is classified into four viscosity grades: K-15, K-30, K-60, and K-90.	Excellent film-forming and binding properties; low-MW PVP forms dense hydrogen-bond networks with PVA to improve gel stability	Enhances cohesive force and mechanical strength of pastes or films; modulates drug-release rate when used as co-polymer or drug loading matrix.	Oral mucoadhesive films; binding adjuvant for toothpaste; carrier for topical oral controlled-release drugs.

(Continued)

Table 1 (continued)

Material	Key properties and parameters	Characteristics	Advantages	Representative applications
Polyvinyl Alcohol (PVA)	The viscosity of PVA typically ranges from 4–6 mPa·s (4% aqueous solution, 20°C).	Outstanding film-forming capacity yielding high-strength, highly transparent films; higher molecular weight increases tensile strength yet reduces film flexibility.	Forms temporary protective barriers on teeth or dental restorations; blending with PVP balances mechanical strength; adhesiveness and salivary dissolution rate.	Oral mucoadhesive films; post-operative dental protective films; gel matrices; sustained-release oral patches.
Polymethyl Methacrylate (PMMA)	Tensile strength: 50–77 MPa; flexural strength: 90–130 MPa.	Amorphous; adjustable color; easy to process; conventional heat-curing or CAD/CAM machining.	Low cost; convenient processing; aesthetically acceptable; good color stability; disadvantages include limited mechanical strength, residual monomer, and polymerization shrinkage.	Denture bases; artificial teeth; obturators; individual trays; temporary crowns and bridges; bleaching trays; occlusal splints; impression trays; record bases.
Polyether Ether Ketone (PEEK)	Tensile strength: 90–170 MPa; flexural strength: 140–170 MPa, reaching approximately 24 MPa at 200°C, and remaining at 12–13 MPa under both flexural and compressive loading at 250°C; compressive strength: 100–160 MPa; elastic modulus: 3.6–4.1 GPa; Vickers hardness: 20–30 VHN.	Semi-crystalline; semi-transparent white; high heat and wear resistance; chemical corrosion resistant; CAD/CAM milling or injection molding.	Mechanical properties significantly superior to PMMA (flexural strength ~106% higher, hardness ~116% higher, impact strength ~125% higher); no polymerization shrinkage; low water absorption; good biocompatibility; low allergenicity.	Removable partial denture frameworks and clasps; crown and bridge substructures; implant superstructures and abutments; prosthetic frameworks; orthodontic attachments; occlusal splints.

(Continued)

Table 1 (continued)

Material	Key properties and parameters	Characteristics	Advantages	Representative applications
Polycarbonate (PC)	Bending strength: 100–120 MPa; tensile strength: 60–70 MPa; elastic modulus: 2.0–2.5 GPa; impact strength: 50–70 kJ/m^2.	Amorphous transparent thermoplastic; excellent toughness.	Impact resistant; transparent and aesthetic; lightweight; thermoformable.	Temporary restorations; transparent retainers and orthodontic attachments; temporary crowns.
Polyethylene (PE)	High wear resistance; chemical erosion resistant; extremely low water absorption; tensile strength: 15–25 MPa; elastic modulus: 0.8–1.4 GPa; impact strength: 30–50 kJ/m^2.	Semi-crystalline; low friction coefficient; surface can be porous or dense.	Excellent biocompatibility; good wear resistance; impact resistant.	Implant healing abutments or temporary components; articulating prosthetic components; orthodontic bracket slot materials.
Polyurethane (PU)	Adjustable elasticity and toughness; good energy absorption performance; tensile strength: 40 MPa; yield strength: 25 MPa; elastic modulus: 0.05 GPa; impact strength: 80–120 kJ/m^2.	Block copolymers ranging from soft elastomers to rigid plastics.	Excellent flexibility; fatigue resistance; energy absorption capacity (beneficial for occlusal splint damping).	Occlusal splints; mouthguards; flexible temporary restorative materials; temporary soft lining materials.
Polylactic Acid (PLA)	Tunable crystallinity and molecular weight; ester bonds readily hydrolyse in aqueous environments; degrades over months or longer; high modulus but relatively brittle; high crystallinity; tensile strength: 60 MPa; flexural strength: 87 MPa; elastic modulus: 3.30 GPa; elongation at break: 4%; impact strength: 2.45 kJ/m^2.	High mechanical strength and rigidity; degradation product is lactic acid with good biosafety.	Suitable for oral therapeutic scenarios requiring fast drug release or short-term mechanical support; processable into microspheres, fibres or scaffolds; structurally stable during support period; toughness can be improved by copolymerization.	Oral tissue-engineering scaffolds; absorbable dental sutures; short-term sustained-release drug microspheres; 3D printed surgical guides; absorbable fixation devices.

(Continued)

Table 1 (continued)

Material	Key properties and parameters	Characteristics	Advantages	Representative applications
Polycaprolactone (PCL)	Semi-crystalline hydrophobic polymer; low glass transition temperature (Tg ≈ −60°C); extremely slow degradation; high chain segment mobility; tensile strength: 15–25 MPa; elongation at break: 300–1000%.	Very slow degradation (months to years); flexible chain segments; readily processed by electrospinning or extrusion; long-term structural integrity; semi-crystalline; hydrophobic; low softening point; easily formed into soft matrices.	Suitable for periodontal regeneration therapies requiring long-term drug release or mechanical support; co-blendable with PLA/PLGA to tailor degradation cycles; excellent flexibility; suitable for long-term scaffolds requiring soft tissue.	Periodontal regeneration membranes; drug-release fibrous scaffolds; long-term antibacterial sustained-release systems for chronic periodontitis; soft tissue regeneration scaffolds; electrospun fiber membranes; flexible implants.
Poly(lactic-co-glycolic acid) (PLGA)	Lactide/glycolide (LA/GA) ratio and molecular weight can be precisely controlled, governing crystallinity; degradation rate adjustable by lactide; The 50 : 50 PLGA with Mn = 50 kDa exhibits a tensile strength of 35 MPa and a degradation period of 6 months; the 75 : 25 PLGA with Mn = 80 kDa exhibits a tensile strength of 50 MPa and a degradation period of 12 months; The glass transition temperature ranges from 40–60°C.	Degradation tunable from weeks to months by adjusting LA/GA ratio; exhibits both surface and bulk erosion characteristics; bulk erosion-type hydrolysis; release curve approximates linearity; flexibility superior to PLA.	One of the most clinically mature biodegradable polymers; processable into microspheres, nanoparticles, porous scaffolds and implant coatings for site-specific local drug delivery; precisely adjustable degradation period; excellent biocompatibility; degradation products; strong drug loading capacity.	Dental drug-delivery systems (microspheres/nanoparticles); bone tissue-engineering scaffolds; sustained-release coatings for dental implants; absorbable controlled-release microspheres/nanoparticles; electrospun fiber membranes; local anti-inflammatory and antibacterial sustained-release carriers.

(Continued)

Table 1 (continued)

Material	Key properties and parameters	Characteristics	Advantages	Representative applications
Polyhydroxyalkanoate (PHA)	Molecular weight and co-monomer composition adjustable; the tensile strength was 66.8 ± 1.8 MPa, the Young's modulus was 1322 ± 21 MPa, and the elongation at break under maximum stress was 5.0 ± 0.5%.	Microbially fermented; monomers are natural 3-hydroxy fatty acids; excellent biocompatibility and cell affinity; degradation rate slow to moderate; degradation products are natural metabolites with no cumulative *in-vivo* toxicity; semi-crystalline.	Natural origin and eco-friendly; mechanical properties range from brittle rigid to soft elastomeric materials to satisfy diverse requirements for dental-bone regeneration. good biocompatibility; mild local acidic environment.	Alveolar-bone regeneration scaffolds; oral soft-tissue-engineering scaffolds; absorbable oral restorative materials; Regeneration scaffolds; drug delivery carriers; absorbable barrier membranes (as alternatives or blended components to PLA/PLGA).

2.1. Natural polymeric materials

Natural polymers are primarily derived from plants, animals, or microorganisms and generally exhibit favorable biocompatibility, low toxicity, and renewability [22]. Among these, polysaccharide-based polymers are the most widely employed in oral care. Sodium carboxymethyl cellulose (CMC) is one of the most classic macromolecular additives in the toothpaste industry; the carboxymethyl groups introduced along its molecular chain impart excellent water solubility and rheological control, enabling it to effectively enhance the viscosity and yield value of the system at low dosages, thereby preventing the separation of abrasives from the liquid phase [23,24]. In the presence of divalent cations such as calcium ions, sodium alginate rapidly forms thermo-irreversible gels, a property that makes it an ideal matrix-forming agent for oral ulcer films and periodontal dressings [25,26]. Xanthan gum, a microbial fermentation polysaccharide, exhibits pronounced pseudoplastic rheological behavior: its viscosity decreases sharply under high shear and recovers rapidly at rest. This characteristic complements the thickening action of CMC, and when the two are combined, they produce synergistic effects that significantly improve the stability and texture of the toothpaste. Chitosan, the product of chitin deacetylation, possesses abundant free amino groups along its molecular chain. It not only demonstrates good film-forming ability and mucoadhesion but also displays broad spectrum antibacterial activity by disrupting bacterial cell membranes through electrostatic interactions, thus showing unique advantages in oral antibacterial and wound care applications [27,28]. Nevertheless, natural polymers are generally subject to inherent limitations, including considerable batch-to-batch quality variation, susceptibility to microbial degradation, and the requirement that certain materials be dissolved within specific pH ranges. To some extent, these factors limit their large-scale application in high-end oral care products.

2.2. Synthetic polymeric materials

Synthetic polymeric materials are prepared through precise chemical polymerization reactions and offer outstanding advantages such as narrow molecular weight distribution, controllable structures, and stable performance. Polyacrylic acid compounds, such as carbomer, are highly cross-linked acrylic polymers; their molecular networks contain numerous carboxyl groups that dissociate in alkaline environments, generating like charge repulsion that causes the initially coiled molecular chains to fully extend, thereby leading to an exponential increase in system viscosity [29]. This unique pH-dependent thickening mechanism makes carbomer an irreplaceable excipient in toothpaste gel systems and oral mucoadhesive drug delivery systems [30]. Water-soluble synthetic polymers such as polyethylene glycol (PEG) and polyvinylpyrrolidone (PVP) are often employed as humectants, solubilizers, and crystallization inhibitors. They can form stable hydrogen bond networks with water molecules, effectively maintaining the moisture balance of oral care products during storage and use [25]. Polyvinyl alcohol (PVA), owing to its excellent film-forming properties, mechanical strength, and transparency, is frequently selected as the substrate for oral ulcer patches, forming a dense, flexible, and well-adherent protective layer on the mucosal surface [31]. While providing stable performance, synthetic polymeric materials also face challenges such as residual monomers, potential allergenicity, and environmental persistence. In recent years, related research has gradually shifted toward low-toxicity and environmentally friendly approaches [32].

In addition, easily biodegradable polymeric materials represent an important direction in the development of oral care materials, primarily including polyesters, poly(ether ester)s, and microbially synthesized polymers. Polyester materials such as polylactic acid (PLA), polycaprolactone (PCL), and poly(lactic-co-glycolic acid) (PLGA) contain ester bonds in their main chains, which can undergo hydrolysis and gradual degradation in aqueous environments or under enzymatic action; their degradation rate can be precisely regulated by adjusting the monomer ratio, molecular weight, and crystallinity. These materials have been widely applied in absorbable sutures, drug sustained release microspheres, guided bone regeneration membranes, and other fields [33,34]. Polyhydroxyalkanoates (PHA), synthesized by microorganisms through the fermentation of sugars or lipids, possess physical

properties similar to those of petrochemical plastics and are completely biodegradable. As novel eco-friendly materials, they show promising application prospects in oral care product packaging and disposable devices [34]. It should be emphasized that the selection of oral care materials depends not only on the fundamental properties of the polymers themselves but also requires thorough consideration of their comprehensive performance in the specific oral environment. The moist oral environment demands that materials possess good hydrolysis resistance and an appropriate degree of swelling; enzymes and ions in saliva may cause degradation or crosslinking of certain materials; and the adhesion time on the oral mucosal surface, mechanical strength, and interfacial bonding capability with dental tissues are all key factors determining their clinical applicability [35]. Therefore, an ideal polymeric material for oral care should achieve a delicate balance among biocompatibility, functionality, and environmental adaptability.

2.3. Polymeric materials basic properties and differences

2.3.1. Thickening efficiency

Carbomers serve as the predominant thickeners in modern toothpastes with outstanding thickening efficiency. At a usage concentration of merely 0.5% w/w, carbomers can generate viscosities ranging from 10,000 to 60,000 cP in water. They possess a high yield value, endowing toothpaste with characteristic thixotropy, elasticity, and stand up properties [36]. In an anhydrous toothpaste system thickened with 0.75% carbomer, the addition of only 0.2% crospovidone doubles the system viscosity, demonstrating a prominent synergistic viscosity enhancing effect [37]. Carbomers exhibit optimal performance within pH 6–9 (pKa $\approx 6.0 \pm 0.5$). Below pH 5, microgel collapse triggers a sharp loss of viscosity and yield value. Such pH dependent behavior must be considered in the formulation design of oral care products.

In contrast, chitosan contributes far less to thickening than carbomers. Within the high viscosity non Newtonian matrix of toothpaste, chitosan shows limited thickening capacity at typical application dosages and requires combination with dedicated thickeners such as carrageenan, xanthan gum, or cellulose derivatives. Its overall rheological contribution is strongly governed by molecular weight and formulation pH: high molecular weight chitosan delivers greater viscosity and film forming strength, whereas low molecular weight chitosan displays better dispersibility yet marginal thickening effects [38].

2.3.2. Mucoadhesion strength

Direct comparative studies on toothpaste formulations reveal remarkable differences in peak detachment force and total work of adhesion for toothpastes formulated with distinct bioadhesive polymers on mucin coated substrates. Carbopol-based formulations exhibited the highest peak separation force, ranging from approximately 5 to 6 N; the chitosan formulation exhibited moderate peak separation force (approximately 3–4 N); while commercial toothpastes (Colgate, Aquafresh) generally had lower peak separation forces (approximately 1–2 N) (The contact time was 30 s with 500 g of force, 37°C) [39]. In a comparison of pure polymer films, Carbomer has an average peak adhesion strength of approximately 0.9 N/cm^2, which is the highest among all adhesive polymers commonly used in oral care; the chitosan-PEGAc (10 kDa) derivative follows closely behind (approximately 0.85 N/cm^2); Gantrez MA (methacrylate polymer) is approximately 0.70 N/cm^2; unmodified chitosan is approximately 0.60 N/cm^2; and CMC has the lowest value, at approximately 0.45 N/cm^2. After derivatization, the adhesion strength of chitosan decreases by approximately 13.5%, but it remains about 12% stronger than that of pectin [40].

2.3.3. Degradation cycle

In the moist oral environment, which is characterized by high enzyme activity, there are order-of-magnitude differences in the degradation rates of the three major aliphatic polyesters. The degradation rate of PLGA is relatively controllable: PLGA exhibits the fastest degradation rate at an LA : GA ratio

of 50:50, whereas the degradation time is prolonged at ratios of 75:25 and 85:15. The degradation rate of PLA depends on crystallinity and molecular weight: amorphous, low molecular weight PLA scaffolds can degrade within 4–8 weeks, while highly crystalline, high molecular weight PLA can persist for 6 months to several years [17,41]. PCL degrades extremely slowly, typically requiring 12–24 months to achieve comparable mass loss, making it suitable only for long term scaffolds or ultra-long-acting drug delivery devices.

2.3.4. Antibacterial activity

In terms of inhibiting oral pathogens, microorganisms were grown in Trypticase soy broth, blood agar at 37°C in a atmosphere containing 5% CO_2 high molecular weight (HMW) chitosan presents an minimum inhibitory concentration (MIC) of 0.62 mg/mL and an minimum bactericidal concentration (MBC) of 1.25 mg/mL against *Streptococcus mutans*, and an MIC of 0.62 mg/mL and an MBC of 1.25 mg/mL against *Streptococcus sobrinus*. Low molecular weight (LMW) chitosan yields an MIC of 0.62 mg/mL and an MBC of 1.25 mg/mL against *S. mutans*, and an MIC of 1.25 mg/mL and an MBC of 2.5 mg/mL against *S. sobrinus* [42]. Overall, HMW chitosan possesses superior antibacterial activity relative to LMW chitosan, with more pronounced differences observed against *S. sobrinus*.

Chitosan also exerts potent inhibitory effects on periodontal pathogens. HMW chitosan shows an MIC of 0.5 mg/mL against *Porphyromonas gingivalis*, *Tannerella forsythia*, *Prevotella intermedia*, and *Prevotella buccae*, and an MIC of 2.5 mg/mL against *Aggregatibacter actinomycetemcomitans*. The MIC values of LMW chitosan against these species range from 0.5 to 1.5 mg/mL. Importantly, sub-MIC concentrations of chitosan substantially reduce violacein production by interfering with quorum-sensing signaling pathways ($p < 0.005$), thereby disrupting biofilm formation and maturation. This indicates that chitosan can mediate anti-biofilm effects via quorum-sensing inhibition even at non-bactericidal concentrations [43].

2.3.5. Mechanical properties

PMMA possesses a tensile strength of 50–77 MPa, and a flexural strength of 90–130 MPa. PEEK has an elastic modulus of 3.6–4.1 GPa, which is closer to cortical bone and offers far better stress transfer matching than PMMA. Its Izod impact strength is significantly higher than that of PMMA, making it less prone to fracture in removable partial denture frameworks with retainer notches. For provisional restorations, PEEK demonstrates superior marginal gap (56.00 ± 4.67 μm) and internal gap (128.90 ± 8.39 μm) values compared with PLA and PMMA, together with the highest fracture resistance (840.90 ± 13.23 N). In terms of bonding strength, the shear bond strength between PMMA and acrylic resin teeth is approximately 2–3 MPa (conventional chemical primers yield about 2.5 MPa, while optimized formulations can exceed 3 MPa). Untreated PEEK surfaces only achieve a bonding strength of 0.4–0.8 MPa; nevertheless, it can be increased to 1.8–2.2 MPa after aluminum oxide sandblasting followed by a silica sol-silane coating treatment, approaching the bond strength level of PMMA [44,45].

3. Application of polymeric materials in oral hygiene products

Oral hygiene products represent the most fundamental means of maintaining oral health. Polymeric materials play multiple roles in daily essentials such as toothpaste, toothbrushes, and dental floss, including thickening, stabilizing, film forming, moisturizing, and lubricating. They serve as the core technical support for the realization of product functions and the optimization of user experience.

3.1. Toothpaste

In toothpaste formulation systems, polymeric additives constitute the structural backbone and functional essence of the paste. The thickening and stabilizing system directly determines the formability, extrudability, and shelf stability of toothpaste. Traditional formulations predominantly employ a combined system of CMC and xanthan gum. CMC endows the paste with the necessary consistency

and yield value, enabling abrasive particles to be uniformly suspended without sedimentation; xanthan gum, in contrast, provides high viscosity at low concentrations and exhibits excellent shear-thinning behavior under high shear rates, allowing the paste to be easily extruded from the tube and to break cleanly. The synergistic effect of these two components can effectively prevent liquid-phase separation and flavor volatilization [23,24]. In recent years, with the growing consumer preference for natural and green products, novel microbial polysaccharides such as gellan gum and welan gum have gradually been introduced as bio-colloids into high-end toothpaste formulations. These polysaccharides demonstrate superior stability compared to conventional cellulose ethers in oral care systems subject to considerable variations in ionic strength and pH [22,24]. The humectant system is critically important for preventing water loss and hardening of toothpaste during storage and use. In addition to small molecule polyols such as glycerol, sorbitol, and propylene glycol, polyacrylate-based polymeric humectants can bind substantial amounts of free water through the hydrophilic groups on their molecular chains, thereby maintaining the softness and lubricating sensation of the paste across a wide humidity range. With respect to the loading of flavors and active ingredients, microencapsulation technology has drawn increasing attention. By encapsulating menthol, antibacterial agents, and other substances within polymeric wall materials via gelatin gum arabic complex coacervation or interfacial polymerization, controlled sustained release can be achieved during toothbrushing, significantly prolonging the duration of oral freshness while simultaneously protecting sensitive active ingredients from oxidative or hydrolytic degradation [28,29].

Furthermore, HMW chitosan (>200 kDa) can significantly increase the viscosity of the system and enhance film strength even at a concentration of 0.5%, but it dissolves slowly and requires thorough swelling at a low pH beforehand, which increases process complexity; LMW chitosan (<50 kDa) exhibits excellent dispersibility and dissolves rapidly; however, at conventional addition concentrations (0.5%–1.0%), its contribution to the overall rheology of toothpaste is minimal, making it difficult to replace specialized thickeners such as carbomer or CMC [38]. Studies based on the tensile strength method have shown that toothpaste formulations containing different polymers exhibit significant differences in peak separation force on a mucin-coated substrate. Aspinall et al. used texture analysis techniques to measure the peak separation force and total adhesion work of toothpaste formulations containing different mucosal-adhesive polymers under conditions simulating the oral mucin environment. Taking Carbopol ETD 2020 as an example, this formulation exhibited the highest peak separation force and the greatest total adhesion work, significantly outperforming xanthan gum, Carbopol 974, and the commercially available toothpaste control group. The peak separation forces of most toothpaste formulations tested in the study fell within the range of 0.3–0.6 N, whereas the Carbopol ETD 2020 formulation was significantly higher than this range, indicating stronger mucosal adhesion properties. At the same time, the study also found that there is not a simple linear relationship between formulation viscosity and oral retention: although the Carbopol Ultrez formulation exhibited lower viscosity in rheological testing, its peak separation force and total adhesion energy remained high, suggesting that low viscosity does not necessarily result in low oral retention. Chitosan-based toothpaste was used as a positive control in the study by Aspinall et al., and texture analysis results showed that it possessed an extremely high peak separation force. However, this formulation dried and caked rapidly after simulated brushing, limiting its feasibility in practical applications and preventing further oral retention time testing. This phenomenon is consistent with the mechanism by which chitosan forms electrostatic interactions and hydrogen bonds with oral mucin glycoproteins through protonated amino groups, thereby exhibiting excellent mucosal adhesion. Although chitosan possesses outstanding theoretical adhesion properties, its rapid drying and high viscosity limit its practical application in toothpaste formulations. Studies have shown that chemical modification (such as thiolation) can significantly improve the rheological properties of chitosan: the dynamic viscosity of a 1% thiolated chitosan-mercaptoacetic acid solution can increase up to 16,500-fold under the action of an oxidizing agent, and its apparent viscosity also increases significantly when mixed with mucin. This provides a direction for improving the practicality of chitosan-based oral care products [38–40,46]. Abusrewil

et al. reported that 0.5 wt% high concentration chitosan (300–700 kDa, degree of deacetylation ≥ 92.6%) reduced the viable colony forming unit (CFU) counts of biofilms of four bacterial species by approximately 79% and 90%, respectively [47].

3.2. Toothbrushes and dental floss

Advances in toothbrush and dental floss materials also profoundly reflect the revolution that polymer science has brought to oral cleaning tools. Toothbrush bristles have undergone a transition from natural hog bristles to synthetic polymeric materials; currently, the mainstream materials include nylon 6, nylon 66, polybutylene terephthalate (PBT), and polyethylene terephthalate (PET). Nylon bristles exhibit good resilience and strong abrasion resistance but are prone to dimensional changes upon moisture absorption; PBT bristles possess moderate stiffness and low water absorption, making them more suitable for soft bristle toothbrushes; PET, owing to its cost advantage, is often used in mass market products. The tapering process of bristle tips, accomplished through chemical etching or physical abrasion, gradually reduces the filament ends to micrometer scale dimensions, facilitating deep cleaning along the gingival margin and into interdental spaces while significantly reducing mechanical trauma to the gums [48]. The design of toothbrush handles also exemplifies the ergonomic application of polymeric materials: the thermoplastic elastomer (TPE) overmolded on the outer surface not only provides a soft and comfortable grip but also, through its anti-slip surface texture, maintains stable maneuverability under wet-hand conditions, thereby reducing the risk of slipping during brushing. As an effective tool for mechanically removing interproximal dental plaque, dental floss has seen its base material evolve from natural silk to synthetic polymeric fibers. Ultra high molecular weight polyethylene fiber has become the mainstream choice for premium dental floss due to its extremely high tensile strength, low elongation at break, and excellent lubricity, while the introduction of polytetrafluoroethylene (PTFE) coating further reduces the surface friction coefficient of the fiber, enabling the floss to pass smoothly through tight interproximal contacts without readily breaking [14].

3.3. Mouthwash

Polymeric materials perform multiple functions in mouthwashes. Natural or semi-synthetic polysaccharides such as sodium carboxymethyl cellulose, hydroxyethyl cellulose, and xanthan gum serve primarily as thickening and stabilizing agents. By modulating the viscosity and rheological properties of the system, they improve the mouthfeel, spreadability, and storage stability of mouthwashes, enabling uniform distribution within the oral cavity without being washed away too rapidly. Polymeric materials such as carbomer and chitosan confer mucoadhesive properties to mouthwashes; the former forms hydrogen bonds with mucosal glycoproteins through its carboxyl groups, while the latter generates electrostatic attraction with the mucosa via its positively charged amino groups. Both can form a transient retentive film on tooth and mucosal surfaces, thereby prolonging the residence time of active ingredients [27,29,31]. Drug delivery mouthwashes frequently utilize PLGA, chitosan, and sodium alginate to construct microspheres, nanoparticles, or thermosensitive *in situ* gels. The former achieve sustained release of components such as fluoride and antimicrobial peptides through degradation or diffusion mechanisms, while the latter undergo a liquid-to-gel transition at oral temperature, thereby prolonging drug retention at local lesion sites [30]. In addition, hydrophilic polymers such as PEG, PVP, and HA possess excellent hydration and moisturizing capabilities. They can form a hydrated protective layer on the mucosal surface, mimicking the lubricating function of natural saliva and effectively alleviating dry mouth and associated mucosal discomfort [22].

3.4. Green transition products

Traditional oral cleaning products have also brought about environmental concerns that cannot be overlooked, particularly the PE microbeads once widely incorporated into toothpaste. These plastic particles, typically less than 1 mm in diameter, serve as colorants, abrasives, or stabilizers; however, due to their poor biodegradability, they enter water bodies through municipal wastewater systems and

cause long-term cumulative harm to aquatic ecosystems. Currently, multiple countries and regions, including China, have introduced policies to restrict or ban the use of microbeads in cosmetics and oral care products, which has substantially accelerated the research, development, and industrialization of alternative materials such as cellulose microbeads, jojoba ester microbeads, natural minerals, and biodegradable polyester microbeads [49]. This shift not only reflects a positive response of polymeric materials science to environmental protection but also signifies that oral cleaning products are transforming and upgrading toward a greener and more sustainable direction.

The tensile strength of conventional polypropylene (PP) toothbrush handles generally ranges from 30–40 MPa. By comparison, PLA formulations can reach a tensile strength of 35–50 MPa with a flexural modulus of 2.5–3.5 GPa, which is higher than the typical value of 1.5–2.0 GPa for PP. This indicates that PLA handles can match or even outperform PP handles in terms of stiffness and strength, and are capable of withstanding normal brushing forces [50]. Nevertheless, PLA exhibits an elongation at break of only approximately 6%, far lower than the high ductility of PP, which confers a higher risk of brittleness during daily use. In particular, PLA handles are more prone to fracture under impact loads. Furthermore, the glass transition temperature (T/g) of PLA is approximately 55–60°C, close to the hot water washing temperatures that may be encountered in oral care scenarios, which may cause handle deformation in hot water. In contrast, the T/g of PP is approximately 0°C (melting point ≈ 160°C) and exhibits superior thermal stability under identical conditions [51]. Therefore, PLA handles can serve as alternatives to PP in terms of basic mechanical load-bearing capacity, yet they still show obvious shortcomings in toughness, heat resistance and moisture resistance. Modification strategies such as PLA/PBAT blending and natural-fiber reinforcement are required to bridge these performance gaps.

Traditional nylon (primarily nylon 6 and nylon 6,6) has long been the irreplaceable standard for toothbrush bristles due to its excellent resilience, wear resistance, and appropriate stiffness. Although PLA bristles are marketed as plant-based, their performance has significant shortcomings-PLA's brittleness causes the bristles to bend and deform easily during use and results in poor recovery, leading to a service life far shorter than that of nylon bristles. To date, no studies have demonstrated that PLA bristles meet the effectiveness standards for dental cleaning [52]. In a randomized, controlled, multicenter trial, the modified PLGA membrane produced results similar to those of the e-PTFE membrane in terms of extraction site healing and bone formation; however, it was less effective than the titanium-reinforced e-PTFE membrane in maintaining horizontal bone thickness, and the latter performed better in preventing soft tissue invasion [53].

PLA has a tensile strength of 60 MPa, a flexural strength of 87 MPa, and an elastic modulus close to 3.30 GPa, providing high stiffness; however, its elongation at break is only about 6%, indicating significant brittleness [51,54]; PCL has a tensile strength of only 15–25 MPa, but it exhibits extremely high ductility and toughness, with a degradation period as long as 2–4 years. It also has significant surface hydrophobicity (water contact angle of 80–90°) and lacks intrinsic bioactivity, which limits its application in scenarios involving load-bearing and close cell material interaction [54]. PLA/PCL blends are designed to balance stiffness and toughness; however, even these composites face challenges such as difficulty in precisely controlling degradation rates, insufficient dimensional stability, inadequate mechanical strength under high-stress oral functions, and relatively high material costs and processing complexity.

In summary, biodegradable membranes do not offer "equivalent performance" in all clinical settings; their suitability depends on the morphology of the defect, the required barrier duration, and the mechanical support requirements. Although these materials are designed to be biodegradable, they often lose their load-bearing capacity or barrier function before the end of their intended service life-this is the most fundamental contradiction faced by biodegradable polymers in dentistry [55].

4. Application of polymeric materials in oral treatment and care products

As oral medicine transitions from a purely disease treatment paradigm to a comprehensive model integrating prevention, aesthetics, rehabilitation, and health management, the application of polymeric materials in professional oral treatments and care products is becoming increasingly profound. These applications broadly encompass multiple specialized fields, including prosthodontics, orthodontic treatment, mucosal care, and surgical assistance (Figure 2).

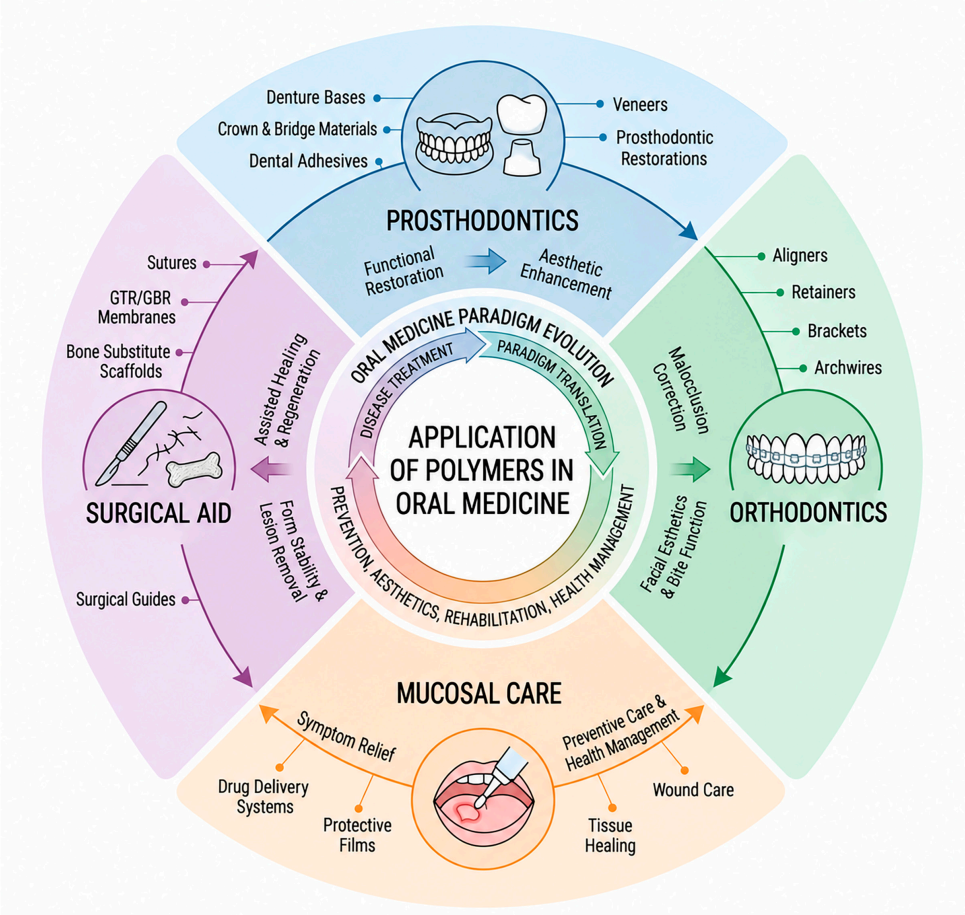

Figure 2. Applications of polymeric materials in oral treatment and care products

4.1. Dental restorative materials

In the field of prosthodontics, polymeric materials serve as the core constituents of modern dental composite resins and adhesives [56]. Dental composite resins typically comprise three main components: an organic resin matrix, inorganic fillers, and a coupling agent. The organic matrix is predominantly composed of bisphenol A-glycidyl methacrylate (Bis-GMA) and its diluent monomer, triethylene glycol dimethacrylate (TEGDMA) [57]. The Bis-GMA molecule contains rigid aromatic rings and multiple polymerizable groups, which endow the cured material with favorable mechanical properties, wear resistance, and dimensional stability. However, due to its high viscosity at room temperature, diluent monomers such as TEGDMA are incorporated to reduce viscosity and facilitate clinical handling [58]. Nevertheless, volumetric shrinkage inevitably occurs during the light-curing polymerization process of resins. This shrinkage may lead to marginal gaps between the restoration and the tooth structure, thereby causing microleakage and secondary caries [59]. To improve the interfacial bonding between the organic

matrix and inorganic fillers, silane coupling agents are widely applied to treat the filler surfaces. By forming chemical bonds, they bridge the polymer matrix and the inorganic phase, thus enhancing the overall performance and durability of the composite resin [60].

The development of dental adhesives has undergone a technological transition from total-etch to self-etch systems. Adhesives containing functional monomers, such as 4-methacryloyloxyethyl trimellitate anhydride (4-META) or 10-methacryloyloxydecyl dihydrogen phosphate (10-MDP), can form stable complex layers with calcium ions in dentin, significantly improving bond strength and hydrolytic stability [56]. Furthermore, PMMA as a classic material for denture bases, possesses excellent processability and aesthetics. However, it is associated with issues of polymerization shrinkage and sensitization caused by residual monomers; prolonged use may induce mucosal irritation and allergic reactions [21].

4.2. Orthodontic materials

The field of orthodontics represents another crucial arena where polymeric materials demonstrate their unique advantages [61]. In recent years, bracketless clear aligner therapy has gained widespread popularity among both patients and clinicians due to its aesthetic appeal, comfort, and removability. The core of this technology lies in the application of precision-engineered thermoplastic polymeric materials. Currently, the primary materials used for clear aligners include polyethylene terephthalate glycol (PETG) and thermoplastic polyurethane (TPU). PETG exhibits excellent transparency, dimensional stability, and processability, providing relatively stable orthodontic force expression at oral temperatures. Conversely, TPU, owing to its superior elasticity and flexibility, demonstrates distinct advantages in expressing complex tooth movements, particularly intrusion and rotation [62,63]. These thermoplastic materials undergo stress relaxation in the oral thermal environment; therefore, the retention duration of their mechanical properties directly dictates aligner efficiency and the required wear cycle.

Furthermore, orthodontic adhesive resins, serving as the connecting medium between fixed appliances and tooth surfaces, typically employ a light cured polymer matrix. By incorporating fluoride releasing glass fillers into a Bis-GMA/TEGDMA matrix, these adhesives can achieve mechanical retention while simultaneously providing a degree of caries preventive effect [64]. The materials used for retainers following orthodontic treatment are also predominantly thermoplastic polymers. The transparency, stain resistance, fracture resistance, and biosafety of these materials are critical considerations for clinical selection and long-term patient compliance [61].

4.3. Periodontal and mucosal care materials

In the realm of periodontal and mucosal care, polymeric materials provide an efficient technological platform for local drug delivery and wound protection. Oral ulcers are among the most common oral mucosal diseases. Ulcer patches or gels formulated with film forming matrices such as chitosan, CMC, PVA, or hydroxypropyl methylcellulose (HPMC) can form a protective film over the ulcer surface. This effectively isolates the wound from mechanical and chemical damage caused by saliva, food, and thermal stimuli [65].

The design of bilayer or multilayer oral patches exemplifies the ingenious combination of polymeric materials. Typically, a highly mucoadhesive polymer is used as the base layer to prolong mucosal retention time, while a hydrophobic polymer serves as the outer layer to control the drug release rate and prevent rapid salivary erosion. For the local pharmacotherapy of periodontal diseases, drug delivery systems represented by *in situ* gels demonstrate unique advantages. Utilizing thermosensitive polymers, these systems remain liquid at room temperature for easy injection but undergo a rapid sol-gel phase transition at body temperature. This allows antimicrobial agents to be retained within periodontal pockets for sustained release, significantly increasing local drug concentrations while minimizing the side effects associated with systemic administration [66,67]. Furthermore, for patients suffering from xerostomia (dry mouth), oral moisturizing gels containing polyacrylic acid or modified cellulose can

form a stable hydration layer on the oral mucosa. By mimicking the lubricating function of natural saliva, these gels effectively alleviate the burning sensation and dysphagia caused by insufficient salivary secretion [25].

4.4. Oral surgical auxiliary materials

The development of oral surgical auxiliary materials also profoundly reflects the clinical value of polymeric technology. Hemostatic sponges commonly used after tooth extraction or periodontal surgery are typically based on gelatin, oxidized regenerated cellulose, or chitosan, achieving rapid hemostasis through a dual mechanism of physical compression and chemical coagulation promotion. Among these, chitosan-based hemostatic materials have been increasingly utilized in recent years due to their natural antibacterial properties, biodegradability, and positive charge, which effectively facilitates the aggregation of red blood cells and platelets [68].

In guided bone regeneration (GBR) procedures, the selection of barrier membrane materials directly impacts the repair outcomes of bone defects. Absorbable collagen membranes and PLGA membranes can avoid the trauma associated with secondary surgical removal; however, their degradation rates must be precisely matched to the bone regeneration cycle. Premature degradation results in the loss of barrier function and the ingrowth of fibrous connective tissue, whereas delayed degradation may impair new bone remodeling [69]. Furthermore, absorbable sutures are gradually replacing traditional silk sutures in oral soft tissue closure. PGA and PLGA sutures can provide sufficient initial tension and subsequently degrade and absorb on their own after wound healing, thereby eliminating the pain, anxiety, and risk of infection associated with suture removal [33,68].

5. Design and application of functionalized polymeric materials

With the deepening interdisciplinary integration of materials science and dentistry, the design of functionalized polymeric materials has become a key strategy for enhancing the therapeutic attributes and clinical value of oral care products. Through molecular structure modification, the incorporation of nanotechnology, and the construction of composite systems, Figure 3 illustrates the functional applications of polymeric materials in modern oral care, which primarily include antibacterial activity, repair promotion, desensitization, and precise drug delivery [28].

5.1. Antimicrobial functional materials

Antibacterial function is the most fundamental and also the most important direction for functional expansion in oral care materials [70]. The oral cavity is one of the body cavities with the richest microbial colonization, and the formation of plaque biofilms is the primary etiological factor for dental caries and periodontal diseases [71]. Traditional antibacterial strategies have largely relied on the release of free antimicrobial agents such as chlorhexidine, triclosan, and essential oils; however, long term exposure can readily lead to bacterial resistance and oral microbiota dysbiosis [72]. To overcome this limitation, researchers have developed contact-active antibacterial polymeric materials. By covalently grafting antibacterial groups onto polymer backbones, these materials exert bactericidal effects upon contact with bacteria through disruption of the cell membrane or interference with metabolism, without releasing antimicrobial agents into the surrounding environment, thereby effectively reducing the problem of drug resistance [71,73]. Quaternized chitosan is a typical representative of this class of materials; the quaternary ammonium cations along its molecular chain can adsorb onto negatively charged bacterial membranes through electrostatic interactions, leading to leakage of cellular contents and cell death [27]. Guanidine-based polymers such as polyhexamethylene biguanide (PHMB) have also been demonstrated to possess favorable oral antibacterial activity, and immobilization onto polymeric carriers can prolong their retention time on the oral mucosa while reducing systemic absorption [72]. Furthermore, uniformly loading inorganic antimicrobial agents such as silver nanoparticles and zinc oxide nanoparticles into polymeric matrices can yield synergistic organic–inorganic antibacterial effects; the sustained release

characteristics of the nanoparticles help maintain a long-lasting antibacterial concentration locally in the oral cavity without producing systemic toxicity [74].

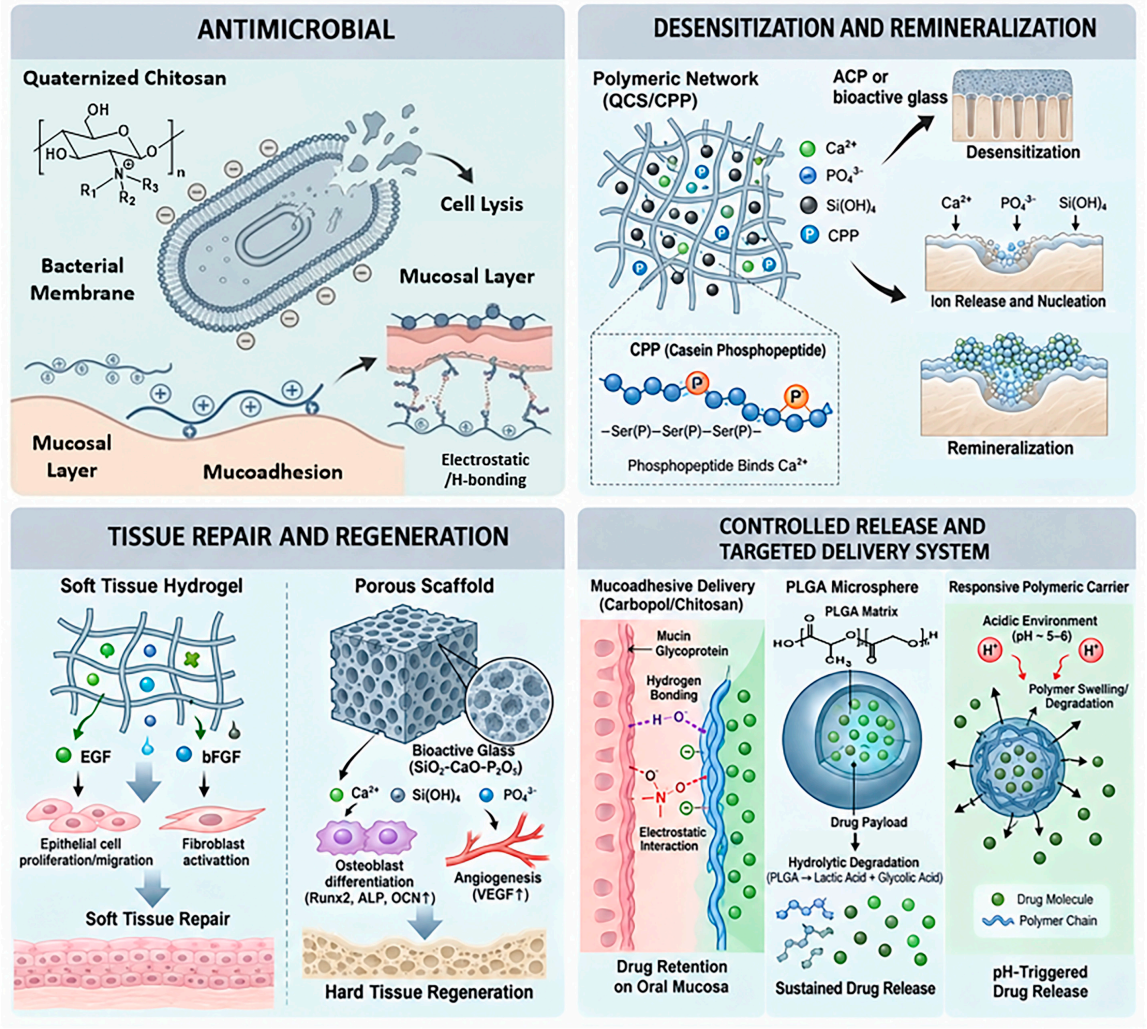

Figure 3. Functional applications of polymeric materials in oral care

Kaur et al. pointed out that contact killing antibacterial surfaces can kill bacteria via physical damage to bacterial cell membranes, without relying on chemical biocides that could be metabolically evaded or enzymatically degraded. Theoretically, this approach does not generate selective pressure that drives adaptive mutations [75]. Qiu et al. systematically illustrated the bactericidal mechanism of cationic polymers: these polymers adsorb onto negatively charged bacterial cell membranes through electrostatic attraction. Subsequently, hydrophobic segments insert into the lipid bilayer and disrupt membrane integrity, resulting in the leakage of cellular contents. This contact-killing process may not release toxic ions or small molecule substances, thereby minimizing the risk of sublethal exposure [76]. Furthermore, multiple long term studies on contact-killing antibacterial polymer coatings have reported sustained antibacterial performance with no emergence of detected drug resistant subpopulations [77]. Nevertheless, it should be noted that most of these studies were performed under controlled laboratory conditions with limited observation periods.

Strategies to reduce the risk of antimicrobial resistance on contact-activated antimicrobial surfaces include: (1) ensuring coating uniformity and durability-by optimizing surface grafting density and cross-linking strategies to avoid sublethal exposure zones caused by coating defects; (2) developing synergistic antimicrobial systems-combining contact-activated bactericidal activity with anti-adhesion functions to reduce initial bacterial colonization and biofilm formation, thereby reducing the selection pressure for resistance at the source; (3) Avoiding reliance on a single target-adopting multi-mechanism synergistic antimicrobial strategies to reduce the likelihood of bacteria escaping bactericidal effects through a single mutation; (4) Establishing a long-term clinical monitoring system-monitoring antibiotic resistance associated with the long term use of contact-antimicrobial oral care products to assess the risk of resistance evolution in real world settings [75–78].

5.2. Materials for promoting tissue repair and regeneration

Promoting the healing of oral soft tissues and the regeneration of hard tissues represents another important development direction for functionalized polymeric materials [79]. Rapid healing of wounds following oral surgery or mucosal ulceration is crucial for alleviating patient discomfort, preventing infection, and restoring function [65]. Hydrogel dressings prepared from natural polymers such as hyaluronic acid, sodium alginate, and chitosan can provide a moist healing environment for the wound, promoting epithelial cell migration, proliferation, and granulation tissue formation [25,26]. By incorporating bioactive factors such as epidermal growth factor (EGF), basic fibroblast growth factor (bFGF), or growth-factor-rich platelet lysate into these hydrogel networks, the tissue repair process can be further accelerated [79]. With regard to hard tissue regeneration, polymeric scaffolds used for guided bone regeneration are typically fabricated from materials such as polylactic acid or polycaprolactone. Through techniques including freeze-drying, salt leaching, or 3D printing, scaffolds with interconnected porous structures are created, providing a biomimetic microenvironment for osteoblast adhesion, proliferation, and differentiation. In the repair of periodontal and oral and maxillofacial tissue defects, polymer-based scaffolds not only provide mechanical support to stabilize blood clots and the defect site but also regulate host cell behavior by mimicking the extracellular matrix (ECM) microenvironment, thereby promoting sequential regeneration of soft and hard tissues. Natural polymers such as collagen, chitosan, and filaggrin, due to their content of cell adhesion sequences such as RGD, can effectively promote the adhesion, spreading, and osteogenic/fibroblastic differentiation of periodontal ligament stem cells (PDLSCs) and bone marrow mesenchymal stem cells (BMSCs); Synthetic biodegradable polymers such as PLA, PCL, and their copolymers can achieve a balance between degradation rate and mechanical strength by adjusting molecular weight and monomer ratios, thereby meeting the mechanical requirements of different defect sites. In recent years, design strategies for functional composite scaffolds have gradually shifted from providing simple structural support to an integrated "structure-activity" approach. For example, loading bioactive factors or antimicrobial agents onto PLGA microspheres and embedding them within a PCL nanofiber scaffold enables the programmed controlled release of these factors, promoting bone regeneration while inhibiting postoperative infection. Furthermore, biomimetic mineralization strategies (such as polymer-induced enamel-like/dentin-like deposition) offer new avenues for the *in situ* regeneration of dental hard tissues. It is noteworthy that in recent years, combining inorganic components such as bioactive glass and nano-hydroxyapatite with biodegradable polymers has significantly enhanced the osteoconductivity and osteoinductivity of the materials. This organic inorganic composite strategy has become a cutting-edge hotspot in the research of materials for oral bone defect repair.

5.3. Tooth desensitization and remineralization materials

For dentin hypersensitivity and early carious lesions, functionalized polymeric materials offer a new avenue for non-surgical intervention [80]. Dentin hypersensitivity primarily results from gingival recession or enamel wear leading to the exposure of dentinal tubules. Functionalized polymeric sealers

form a dense coating on the dentin surface, physically sealing the open dentinal tubules and thereby blocking the transmission of external stimuli to pulpal nerves. Studies have demonstrated that polymers such as modified polyacrylic acid and polylysine containing phosphate groups or carboxyl groups can specifically bind to hydroxyapatite in dentin or undergo *in situ* mineralization, forming a stable protective layer that significantly reduces dentin permeability [81]. With regard to promoting tooth remineralization, polymeric carriers loaded with amorphous calcium phosphate (ACP) or bioactive glass can continuously release calcium and phosphate ions in the oral environment, forming a supersaturated mineralizing solution on the tooth surface and inducing the reprecipitation of demineralized enamel and dentin. Some studies have further explored combining casein phosphopeptide (CPP) with polymeric carriers, utilizing CPP to stabilize calcium phosphate clusters and deliver them specifically to the tooth surface, thereby further enhancing remineralization efficiency and providing new material-based strategies for the non-invasive treatment of early caries [82,83].

Regarding the desensitizing effects of polymer materials on dentin, in a randomized controlled split-mouth clinical trial, 33 participants (99 teeth) were randomized and treated; 31 participants (93 teeth) were analyzed (mITT). Participants received three protocols: GLU: 3% potassium nitrate + glutaraldehyde/HEMA desensitizer (control); GAS: 3% potassium nitrate + glutaraldehyde/HEMA desensitizer + universal adhesive; and GAR: 3% potassium nitrate + glutaraldehyde/HEMA desensitizer + universal adhesive + bulk-fill flowable resin. The results showed that glutaraldehyde/2-hydroxyethyl methacrylate (HEMA) based desensitizing agents combined with potassium nitrate could effectively alleviate hypersensitivity symptoms up to 180 days. However, additional application of universal adhesives or flowable resin sealing layers did not yield statistically or clinically significant incremental improvements [84]. A plausible underlying mechanism is that the glutaraldehyde/HEMA system forms precipitates to occlude dentinal tubules via covalent cross linking between glutaraldehyde and plasma proteins (particularly albumin) within dentinal tubular fluid. HEMA facilitates deeper penetration of glutaraldehyde into tubules and stabilizes protein plugs. The composite structure of poly HEMA and glutaraldehyde cross linked albumin effectively blocks hydrodynamic stimulation arising from dentinal fluid movement [85]. Nevertheless, the outermost polymeric thin film is directly exposed to the oral milieu and is therefore more susceptible to degradation by mechanical wear and acidic challenges. By contrast, protein plugs formed solely by glutaraldehyde/HEMA desensitizing agents reside inside the tubules and may exhibit superior clinical durability [84].

The effects of polymer materials on tooth remineralization include results from both laboratory and clinical studies. Based on a 12-month randomized, single-blind, prospective, controlled clinical trial, Rechmann et al. evaluated the effect of CPP-ACP on the prevention and reversal of white spot lesions(WSLs) during orthodontic treatment with fixed appliances. And the results demonstrated that the enamel decalcification index (EDI) was lower in the experiment group (minimally invasive (MI) Varnish and MI Paste Plus (containing CPP-ACP and fluoride)) than in the control group, and the combination of these substances was able to effectively release fluoride ions in the oral environment and increase their concentration [86]. Another study indicated that both CPP-ACP and CPP-ACPF can decrease the prevalence and increase the remineralization of white spot lesions during/after orthodontic treatment [87]. Their results confirmed the advantages of CPP-ACP containing formulations in terms of visual lesion improvement and elevated mineral content. When CPP-ACP is used especially in combination with fluoride, it exerts superior remineralizing effects on white spot lesions compared with fluoride monotherapy or other remineralizing agents, representing a clinically relevant remineralization option for orthodontic patients and individuals at high caries risk [88]. Milly et al. divided 32 human enamel samples with artificial white spot lesions into 4 experimental groups (n = 8); (a) bioactive glass (BG) slurry, (b) bioactive glass slurry containing polyacrylic acid (BG-PAA) slurry, (c) “standardised” remineralisation solution (positive control) and (d) de-ionised water (negative control). The results showed that BG and BG-PAA surface treatments enhance enamel white spot lesions remineralisation. Although the reduction in lesion depth was not statistically significant, remarkable increases in

phosphorus contents were detected by both Raman spectroscopy and scanning electron microscopy [89]. By constructing a biological model that more closely reflects clinical conditions, this study evaluated the role of chitosan-pretreated enamel surfaces in promoting the remineralization process. Artificial white spot models were fabricated using an acidic gel. The experiment was divided into 6 groups (n = 8): (1) bioactive glass slurry, (2) bioactive glass containing polyacrylic acid (BG + PAA) slurry, (3) chitosan pre-treated WSLs with BG slurry (CS-BG), (4) chitosan pre-treated WSLs with BG + PAA slurry (CS-BG + PAA), (5) "standard" remineralisation solution (RS) and (6) de-ionised water (negative control, NC). And the remineralization treatment was conducted using a pH cycling method for 7 days. Prior to each application of the remineralizing agent, a dental pellicle was allowed to form on the model surface for 3 min. Changes in mineral content, as well as the microscopic hardness and structural characteristics of the surface and subsurface layers, were evaluated using Raman spectroscopy, Knoop microhardness testing, and scanning electron microscopy, respectively. The results indicated that pretreatment with 2.5 mg/mL chitosan solution prior to application of BG or BG + PAA slurry could further boost remineralization efficacy. This phenomenon is attributed to the cationic nature of chitosan, which promotes its adsorption onto negatively charged demineralized enamel surfaces and prolongs the local retention time of bioactive glass derived ions [90].

5.4. Controlled drug release and targeted delivery system

Drug controlled release and targeted delivery systems represent one of the most promising application areas for functionalized polymeric materials [91]. Oral mucoadhesive drug delivery systems exploit the properties of mucoadhesive polymers such as carbomer, chitosan, and polycarbophil to prolong the residence time of formulations on the oral mucosa, thereby enhancing the bioavailability of locally or systemically acting drugs. Microsphere and nanoparticle technologies further improve delivery precision; for example, drug-loaded microspheres fabricated from PLGA can achieve sustained release over weeks or even months, making them particularly suitable for periodontal maintenance therapy that requires long term maintenance of local drug concentrations [92]. Owing to their positive charge and bioadhesiveness, chitosan nanoparticles readily bind to negatively charged mucosal surfaces and have been investigated for delivering macromolecular bioactive substances such as antimicrobial peptides and DNA vaccines [27,74]. At the forefront is the design of stimuli responsive delivery systems. For instance, by taking advantage of the local pH decrease caused by acid production from plaque metabolism during caries, pH-responsive polymeric carriers can be designed to maintain structural stability at normal pH to store the drug, while undergoing conformational changes or degradation when the pH drops to a pathogenic level, thereby rapidly releasing the encapsulated antimicrobial agents and achieving precise targeting of diseased sites [93,94]. This intelligent responsive strategy points the way forward for topical oral drug delivery systems, holding great promise for substantially increasing the therapeutic index of drugs while reducing systemic side effects.

5.4.1 Encapsulation efficiency

The encapsulation efficiency of PLGA nanoparticles is strongly governed by drug hydrophobicity, drug to polymer ratio, and polymer composition. For hydrophobic small molecule drugs, PLGA systems can achieve high encapsulation efficiency. When clindamycin hydrochloride is loaded into PLA and PLGA nanoparticles at an optimal drug to polymer ratio of 1:10, the PLGA system delivers superior encapsulation efficiency and a prolonged release profile owing to its higher hydrophilicity, with sustained drug release lasting up to 144 h [95]. For temozolomide (TMZ) loaded PLGA nanoparticles, the encapsulation efficiency ranges from 52% to 69.67% with particle sizes of 164.4–235.5 nm. These nanoparticles are further incorporated into an 18% Pluronic F-127 hydrogel matrix to construct a composite delivery system [96].

Encapsulation efficiencies of chitosan based nanoparticles are also highly system dependent. For oxaliplatin loaded chitosan nanoparticles ionically cross linked with sodium tripolyphosphate (TPP), the

encapsulation efficiency reaches 89.1 ± 4.7% with a drug-loading content of 44.5 ± 2.4%, exhibiting acceptable physicochemical stability [97]. Chitosan coated lipid core nanocapsules achieve nearly 100% encapsulation efficiency. When combined with chitosan based thermosensitive hydrogels, this composite system presents a narrow particle size distribution with an average diameter of 218–224 nm, and the zeta potential shifts from negative to positive. *In situ* gelation occurs at oral temperature to improve local retention and reduce drug loss caused by salivary washing. Nevertheless, high encapsulation efficiency cannot be guaranteed for all chitosan nanocapsules; some formulations only attain an encapsulation efficiency of 21.5 ± 2.7% and a final drug loading content of 3.5 ± 0.5%. This indicates that the encapsulation performance of chitosan based systems is highly dependent on drug polymer compatibility and fabrication parameters [98]. After chitosan coating, fluconazole nanostructured lipid carriers had an average particle size of approximately 335 nm and an encapsulation efficiency of approximately 73.1%. They exhibited sustained release characteristics *in vitro* and caused only mild histological changes in rabbit oral mucosa; the chitosan coating further enhanced adhesion and prolonged the duration of antifungal activity against Candida [97].

5.4.2 Cumulative release and release duration

PLGA nanoparticles generally exhibit a biphasic release pattern: an initial burst release phase followed by sustained controlled release. Under optimal formulation conditions, clindamycin hydrochloride loaded PLGA nanoparticles sustain drug release for up to 144 h, and display significantly lower MIC values against *Enterococcus faecalis* and *Bacillus cereus* compared with the free drug [95]. After temozolomide loaded PLGA nanoparticles are embedded within Pluronic F-127 hydrogel, only approximately 25% of the total loaded drug is released within 60 days, showing extremely slow sustained release. The release kinetics fit the Korsmeyer-Peppas model, suggesting that PLGA hydrogel composite systems are suitable for oral scenarios requiring long term (month scale) low dose continuous administration, yet are not appropriate for acute infection treatment where rapid achievement of effective drug concentration is demanded [96].

Chitosan coating not only strengthens mucoadhesive properties but also markedly modulates release profiles. In the clinical study reported by Arafa et al., two formulations of ciprofloxacin hydrochloride loaded chitosan coated PLGA nanoparticles (CIP-CS-PLGA-NPs) achieved cumulative release percentages of 50.03% ± 0.7345% and 77.98% ± 3.122% over 72 h, respectively. These formulations yielded superior healing rates, reduced bacterial counts, and enhanced biofilm inhibitory effects compared with conventional approaches during root canal retreatment [99].

Amoxicillin loaded chitosan nanogels achieve an encapsulation efficiency of 78%. The cumulative drug release reaches 86% after 48 h at pH 5.5 (mimicking the acidic microenvironment of inflamed tissues), whereas only 31% is released at neutral pH 7.4 nearly a three fold difference, which validates the trigger efficiency of pH responsive release. Similarly, fluoride loaded PLGA nanoparticles have an encapsulation efficiency of 82%; the cumulative release reaches 92% within 72 h at pH 6.2 but only 35% at pH7.4 [100]. These findings demonstrate divergent release behaviours of stimulus responsive carriers between pathological and normal oral microenvironments, providing experimental evidence for the “on-demand release” design concept.

Mesoporous silica nanoparticles (MSNs), characterised by uniform pore structures (2–10 nm) and high specific surface areas, possess unique loading and release properties for oral drug delivery. Typical small molecule therapeutic agents (fluoride, chlorhexidine, ibuprofen, curcumin, etc.) generally exhibit drug loading contents of 10–30 wt% and encapsulation efficiencies above 70%–90%. For curcumin within silver mesoporous silica hybrid systems, the drug loading content is 25% and the encapsulation efficiency is 77% [101]. Regarding oral release performance, the acidic microenvironment of dental plaque (pH ≈ 5.5) accelerates drug liberation: roughly 20% of loaded agents are released within the initial 24 h, followed by continuous release at a rate of 5%–10% per day, attaining 80% cumulative release within 1–2 weeks. Under neutral pH conditions, degradation is retarded and the release period

can be extended to 4 weeks [102]. When functionalized with pH-responsive polymers via "molecular gating," the gate opens at pH values below 6 to enable rapid drug release during cariogenic attacks, while under normal conditions the gate remains closed to maintain sustained, low-level release, making MSNs suitable for the controlled delivery of antimicrobial agents, anti-inflammatory agents, and remineralizing ions in dental restorations and topical formulations [102].

5.5. Comparison of different functionalized polymeric materials

In terms of functional mechanisms: The core advantage of antibacterial polymers lies in their ability to provide long-term, non-releasing contact-killing activity. Taking quaternary ammonium salt monomers as examples, their cationic groups can be covalently bonded to the resin network to achieve time-independent surface antibacterial effects. At low-to-moderate loading levels (typically $\leq$ 1–2 wt%), the flexural strength, modulus and surface hardness of composite resins can be preserved or even enhanced [103,104]. Zalega and Bociong demonstrated that the incorporation of 7.5% dimethyl-hexadecyl-methacryloxyethyl-ammonium iodide (DHMAI) not only improved the conversion rate but also yielded potent antibacterial effects, while maintaining acceptable mechanical properties. Nevertheless, they possess notable intrinsic limitations: antibacterial activity is largely confined to the material surface and cannot penetrate deep-seated biofilms; high concentrations of quaternary ammonium salts (QAMs) may increase cytotoxicity and water absorption, and compromise polymerization shrinkage and long-term strength. Furthermore, contact-mediated bactericidal action does not equate to zero risk of antimicrobial resistance; bacterial evolution under sublethal exposure and within biofilm microenvironments may still drive resistance development [103–105].

Functional mechanisms of tissue-repair polymers are distinctly different. By constructing a highly hydrated extracellular matrix-mimetic microenvironment, these polymers promote angiogenesis, re-epithelialization, type-III collagen deposition and upregulation of growth factors such as vascular endothelial growth factor (VEGF), thereby accelerating mucosal closure and tissue regeneration [106,107]. Zhao et al. confirmed that gelatin/hyaluronic-acid-based in-situ-forming hydrogels facilitate wound healing by accelerating epithelial reconstruction, angiogenesis and collagen deposition [108]. However, major limitations of such materials include insufficient mechanical load-bearing capacity and difficulties in precisely tuning degradation rates to match the rhythm of tissue regeneration. Moreover, available clinical evidence is mostly derived from small-size cohorts with heterogeneous formulations and short term follow up, and long term efficacy data remain scarce.

Controlled-release polymers feature site-specific, time-programmed and dose-quantified drug release on demand, enabling a precise "trigger-release" therapeutic paradigm. For instance, PMs@NaF-SAP polymeric micelles release 70% tannic acid and 80% sodium fluoride within 24 h at pH 5.0, conferring dual anti-biofilm and remineralizing functions. Their drawbacks include limited drug-loading capacity and complicated fabrication procedures. Additional external triggering devices increase clinical operational complexity. Premature degradation or burst release may occur in the moist, enzymatically active oral environment, shortening the effective therapeutic window [109,110].

In terms of performance: Among antibacterial polymers, quaternary ammonium salt monomers can be copolymerized with resin matrices at low loading while retaining mechanical performance. By contrast, releasing type antibacterial fillers such as silver nanoparticles act as stress concentrating sites at high loadings, reducing flexural strength and wear resistance; additionally, silver nanoparticles may cause tooth discoloration. This has driven researchers to adopt ZnO particles as alternatives with superior optical properties yet inferior antibacterial potency [103,111].

Tissue-repair hydrogels rely on low crosslinking density to maintain high water content and channels for cell migration, which intrinsically limits their mechanical strength and renders them unsuitable for load-bearing sites. This constitutes a fundamental barrier to their application in the mechanically complex oral cavity [106,107]. The incorporation of controlled release polymeric microspheres or capsules may also introduce defects within resin matrices. Although the amorphous calcium phosphate

nanoparticles-dimethylaminohexadecyl methacrylate (NACP-DMAHDM) composite system integrates self healing, antibacterial and remineralizing functions, high microcapsule fractions can impair mechanical performance [109].

In terms of biocompatibility and safety: Safety concerns for antibacterial polymers centre on potential damage to host cell membranes induced by cationic groups and cumulative cytotoxicity upon long-term exposure. In particular, cytotoxicity of high dose QAMs needs to be mitigated via optimization of alkyl chain length and loading level control [103,104]. Silver nanoparticles raise risks of systemic exposure originating from ion leaching as well as ecotoxicological hazards [111].

Safety challenges for tissue-repair polymers are relatively moderate. Natural polymers such as gelatin and hyaluronic acid generally exhibit favourable biocompatibility and low immunogenicity. Nevertheless, animal derived sources carry potential risks of pathogen transmission, and residual crosslinking agents (e.g., glutaraldehyde) may trigger cytotoxicity [106,107]. Safety issues of controlled-release polymers are multifaceted, covering cytotoxicity originating from carrier degradation products, acute toxicity induced by drug burst release, as well as systemic distribution and accumulation of nanocarriers.

6. Biocompatibility and safety evaluation

Although polymeric materials have brought numerous conveniences and advancements to oral care, their prolonged or repeated contact with human tissues has sparked extensive discussions regarding biosafety. A scientific and comprehensive evaluation of the safety of polymeric materials used in oral care is a fundamental prerequisite for ensuring their harmlessness and efficacy in clinical applications [112].

6.1. Evaluation of oral mucosal irritation and sensitization

The evaluation of oral mucosal irritation and sensitization serves as the primary step in assessing material safety. As the first line of defense in direct contact with oral care products, the integrity and health status of the oral mucosa directly influence user comfort and overall systemic health.

In *in vitro* evaluations, besides the conventional cytotoxicity assays using mouse fibroblasts (L929) or human gingival fibroblasts [113], three-dimensional tissue-engineered models such as reconstructed human gingival epithelium (RhGE) and reconstructed human oral epithelium (RhOE) have been increasingly applied in recent years. These models simulate the real oral mucosal environment and can more accurately reflect the impact of materials on epithelial barrier function and the release of inflammatory cytokines [35]. Furthermore, the Hen's Egg Test-Chorioallantoic Membrane (HET-CAM) assay, characterized by its rapid, simple procedure and lack of requirement for complex equipment, is frequently utilized as a preliminary screening tool for material irritation [10].

For *in vivo* evaluations, oral mucosal irritation tests are typically conducted using animal models such as hamsters and rabbits. The assessment involves observing erythema, erosion, and ulceration of the mucosa at the contact site, combined with histopathological examinations to evaluate the degree of inflammatory cell infiltration and the depth of tissue damage [112]. For suspected sensitizers, human patch tests or repeated patch tests are further required to determine their sensitization potential.

For oral polymeric materials classified as medical devices, biocompatibility assessment shall follow the combined framework of ISO 7405:2025 and the ISO 10993 series. As the dedicated standard for dentistry, ISO 7405:2025 complements the general ISO 10993 framework with dentistry specific test protocols and exposure scenario considerations. For consumer grade oral care products, safety evaluation is mainly governed by cosmetic related regulations and pharmacopoeia standards; the testing depth and contact duration classification system differ from those applied to medical devices [114,115].

According to the duration of material contact with oral tissues, three categories are defined: limited contact ($\leq$24 h), prolonged contact ($>$24 h to 30 days), and long term contact ($>$30 days). Assessment for limited contact devices typically requires a 2 to 3 month testing cycle covering three

core endpoints: cytotoxicity, sensitisation, and irritation. For prolonged contact and long term contact devices, the evaluation cycle extends beyond six months. In addition to the three above mentioned endpoints, supplementary tests including genotoxicity, subchronic toxicity and even chronic toxicity are required [116,117]. ISO 10993-5:2009 specifies quantitative criteria for *in vitro* cytotoxicity assessment. Cytotoxicity is considered positive when material extracts or directly exposed samples induce more than 30% reduction in cell viability relative to the negative control [113].

6.2. Safety of degradation products and long-term contact

The degradation products and long-term contact safety are core issues that must be addressed for biodegradable and long term indwelling materials. Degradable polymeric materials, such as PLA and PLGA, generate acidic byproducts like lactic acid and glycolic acid during degradation. Although these metabolites can be further metabolized and cleared by the body under physiological conditions, their accumulation in the localized, enclosed, or poorly vascularized oral environment may lead to a decrease in local pH. This not only compromises the viability of surrounding tissue cells but also poses a risk of demineralization to adjacent dental hard tissues. Therefore, during the material design phase, it is essential to precisely control the degradation rate by adjusting the molecular weight, comonomer ratio, and crystallinity, ensuring it matches the tissue repair cycle.

For synthetic polymers, the issue of residual monomers and oligomers cannot be overlooked. For instance, acrylate monomers in dental composite resins may slowly leach from the material if polymerization is incomplete. These monomers not only possess potential cytotoxicity, but some studies also suggest they may exhibit endocrine-disrupting activity [57]. Similarly, residual acrylic acid monomers in polyacrylic acid polymers may induce mucosal irritation [29]. Consequently, pharmacopoeias and relevant standards worldwide have established stringent limit criteria for the content of extractable and residual monomers in oral care products.

The primary biocompatibility concern of PMMA based dental materials stems from leaching residual methyl methacrylate (MMA) monomers. Rus et al. demonstrated dose dependent toxic effects of varying residual monomer levels on gingival fibroblasts. The PMMA formulation with the highest residual monomer content reduced fibroblast metabolic activity by approximately 33% at 24 h ($p < 0.001$) and 48% reduction of roughly 48% at 48 h ($p < 0.001$). The intermediate residual monomer formulation caused around 20% loss in cell viability at both 24 h and 48 h. The formulation with the lowest residual monomer content yielded approximately 14% viability loss at 24 h and 48 h, respectively [118]. A 3D printed scaffolds fabricated from medium molecular weight chitosan achieved a dental pulp stem cell (DPSC) wound healing rate of 97.1% within 24 h. Cytotoxicity assays confirmed non-cytotoxic behaviour, supporting the survival and proliferation of DPSCs as well as osteoblast differentiated DPSCs [119]. Chitosan based biological coatings also exhibit favourable biocompatibility without cytotoxicity, and deliver superior antibacterial activity compared with uncoated controls [120]. Nevertheless, chitosan nanoparticles present a more complex safety profile. Despite low toxicity towards normal cells, these nanoparticles can be internalised by human dental pulp cells and markedly trigger the formation of micronuclei, nuclear buds and nuclear bridges, indicating potential genotoxic effects. This finding alerts that long term safety should be carefully evaluated before deploying chitosan nanoparticles in root canal therapy and dental pulp tissue engineering [121]. The cell survival rates for the pectin-chitosan composite (in a 1:3 ratio) at 24, 48, and 72 h were 93.42%, 89.88%, and 86.85%, respectively, all well above the 70% cytotoxicity threshold specified in ISO 10993-5, demonstrating the robust biocompatibility of this natural polymer composite system [122].

After conventional polymeric materials enter the oral environment, they gradually degrade into micro- and nanoplastics via mechanical abrasion, thermal stress, hydrolysis or pH fluctuations. High molecular weight polymers (e.g., PEEK and high-density PE) possess tight chain entanglement and high crystallinity, leading to extremely slow degradation, with microplastic generation occurring over timescales measured in years. In contrast, low-molecular-weight or amorphous polymers (such as

low molecular weight PLGA and certain plasticiser-modified PMMA) are more prone to generating micro/nano fragments within a short period (weeks to months). Nevertheless, whether microplastic production during degradation inevitably leads to health hazards remains debated among the scientific community. Spirito and Folliero et al. reported that dental origin micro/nanoplastics are not confined locally within the oral cavity; they possess considerable potential for systemic dissemination and toxicity. Nanoplastics can cross epithelial and endothelial barriers to enter blood circulation and have been detected in distant organs including the liver and heart, raising concerns regarding long term accumulation and systemic toxic consequences [123].

From an occupational exposure perspective, dental practitioners may chronically inhale polymer derived micro/nanoparticles generated during grinding of composite resin restorations. Although acute cytotoxicity may be low, repeated inhalation of nanoparticles capable of deep alveolar penetration may increase risks of chronic bronchial irritation, persistent oxidative damage and inflammatory pulmonary injury. Comparable concerns exist in orthodontics: long term polymeric appliances such as clear aligners and retainers continuously release microplastics under oral mechanical forces and temperature cycling, and their long term health consequences remain unclear [123–125].

Although multiple *in vitro* studies demonstrate that microplastics can act as attachment substrates for bacterial biofilms, facilitating colonisation of pathogenic bacteria and triggering oral dysbiosis, other studies indicate that the type of biofilm formed on microplastic surfaces is strongly dependent on material composition, surface roughness and exposure concentration. For example, long-term exposure to polyethylene (PE) microplastics at low concentrations (<0.1 mg/L) exerts no notable influence on the community structure of resident oral microbiota (e.g., *Streptococcus salivarius*). A marked increase in the proportion of pathogenic bacteria (such as *Porphyromonas gingivalis*) is only observed once the concentration exceeds a certain threshold (>10 mg/L) [124,125]. This suggests that microplastic risks may exhibit a clear dose-threshold effect. In addition, some researchers argue that the actual retention time of microplastics in the oral cavity is constrained by salivary flow and mechanical flushing from mastication, far shorter than exposure durations in sites such as the gastrointestinal tract. Accordingly, their capacity to serve as long-term carriers for oral bacteria requires re-evaluation.

It's worth noting that combining dental nanomaterials with polymeric materials can significantly boost mechanical strength and antibacterial properties, but their high surface area and surface reactivity also bring unique safety challenges. *In vitro* studies indicate prolonged exposure can induce oxidative stress, cytotoxicity and DNA damage in oral epithelial cells and gingival fibroblasts. Animal model data suggest both metal based and polymer based nanomaterials may cross mucosal barriers into systemic circulation. Although most commercially available dental nanomaterials satisfy current biocompatibility standards and exhibit low acute toxicity, their long term effects require continuous attention and thorough comparative investigation [126].

Based on the above findings, risk assessment of microplastics and nanomaterials is related to the following factors: test models (*in vitro* monolayer cell cultures versus *in vivo* multi-barrier systems), exposure conditions (single acute exposure versus repeated long-term exposure, high concentrations versus environmentally relevant concentrations), and material characterisation (particle size, molecular weight, surface charge, and ageing status). Most available evidence originates from short-term high-dose *in vitro* experiments, which limits the validity of extrapolating these findings to real-world scenarios of long-term oral exposure. Future studies should adopt standardised, reproducible animal models for low-dose long-term exposure, combined with multi-omics techniques (genomics, metabolomics, and microbiomics), to construct quantitative predictive models correlating polymer molecular weight, concentration and long-term toxicity.

6.3. Interactions between degradation products and the oral microbiome

Conventional biocompatibility assessments focus solely on cytotoxicity endpoints of individual components and fail to capture the complex influences of degradation products on microbial community

dynamics and ecological homeostasis. Resin based restorative materials, which are based on methacrylate monomers undergo biodegradation in the oral environment via hydrolysis and catalysis by bacterial esterases (notably those produced by *Streptococcus mutans* and *Enterococcus faecalis*). This process releases degradation products such as bishydroxy-propoxy-phenyl propane (Bis-HPPP), methacrylic acid, and triethylene glycol (TEG). These degradation products exert pronounced strain dependent effects on oral bacteria. At physiologically relevant *in vivo* achievable concentrations, TEG promotes the proliferation of *Streptococcus mutans* and up regulates the expression of glucosyltransferase B (gtfB, a key virulence factor involved in biofilm formation) together with the transcriptional regulator yfiV, thereby enhancing bacterial biofilm forming capacity and virulence expression. In contrast, Bis-HPPP and methacrylic acid have been documented to exert mild inhibitory effects on the growth of *Streptococcus mutans*. These findings further indicate that degradation products do not simply "promote" or "inhibit" bacteria; instead, they may remodel the community structure of oral biofilms by differentially modulating the growth and virulence of distinct bacterial species, consequently elevating the risk of secondary caries [127–129].

From the perspective of interactions between polymeric degradation products and oral microorganisms, degradation products modulate bacterial growth and virulence on one hand. On the other hand, esterases produced by species such as *Streptococcus mutans* and *Enterococcus faecalis* in turn accelerate the degradation of resin matrices, weaken the resin dentin interface, and facilitate bacterial infiltration into the interface and root canal system. This gives rise to a positive feedback loop of "degeneration-bacterial enrichment-further degradation" [130]. Gregson et al. confirmed that three oral bacterial strains can substantially alter the surface and mechanical properties of dental resin materials [131]. Similarly, we shouldn't overlook that dental composites, resins, and oral care products can release micro- and nanoplastic particles during daily use and wear; these particles can serve as surface substrates for bacterial colonization and biofilm formation in the oral cavity and may cause micro-injuries to the mucosa due to mechanical abrasion [132]. Analogous evidence from gut microbiome studies and *in vivo* animal studies suggests that long term exposure to microplastics can lead to a reduction in beneficial commensal bacteria and an expansion of pro-inflammatory bacterial populations, resulting in oral dysbiosis. This is associated with oxidative stress, DNA damage responses, and a persistent inflammatory environment in gingival tissue, which may accelerate the progression of periodontal disease [123,133,134].

Systematic evaluation of the effects of degradation products on the community structure and functions of the whole oral microbiome (e.g., microbiota diversity, metabolome, and gene expression profiles), the "cumulative effects" of degradation products, their actual oral exposure levels and residence time, as well as the dose response relationships governing host microbe interactions remain key directions for future investigations.

6.4. Potential risks and management measures

The application of nanomaterials in oral care has also introduced novel safety challenges and regulatory dilemmas. Nanoparticles such as silver nanoparticles, titanium dioxide nanoparticles, and nano-hydroxyapatite exhibit exceptional functional properties due to their size effects. However, their small size also facilitates the penetration of biological barriers, allowing them to enter the bloodstream or lymphatic system and accumulate in distal organs such as the liver and kidneys [135]. Further systematic toxicological research and evidence-based medical data are required to elucidate their long-term chronic toxicity, potential genotoxicity, and impacts on the oral microecosystem. Additionally, certain polymeric materials may release microplastic particles during prolonged mastication, friction, and aging processes. The oral exposure dose, cumulative effects, and potential impacts on systemic health of these particles have become major research focal points in the global environmental and health fields.

Regarding standards and regulations, oral care products are generally categorized and managed as either cosmetics or medical devices based on their intended use and risk levels. Daily hygiene

products, such as toothpaste and mouthwash, primarily comply with safety and technical specifications for cosmetics. Conversely, dental restorative materials, orthodontic materials, and implantable materials are regulated as medical devices, necessitating more rigorous biocompatibility evaluations, preclinical validations, and clinical trials [114,116].

Regulatory classification and approval pathways: Regulatory approval pathways for the application of polymeric materials in oral care vary across different regions [136,137]. In China, the National Medical Products Administration (NMPA) implements a filing regime for conventional polymers such as carbomer, CMC and chitosan when used as cosmetic raw materials. However, products with new functional claims (e.g., antibacterial and desensitising effects) or novel polymeric structures (e.g., modified chitosan nanoparticles and dendrimers) require submission of safety assessment data and registration applications. Furthermore, there are clear concentration limits for polymeric thickeners used in toothpaste. For example, the practical concentration of carbomer as a general thickener in formulations is typically no more than 1.0%. If carbomer is filed as a functional ingredient (e.g., film-forming agent or carrier for active ingredients), toxicological data based on molecular weight distribution must be provided. For nano-sized polymers (particle size $<$ 100 nm), additional data on percutaneous absorption, long-term exposure and genotoxicity are required. The U.S. Food and Drug Administration (FDA) adopts a dual regulatory model for oral care products. For most products such as toothpastes and mouthwashes containing active ingredients specified in the Over-the-Counter (OTC) Monograph (e.g., sodium fluoride), polymeric excipients in formulations (e.g., carbomer and CMC) can be marketed directly via the OTC monograph pathway without pre-market approval. Nevertheless, once a polymer itself is identified as a new active ingredient or a new excipient (e.g., novel nano-antimicrobial polymers), a new drug application (NDA) or abbreviated new drug application (ANDA) is mandatory, with full chemistry, manufacturing, and controls (CMC) documentation including concentration-effect relationships, molecular weight distribution and biodegradability. The European Union has specific provisions for nano-polymeric materials. All products containing nanomaterials must be explicitly labelled in the ingredient list (e.g., “Chitosan (nano)”), and a safety assessment report from the Scientific Committee on Consumer Safety (SCCS) must be submitted to the European Commission. The report shall contain complete characterisation data of nanoparticles (including particle size distribution, specific surface area, surface charge, etc.) and concentration-based exposure assessment. Under the medical device pathway, the Medical Device Regulation (MDR) mandates that polymeric materials for Class II and higher oral medical devices meet biocompatibility requirements specified in the ISO 10993 series of standards. For biodegradable polymers (e.g., PLGA bone repair materials), long-term data on degradation timescale, degradation products and local tissue responses are required.

Hierarchical application of the ISO standard system: The ISO 10993 series provides a general biological evaluation framework for all medical devices. The parts most relevant to dental materials include Part 5 (*in vitro* cytotoxicity), Part 6 (local effects after implantation) [138], Part 10 (irritation and skin sensitization) [139], Part 11 (systemic toxicity) [140], Part 17 (toxicological risk assessment) [141], and Part 18 (chemical characterization) [142]. ISO 7405 is a supplementary standard specifically developed for dentistry, intended to be used alongside the ISO 10993 series. Its core value lies in acknowledging dentistry specific requirements, including pronounced influences on material biological behaviour exerted by unique oral environmental conditions such as persistent moisture, temperature fluctuations, masticatory stress, and dynamic pH shifts; it also provides dentistry specific test methods and evaluation criteria.

Differences in preclinical evaluation: Biocompatibility assessment of dental medical materials shall follow the risk management principles specified in ISO 14971 [143]. Appropriate test batteries are selected according to the nature of device human contact (surface contact, external communicating, implanted) and contact duration (transient, short term, long term). For Class II oral devices (e.g., clear aligners worn for up to 22 h per day), a comprehensive biocompatibility data package must be submitted, covering chemical characterization (ISO 10993-18), justification for test waivers based

on chemical characterization and toxicological risk assessment (ISO 10993-17), *in vitro* cytotoxicity (ISO 10993-5/ISO 7405), sensitization (ISO 10993-10), and systemic toxicity and implantation tests where necessary. The margin of safety framework in Annex E of ISO 7405:2025 specifically mandates that for dental devices containing nanomaterials, manufacturers shall compare estimated nanoparticle exposure levels against identified toxicological thresholds and demonstrate, via quantitative margin of safety calculations, that exposure is substantially below adverse effect thresholds. By comparison, safety evaluation for daily oral care products generally does not require the full test matrix of ISO 10993/ISO 7405; assessments are performed under cosmetic or OTC drug related regulations, with markedly lower levels and systematic rigour of evidence compared with medical device standards.

7. Research frontiers and development trends

Research on polymeric materials for oral care is advancing toward greener, smarter, more precise, and more digital approaches. The deep integration of materials science with life sciences, nanotechnology, and information technology continually gives rise to new research hotspots and technological breakthroughs.

7.1. Eco-friendly oral care materials

The development of environmentally friendly polymeric materials has become a global consensus and an inevitable trend in industrial advancement. Traditional oral care products, especially toothpaste containing non-degradable microplastic beads and synthetic fiber dental floss, have imposed a heavy burden on the ecological environment. In recent years, bio-based polymeric materials prepared from renewable resources have attracted widespread attention. For example, cellulose nanofibers extracted from plants and subjected to nanoscale treatment can serve not only as mild abrasives and thickeners in toothpaste but also possess complete biodegradability [144]. Biodegradable polymers such as PLA and PHA have been progressively applied in toothbrush handles, dental floss base materials, and product packaging [145,146]. However, the practical adoption of these materials still faces challenges including insufficient mechanical properties, relatively high production costs, and the difficulty of controlling degradation conditions within the oral environment. Future research needs to enhance their overall performance through approaches such as blend modification, nano-reinforcement, and surface functionalization while preserving their eco-friendly characteristics, in order to meet the application requirements of oral care products.

7.2. Smart responsive oral care materials

Stimuli-responsive polymeric materials represent the future direction of oral care material design. These materials are capable of sensing specific stimuli in the oral environment, such as pH, temperature, enzyme activity, or mechanical force and responding with corresponding structural and functional changes [147]. In the field of topical oral drug delivery, pH-responsive polymeric carriers can precisely release antimicrobial agents within the carious lesion microenvironment where plaque acid production causes a pH decrease, while remaining stable at the neutral pH of healthy tissues, thereby enabling targeted therapy for diseased sites [148]. Temperature-responsive materials such as poloxamer 407 can exist in a liquid state at room temperature for easy instillation and rapidly transform into a gel state at oral temperature, thus prolonging drug retention time in periodontal pockets or on mucosal surfaces [149]. Moreover, the concept of mechano-responsive materials has also been proposed. By incorporating mechanically sensitive groups into the material, the release of flavors or drugs can be triggered under masticatory pressure. Although currently mostly at the laboratory research stage, this concept offers highly imaginative and innovative ideas for the development of novel oral care products [150].

The most prominent challenge for oral mucosal drug delivery stems from continuous salivary secretion and clearance. The resting salivary flow rate is approximately 0.3–0.5 mL/min, and can surge to 5–7 mL/min under stimulated conditions. Such a dynamic environment causes conventional pH responsive drug carriers and dressings to be rapidly diluted and eliminated before they exert therapeutic

effects, often resulting in an effective residence time ranging from only several minutes to a few hours [151]. As highlighted in the review by Conte et al., hydrogels based on natural polymers such as gelatin and hyaluronic acid can be specifically degraded by bacterial proteases (e.g., gingipains, streptococcal proteases), enabling selective release of antibacterial and pro-healing compounds at infection sites [152]. Saliva in a healthy oral cavity is generally maintained within a neutral pH range of 6.5–7.5 and possesses strong buffering capacity. Local pH fluctuations mainly originate from dietary intake (acidic beverages can rapidly lower plaque pH below 4.5), acid production via bacterial fermentation (plaque pH can drop below 5.5 at caries active sites), and inflammatory responses [153]. Saliva is abundant in enzymes such as α-amylase, lysozyme and proteases, together with inorganic salts of high ionic strength. These constituents can substantially alter the ionization state and structural integrity of pH responsive polymers, and may even accelerate premature polymer degradation and premature drug release [153,154]. Meanwhile, mucin and other macromolecules present in saliva may trigger non specific adsorption and aggregation of polymers, giving rise to biofouling that interferes with the functional performance of responsive moieties. Notably, for dental restorative materials and hydrogel dressings, the realization of pH responsiveness or self healing functions usually relies on the introduction of dynamically reversible chemical bonds, including imine bonds, disulfide bonds, metal ligand coordination bonds and hydrogen bonds. Nevertheless, these dynamic bonds are frequently achieved at the cost of cross linking density and mechanical strength [155]. Furthermore, the successful performance of pH responsive polymers in the gastrointestinal tract relies on pronounced and relatively stable pH gradients (gastric pH $\approx$ 1.2 versus intestinal pH $\approx$ 7.4). In contrast, oral pH fluctuates rapidly within a much narrower range, so many responsive systems designed for gastrointestinal pH gradients cannot be reliably triggered in the oral cavity [153]. Accordingly, pH responsive polymeric materials intended for oral applications must exhibit robust mucoadhesive properties or biofilm penetrating capability to counteract salivary clearance and prolong residence time; otherwise, their responsive release functions cannot be realized.

In additon, Chen et al. summarized the latest advances of stimuli-responsive antibacterial nanocarriers for oral biofilm control and microbiome restoration. They pointed out that pH-sensitive polymeric micelles and nanoparticles can utilize local acidification of cariogenic dental plaques (pH $<$ 5.5) as a triggering signal to achieve precise release of antibacterial and remineralizing agents. Such "biofilm-responsive" nanoformulations possess unique advantages because they can sense dynamic microenvironmental changes within polymicrobial biofilms [110]. Liu et al. also outlined cutting-edge progress of smart stimuli responsive materials in oral wound healing, including the nanozyme system of chitosan-iron-nitrogen co-doped carbon dots (CS@Fe-N CDs), which converts H_2O_2 into ·OH to realize dual antibacterial and pro-healing functions. Other innovative designs include semi-interpenetrating network hydrogels (poly(NIPAAm)-chitosan) loaded with papain to achieve controlled release and synergistic antibacterial-regenerative effects [155].

7.3. Nano-engineered oral care materials

Advances in nanotechnology and composite materials have opened up new possibilities for the leap in performance of oral care materials. Nanofiller-reinforced dental composite resins not only exhibit higher mechanical strength and wear resistance, but their nanoscale dimensions also impart superior aesthetic properties and polishability to the material [156]. Nanofibrous membranes fabricated by electrospinning technology, owing to their high specific surface area and biomimetic three-dimensional network structure, demonstrate favorable cytocompatibility and tissue integration capability in oral wound dressings and guided tissue regeneration membranes [157]. By uniformly dispersing bioactive nanoparticles, such as bioactive glass, nanosilver, or hydroxyapatite, within a polymeric matrix, multifunctional composite materials can be constructed that combine antibacterial, osteogenesis-promoting, and mineralization-inducing functions [158]. It is noteworthy that the standardization of large-scale preparation processes, the accumulation of long-term biosafety data, and the compatibility

with existing clinical workflows remain key bottlenecks for nanomaterials transitioning from the laboratory to clinical application.

Nanoparticles are capable of crossing biological barriers, accumulating in oral tissues as well as systemic compartments, and may trigger cytotoxicity, genotoxicity and immunotoxicity via mechanisms including oxidative stress, mitochondrial damage, apoptosis, necrosis and autophagy [159,160]. Compared with conventional polymeric materials, toxicological evaluation of nanomaterials requires additional attention to decisive parameters governing their biological behaviors, such as particle size, shape, specific surface area, surface charge, aggregation state and surface chemistry. The FDA mandates that any material exhibiting nanoscale dependent properties must submit non clinical safety studies compliant with Good Laboratory Practice (GLP), covering endpoints including biodistribution, *in vivo* persistence, oxidative stress mediated cytotoxicity, inflammatory responses and immune activation [161]. Similarly, the EMA requires detailed documentation of manufacturing controls, scale up processes and reproducibility evidence for nanomaterials within the Common Technical Document (CTD) [162].

Nano oral materials serve long term in the moist, dynamic, enzyme rich oral environment subjected to continuous salivary clearance. Their release, degradation and exposure profiles differ markedly from those in other *in vivo* sites, which necessitates regulatory assessment based on *in vitro* and *in vivo* models that recapitulate physiologically relevant oral conditions [163,164]. Nevertheless, current standardized testing systems are mostly designed for general medical scenarios and lack dedicated evaluation specifications tailored to the oral microenvironment (e.g., salivary flow, pH fluctuations, multispecies biofilms and mechanical forces). This places regulatory agencies in a position with insufficient evidence when assessing the long term safety and ecotoxicological impacts of nano based oral products.

From the standardization perspective, characterization and testing methods for nanomaterials in oral applications remain fragmented, and universally accepted standardized protocols are lacking, particularly for particle size distribution, surface chemistry, aggregation behavior, release kinetics and analysis of degradation products [163]. The ISO 10993 series provides a fundamental framework for biocompatibility assessment of medical devices containing nanomaterials. The newly revised ISO 7405 also incorporates preclinical assessment protocols for nanomaterials (including particle size analysis and surface characterization) [165]. Even so, these standards are still continuously evolving and lack sufficient specificity for oral use specific scenarios. In summary, the current challenges faced by nanomaterials include manufacturing processes, accumulation of hazardous substances, and standardized performance evaluation methodologies.

7.4. Digital and personalized oral care materials

The development of digital and personalized oral care materials is profoundly transforming the paradigm of oral healthcare [166]. 3D printing technology, particularly stereolithography (SLA)/digital light processing (DLP) and material extrusion (MEX)/fused deposition modeling (FDM) techniques, has been widely applied in the fabrication of dental models, surgical guides, temporary restorations, and orthodontic appliances. Compared with traditional processes, 3D printing enables the rapid prototyping of complex geometric structures and supports personalized customization based on patients' oral anatomical data [167]. For example, based on a patient's cone-beam computed tomography (CBCT) data and intraoral scanning data, personalized guided bone regeneration scaffolds that perfectly match the individual bone defect morphology can be designed and fabricated [168]. In the domain of oral care products, digital technology has also given rise to the concept of personalized precise drug delivery systems, namely, customizing mouthwash or oral gel formulations with specific antibacterial spectra based on individual oral microbiota testing [169]. This shift from "one-size-fits-all" to "tailored to the individual" not only holds promise for enhancing the precision of oral disease prevention and treatment but also poses entirely new challenges to the formulation design, manufacturing processes, and regulatory systems of oral care materials.

The high upfront capital outlay for 3D printers, combined with licensing fees for dedicated software, ongoing material consumption, and subsequent maintenance expenses, substantially raises the adoption barrier for private dental clinics. A scoping review explicitly identified the high initial purchase cost of printers, along with associated maintenance and material expenses, as major obstacles limiting the adoption of this technology in private practices. Furthermore, the quality of printed products varies considerably depending on the materials utilized. The real world costs of 3D printed personalized products extend well beyond raw material consumption, and include labour intensive post processing steps (washing, support removal, photocuring/heat treatment, surface finishing, and quality inspection), as well as extra expenditures stemming from print failures, model rework, and quality control procedures. High precision diagnostic models or patient specific devices frequently require expensive high performance specialty resins, which further drive up overall costs [170–177].

Most 3D printed dental products are classified as custom made medical devices and must satisfy stringent requirements for biocompatibility, sterilization, and CE/ISO certification. Manufacturers and point of care printing facilities are additionally obliged to maintain GMP compliant quality control systems and comprehensive documentation validating post processing workflows [178–180]. These regulatory compliance requirements markedly increase administrative overhead and labour input, raising the amortized compliance cost per unit product. Meanwhile, clinical uptake of personalized 3D printed oral products faces structural constraints within payment systems. Established dental procedure coding systems (e.g., the US CDT codes) do not explicitly cover many additively manufactured products. Consequently, insurers lack corresponding reimbursement codes and frequently categorize such products as “experimental” treatments and deny coverage. Reimbursement may only become available once sufficient evidence regarding cost effectiveness and clinical outcomes is submitted [181]. This implies that even if 3D printing can reduce costs from a technical perspective, patients are still required to bear full out of pocket expenses in the absence of third party payer support, which greatly limits its accessibility.

7.5. Advantages and industrialization constraints of polymeric materials in oral care applications

Polymeric materials have gone through a long period of research and development for oral care, and their clinical application paths are still being explored and improved (Figure 4). Currently, the limitations of new polymeric materials in terms of industrialization are primarily reflected in the following areas: The excellent performance of new polymeric materials demonstrated under laboratory conditions cannot be automatically translated into reliable performance for industrialized products; there exists a systematic gap between the two. Ma et al. pointed out that the clinical translation of oral materials faces multiple challenges, and market demand serves as a key driver for material and technological innovation. The tension between these two factors constitutes the core proposition of translational research [182]. Patel et al. also systematically noted that numerous biomaterials are well validated at the laboratory stage yet remain stalled at the preclinical phase or ultimately fail clinical translation due to limitations in reproduction and scale up manufacturing. Their review focused on recurrent key obstacles reported in the literature, including difficulties in maintaining batch to batch consistency of material properties, economically infeasible scale up of manufacturing processes, and insufficient clinical evidence to persuade clinicians to replace existing well established materials [183].

Furthermore, laboratory scale synthetic approaches (e.g., nanoprecipitation, emulsion method, high pressure homogenization) commonly encounter loss of control of critical quality attributes upon equipment scale enlargement, manifested as uneven heat transfer, reduced mixing efficiency, residual solvents, broadened particle size distribution, nanoparticle aggregation, abnormally elevated dispersion viscosity, and drifts in drug loading capacity and release kinetics [184,185]. Paliwal et al. systematically compared the scale up feasibility and inherent limitations of various nanoformulation manufacturing processes. They indicated that preparation methods adopting organic solvents generally suffer from cumbersome removal of solvent residues and deteriorated product homogeneity after scale up. Aligning target product

profiles (TPP) and critical quality attributes (CQAs) under the quality by design (QbD) philosophy represents the core pathway to achieve scalable and reproducible production [184]. Recent studies have further quantitatively compared the scale up characteristics of five mainstream nanoformulation platforms: liposomes, polymeric micelles, nanoemulsions, nanocrystals, and albumin/biopolymer based systems. It is explicitly demonstrated that controlled micromixing and solvent displacement are essential for obtaining stable particle sizes ranging from 50–200 nm for polymeric nanoparticles, whereas standardization, quantitative comparability, and data driven control are pivotal for scalable and regulable manufacturing [185]. In addition, before new dental polymers can enter the market, they must meet both safety standards for materials and performance requirements for medical devices, and must be supported by accurate research findings, including comprehensive preclinical toxicology data, animal studies, and multi-year clinical trials [186].

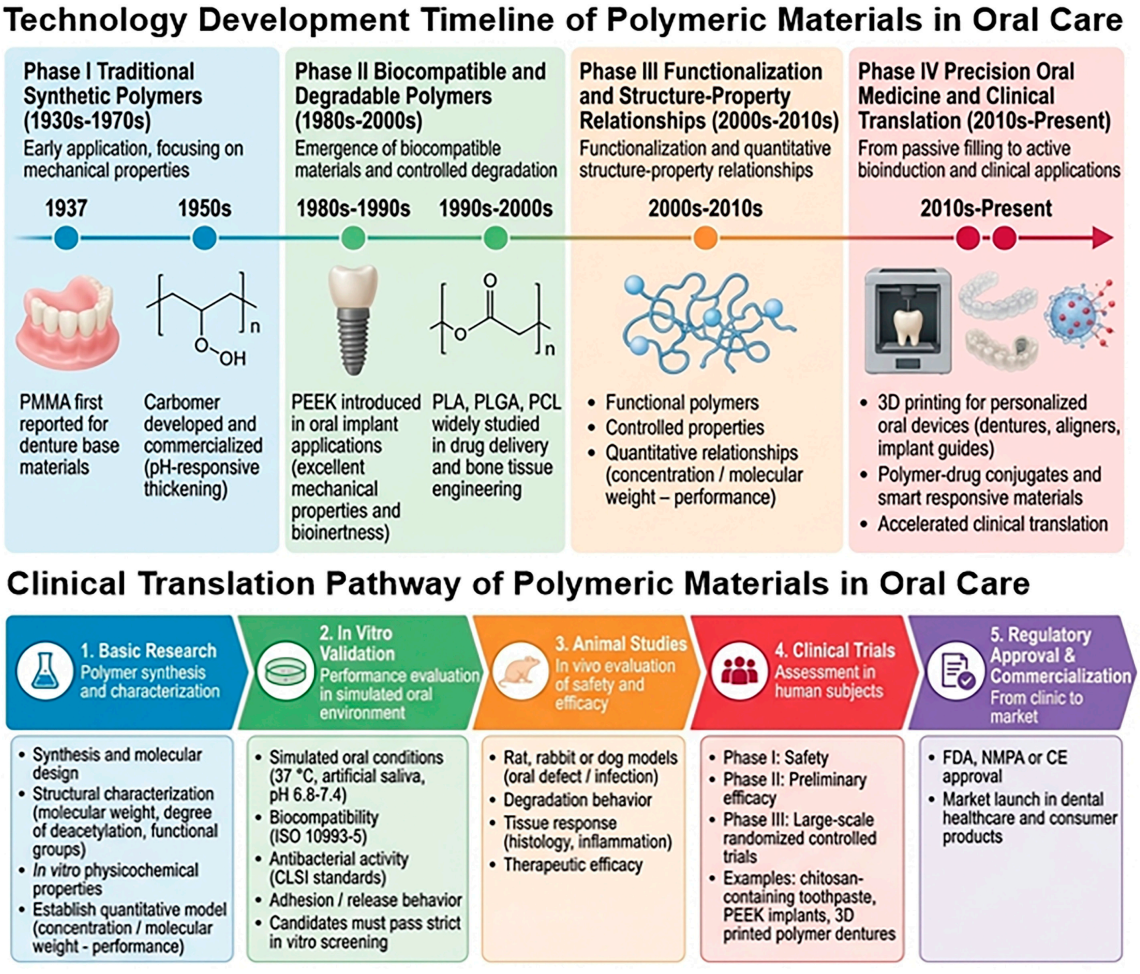

Figure 4. The development history and clinical translation pathways of polymeric materials for oral care

The application of different types of polymers in oral care may face varying degrees of translation barriers. For biodegradable materials, the oral environment is far more complex than *in vitro* PBS systems. Saliva contains hydrolytic enzymes including esterases and proteases, which markedly accelerate polyester bond cleavage, resulting in actual degradation rates 30%–50% faster than predictions obtained in 37°C PBS. Temperature fluctuations during food intake (from 4°C cold drinks to 60°C

hot meals) subject materials to repeated thermal expansion and contraction, accelerating the initiation of microcracks. Frequent pH fluctuations ranging from 5.5 (acidic foods) to 8.0 (alkaline foods) alter the protonation state and hydrophilic–hydrophobic balance of polymer chains [17,41]. Regarding antimicrobial polymers, taking chitosan as an example, the exposure level required to achieve effective antibacterial activity in local tissue or cell culture environments approaches or exceeds the threshold for cytotoxicity. When the chitosan concentration increases from 0.5 mg/mL to 2.0 mg/mL, the antibacterial efficacy (measured by the reduction in bacterial survival rate) rises by only approximately 20%, whereas the viability of human gingival fibroblasts drops sharply from >90% to <60%. From the perspective of molecular weight, HMW chitosan (>200 kDa) possesses long molecular chains and high positive charge density. Although it achieves potent antibacterial effects at low concentrations via a "molecular fence" mechanism, its strong adhesive property impedes clearance from oral mucosal surfaces, prolonging local residence time and increasing mucosal irritation under chronic exposure. LMW chitosan (<50 kDa) readily penetrates bacterial cell walls to exert intracellular antibacterial effects, yet it also more easily crosses host cell membranes and induces stronger intracellular oxidative stress. Nevertheless, when chitosan is fabricated into nanoparticles, antibacterial effects can be achieved at lower concentrations, accompanied by substantially elevated genotoxicity risks including micronucleus formation and DNA damage [40,42,43,121,187]. For 3D printed polymers, 3D printed oral devices feature a "one design per patient" characteristic. Although regulatory pathways for such custom medical devices have been explored by China NMPA, the US FDA and EU MDR, a unified and mature regulatory framework has not yet been fully established. Polymers commonly used in 3D printing may generate low-molecular-weight oligomers, residual monomers, or photoinitiator degradation products during high-temperature or photopolymerisation processes. Toxicological assessment of these substances must be performed separately, and the general safety data sheets from raw material suppliers cannot be simply adopted. In terms of cost, despite the technical advantages of 3D printing for small-batch, customised devices with complex architectures, the per-unit compliance cost (including biocompatibility validation of materials, process validation for printing, software qualification, sterilisation validation, and post-market clinical follow-up) may render the final product far more expensive than conventional alternatives, which represents another critical factor hindering translational application [118,188].

To further explore the differences between polymer materials and traditional materials in terms of health economics, an analysis was conducted from three dimensions: direct medical costs, indirect social costs, and the value of long-term health outcomes [45,124,189,190]. Taking oral maxillofacial bone defect repair as an example, traditional permanent implants such as titanium alloys have manageable material costs, yet secondary surgery for removal or management of complications may be required. PLGA/chitosan composite scaffolds serve as absorbable alternatives; while their material cost may only account for 1/5–1/3 of that of titanium alloys, the comprehensive cost must consider preoperative image-based design, intraoperative scaffold fixation, postoperative degradation monitoring, and expenses for managing incomplete degradation or inflammatory reactions. Conventional toothpastes containing chlorhexidine or triclosan are low-cost, yet they are limited by tooth discolouration, taste alteration and bacterial resistance. Chitosan-based antimicrobial polymeric toothpaste has raw material costs approximately 2–3 times higher than the former, but its physical antibacterial mechanism (membrane disruption) reduces resistance risks. It also functions as a thickener and film-forming agent, decreasing the use of other excipients in formulations. Conventional PMMA compression-moulded dentures have low per-unit manufacturing costs but require multiple manual steps including impression taking, model pouring, heat treatment and polishing, leading to long fabrication cycles (typically 1–2 weeks) and multiple patient visits. 3D printed polymeric dentures can shorten the production cycle to 24–48 h and reduce clinic visits, offering notable benefits in terms of patients' time expenditure and lost social productivity.

8. Conclusions and future perspectives

Polymeric materials, as the core material foundation of modern oral care, have evolved from an initially auxiliary physical role into a multifunctional platform integrating cleaning, treatment, prevention, and repair. This paper systematically reviews the fundamental properties of natural polymers, synthetic polymers, and biodegradable polymers in oral care, elaborates in detail on their extensive applications in oral cleaning products, dental restorations, orthodontics, mucosal care, and surgical assistance, and delves into functional design strategies such as antibacterial activity, repair promotion, desensitization, and controlled drug release. Meanwhile, the continuous improvement of the biocompatibility and safety evaluation systems for materials also provides important assurance for the clinical translation of new materials.

Nevertheless, the research and application of polymeric materials for oral care still face numerous challenges. First, the mechanisms by which the complex dynamic oral environment affects material stability, functional durability, and interfacial behavior have not yet been fully elucidated, and the aging and failure patterns of materials under the combined action of saliva, enzymes, and microorganisms require further in-depth investigation. Second, some functionalized materials, such as stimuli-responsive carriers and nanocomposites, remain largely at the laboratory research stage, lacking large-scale clinical validation and long-term safety data, and are still some distance from genuine clinical translation. Moreover, existing evaluation systems remain insufficient in assessing the environmental health risks of novel materials, particularly biodegradable materials and nanomaterials. Looking forward, the development of polymeric materials for oral care will exhibit several major trends: first, the deepening of the green and sustainable concept, which will promote the application of bio-based and biodegradable materials throughout the entire life cycle of oral care; second, the intelligent design of materials, utilizing stimuli-responsive polymers to achieve precise disease prevention and treatment; and third, the digitalization of manufacturing technology, leveraging technologies such as 3D printing and artificial intelligence to achieve personalized customization and precise adaptation of oral care products. It can be anticipated that with the deep interdisciplinary integration of materials science, dentistry, microbiology, and information technology, novel polymeric materials for oral care will play an even more important and irreplaceable role in improving human oral health.

(1) The most urgent priority in the field of polymeric materials for oral care is to systematically shift the research focus from *in vitro* performance validation toward the evaluation of clinical efficacy and safety. The main areas of research include the following: conducting systematic controlled trials with multi indicator endpoints; refining the research and translational model system progressing sequentially from *in vitro* assays through animal studies to human investigations, in particular adopting animal models and multispecies biofilm models that better recapitulate the real world oral environment; performing long term follow up studies focusing on the degradation behaviour of polymeric materials after prolonged oral service, microplastic release, and their chronic impacts on mucosal barriers, immune responses and microbiome homeostasis, rather than relying merely on acute toxicity assessments.

(2) The lack of standardization constitutes a core bottleneck hindering the transition nanotechnology enabled multi-stimuli responsive polymeric oral materials from laboratory investigations toward standardized manufacturing and regulatory approval. The main areas of research include the following: establishing standardized analytical and characterization methods to guarantee data comparability; developing standardized *in vitro* evaluation models that mimic physiologically relevant oral conditions (temperature cycling, salivary flow, pH fluctuations, multispecies biofilms, and mechanical wear); formulating a unified classification framework and standardized safety assessment system tailored for oral nano/stimuli responsive polymeric materials.

(3) Develop intelligent polymeric material systems capable of specifically recognizing and intervening in pathogenic biofilms within the complex oral microbial ecosystem, while maximally preserving the homeostasis of commensal flora and microbiome diversity. The main areas of research include

the following: constructing multi-stimuli responsive smart release systems that exploit specific microenvironmental signals of pathogenic biofilms (local acidification, over expression of specific proteases, oxidative stress, etc.) as triggering cues, so as to precisely target pathogenic bacteria without off target damage to commensal microorganisms; conducting longitudinal multi omics studies (metagenomics, metabolomics, transcriptomics) to systematically assess the long term influences of polymeric materials (including degradation products and micro/nano plastics) on oral microbiome diversity, community structure and functional metabolic networks, and to unravel underlying mechanisms; advancing selective antibacterial strategies and fundamentally maintain the balance of the oral microbial ecosystem.

Oral health is essentially a reflection of microbial ecological homeostasis. Any material design that disregards microbiome impacts will eventually encounter shortcomings in clinical efficacy and safety. The development of microbiome friendly intelligent materials represents a fundamental paradigm shift in the application mode of polymeric materials for oral care.

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