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From Occlusal Stability to Load Distribution: The Orthodontic-Restorative Continuum in Implant-Supported Fixed Partial Denture Design: Narrative Review

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Abstract

Background: Achieving and maintaining occlusal stability after orthodontic treatment is not merely an aesthetic objective; it constitutes the biomechanical foundation upon which implant-supported Fixed Partial Dentures (FPDs) are planned, fabricated and expected to function over time.

Objective: This narrative review integrates two bodies of literature that are rarely examined together: the biology of post-orthodontic occlusal adaptation and the biomechanics of implant-supported prosthetic design.

Discussion: We examine how settling, relapse and retention failures can compromise the occlusal scheme that governs load distribution in implant-supported FPDs and how prosthetic variables including cantilever length, connector dimensions, pontic design and framework material interact to determine peri-implant stress distribution and long-term complication rates.

Conclusion: A clinical decision-making framework grounded in the available biomechanical evidence is proposed, key evidence gaps are identified and future research directions are outlined. This review is intended for orthodontists, prosthodontists and general dentists engaged in interdisciplinary treatment planning.

Keywords: Orthodontic Relapse; Occlusal Settling; Periodontal Ligament; Supracrestal Fibrotomy; Implant-Supported Fixed Partial Dentures; Prosthetic Design; Biomechanical Stress; Cantilever Geometry; Finite Element Analysis

Introduction

Post-orthodontic treatment stability is one of the most consistently documented concerns in modern clinical practice. The literature reports a high prevalence of relapse following the active phase of orthodontic treatment and this carries a significant negative impact, not only on the patient's aesthetic satisfaction but on the clinical credibility of the treatment itself in the long term [1]. The challenge lies in the fact that the dental position achieved at the end of active treatment is not inherently permanent, which underscores the need for a thorough understanding of the factors that govern the durability of occlusal results. For the purposes of this review, certain terms require precise definition. Relapse refers to the regression of teeth toward their original malocclusion, driven by the elastic memory of periodontal fibers and muscular pressures [2]. Settling, by contrast, describes the physiological process of post-debonding adaptation, in which teeth seek maximum intercuspation through functional micro-movements. Long-term occlusal stability represents the clinical goal: a harmonic interarch balance that resists pathological change over time.

A critical distinction that this work seeks to highlight is the difference between true relapse and the normal adaptive changes associated with aging and residual growth, a confusion that frequently appears in the literature and leads to erroneous diagnoses of therapeutic failure [2]. To quantify these phenomena, clinicians rely on standardized occlusal assessment tools such as the PAR index (Peer Assessment Rating), the ABO-OGS (Objective Grading System), digital occlusal analysis using T-scan and the evaluation of articulation in maximum intercuspation, all of which allow objective measurement of finish quality [3].

Multiple factors influence post-treatment occlusal outcomes, including the type of initial malocclusion, the orthodontic mechanics used, the patient's facial biotype, and, critically, the retention protocol employed. When these outcomes are not adequately maintained, the foundation for future prosthetic rehabilitation is compromised, increasing the risk of technical failures in complex restorative systems such as implant-supported Fixed Partial Dentures (FPDs) [2,3].

The purpose of this narrative review is to analyze how orthodontic occlusal outcomes influence the biomechanical performance of implant-supported FPDs. The central question guiding this review is: how do the quality and stability of post-orthodontic occlusion affect the design requirements and long-term success of implant-supported restorations? The following sections examine the biological mechanisms underlying post-debonding adaptation, the role of retention protocols in preserving the occlusal foundation and the prosthetic variables that determine stress distribution and the risk of complications in implant-supported FPDs.

Scope and Approach to the Literature

This article is a narrative review rather than a systematic review or a case report. Electronic databases, including PubMed, Scopus and Web of Science, were searched for English-language publications addressing post-orthodontic occlusal stability, retention and the biomechanics of implant-supported fixed partial dentures, supplemented by hand-searching the reference lists of relevant articles. Because the aim is to integrate two distinct bodies of evidence into a clinically oriented framework rather than to pool quantitative outcomes, sources were selected for their conceptual and biomechanical relevance; the synthesis is interpretive and no formal risk-of-bias scoring or meta-analysis was undertaken. The review is organized into thematic sections that move from the biological basis of occlusal stability, through retention, to the prosthetic variables and materials that govern load distribution in implant-supported restorations.

Occlusal Settling vs. True Relapse: Biological Mechanisms and the Post-Debonding Adaptation Period

Following the achievement of orthodontic treatment goals, the period immediately after debonding represents a critical physiological transition. Stability should not be understood as a static endpoint but as a dynamic process in which the dentition must adapt to functional loads without the rigid support of the appliance [4]. This section distinguishes between occlusal settling, defined as a beneficial adaptive process and true relapse, which represents a failure of the supporting tissues to maintain the corrected tooth positions. Understanding this distinction is clinically essential, not only for assessing orthodontic outcomes but for planning any subsequent prosthetic rehabilitation [5].

The stability of tooth position depends on the reorganization of the periodontium. During active treatment, the Periodontal Ligament (PDL) is widened and alveolar bone undergoes constant remodeling. Once orthodontic forces are removed, the PDL requires approximately 12 to 16 weeks to regain its normal architecture. However, the supracrestal and transseptal collagen fibers possess an elastic memory that persists considerably longer; these fibers can maintain stretch and exert relapse-driving forces for up to 8 to 12 months after debonding, creating a sustained risk of rotational relapse if not adequately managed [6,7].
In contrast to relapse, occlusal settling is the spontaneous improvement of tooth contacts into a more interdigitated position and it typically occurs within the first 6 to 12 months after appliance removal. Research indicates that the first 24 hours after debonding are particularly significant; as teeth are released from archwire constraints, immediate micro-movements begin to redistribute occlusal forces more evenly. Clinical evidence suggests that allowing for controlled settling can yield a more stable long-term result, provided the changes are physiological rather than pathological [8]. The rationale for this is biomechanical: when the dentition is permitted to seek its own intercuspal position, occlusal contacts redistribute toward a more even and self-stabilizing pattern, which reduces the premature contacts and lateral force vectors that would otherwise sustain the elastic rebound of the periodontal fibers and perpetuate relapse [9].

The primary biological drivers of true relapse are the gingival and transseptal collagen fibers. Unlike the PDL, which remodels relatively quickly, these supracrestal fibers are slow to reorganize and this biological reality provides the rationale for circumferential supracrestal fibrotomy (pericision). By surgically severing these fibers at the time of debonding in cases of severe pretreatment rotations, clinicians can significantly reduce the elastic rebound effect, allowing gingival tissues to heal in a more neutral position relative to the corrected tooth orientation [10,11].

The pattern of post-treatment occlusal change is not uniform across the arch. The anterior segment, particularly the mandibular incisors, is more susceptible to relapse due to the narrowness of the alveolar bone and the sustained influence of soft tissue pressure [11]. Residual mandibular growth in young adults can also cause late incisor crowding, a phenomenon that is frequently and erroneously attributed to orthodontic treatment failure rather than to ongoing skeletal development [12]. In the posterior segment, settling tends to be more predictable, as the vertical forces of mastication encourage teeth toward a stable functional interdigitation [13].

Understanding these biological mechanisms is essential for two reasons. First, it allows the clinician to distinguish expected physiological adaptation from genuine treatment failure. Second and more directly relevant to the purpose of this review, the occlusal scheme that emerges from the post-debonding adaptation period becomes the functional environment in which any future implant-supported restoration must operate [14]. An arch that has settled into stable posterior contacts and harmonious anterior guidance provides a fundamentally different biomechanical substrate for implant prosthetics than one that has relapsed into lateral discrepancies or premature contacts.

Retention Protocols and Their Role in Occlusal Stability: Fixed vs. Removable Retainers

Orthodontic retention preserves tooth position and limits relapse after active treatment. Because periodontal remodeling, soft-tissue pressure, occlusal adaptation and residual growth continue after appliance removal, retention is essential to consolidate treatment outcomes [14,15].

Retainers vary in design and dependence on patient compliance. Fixed lingual retainers, including multi-stranded and single-wire designs, provide continuous anterior stabilization, whereas removable Hawley, vacuum-formed and positioner appliances depend on prescribed wear. Each type differs in its ability to maintain alignment and occlusal relationships over time [15,16].

Fixed and removable retainers have been widely compared [14]. Fixed retainers generally show lower relapse rates, especially in the mandibular anterior region, but findings depend on compliance, initial malocclusion severity, residual growth and differences in study protocols [17].

Despite their alignment benefits, fixed retainers can cause unwanted tooth movements, including torque changes, rotations or displacements that may mimic relapse or create new malpositions [17]. These risks highlight the need for regular follow-up during retention [14,16]. Optimal retention duration remains debated. Some clinicians favor indefinite retention because dental change is lifelong, while others recommend individualized, time-limited protocols. No universal consensus currently exists [15,16,18].

How soon a retainer is fitted once the appliance has been removed can shape both the way occlusal forces settle and the pace at which the periodontal tissues reorganize during the early transitional phase. Prompt placement is thought to help maintain corrected tooth positions during this unstable interval and to reduce the likelihood of relapse in the first weeks after treatment [16,18].

For patients who are candidates for implant-supported fixed partial dentures, the retention phase carries an additional layer of clinical significance. The occlusal scheme that is preserved or lost, during retention directly determines the biomechanical environment in which future implants will function [19]. Retention failure that allows posterior collapse, anterior crowding or midline deviation will introduce load vectors not accounted for in the original prosthetic plan, increasing the risk of non-axial implant loading, screw loosening and marginal bone loss. Clinicians planning implant-supported restorations in orthodontically treated patients should therefore verify retention compliance and confirm occlusal stability before advancing to the restorative phase (Fig. 1).

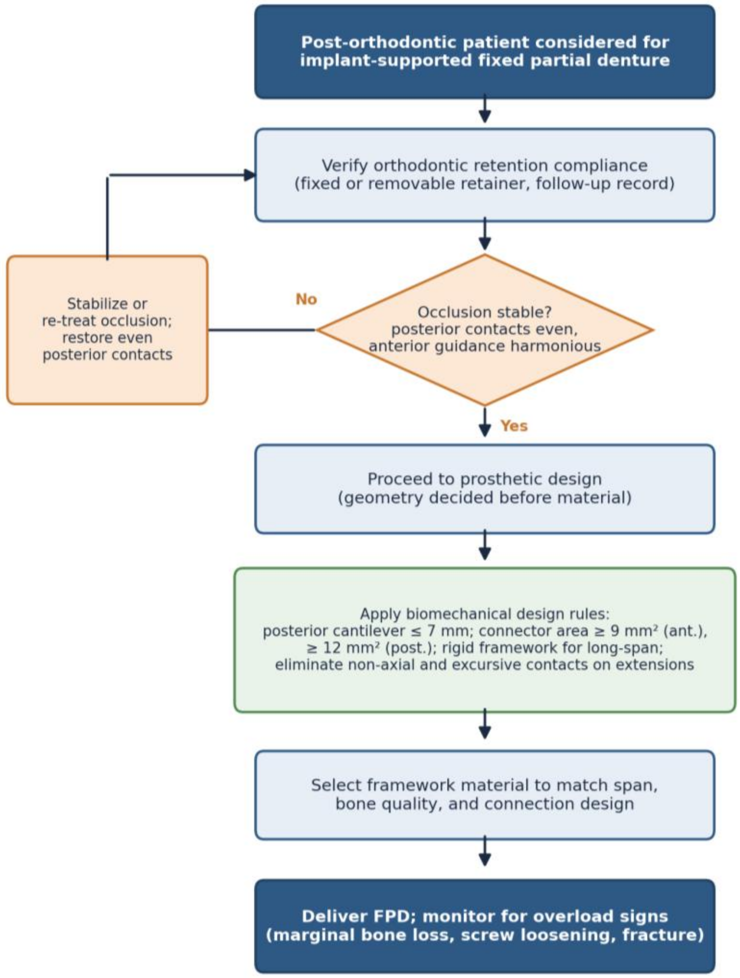

Figure 1: Clinical decision-making workflow for the orthodontic-to-restorative continuum in implant-supported fixed partial dentures. Occlusal stability is verified before prosthetic design proceeds. Geometry is determined before material selection. The restoration is monitored for biomechanical overload after delivery [16-19].

Cantilever Extensions, Connector Dimensions and Pontic Design: Prosthetic Variables with Direct Biomechanical Consequences

Prosthetic geometry is independent of osseointegration, yet it is one of the most decisive factors in how occlusal forces are transmitted through the restoration and into the periimplant tissues [19]. The distribution of biomechanical stress and the risk of mechanical failure are directly influenced by cantilever length, connector dimensions and pontic morphology and each of these variables must be optimized before framework material selection, since geometry determines the stress field within which any material must perform [20].

Cantilever Extensions

The biomechanical rationale for caution with cantilever extensions rests on a straightforward mechanical principle: a cantilever acts as a lever arm that amplifies the bending moment applied to the most proximal implant in direct proportion to the distance between the implant axis and the point of load application [21]. Under 100 N of axial loading, maximum stress at the cortical bone level reaches approximately 117 MPa when the prosthesis extends to the first molar position. When loading is applied at 30° and 45° to the implant axis, von Mises stresses in both the implant body and prosthetic connector exceed those recorded under axial loading, confirming that non-axial force vectors compound the lever-arm effect and generate greater periimplant deformation per unit of load [22].

This relationship has direct clinical consequences. Each additional millimeter of cantilever length beyond the implant axis increases the bending moment transmitted to the adjacent implant, cumulatively raising the risk of screw loosening and crestal bone overload [23]. Clinical data support a threshold of 7 mm as the upper limit for posterior cantilever length; technical failure rates for extensions exceeding this limit have been reported at 36% (95% CI: 8-71%) compared with 25% (95% CI: 13-39%) for shorter extensions, with a pooled failure rate of 13% across seven of eleven clinical studies reviewed [21].

In-vitro cyclic loading studies at 80 N over 240,000 cycles have identified prosthetic screws and abutments as the primary failure sites, with deformation rates of 90.8% and 63.2% respectively, while frameworks remained structurally intact across all groups. Screw-retained prostheses demonstrated the highest failure loads at 663.5 ± 196 N, compared with 428-487 N for cemented designs [23]. Monolithic zirconia cantilever FPDs on titanium-base abutments showed adequate mechanical stability within physiological load ranges, though performance declined with increasing cantilever span, reinforcing the 7 mm clinical threshold and the importance of eliminating lateral excursive contacts on cantilever extensions [24].

Connector Dimensions

Connector dimensions have a direct and measurable effect on fracture resistance. Three-unit all-ceramic posterior FPDs with 3 mm connector height demonstrated maximum principal stresses of 219 MPa at the mesial abutment neck under axial loading, while connectors of 4 to 5 mm height reduced this to 194 MPa with a more uniform stress distribution across the restoration [19]. The gingival embrasure of the distal connector consistently emerges as the primary stress concentration site and the reported origin of ceramic crack initiation under bending loads (Fig. 2).

The cross-sectional area of the connector also modifies how stress is transferred to the abutments. For high-modulus frameworks such as zirconia, increasing the cross-sectional area from 4 to 12 mm² progressively reduced abutment stress; for low-modulus frameworks, the opposite effect was observed, with stress increasing by more than 115% across the same area range [20]. In both cases, connectors and occlusal loading surfaces were the primary stress concentration sites regardless of material. Current recommendations specify a minimum connector cross-sectional area of 9 mm² for anterior implant-supported FPDs and 12 mm² for posterior restorations, where functional forces are highest. Oval connector cross-sections distribute stress more favorably than circular ones under equivalent loading conditions [25].

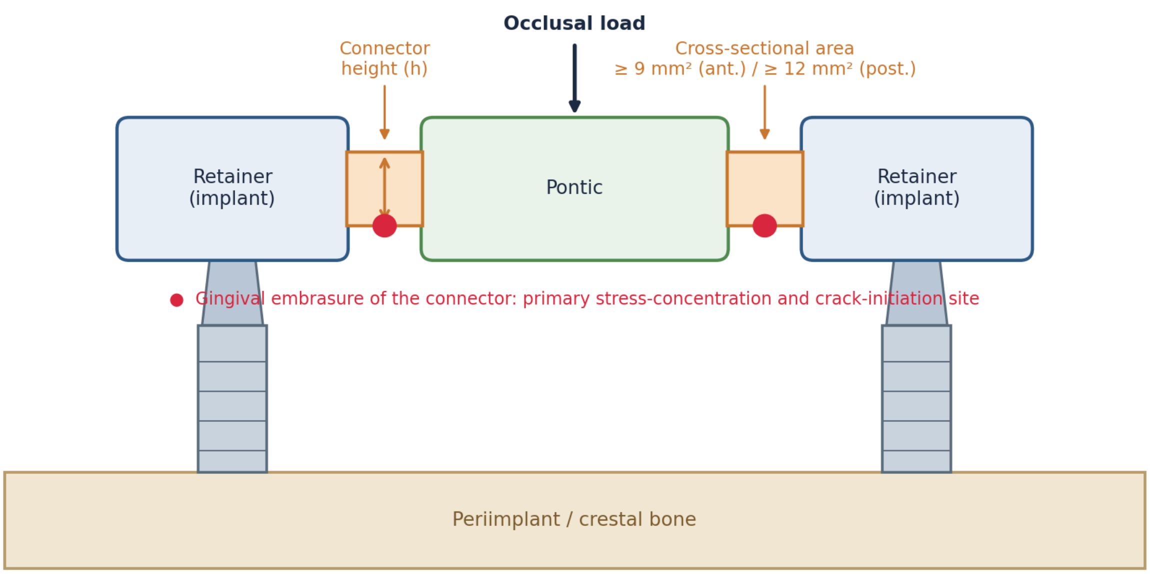

Figure 2: Schematic of a three-unit implant-supported fixed partial denture showing the connector regions between the retainers and the pontic. Connector height and cross-sectional area (at least 9 mm squared anteriorly and 12 mm squared posteriorly) govern fracture resistance; the gingival embrasure of the connector is the primary stress-concentration and crack-initiation site under occlusal load [22,25].

Pontic Design

Pontic morphology influences both the restoration's hygiene profile and its biomechanical behavior. From a hygienic standpoint, modified ridge-lap designs generate less soft tissue contact pressure than saddle morphologies, while ovate pontics, though optimal for esthetic emergence profiles, present the largest tissue contact area and the greatest potential for mucosal pressure [19]. From a biomechanical standpoint, hygienic designs with narrow pontic widths and reduced occlusal table area minimize the bending moment at the connector, which promotes mechanical longevity over esthetic considerations [26,27].

Under symmetric support conditions, the maximum stress in a fully implant-supported FPD under 100 N of axial loading is concentrated at the pontic body, reaching approximately 17 MPa. When support becomes asymmetric, stress shifts toward the connector adjacent to the less rigid abutment, significantly increasing the risk of ceramic fracture at that location [26].

Long-Span FPDs

Long-span FPDs compound all of the risks described above. Four-unit monolithic zirconia FPDs with a connector cross-sectional area of 9 mm^2 demonstrated significantly lower fracture loads than equivalent designs with 12 mm^2 connectors and flexural stress at the connector increased in a dose-dependent manner as span length grew [25]. Clinical outcomes from zirconia FPDs with cantilever extensions confirm that strict occlusal control is a critical determinant of success; screw loosening and ceramic chipping remain the most frequent technical complications in extended-span designs and both are directly related to the geometric variables reviewed in this section [27].

Framework Materials and Prosthetic Components: From Metal Alloys to Zirconia and PEEK

The mechanical properties of framework materials determine how occlusal forces are absorbed, distributed and transmitted through the prosthesis to the implant and the surrounding bone. Before selecting a material, clinicians must understand three key properties: the elastic modulus (Young's modulus), which governs how much a material deforms under load; flexural strength, which defines the threshold before structural failure; and fracture toughness, which describes the material's resistance to crack propagation once initiated [28]. These properties do not act in isolation; they interact with the geometric variables discussed in Section 4 to produce the net stress experienced by the periimplant bone and the prosthetic components themselves.

Metal Alloys: CoCr, PdAg and Titanium

Cobalt-chromium (CoCr) and palladium-silver (PdAg) alloys have a well-established clinical history in implant-supported FPDs. Their high elastic moduli provide rigid frameworks that resist flexion under load, which is advantageous in long-span and cantilever designs where framework deformation would amplify stress at the connectors and implant interfaces [29,30]. Titanium offers a slightly lower stiffness than CoCr, combined with excellent biocompatibility and decades of clinical evidence. Dynamic finite element analysis has demonstrated that CoCr frameworks exhibit lower deformation under functional loading than PEEK-based designs; however, their high rigidity means that less energy is absorbed by the framework itself, resulting in a proportionally greater transfer of stress to the periimplant bone [31].

Zirconia

Among currently available dental prosthetic materials, zirconia possesses the highest tensile strength and elastic modulus, including when compared directly with titanium-based alloys [29]. This rigidity translates into excellent load-bearing capacity for single-unit and short-span FPDs, where the material performs reliably within physiological load ranges. In multi-unit and long-span restorations, however, the same rigidity becomes a liability; zirconia's brittle failure mode means that when the stress threshold is exceeded, fracture tends to be catastrophic rather than gradual, leaving little clinical warning before complete structural failure. Titanium-zirconia assemblies have been shown to outperform pure zirconia implant systems in fracture resistance testing, suggesting that hybrid material combinations may offer superior overall performance in mechanically demanding clinical situations [32].

PEEK

Polyether Ether Ketone (PEEK) has attracted significant research interest because its elastic modulus, approximately 3 to 4 GPa, is closer to that of cortical bone than any traditional dental framework material. This characteristic offers a theoretical advantage in reducing stress shielding and improving shock absorption at the prosthesis level [32]. The clinical picture, however, is more nuanced than the material science alone suggests. PEEK does reduce stress concentration within prosthetic components and dampens impact loads effectively; but its lower stiffness simultaneously allows greater deformation at the implant-prosthesis interface, increasing micromovement and transferring greater stress to the periimplant bone compared with stiffer materials [29]. Put simply, PEEK protects the framework at the expense of the bone, while rigid materials like zirconia protect the bone at the expense of the framework. Neither response is universally desirable; the clinical context, particularly the quality and volume of available bone, should guide material selection. Additionally, PEEK's current aesthetic limitations and relatively lower flexural strength make it less suitable as a definitive framework material in high-load posterior situations, though ongoing research into surface modifications and fiber reinforcement may expand its indications in the future [31].

Fiber-Reinforced Frameworks

Carbon fiber and glass fiber-reinforced composites represent an emerging category of framework materials for implant-supported FPDs. Their mechanical properties occupy the middle ground between the high rigidity of zirconia and CoCr and the flexibility of PEEK, offering elastic moduli that can be tuned depending on fiber orientation and volume fraction [33]. Evidence for their clinical performance in implant-supported FPDs remains limited and largely experimental at this stage and they should be considered investigational rather than established options for routine clinical use [34].

Implant-Abutment Connection Design

The design of the implant-abutment connection independently modifies the stress distribution pattern at the crestal bone level and should be understood as an integral biomechanical variable rather than a hardware detail. Internal connections, particularly conical or Morse taper designs, reduce micromovement at the implant-abutment interface and improve load dissipation compared with external hexagonal configurations [32]. This reduction in micromovement has two important consequences: it decreases the risk of screw loosening under cyclic functional loading and it reduces the concentration of stress at the crestal bone, which is the region most vulnerable to marginal bone loss over time. The choice of connection design interacts directly with framework stiffness and cantilever geometry; a rigid framework on an internal conical connection will behave differently under the same load than the same framework on an external hex and these interactions should inform treatment planning in complex cases.

Tooth-Implant-Supported FPDs

When a natural tooth serves as one abutment and an osseo-integrated implant serves as the other, the biomechanical situation is complicated by the fundamental difference in mobility between the two supports. Natural teeth exhibit a physiological mobility range of approximately 25 to 100 μm due to the compliance of the periodontal ligament; by contrast, osseo-integrated implants are essentially rigid [33]. This mobility mismatch generates differential stress distribution along the span and may contribute to marginal bone loss around the implant abutment over time [33]. Systematic review data suggest that tooth-implant-supported FPDs demonstrate survival rates comparable to fully implant-supported designs, but careful prosthetic planning, including the use of non-rigid connectors in selected cases, is essential to manage the biomechanical mismatch and protect both the natural tooth and the implant from overload [34].

Clinical Decision-Making Framework, Evidence Gaps and Future Perspectives in Implant-Supported FPD Design

This review set out to connect two literatures that are usually kept apart and the preceding sections show that they are in fact continuous. The occlusal scheme handed over at the end of orthodontic treatment is the same scheme that an implant-supported FPD must later load, so the success of the restoration depends on a diagnostic approach that weighs not only the three-dimensional position of the teeth but also their future biomechanical role under function. Read together, the evidence points to a single organizing principle: stress that is not controlled biologically, through stable occlusion and adequate retention, must afterwards be controlled mechanically, through prosthetic geometry and material selection and the two strategies are most effective when they are planned as one. The subsections that follow translate this principle into a working clinical framework, identify where the evidence remains thin and outline the questions most worth pursuing.

Synthesis of Critical Biomechanical Variables

Among the prosthetic variables reviewed, cantilever length exerts the most direct effect on periimplant stress and the 7 mm threshold represents a clinically meaningful limit supported by both *in-vitro* and clinical data [11]. Connector dimensions rank second: inadequate cross-sectional area, particularly in posterior and long-span FPDs, is a primary driver of framework fracture. Framework material modulates these effects: high-modulus materials such as zirconia and CoCr are better suited to long-span and cantilever designs, whereas PEEK may reduce prosthetic stress concentrations but at the expense of increased peri-implant bone loading. Implant-abutment connection design acts as an additional modifier of crestal bone stress and should be selected with these interactions in mind [21].

Screw loosening, once a frequent complication, has become significantly less common with improvements in screw design and torque protocols; reported 5-year rates declined from 28.8% before 2000 to 4.7% thereafter [23]. Overall, 5-year survival rates for fixed implant-retained prostheses are high (approximately 97.1%), but biological and technical complications remain a meaningful source of morbidity [3].

Clinical Decision-Making Framework

Before fabricating an implant-supported FPD in a patient with prior orthodontic treatment, the clinician should verify three conditions: (1) that retention compliance has been maintained and no significant relapse has occurred; (2) that the posterior occlusal contacts are stable, well-distributed and free of premature contacts or lateral interferences; and (3) that anterior guidance is harmonious with the planned prosthetic occlusion.

Prosthetic design decisions should then be guided by a hierarchy of biomechanical risk (Table 1). In the anterior region, the connector cross-sectional area should be at least 9 mm² and cantilever extensions should be avoided where possible [16,35]. In the posterior region, the minimum connector area increases to 12 mm², the cantilever length should not exceed 7 mm and non-axial contacts on extensions should be eliminated. Long-span FPDs require the most rigid frameworks available; zirconia with triangular connector cross-sections has been shown to demonstrate the highest fracture resistance in this scenario [24,36]. Distal cantilever configurations generate higher stress concentrations than mesial ones; while mesial cantilevers in specific clinical scenarios, such as the atrophic posterior maxilla with distally tilted implants, may be biomechanically acceptable, this cannot be generalized without careful case-by-case analysis [25,37].

Clinical signs of biomechanical overload should be recognized early. Progressive marginal bone loss, recurrent screw loosening and ceramic or framework fractures are red flags warranting a systematic review of the occlusal scheme, prosthetic design and implant distribution before any repair or replacement is undertaken.

Design variable	Recommendation / threshold	Biomechanical rationale	Ref.
Posterior cantilever length	7 mm or less	Each additional millimeter amplifies the bending moment on the proximal implant; technical failure rates were 36% above this limit versus 25% below it	[21,23]
Connector cross-sectional area, anterior	9 mm squared or more	Inadequate area is a primary driver of framework fracture; the connector is a principal stress-concentration site	[25,35]
Connector cross-sectional area, posterior	12 mm squared or more	Functional forces are highest posteriorly; larger area lowers connector flexural stress in long-span designs	[20,25]
Connector height	4 to 5 mm	Increasing height from 3 mm to 4 to 5 mm reduced peak stress from 219 to 194 MPa with more uniform distribution	[19]
Connector cross-section shape	Oval or triangular	Distributes stress more favorably than circular sections; triangular geometry showed the highest fracture resistance in long-span FPDs	[25,36]
Pontic design	Narrow, hygienic (modified ridge-lap)	Reduced pontic width and occlusal table minimize the bending moment transmitted to the connector	[19]
Framework for long-span	High-modulus	Rigid frameworks resist the flexion that would	[29,31]

Design variable	Recommendation / threshold	Biomechanical rationale	Ref.
or cantilever	material (zirconia, CoCr)	amplify stress at connectors and implant interfaces	
Implant-abutment connection	Internal conical (Morse taper)	Reduces interface micromovement and crestal stress relative to external hex, lowering screw loosening	[32]
Excursive contacts on cantilevers	Eliminate	Non-axial vectors compound the lever-arm effect and increase periimplant deformation per unit load	[22,24]

Table 1: Summary of key biomechanical design parameters and evidence-based thresholds for implant-supported fixed partial dentures.

Evidence Gaps

Despite the growing body of biomechanical literature, several gaps limit the clinical translation of current evidence (Table 2). Finite element analysis models remain simplified constructs that cannot fully replicate the viscoelastic properties of bone, the variability of patient anatomy or the dynamic complexity of masticatory loading [24,38]. Longitudinal clinical studies with follow-up periods exceeding 5 years are scarce and most available data come from short- to medium-term observational studies. Standardization of loading parameters across *in-vitro* studies remains poor, making direct comparisons difficult. Perhaps most importantly, patient-reported outcome measures are almost entirely absent from the biomechanical literature, leaving a substantial gap in our understanding of how material and design choices translate into patient experience and quality of life [2].

Current Evidence Gap	Corresponding Research Priority
Finite element models cannot fully replicate the viscoelastic properties of bone, anatomic variability or dynamic masticatory loading	Patient-specific finite element models derived from CBCT data to improve clinical relevance
Longitudinal clinical studies with follow-up beyond 5 years are scarce; most data are short to medium term	Long-term longitudinal studies that integrate biomechanical parameters with clinical outcomes
Loading parameters across *in-vitro* studies are poorly standardized, making direct comparison difficult	Standardized loading protocols to enable direct cross-study comparison
Patient-reported outcome measures are almost entirely absent from the biomechanical literature	Integration of patient-centered and quality-of-life outcomes into biomechanical research
The effect of parafunctional habits, particularly bruxism, on long-term fatigue behavior is inadequately studied	Dedicated investigation of parafunctional loading on the fatigue behavior of implant-supported FPDs
Emerging bioactive materials and AI-assisted prosthetic design are not yet clinically validated	Clinical evaluation of bioactive materials and AI-assisted prosthetic design workflows

Table 2: Principal evidence gaps in the biomechanical literature on implant-supported fixed partial dentures and the corresponding research priorities [24,38].

Conclusion

The biomechanical evidence reviewed in the preceding sections converges on a consistent message: the success of implant-supported FPDs is determined not by any single variable but by the interaction among multiple factors, beginning with the occlusal foundation established by orthodontic treatment and ending with the framework's material properties. A clinician who addresses only one of these domains while neglecting the others is working with an incomplete picture. The practical message is interdisciplinary: the osseous and occlusal substrate that receives an implant-supported FPD is shaped by the orthodontic and biological history preceding it and ignoring that history is a clinical risk. When retention, settling and prosthetic design are planned as a single continuum grounded in biomechanical evidence, durable and complication-free restorations become a realistic goal.

Future research should prioritize longitudinal studies that integrate biomechanical parameters with clinical and patient-centered outcomes. Patient-specific finite element models derived from CBCT data represent a promising direction for improving the clinical relevance of computational biomechanics. The impact of parafunctional habits, particularly bruxism, on the long-term fatigue behavior of implant-supported FPDs is an area of critical importance that remains inadequately studied. Emerging bioactive materials and AI-assisted prosthetic design represent additional frontiers that may substantially change clinical practice over the next decade.

Conflict of Interest

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Data Availability Statement

The data supporting the findings of this study are available from the corresponding author upon reasonable request.

Ethical Statement

The project did not meet the definition of human subject research under the preview of the IRB according to federal regulations and therefore was exempt.

Informed Consent Statement

Not applicable.

Authors' Contributions

All authors contributed equally to this paper.

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