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Porównanie ścieralności materiałów odtwórczych na zębach naturalnych i implantach stomatologicznych – badanie in vitro

Rozprawa na stopień doktora nauk medycznych i nauk o zdrowiu

w dyscyplinie nauki medyczne

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Obrona rozprawy doktorskiej przed Radą Dyscypliny Nauk Medycznych

Warszawskiego Uniwersytetu Medycznego

Warszawa, 2026 r.

Słowa kluczowe: starcie zębów, więzadło ozębnej, biomechanika, implanty stomatologiczne, badania in vitro, ceramika stomatologiczna

Key words: tooth wear, periodontal ligament, biomechanics, dental implants, in vitro techniques, dental ceramics

Spis Treści

Wykaz skrótów

- **CAD/CAM** – computer-aided design / computer-aided manufacturing
- **PDL** – periodontal ligament (więzadło ozębnej)

Wykaz publikacji stanowiących rozprawę doktorską

Publikacja nr 1

Kosewski, P., Kosewski, J., & Mielczarek, A. (2025). Simulation of the Periodontal Ligament in Dental Materials Research: A CAD/CAM-Based Method for PDL Modeling. *Journal of Functional Biomaterials*, *16*(12), 429. https://doi.org/10.3390/jfb16120429

Punkty MNiSW: 100

JIF: 5.2

JIF Quartile: Q1

Publikacja nr 2

Kosewski, P., De Angelis, F., Sorrentino, E., Mielczarek, A., Buonvivere, M., & D'Arcangelo, C. (2023). Effect of the Abutment Rigidity on the Wear Resistance of a Lithium Disilicate Glass Ceramic: An In Vitro Study. *Journal of Functional Biomaterials*, *14*(8), 395. https://doi.org/10.3390/jfb14080395

Punkty MNiSW: 100

JIF: 5.0

JIF Quartile: Q1

Streszczenie

Biomechaniczne różnice pomiędzy zębami naturalnymi a implantami stomatologicznymi, wynikające przede wszystkim z obecności więzadła ozębnej, wpływają na sposób przenoszenia obciążeń okluzyjnych oraz na zachowanie materiałów protetycznych w warunkach czynnościowych. W badaniach laboratoryjnych różnice te bywają pomijane, co może ograniczać trafność interpretacji wyników dotyczących zużycia materiałów odtwórczych.

Celem niniejszej rozprawy doktorskiej było porównanie ścieralności materiału protetycznego w rekonstrukcjach opartych na implantach stomatologicznych oraz w układzie odtwarzającym biomechanikę zęba naturalnego. W tym celu opracowano standaryzowany model eksperymentalny umożliwiający kontrolowane odwzorowanie elastyczności podparcia, charakterystycznej dla uzębienia naturalnego.

Rozprawa została oparta na cyklu dwóch prac oryginalnych. Pierwsza z nich dotyczyła opracowania autorskiego modelu fizycznego odtwarzającego właściwości podporowe zęba naturalnego z wykorzystaniem technologii CAD/CAM oraz krytycznej analizy dotychczas stosowanych metod w tym obszarze. W pracy tej wskazano najważniejsze ograniczenia metodologiczne wcześniejszych modeli, określono sytuacje badawcze, w których uwzględnienie symulacji więzadła ozębnej ma istotne znaczenie, a także sformułowano zalecenia dotyczące dokumentacji i standaryzacji badań laboratoryjnych wykorzystujących takie modele.

Druga publikacja obejmowała ocenę wpływu rodzaju podparcia na ścieralność ceramiki dwukrzemianowo-litowej w warunkach symulowanego obciążenia okluzyjnego. Wykazano, że korony oparte na sztywnym podparciu implantologicznym ulegały większemu starciu niż rekonstrukcje badane w układzie odtwarzającym biomechanikę zęba naturalnego. Wyniki te wskazują, że charakter podparcia istotnie wpływa na przebieg procesu ścierania materiału protetycznego w warunkach laboratoryjnych.

Przeprowadzone badania sugerują, że rodzaj filaru rekonstrukcji protetycznej może mieć istotne znaczenie dla ścieralności materiałów protetycznych. Opracowany model laboratoryjny może ponadto stanowić narzędzie badawcze w analizach in vitro w różnorodnych scenariuszach badawczych z zakresu stomatologii odtwórczej.

Abstract

Title: Comparison of the Wear of Restorative Materials on Natural Teeth and Dental Implants: An In Vitro Study

Biomechanical differences between natural teeth and dental implants, arising primarily from the presence of the periodontal ligament, affect both the transmission of occlusal loads and the behavior of prosthetic materials under functional conditions. In laboratory studies, these differences are sometimes overlooked, which may limit the validity of the interpretation of results concerning the wear of restorative materials.

The aim of this doctoral dissertation was to compare the wear of prosthetic materials in implant-supported restorations with that observed in a model reproducing the biomechanical behavior of a natural tooth. For this purpose, a standardized experimental model was developed to enable controlled simulation of the resilience characteristic of natural tooth support.

The dissertation was based on a series of two original studies. The first concerned the development of an original physical model reproducing the supporting properties of a natural tooth using CAD/CAM technology, together with a critical analysis of the methods previously used in this field. This study identified the key methodological limitations of earlier models, defined the research scenarios in which simulation of the periodontal ligament is of particular importance, and formulated recommendations regarding the reporting and standardization of laboratory studies employing such models.

The second study evaluated the influence of the type of support on the wear of lithium disilicate ceramic under simulated occlusal loading. It was demonstrated that crowns supported by rigid implant-based support underwent greater wear than reconstructions tested in a system reproducing the biomechanics of a natural tooth. These findings indicate that the nature of the support significantly affects the wear behavior of prosthetic material under laboratory conditions.

The conducted studies indicate that the type of abutment supporting a prosthetic reconstruction may play an important role in the wear of prosthetic materials. Moreover, the developed laboratory model may serve as a research tool for laboratory analyses across a wide range of investigative scenarios in restorative dentistry.

Wstęp

Układ stomatognatyczny stanowi złożony system funkcjonalny obejmujący zęby, przyzębie, stawy skroniowo-żuchwowe, mięśnie żucia oraz struktury nerwowe[1], w którym kontakty zwarciowe pomiędzy zębami przeciwstawnych łuków ulegają ciągłym fizjologicznym zmianom w odpowiedzi na obciążenia czynnościowe. Ścieranie się twardych tkanek zębów jest procesem nieodłącznie związanym z funkcją żucia. Proces ten pełni istotną rolę adaptacyjną, umożliwiając stopniowe dostosowywanie relacji zwarciowych do warunków zmieniających się w czasie[2]. Nawet szkliwo, mimo iż jest najbardziej zmineralizowaną tkanką organizmu ludzkiego i charakteryzuje się wysoką odpornością na zużycie, w warunkach fizjologicznych ulega powolnemu ścieraniu. Proces ten, przebiega równolegle z ciągłą, fizjologiczną migracją zębów oraz przebudową tkanek przyzębia, co umożliwia zachowanie stabilnej i funkcjonalnie zharmonizowanej okluzji, pomimo stopniowo ewoluujących warunków biomechanicznych i anatomicznych[3]. Okluzja nie stanowi zatem stanu statycznego, lecz jest wynikiem dynamicznych interakcji pomiędzy siłami zwarciowymi, właściwościami mechanicznymi tkanek twardych zębów oraz zdolnością układu stomatognatycznego do adaptacji[2].

Kluczowym elementem umożliwiającym tę adaptację jest więzadło ozębnej (periodontal ligament, PDL). Stanowi ono wyspecjalizowaną, włóknistą strukturę łącznotkankową, zakotwiczoną zarówno w cemencie korzeniowym jak i blaszce kostnej zębodołu[4]. Ozębna pełni funkcję aparatu zawieszeniowego zęba, zapewniając jego fizjologiczną ruchomość oraz zdolność do tłumienia i rozpraszania obciążeń mechanicznych. Jej właściwości biomechaniczne wynikają nie tylko z obecności włókien kolagenowych, lecz także płynu śródmiąższowego oraz bogatego unaczynienia, co nadaje jej charakter wiskoelastyczny[5]. Opisane cechy budowy umożliwiają częściową absorpcję energii generowanej podczas kontaktów zwarciowych i stopniowe przekazywanie jej do otaczającej tkanki kostnej. Tym samym dochodzi do obniżenia koncentracji naprężeń w obrębie korzenia zęba[6].

W sytuacji, gdy braki zębowe są uzupełniane z wykorzystaniem implantów stomatologicznych, pojawiają się istotne różnice w reakcji układu stomatognatycznego na obciążenia czynnościowe. W wyniku osteointegracji implanty tworzą bezpośrednie i sztywne połączenie z kością, przez co nie wykazują fizjologicznej ruchomości charakterystycznej dla zębów naturalnych. Konsekwencją tego jest również utrata funkcjonowania sieci mechanoreceptorów związanych z więzadłami ozębnej, co znacząco zmienia sposób percepcji oraz kontroli obciążeń działających w układzie stomatognatycznym. Eliminacja tych struktur pozbawia układ stomatognatyczny nie tylko proprioceptywnej informacji zwrotnej, ale również istotnego elementu neuromotorycznej regulacji obciążeń okluzyjnych. W konsekwencji prowadzi to do zasadniczo odmiennej charakterystyki biomechanicznej funkcjonowania implantów w porównaniu z zębami naturalnymi[5]. Z punktu widzenia biomechaniki okluzji oznacza to, że mechanizmy adaptacyjne, funkcjonujące w naturalnym uzębieniu są w przypadku implantów znacząco ograniczone[7]. Brak elastycznego aparatu zawieszeniowego ma istotne konsekwencje kliniczne, zwłaszcza w aspekcie długoterminowej trwałości odbudów implantoprotetycznych[5,8]. Z tego względu, kompensację braku ruchomości fizjologicznej implantu uzyskuje się poprzez redukcję kontaktów okluzyjnych na koronie implantoprotetycznej oraz systematyczną kontrolę i korektę relacji zwarciowych[9].

Dodatkowym czynnikiem wpływającym na funkcjonowanie układu stomatognatycznego w przypadku obecności prac osadzonych na implantach są właściwości materiałów stosowanych do rekonstrukcji koron protetycznych. Współcześnie do tego typu odbudów najczęściej wykorzystywane są ceramiki na bazie dwukrzemianu litu, ceramiki szklane lub tlenek cyrkonu. Materiały te charakteryzują się wysoką twardością, odpornością na zużycie oraz korzystnymi cechami estetycznymi. Ich właściwości mechaniczne znacząco odbiegają jednak od parametrów szkliwa i zębiny[10]. Ceramiki, a w szczególności tlenek cyrkonu, wykazują ograniczoną zdolność do ścierania się oraz wysoki moduł sprężystości, co prowadzi do odmiennego zachowania tribologicznego w kontakcie z antagonistami[11]. Sztywne zakotwiczenie implantu w kości sprzyja zwiększonemu obciążeniu struktur przeciwstawnych oraz występowaniu powikłań natury mechanicznej w obrębie odbudów protetycznych[12,13].

Ocena ścieralności materiałów stomatologicznych w warunkach klinicznych jest zadaniem złożonym i obarczonym licznymi ograniczeniami. Na zużycie tkanek i materiałów odtwórczych wpływa jednocześnie wiele czynników, takich jak siły żucia, schemat okluzji, obecność parafunkcji, choroby współistniejące, dieta czy indywidualne cechy pacjenta[14,15]. W efekcie, badania *in vivo* rzadko umożliwiają jednoznaczne określenie wpływu pojedynczych zmiennych, w tym rodzaju filaru protetycznego. W odpowiedzi na te ograniczenia istotną rolę odgrywają badania *in vitro*, które pozwalają na kontrolę warunków eksperymentalnych oraz wyizolowaną ocenę poszczególnych zmiennych[16]. Z drugiej strony, modele laboratoryjne o nadmiernie uproszczonej konstrukcji mogą nie odzwierciedlać w pełni warunków klinicznych, zwłaszcza jeśli nie uwzględniają obecności więzadła ozębnej jako kluczowego komponentu funkcjonalnego.

Od wielu lat w celu zwiększenia możliwości translacji wyników badań laboratoryjnych na warunki kliniczne podejmowane są próby symulacji PDL w modelach *in vitro*. Najczęściej koncentrują się one na wprowadzeniu elastycznej warstwy materiału (np. materiału silikonowego lub polieterowego) pomiędzy korzeń zęba a materiał imitujący kość[17]. Mimo licznych doniesień, proponowane rozwiązania charakteryzują się istotną zmiennością metodologiczną i trudnością w zapewnieniu powtarzalnej grubości oraz jednorodności warstwy symulującej więzadło ozębnej[17-19].

Brak standaryzacji w tym obszarze utrudnia porównywanie wyników badań i prowadzi do istotnych ograniczeń interpretacyjnych. Rozwój technologii cyfrowych stworzył jednocześnie nowe możliwości projektowania i wytwarzania modeli eksperymentalnych, umożliwiając precyzyjne definiowanie ich geometrii oraz zwiększenie powtarzalności procesu. W efekcie, modele te mogą lepiej odzwierciedlać rzeczywiste warunki kliniczne, co zwiększa wiarygodność uzyskiwanych danych i ułatwia ich porównywalność między różnymi ośrodkami badawczymi.

Cykl publikacji stanowiący podstawę niniejszej rozprawy doktorskiej wpisuje się w nurt badań ukierunkowanych na pogłębienie zrozumienia biomechanicznych różnic pomiędzy zębami naturalnymi a implantami stomatologicznymi oraz ich wpływu na proces zużycia materiałów protetycznych.

Ujęcie obu prac w jednym cyklu badawczym wynika ze wspólnej problematyki, obejmującej zarówno analizę zjawisk związanych ze ścieralnością, jak i tworzenie metod pozwalających na ich odtwarzanie i ocenę w warunkach laboratoryjnych.

Pierwsza publikacja koncentruje się na opracowaniu i walidacji standaryzowanego, cyfrowego modelu *in vitro* symulującego więzadła ozębnej z wykorzystaniem technik CAD/CAM. Zaproponowana metoda umożliwia precyzyjne zaprojektowanie przestrzeni dla warstwy elastycznej, jej powtarzalne odtworzenie oraz funkcjonalną kalibrację opartą na pomiarach stopnia ruchomości badanego filaru. Proponowany model stanowi odpowiedź na zidentyfikowane w literaturze ograniczenia metodologiczne. Umożliwia jednocześnie stworzenie układu testowego, pozwalającego na wiarygodne porównanie zachowania materiałów protetycznych osadzonych na zębach i na implantach, przy jednoczesnym ograniczeniu wpływu innych zmiennych.

Druga publikacja koncentruje się na ocenie wpływu rodzaju filaru protetycznego na ścieralność ceramiki na bazie dwukrzemianu litu. W pracy tej przeanalizowano różnice w poziomie ścieralności odbudów osadzonych na implantach i rekonstrukcji opartych na zębach, z wykorzystaniem symulowanego więzadła ozębnej. Uzyskane wyniki sugerują, że charakter podparcia odbudowy istotnie wpływa na przebieg procesu zużycia materiału - korony osadzone na implantach wykazywały większą intensywność starcia niż rekonstrukcje oparte na zębach z symulacją PDL, co podkreśla znaczenie biomechaniki filaru w ocenie właściwości tribologicznych ceramik stomatologicznych.

Prezentowane osiągnięcia naukowe mogą stanowić podstawę do dalszych prac nad optymalizacją materiałów oraz protokołów badawczych w implantoprotetyce.

Założenia i cel pracy

Cel główny:

Głównym celem niniejszej rozprawy jest ocena, w warunkach in vitro, wpływu rodzaju filaru protetycznego - w postaci implantu stomatologicznego lub modelu odzwierciedlającego biomechanikę zębów naturalnych - na ścieralność materiału stosowanego do rekonstrukcji protetycznej.

Cele pośrednie:

1. Przegląd i krytyczna analiza dotychczasowych metod symulowania więzadła ozębnej stosowanych w laboratoryjnych badaniach właściwości mechanicznych materiałów stomatologicznych.
2. Opracowanie i walidacja modelu eksperymentalnego umożliwiającego badanie właściwości mechanicznych materiałów protetycznych z wykorzystaniem imitacji zębów naturalnych z odtworzonymi właściwościami mechanicznymi więzadła ozębnej.
3. Ocena ścieralności koron na bazie dwukrzemianu litu osadzonych na implantach stomatologicznych i na replikach zębów z symulacją więzadła ozębnej w warunkach laboratoryjnych.

Materiał i metody

Badania stanowiące podstawę niniejszej rozprawy doktorskiej miały charakter badań laboratoryjnych i zostały zrealizowane w ramach dwóch komplementarnych etapów, których szczegółowy opis zawarto w publikacjach wchodzących w skład cyklu.

W pierwszym etapie przeprowadzono przegląd piśmiennictwa dotyczący metod symulacji więzadła ozębnej, a następnie opracowano i zastosowano model symulujący obecność PDL, umożliwiający przybliżone odtworzenie warunków biomechanicznych charakterystycznych dla zęba naturalnego. Model ten został zaprojektowany i wykonany z wykorzystaniem technologii cyfrowych CAD/CAM, co umożliwiło standaryzację geometrii próbek oraz zapewniło powtarzalność warunków badawczych.

W drugim etapie przeprowadzono badania ścieralności materiału ceramicznego na bazie dwukrzemianu litu, osadzonego na filarach o zróżnicowanej sztywności, odpowiadających warunkom biomechanicznym zęba naturalnego oraz implantu stomatologicznego. Próbki poddano cyklicznym obciążeniom mechanicznym w symulatorze żucia.
W części eksperymentalnej nie wykorzystywano naturalnych zębów ludzkich, lecz standaryzowany model repliki zęba z fizyczną symulacją więzadła ozębnej. Takie rozwiązanie przyjęto celowo w celu ograniczenia zmienności próbek oraz zapewnienia jednorodnych warunków badania.

Analizę zużycia materiału ceramicznego przeprowadzono stosując metody ilościowe, oparte na pomiarach wykonanych z użyciem skanera optycznego. Ślady starcia poddano również analizie jakościowej z wykorzystaniem skaningowego mikroskopu elektronowego. Uzyskane dane poddano analizie statystycznej adekwatnej do charakteru badania, zgodnie z metodyką opisaną w publikacjach stanowiących podstawę rozprawy.

Kopie opublikowanych prac

Publikacja nr 1

Journal of
Functional Biomaterials

MDPI

Article

Simulation of the Periodontal Ligament in Dental Materials Research: A CAD/CAM-Based Method for PDL Modeling

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Abstract

The periodontal ligament (PDL) is essential for the physiological mobility and load distribution of natural teeth, yet its simulation in mechanical testing remains inconsistent and insufficiently standardized. The absence of a resilient suspension system can alter force transmission, affect failure patterns, and reduce the clinical relevance of in vitro outcomes. This study aimed to develop a reproducible CAD/CAM-based model for PDL simulation that provides elastic suspension of a tooth replica under laboratory conditions. A digitally defined offset was applied around a tooth replica to create a controlled PDL space, which was filled with polyether. To ensure precise seating of the specimens, a 3D-printed positioning device was used. Functional calibration was performed using Periotest measurements to identify the offset that reproduced physiological tooth mobility. A digital offset of 0.85 mm produced a radiographically confirmed polyether layer of 0.86 $\pm$ 0.05 mm and yielded Periotest values comparable to natural teeth in the horizontal direction (mean PTV = 2.99 $\pm$ 0.92). Vertical measurements demonstrated higher damping (mean PTV = $-$4.02 $\pm$ 0.56), consistent with the anisotropic behavior of natural PDL. The model showed high fabrication accuracy and predictable mechanical behavior, providing a physiologically relevant method for incorporating PDL simulation into laboratory mechanical testing.

Keywords: computer-aided design; experimental model; fixed partial denture; in vitro testing; mechanical testing; periodontal ligament

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Academic Editor: Pankaj Vadgama

Received: 21 September 2025
Revised: 26 October 2025
Accepted: 21 November 2025
Published: 24 November 2025

Citation: Kosewski, P.; Kosewski, J.; Mielczarek, A. Simulation of the Periodontal Ligament in Dental Materials Research: A CAD/CAM-Based Method for PDL Modeling. *J. Funct. Biomater.* **2025**, *16*, 429. https://doi.org/10.3390/jfb16120429

1. Introduction

Natural teeth are anchored in the alveolar bone via the periodontal ligament (PDL)—a thin layer of connective tissue approximately 0.1–0.3 mm thick that fills the space between the tooth root and the alveolar bone [1]. It is composed of collagen fibers embedded in a ground substance and contains interstitial fluid, which imparts its viscoelastic properties [2]. This means that the PDL undergoes slight deformation under applied forces, but its stiffness increases with the magnitude and rate of loading. Under typical masticatory forces of approximately 70 N, the PDL allows for root deflection in the range of 0.03–0.15 mm [1]. Under increasing loads, the PDL exhibits strain-dependent stiffening, as collagen fibers progressively stretch, leading to enhanced stress transfer to the surrounding alveolar bone. [3] This damping mechanism protects the tooth and its surrounding structures from trauma, as the PDL absorbs energy and prevents stress concentration within the root itself [4].

Modern dental treatment concepts are increasingly based on biomimetic principles, aiming to restore the function of the stomatognathic system in a manner that closely replicates natural conditions. A fundamental principle in the rehabilitation of the stomatognathic system is the use of restorative materials that mimic the behavior of natural tissues during oral functions—such as chewing, occlusion, parafunctions, and voice articulation—and that can adapt over time to changing occlusal conditions, similarly to enamel, dentin, or bone tissue [5–7]. An additional challenge arises in replicating the function of natural teeth with dental implants used to replace missing teeth. Unlike natural teeth, implants osseointegrate without a PDL, resulting in markedly different biomechanical properties. Lacking the PDL, the implant is rigidly anchored and virtually immobile. Studies have shown that a loaded implant generates higher stress levels in the surrounding bone compared to a natural tooth suspended by the periodontal ligament apparatus [2]. Therefore, the design of biomimetic implant-supported restorations aims to compensate for the reduced ability to dissipate energy in comparison to natural teeth [2].

Some in vitro studies of stomatognathic function have shown that the laboratory model is often simplified by embedding tested teeth or prosthetic materials in a rigid medium (e.g., acrylic resin or plaster) without simulating the PDL. Such models behave similarly to ankylosed teeth, exhibiting no physiological mobility [8]. Omitting the role of the PDL in such studies may lead to distorted mechanical test results.

Previous in vitro investigations have shown that the influence of PDL simulation varies considerably depending on the type of mechanical test performed—being most pronounced in fixed partial dentures (FPDs) [9–11], post-and-core systems [12,13] and wear [14,15], while generally less significant in single-crown fracture resistance tests [16,17].

In light of these discrepancies, there is a growing trend in laboratory research to incorporate PDL simulation [18]. The aim is to replicate, within experimental models, the tooth's ability to undergo slight displacement under load, as observed under in vivo conditions [18]. Accurate replication of these characteristics should result in a stress distribution similar to that observed in nature, thereby enhancing the clinical relevance and translatability of in vitro findings.

Various techniques have been described in the literature to mimic the function of the PDL. However, a standardized protocol for fabricating experimental models that replicate the PDL is still lacking, and the existing approaches show considerable heterogeneity.

In view of these limitations, there is a need for experimental models that reproduce the elastic behavior of the PDL in a standardized and reproducible manner, while remaining compatible with common mechanical testing workflows.

Therefore, the primary objective of this study was to develop and functionally validate a CAD/CAM-based experimental model for PDL simulation around a single-rooted tooth replica, using a polyether layer with digitally controlled thickness calibrated to physiological tooth mobility. A focused overview of current PDL simulation approaches, summarized in Appendix A, was used to contextualize key design choices and to derive practical recommendations for implementing PDL analogs in future in vitro studies.

2. Materials and Methods

2.1. CAD/CAM-Based Method for the PDL Simulation

A tooth model was digitally designed in SolidWorks 2016 SP 5.0 (Dassault Systèmes SolidWorks Corporation, Vélizy-Villacoublay, France) based on the average dimensions of a maxillary second premolar [19]. The abutment was digitally designed to replicate a tooth prepared for a full lithium disilicate crown, following manufacturer guidelines with a 2 mm occlusal reduction and 1.5 mm axial reduction at the level of crown margin. The convergence angle was set to 8°. The tooth replica was milled from a hybrid ceramic material with mechanical properties comparable to natural dentin [20–23] (Ambarino High Class, Creamed GmbH & Co., Marburg, Germany). Comparative mechanical data for Ambarino High Class and human dentin are provided in Table 1.

Table 1. Comparative mechanical properties of human dentin and Ambarino® High Class hybrid ceramic. Data for Ambarino High Class were obtained from the manufacturer's technical documentation (Creamed GmbH & Co., 2023). Dentin data represent mean ranges reported in the cited studies.

Property	Human Dentin	Ambarino High Class
Elastic modulus (GPa)	16–25 GPa [21–23]	10 GPa
Hardness (GPa)	0.4–0.7 GPa [20,23]	0.71 GPa (≈710 MPa)
Compressive strength (MPa)	190–300 MPa [21,22]	500 MPa

Cubic PMMA (polymethyl methacrylate) (Aidite Clear PMMA, Aidite Technology Co., Ltd., Qinhuangdao, China) blocks measuring 20 × 12 × 12 mm were milled with a cavity left for an artificial socket, generated by digitally enlarging the tooth model and subtracting it from the PMMA block. PMMA was selected because its elastic modulus (mean 2.88 ± 0.01 GPa) [24] is comparable to that of mandibular bone (approximately 3 GPa) [25], ensuring mechanical behavior similar to the supporting structures in vivo.

All components were fabricated using a 5-axis milling unit (RS-Team RS5, RS-Team S.C., Otwock, Poland). Hybrid ceramic specimens were milled under wet conditions with a 1 mm diamond bur at a spindle speed of 60,000 rpm, feed rate of 2.0 mm/s, and a layer thickness of 0.02 mm. PMMA blocks were milled under dry conditions using a carbide bur, with a spindle speed of 40,000 rpm, a feed rate of 3.0 mm/s, and a layer thickness of 0.05 mm.

The positioning devices were fabricated with a stereolithography printer (Formlabs 3, Formlabs Inc., Somerville, MA, USA) using NextDent Model 2.0 resin (Vertex-Dental B.V., Soesterberg, The Netherlands) at 50 µm layer thickness, followed by washing in isopropyl alcohol for 10 min and UV post-curing for 30 min at 60 °C.

The CAD design files for the tooth, socket, and positioning device are openly available on Zenodo (Dataset, DOI: 10.5281/zenodo.17392949) to ensure reproducibility of the CAD/CAM workflow.

According to established methodology, a polyether material (Impregum Penta, 3M ESPE, St. Paul, MN, USA) was used to replicate the viscoelastic properties of the PDL [11,26].

The appropriate space for the polyether layer was determined by fabricating a series of blocks with increasing socket dimensions to achieve vibration-damping properties comparable to natural teeth. Five blocks were fabricated for each of the following ligament space dimensions: 0.25 mm, 0.5 mm, 0.75 mm, and 0.85 mm. The damping capacity of tooth replicas seated in their sockets was evaluated using a Periotest Classic device (Medizintechnik Gulden, Modautal, Germany).

Periotest values (PTVs) range from −8 to +50, where lower or negative readings indicate higher damping capacity and greater stability (as in osseointegrated implants or ankylosed teeth), whereas higher positive readings correspond to reduced damping and increased tooth mobility [27].

The objective was to obtain PTVs within the physiological range of 2–3, corresponding to the mobility of natural maxillary teeth (mean value approximately 2.5 PTV) [27]. A PDL space of 0.85 mm provided vibration damping within this range, confirming its suitability for the experimental model. The relationship between simulated PDL thickness and PTV is presented in Figure 1.

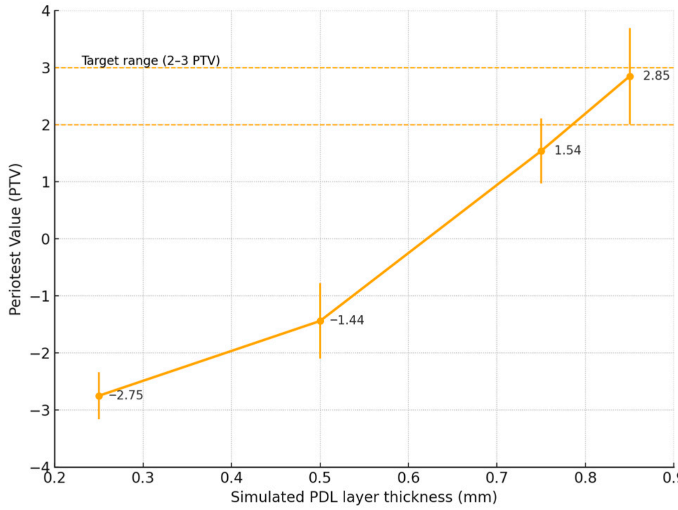


Figure 1. Calibration curve showing the relationship between simulated PDL layer thickness and Periotest values, with error bars representing standard deviations (SD) and the dashed lines indicating the target physiological range of 2–3 PTV.

To position the tooth roots within the blocks, 3D-printed resin positioning devices were used (NextDent Model 2.0, Vertex-Dental B.V., Soesterberg, The Netherlands). The positioning cube included the following:

- A central socket for the coronal part of the replica, ensuring fixed orientation and maintaining the root at a specific distance from the socket base;
- A 2 mm vertical clearance around the socket for excess polyether material during placement;
- An outer flange matching the block's contour with a 0.05 mm offset, providing alignment and ensuring the root axis was parallel to the socket's long axis. The 10 mm-high flange acted as a guide for reproducible root placement within the polyether mass.

The internal surfaces of the socket and the root were coated with polyether adhesive (Polyether Adhesive, 3M ESPE, St. Paul, MN, USA) according to the manufacturer's instructions. The polyether material was injected into the socket with slight excess. During injection, the dispenser tip was positioned at the bottom of the socket and kept immersed in the material while filling, to prevent air entrapment and ensure uniform layer formation. The tooth replica was fixed in the positioning device, which was then pressed onto the PMMA block until full contact was achieved. Setting of the elastic material was carried out under standard room conditions (approximately 22–24 °C, without controlled humidity). Afterwards, the positioning device was removed, and excess polyether was trimmed flush with the PMMA surface using a scalpel blade.

The prepared models were scanned. Lithium disilicate crowns (IPS e.max CAD, Ivoclar Vivadent, Schaan, Liechtenstein) were designed digitally, milled, and crystallized according to the manufacturer's instructions.

Before cementation, the intaglio surfaces of the lithium disilicate crowns were etched with 4.5% hydrofluoric acid for 20 s, rinsed with water, and treated with a universal adhesive (Single Bond Universal, 3M ESPE, St. Paul, MN, USA). The hybrid ceramic abutments were air-abraded with 50 µm aluminum oxide (Al_2O_3) particles at 2 bar pressure from a distance of approximately 10 mm for 10 s, then thoroughly rinsed with water and air-dried before applying the same adhesive system. A dual-cure resin cement (RelyX Ultimate, 3M ESPE, St. Paul, MN, USA) was applied to the internal surface of each crown, which was then seated under constant finger pressure. Excess cement was carefully removed, followed by light curing for 20 s from each surface, and final polymerization was allowed to occur chemically. Care was taken to prevent air entrapment during cement application to minimize potential damping variability associated with interfacial voids. The experimental model is presented in Figure 2.

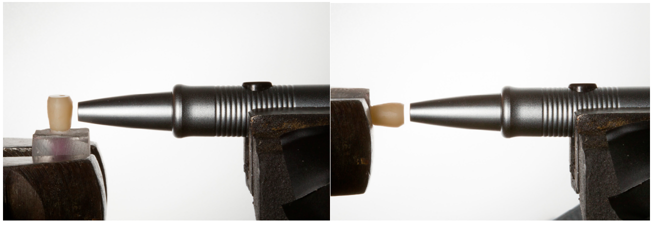
(a) (b) (c) (d)

Figure 2. (**a**) PMMA block with a milled socket and a tooth replica positioned using a 3D-printed positioning device; (**b**) Tooth replica inserted into the socket with a polyether layer simulating the PDL; (**c**) Tooth replica with a lithium disilicate crown cemented in place; (**d**) Cross-sectional view of the PDL-mimicking layer.

A total of 15 specimens were fabricated using the described CAD/CAM-based method. The specimens originated from the same experimental set used in our previous investigation on the effect of abutment rigidity on the wear resistance of lithium disilicate ceramics [15]. The number of specimens per group was originally determined through a power analysis (Statistica 13.3, StatSoft GmbH, Hamburg, Germany), using an independent samples t-test, assuming a 20% intergroup difference based on previously published data, with $\alpha = 0.05$ and a statistical power of 0.8.

The completed models were evaluated using the Periotest device. Measurements were performed horizontally from the buccal, palatal, mesial, and distal directions, and vertically from the occlusal surface, with three repetitions per direction. The average value was calculated for each plane.

During Periotest evaluation both the device and the tested specimens were secured in a metal vise to ensure reproducible alignment and eliminate operator-dependent variability. For horizontal measurements, the tapping rod was oriented perpendicular to the long axis of the tooth replica (Figure 3a), whereas for vertical measurements, the specimen was rotated by 90°, and the handpiece was positioned perpendicular to the occlusal surface (Figure 3b). The tip of the Periotest rod was maintained at a constant distance of 1.5 mm from the tested surface. The device was functionally tested prior to use according to the manufacturer's recommendations. Each measurement consisted of 16 impacts performed over 4 s, following the standard device settings. All measurements were conducted by the same trained operator to avoid inter-operator variability.



(**a**) (**b**)

Figure 3. Periotest measurement setup. (**a**) Horizontal orientation: the tapping rod positioned perpendicular to the long axis of the tooth replica. (**b**) Vertical orientation: the specimen rotated by 90°, with the tapping rod perpendicular to the occlusal surface.

Radiographic evaluation was performed using the paralleling technique, with radiographs obtained using a RXDC Hypersphere (Cefla s.c., Imola, Italy) and Rinn XCP-DS FIT positioners (Dentsply Sirona, Charlotte, NC, USA). Exposure parameters were set to 60 kV and 7 mA, with an exposure time of 0.16 s and a source-to-object distance of 10 cm. Images were obtained in both the sagittal and coronal planes. The periodontal space was measured at three points along the root and at the apex (Figure 4) using RadiAnt DICOM Viewer 2025.1 (Medixant, Poznań, Poland). All measurements were calibrated based on the actual specimen dimensions.

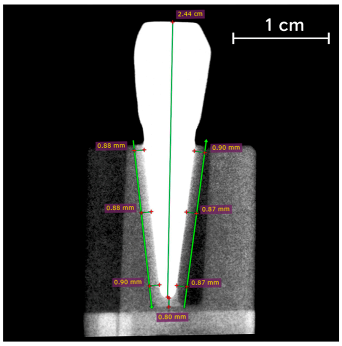


Figure 4. X-ray measurements of the PDL-simulating layer in the CAD/CAM-based experimental model.

2.2. *Statistical Analysis*

Statistical analysis was performed using IBM SPSS Statistics version 29.0 (IBM Corp., Armonk, NY, USA). The Shapiro–Wilk test was used to assess the normality of data distribution. Because the data were not normally distributed, the Kruskal–Wallis test was used to evaluate intergroup differences. Statistical significance was set at $\alpha = 0.05$. Effect size (ε^2) and 95% confidence intervals were calculated. Results are reported as descriptive and inferential statistics.

3. Results

Experimental Model Measurements

The mean PTV in the horizontal plane was 2.99 (SD = 0.92), while in the vertical direction it was −4.02 (SD = 0.56). Average values for each measurement direction are presented in Table 2 and Figure 5 The Kruskal–Wallis test revealed no statistically significant differences between horizontal directions ($p = 0.460$, H(3) = 2.588, $\varepsilon^2 = 0.00$). When the vertical direction was included, a significant overall difference was detected among all orientations (H(4) = 109.272, $p < 0.001$, $\varepsilon^2 = 0.48$). Dunn–Bonferroni post hoc analysis confirmed that vertical PTV values were significantly lower than those in each horizontal direction ($p < 0.001$). The results are presented in Table 2.

Table 2. Mean Periotest values in each direction.

Direction	Mean (PTV)	SD (PTV)	Standard Error (PTV)	95% CI (PTV)	Minimum (PTV)	Maximum (PTV)
Buccal	2.88	1.11	0.17	[2.54, 3.21]	1.00	4.90
Palatal	2.91	1.03	0.15	[2.60, 3.22]	1.10	4.90
Mesial	3.04	0.74	0.11	[2.82, 3.26]	1.70	4.50
Distal	3.14	0.75	0.11	[2.91, 3.37]	1.50	4.50
Overall horizontal	2.99	0.92	0.07	[2.86, 3.13]	1.00	4.90
Vertical	−4.02	0.56	0.08	[−4.19, −3.85]	−5.60	−3.30

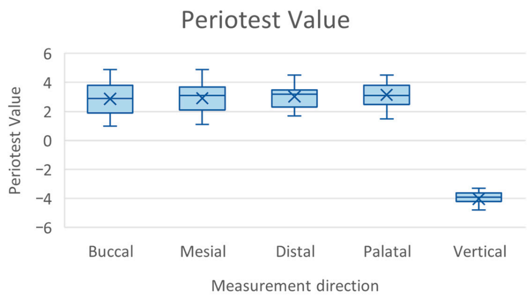


Figure 5. Mean Periotest values in each direction.

Measurements of polyether layer thickness simulating the PDL around the root in each model are shown in Table 3 and Figure 6. The mean layer thickness was 0.86 mm (SD = 0.052 mm).

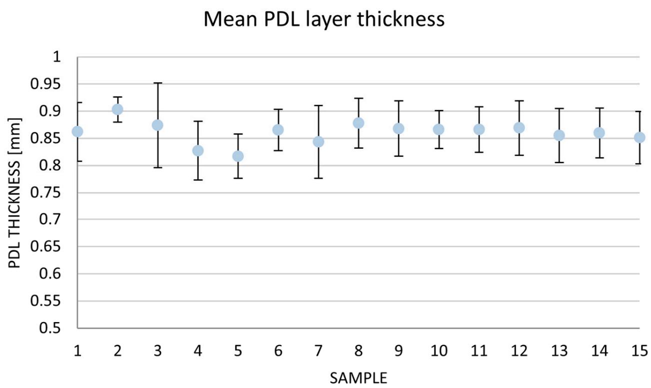


Figure 6. Mean PDL layer thickness.

Table 3. Mean PDL layer thickness.

Specimen	Mean (mm)	SD (mm)	Variance (mm^2)	95% CI (mm)	Minimum (mm)	Maximum (mm)
1	0.86	0.05	0.003	[0.83, 0.89]	0.73	0.92
2	0.90	0.02	0.001	[0.89, 0.92]	0.86	0.95
3	0.87	0.08	0.006	[0.83, 0.92]	0.72	0.99
4	0.83	0.05	0.003	[0.80, 0.86]	0.77	0.94
5	0.82	0.04	0.002	[0.79, 0.84]	0.73	0.86
6	0.87	0.04	0.001	[0.84, 0.89]	0.81	0.92
7	0.84	0.07	0.005	[0.80, 0.88]	0.73	0.95
8	0.88	0.05	0.002	[0.85, 0.90]	0.81	0.94
9	0.87	0.05	0.003	[0.84, 0.90]	0.79	0.95
10	0.87	0.04	0.001	[0.85, 0.89]	0.81	0.92
11	0.87	0.04	0.002	[0.84, 0.89]	0.80	0.92
12	0.87	0.05	0.003	[0.84, 0.90]	0.80	0.93
13	0.86	0.05	0.002	[0.83, 0.88]	0.77	0.93
14	0.86	0.05	0.002	[0.83, 0.89]	0.80	0.94
15	0.85	0.05	0.002	[0.82, 0.88]	0.77	0.93
Overall	0.86	0.05	0.003	[0.85, 0.87]	0.72	0.99

4. Discussion

4.1. The Influence of PDL Simulation on Mechanical Testing Outcomes

A focused review of the literature was performed to contextualize currently used methods of periodontal ligament simulation. The search strategy, selection criteria, and full summary of findings are presented in Appendix A. Previous studies have shown substantial heterogeneity in the methods used to simulate the PDL. This variation concerned not only the diversity of fabrication techniques and materials employed, but also significant inconsistencies in the documentation of these methods. Many studies lacked a detailed description of model preparation protocols [12,13,18,28–30], limiting reproducibility and complicating cross-study comparisons. The inclusion of our previous study [15] carries an inherent risk of bias; therefore, its findings were interpreted with caution to maintain objectivity in the comparative synthesis.

Only a limited number of studies in the available literature included direct comparisons between models with and without PDL simulation, showing that its inclusion significantly influenced the outcomes of selected types of in vitro tests—particularly those assessing fixed partial dentures (FPD) [9–11], post-and-core restorations [13,26], material wear [14,15], and fatigue performance [10,28]. The presence of an elastic layer affected both force distribution and the predominant failure mechanisms. Studies including control groups without simulated PDL demonstrated that the absence of a resilient suspension system resulted in significantly different outcomes.

In fracture resistance testing of single crowns, the influence of simulated PDL appears to be less pronounced, although the number of studies directly comparing models with and without PDL remains limited [16,17].

In contrast, tests evaluating the load-bearing capacity and fracture resistance of fixed partial dentures—both retentive and adhesive types—consistently show that the inclusion of a simulated PDL leads to significantly reduced structural strength. This finding has been reported by multiple authors. Rosentritt et al. demonstrated that simulating the PDL reduced the fracture resistance of all-ceramic FPDs by 40–70%. Moreover, the presence of an elastic foundation caused the fracture patterns to more closely resemble those observed under clinical conditions [10,11]. These results were further corroborated by Waldecker et al., who combined physical experimentation with finite element validation, demonstrating that the absence of an elastic PDL led to an overestimation of adhesive FPD strength by 50–95% [9]. This suggests that testing FPDs on rigid abutments without simulating the PDL yields unrealistically high strength values and produces fracture locations and patterns that deviate from those occurring in vivo. The inclusion of a resilient foundation modifies stress transmission pathways by allowing limited bending and torsional deformation of the supporting structures, thereby producing fracture patterns more consistent with clinical failures [10]. Consequently, drawing conclusions from in vitro tests using rigidly fixed abutments may result in clinical applications where the inherent weakness of the construction only becomes apparent upon failure in the oral environment.

In studies on post-and-core restorations, the presence of simulated PDL resulted in lower measured fracture resistance and different fracture patterns compared to specimens mounted in rigid substrates. Hayashi et al. observed that embedding teeth directly in acrylic resin created an artificial "ferrule effect," which led to overestimated fracture resistance and fracture patterns not representative of clinical conditions—predominantly affecting the coronal region rather than the root [13]. Soares et al. found that PDL simulation significantly influenced the type of observed fractures (with higher incidence of root fractures), but had less impact on the maximum load recorded [26]. Some studies, however, did not report significant differences in fracture resistance between models with and without PDL simulation [31,32]. Nevertheless, most authors emphasize that stress distribution within the root is altered by the presence of a simulated PDL and, therefore, recommend its inclusion in the mechanical testing of post-and-core systems [12,13,26,31,33–36].

Evidence related to the wear resistance of prosthetic materials in the presence of simulated PDL is limited. Rosentritt et al. experimentally demonstrated that implant-supported crowns may exhibit greater wear and cause more material loss on opposing teeth than crowns supported by teeth with simulated PDL [14]. The findings from our 2023 study are consistent with these observations [15]. A larger sample size enabled quantitative assessment, confirming that differences in material wear may be significant. The comparison included lithium disilicate crowns on tooth replicas with simulated PDL, crowns on implants, and pure material specimens embedded in acrylic resin. The volumetric material loss was measured at 0.107 mm^3, 0.166 mm^3, and 0.322 mm^3, respectively. These results indicate that prosthetic material wear is influenced by the damping properties of the PDL and likely also by other components of the reconstruction, such as the cement layer. Considering the role of the PDL is important in the design of prosthetic materials intended to biomechanically approximate implant-supported restorations to natural teeth. Accounting for this factor may help develop materials that compensate for the reduced energy dissipation associated with the rigid abutment represented by the implant [15]. Moreover, the inclusion of PDL simulation is essential when testing prosthetic materials on natural teeth, as it enables a more accurate replication of the physiological load distribution and tooth mobility, thereby allowing materials to better mimic the natural wear behavior of enamel.

In fatigue tests (thermocycling and mechanical loading, TCML), the presence of the PDL affected not only the total number of cycles to failure but also the type of damage observed. Aboushelib et al. reported that elastic fixation prevented the occurrence of cone cracks and concentric damage patterns typically associated with overly rigid models—an effect attributed to the realistic replication of clinical conditions [28]. Likewise, Rosentritt et al. demonstrated that PDL simulation introduced additional bending and torsional forces on prosthetic bridge structures, which contributed to the material aging effect observed during thermomechanical loading; these effects were less pronounced in rigid configurations [10]. In contrast, Nawafleh et al. found no statistically significant differences between models with and without PDL simulation for zirconia crowns [17]. Therefore, the aging effect appears to be more pronounced in the presence of an elastic foundation in FPDs, where it accelerates the material degradation. In the case of single crowns, some authors have suggested that PDL simulation may have a protective effect against extreme overload, although its impact on overall strength appears less evident. While many researchers investigating cyclic loading in restorations involving post-and-core systems have implemented PDL simulation, its influence in this context has not been directly compared between models with and without resilient fixation.

Regarding fracture resistance testing of crowns under static load, the number of studies comparing models with and without PDL simulation is limited. In the study by Preis et al., fracture resistance was tested in crowns placed on implants and on teeth with simulated PDL, with no statistically significant differences in material strength observed between groups [16]. Similarly, Nawafleh et al. reported no significant differences in cyclic load resistance or fracture strength when comparing zirconia crowns on models with and without PDL simulation [17]. These findings suggest that, for single crowns, the influence of PDL simulation on fracture resistance testing is minimal.

4.2. Methods for Simulating the PDL

The most commonly used method for simulating PDL involves manually creating space around the tooth root by dipping the root into molten wax. This technique was first described in 1998 [37]. After placing the tooth in a block and removing the wax, the resulting space was filled with an elastomeric material such as polyether or silicone. The primary advantage of this method is its simplicity; however, the main limitation is the lack of control over the uniformity of the simulated PDL layer. According to various authors, the thickness of the simulated PDL layer may range from 0.00 mm to 0.42 mm (with a planned thickness of 0.25 mm [38]), or between 0.3 mm and 0.7 mm as reported by other researchers [39]. Additionally, as noted by the authors themselves, this method is time-consuming and may lead to inconsistencies in specimen preparation [10].

Higher accuracy and continuity of the PDL layer have been achieved through the direct application of the elastic material onto the root surface. Al-Zahrani et al., by directly applying a latex separator, obtained a consistent and continuous PDL layer with a controlled thickness of 0.25 mm (SD = 0.02 mm), compared to an SD of 0.14 mm using the wax method [38]. Similar techniques using different materials were employed by other researchers [34,36,40–46], applying the elastic material directly to the tooth root. However, in most studies using this approach, there was a lack of data regarding the elastic properties of the materials used, and only one author performed functional validation of the models by measuring their mobility [43]. The main advantages of this approach include its precision and reproducibility. Potential limitations, however, involve the uncertain elastic properties of rubber-based insulating materials or technical elastomers intended for prosthetic models, as well as the time-consuming nature of manually applying multiple material layers when the desired PDL thickness exceeds that of a single application.

In recent years, models based on CAD/CAM technology have been developed, enabling precise control over the design of PDL dimensions in accordance with its expected mechanical characteristics [15,47]. Digital design and manufacturing automate the entire process, reduce production time, and ensure high reproducibility of experimental models. The protocol presented in this study allows for the use of previously validated elastomeric materials to replicate the elastic behavior of the PDL. Tooth replicas can be either fully digitally designed or based on scans of natural teeth. The limitation of this approach, however, is that the use of natural teeth would require the design and fabrication of a custom alveolar socket replica and corresponding positioners for each individual tooth, which significantly increases the time and resources needed to produce such an experimental model.

New approaches are also emerging that utilize 3D printing to fabricate the PDL layer, potentially offering a promising alternative to conventional elastomeric materials. However, these methods have not yet been validated in terms of their mechanical properties, applicability in strength or fatigue testing, nor have they been used in actual studies of prosthetic materials [48].

The most widely used technique for fabricating PDL-simulating models involves creating space around the tooth using an intermediate material (such as wax or foil) or by digitally designing the space for the PDL, followed by inserting the tooth replica into a socket filled with elastic material. Besides the method of creating the space itself, the process of placing the root into the socket also affects the precision and reproducibility of the model. In the vast majority of studies, this step was performed manually [1,4,11,14,16,17,26,32,33,35,44,49–58], which introduces the risk of misaligning the root axis and generating an uneven PDL layer in the final model. Several studies used positioning devices to ensure precise insertion of the tooth into the socket. These included silicone-based guides [38,41,59], gypsum [10], and 3D-printed positioners [15], the latter of which are described in detail in this article. The use of positioners during tooth insertion is critical for achieving a homogeneous elastic material layer around the root, thereby ensuring a reproducible experimental model. The direct application technique—where elastic material is applied directly onto the tooth root surface eliminates this technical issue.

4.3. Materials Used for PDL Simulation, PDL Layer Thickness, and Model Validation

The studies analyzed revealed a lack of a standardized protocol for selecting materials used in physical simulation of the PDL, resulting in a wide variety of approaches and limiting the comparability of results across investigations. The most commonly used category of materials consisted of elastomeric impression materials, with polyether (Impregum, 3M ESPE) being the most frequently applied [10–12,14–16,18,28,32,33,35,50,51,53,55–58,60]. Other elastomeric impression materials were also widely used, most commonly condensation silicones and polyvinyl siloxanes. Standard-viscosity formulations were employed in the majority of cases, although some studies utilized low-viscosity (light body) variants [4,13,17,30,47,49,52,54,61,62]. While often simplified in experimental models as linear-elastic, elastomeric impression materials actually exhibit complex viscoelastic and partially nonlinear stress–strain behavior, with polyether being among the stiffest formulations [63]. These materials are designed for high elastic recovery (low permanent set) and display limited strain tolerance. However, their mechanical response differs substantially from that of the natural PDL, which exhibits viscoelastic, anisotropic, and strain rate-dependent behavior [64]. Another significant category of materials comprised technical elastomers such as “gum resin” or “latex rubber milk” [34,36,40–42,45], as well as prosthetic insulators [38,43,44,46] and denture relining materials [31].

In a study by Soares et al., the effects of different elastomeric materials on tooth fracture resistance were compared. The results indicated that, while the differences between materials were relatively minor, the presence of the PDL-simulating layer itself had a significant impact on fracture resistance. The authors also reported that polyether exhibited mechanical properties most closely resembling the physiological characteristics of the natural PDL, particularly in terms of elasticity and its capacity to dampen dynamic load [26]. Given its favorable biomechanical behavior and extensive characterization in the literature, polyether appears to be the most suitable material for simulating the PDL in in vitro studies.

In contrast, Sterzenbach et al. evaluated model mobility using different materials —technical polyurethane (Anti-Rutschlack, Kaddi-Lack, Dortmund, Germany), polyether (Impregum Penta, 3M ESPE, Seefeld, Germany), and polyvinyl siloxane (Mollosil, DETAX, Ettlingen, Germany)—and reported significantly different outcomes. Under a perpendicular load of 30 N, the measured displacement values were 24 µm, 246 µm, and 210 µm, respectively. Corresponding Periotest values also varied notably: −5 to −3 for polyurethane, −1 to +5 for polyether, and +3 to +15 for polyvinyl siloxane [1]. These results indicate that polyurethane, which is frequently used in studies involving the direct application of a root-coating layer, may overly stiffen the model and fail to adequately replicate the PDL's elasticity. A further conclusion drawn from these findings is that elastomeric materials can differ significantly in their elastic and damping properties. This underscores the importance of functional validation of PDL-simulating models prior to their use in mechanical testing.

Most authors determined the thickness of the simulated PDL layer based on anatomical dimensions of the PDL, typically ranging from 0.2 to 0.3 mm. While these values correspond to the physiological width of PDL observed in histological studies, they may not accurately replicate its biomechanical behavior when elastomeric materials are used. Several experimental studies evaluating tooth mobility through functional testing have demonstrated that a polyether layer of this thickness is excessively rigid compared to the natural PDL. Rosentritt et al., using a universal testing machine, reported that the optimal thickness of the polyether layer should be approximately 1 mm [10]. In our own study, conducted with the Periotest device, we confirmed this characteristic and established that a polyether layer thickness of 0.85 mm is required for a single-rooted tooth to achieve physiological mobility and damping capacity [15]. Therefore, the selection of PDL layer thickness in experimental models should not be based solely on anatomical dimensions but rather on functional criteria—specifically the mechanical properties of the used material. Different elastomeric materials require different layer thicknesses to achieve comparable elastic characteristics. Consequently, the optimal width of simulated PDL must be defined individually for each experimental model.

The majority of studies did not include validation of the experimental models against the actual biomechanical properties of the PDL. Only a limited number of authors assessed tooth displacement using a universal testing machine [1,4,8,10] or an electronic displacement transducer [43], or evaluated vibration damping capacity using the Periotest device [1,15,49,61,62]. The lack of such measurements prevents objective assessment of the biomechanical fidelity of the models and substantially limits both the comparability of the findings and their clinical relevance.

Many authors adopted a simplified approach to the selection of materials for simulating the PDL, often basing their choice solely on the elastic modulus of the material. However, direct comparison of the elastic modulus of synthetic materials with the estimated value for the PDL constitutes a significant oversimplification, as the PDL is neither a homogeneous nor a linearly elastic structure [64]. As a biological tissue, the PDL exhibits anisotropic, nonlinear, and viscoelastic properties, which cannot be adequately described by a single numerical value [65]. Furthermore, most data on PDL elasticity is derived from indirect methods such as inverse finite element analysis (FEA), rather than from direct mechanical testing. Consequently, the elastic modulus of the PDL should be regarded as a context-dependent parameter, influenced by loading conditions, temperature, and measurement methodology [66].

Therefore, the selection of a PDL-analog material should not rely solely on its elastic modulus but should also incorporate functional validation within the context of the specific experimental setup. This may include assessments of tooth mobility under load (e.g., using a universal testing machine) or assessing damping capacity (e.g., using a Periotest device), in order to more accurately replicate the in vivo biomechanical behavior of the PDL. Ultimately, experimental models can only approximate the complex interactions among bone, the periodontal apparatus, dental hard tissues, and prosthetic components and should not be regarded as exact reproductions of clinical conditions.

4.4. Reporting Checklist for Studies Including PDL Simulation

Despite the growing number of in vitro studies incorporating periodontal ligament (PDL) simulation, there is currently no standardized framework for reporting experimental conditions. The absence of unified methodological reporting hinders cross-study comparison, reproducibility, and meta-analytical synthesis.

Based on the methodological considerations discussed above, a set of core parameters is proposed to serve as a minimum reporting checklist for studies involving physical PDL simulation. These parameters encompass essential aspects of model design, material selection, and validation procedures, which collectively determine the biomechanical fidelity of the experimental setup.

Recommended minimum reporting items include the following:

- PDL fabrication method: method of creating the PDL space and applying the analog (e.g., lost-wax technique, spacer foil, direct coating, or CAD/CAM-based digital offset).
- PDL analog: material type, brand, viscosity
- PDL layer thickness: target value, tolerance, and method of verification (e.g., μCT, optical measurement, or radiograph).
- Functional validation: validation method (e.g., Periotest, static deflection, or mobility assessment) and target range representing physiological tooth mobility.
- Positioning and alignment: description of root-axis alignment procedure, use of positioning devices or jigs.
- Tooth and substrate: type of tooth, number of roots, arch, and substrate material (e.g., natural tooth, PMMA, resin, gypsum).
- Socket material: composition, manufacturer, and method of cavity preparation.
- Cementation and restoration: type of luting agent, restorative material, and surface treatment.
- Aging and loading protocol: thermocycling, mechanical load magnitude, and number of cycles.
- Outcome measures: parameters evaluated (e.g., fracture load, load bearing capacity, wear).
- Uncertainty assessment: identification and quantification of potential error sources (e.g., layer thickness variability, positioning deviation, or material property variation); report standard deviation, confidence intervals, or coefficient of variation (CV) for key parameters.

These items are summarized in Supplementary Table S1, which may serve as a practical reference for future research aiming to standardize the reporting and validation of experimental PDL-simulation models.

4.5. A Novel CAD/CAM Model for PDL Simulation

The model presented in this study, based on CAD/CAM technology, demonstrated high geometric precision. The mean simulated thickness of the PDL layer was 0.86 mm (SD = 0.052 mm), closely matching the digitally designed target of 0.85 mm.

The experimental model was intentionally simplified, featuring a uniform PDL layer and an oval root cross-section, in contrast to the variable ligament thickness and irregular root morphology of natural teeth. This simplification was introduced to facilitate model fabrication and to enable precise, repeatable manufacturing through a standardized digital offset procedure. Despite these simplifications, the model accurately reproduced the global elastic response of the periodontal ligament relevant to mechanical testing applications. Future adaptations in different scenarios should account for specific tooth morphologies, e.g., multi-rooted teeth or variable bone levels, which may influence local stress transfer and mobility calibration.

The established 0.85 mm polyether layer thickness represented the optimal configuration for this specific material (Impregum Penta) and the geometry of a single-rooted maxillary premolar. However, the relationship between the PDL analog thickness and resulting damping behavior is both material- and geometry-dependent. Variations in tooth type, root morphology, or the elastic modulus of the selected material may require individual adjustment of the simulated PDL layer to achieve physiologic mobility. The target calibration range of 2–3 PTV used in this study corresponds to clinical values for natural maxillary teeth, as reported by Winkler et al. [27], where the overall average Periotest value was approximately 2.5, with a range of 2.1–3.1 PTV. For mandibular teeth, lower values (0.8–1.8 PTV) have been observed due to differences in supporting bone quality. Therefore, when developing analogous models, laboratories should calibrate their systems to achieve PTVs within the physiologic range of 1–3, adjusting the PDL layer thickness according to the material properties and tooth geometry.

Functional validation using the Periotest device, which quantifies damping capacity, produced values consistent with those of healthy periodontium. The mean PTV in the horizontal plane was 2.99, within the target range of 2.0–3.0. No statistically significant differences were found among the various horizontal directions ($p = 0.460$), confirming the uniformity of the elastic layer surrounding the roots. In contrast, measurements in the vertical orientation showed markedly higher damping, with a mean PTV of -4.02, compared to the horizontal average, indicating significant anisotropy in the model's mechanical response ($H(4) = 109.272$, $p < 0.001$, $\varepsilon^2 = 0.48$). This behavior corresponds closely with in vivo findings reported by Berthold et al., who observed a median PTV of 1.1 for maxillary canines in the horizontal plane (range: -2.6 to +5.9) and a median of -2.5 in the vertical plane (range: -5.8 to +10.7) [67].

The high reproducibility of the obtained results can be attributed to the application of CAD/CAM technology, which allows precise digital design and control of all model components. A key factor ensuring consistency of the PDL-simulating layer was the dedicated 3D-printed positioning device, which ensured uniform placement of the simulated layer around the tooth roots. This approach minimized inter-sample variability and allowed for a reduction in the required sample size for laboratory testing [15].

Moreover, the model offers advantages in terms of cost-effectiveness and reduced preparation time compared with conventional analog methods. A current limitation lies in the use of tooth replicas fabricated from materials different from natural dental tissues. Although this workflow can be adapted for use with extracted human or animal teeth, it would require the design and fabrication of customized models and positioning devices for each specimen.

5. Conclusions

Simulation of the PDL is a critical factor for in vitro studies assessing the mechanical behavior of prosthetic materials and structures. The presence or absence of appropriate PDL simulation in experimental models can substantially impact study outcomes. Despite increasing interest in this topic, significant methodological variability persists in the literature, limiting the comparability of results across studies.

Based on the development of the proposed CAD/CAM-based model, interpreted within the context of relevant literature, the following conclusions can be drawn:

1. The inclusion of a PDL analog is essential when evaluating mechanical properties such as the fracture resistance of fixed partial dentures, the strength of post-and-core restorations, and the wear resistance of prosthetic materials. Omission of this element may distort test outcomes, resulting in either overestimation or underestimation of the actual clinical performance of the material or the prosthetic structure.

2. A comprehensive description of methodologies and materials used in in vitro studies is critical to ensure reproducibility, enable meaningful comparisons across studies, and support the development of reliable and clinically relevant conclusions.
3. Experimental models incorporating PDL simulation should undergo functional validation in addition to geometric verification. This step is essential to confirm that the model accurately replicates the mechanical behavior of the natural PDL, thereby preventing biased or misleading results.
4. The width of the simulated PDL space in experimental models should not be determined solely based on anatomical dimensions. Accurate reproduction requires calibration through functional testing, considering the mechanical properties of the material used to mimic PDL resilience and the anatomical configuration of the tooth replica.
5. The CAD/CAM-based PDL simulation model proposed by the authors demonstrates high reproducibility, ease of fabrication, and accurate emulation of the elastic properties of the natural PDL. This model offers a viable alternative to conventional techniques and may be integrated into various experimental protocols.

Supplementary Materials: The following supporting information can be downloaded at https://www.mdpi.com/article/10.3390/jfb16120429/s1, Table S1. Minimum Reporting Checklist for Studies Including PDL Simulation; Table S2. Data Extraction Template for PDL Simulation Studies; Table S3. Studies Directly Assessing the Influence of PDL Simulation on Mechanical Test Outcomes.

Author Contributions: Conceptualization, P.K., A.M. and J.K.; methodology, P.K. and J.K.; software, P.K. and J.K.; validation, P.K., A.M. and J.K.; formal analysis, P.K. and A.M.; investigation, P.K. and J.K.; resources, P.K.; data curation, P.K. and J.K.; writing—original draft preparation, P.K.; writing—review and editing, A.M. and J.K.; visualization, P.K.; supervision, A.M.; project administration, P.K.; funding acquisition, P.K. All authors have read and agreed to the published version of the manuscript.

Funding: This research received no external funding.

Institutional Review Board Statement: Not applicable.

Informed Consent Statement: Not applicable.

Data Availability Statement: The datasets generated and analyzed during the current study, including the STL design files of the tooth, socket, and positioning device, as well as the raw Periotest measurement data used for calibration and functional validation, are openly available on Zenodo at https://doi.org/10.5281/zenodo.17392949 (accessed on 20 September 2025).

Acknowledgments: The authors would like to express their sincere gratitude to Camillo D'Arcangelo for his inspiration to explore this topic and his valuable contribution to the conceptualization of the experimental model. The authors express their gratitude to MaxLab Laboratorium Protetyczne and RS-TEAM S.C. for their support in fabricating the experimental models. Grateful acknowledgments are also extended to Maciej Kalinowski for his assistance in creating the 3D models. The authors acknowledge the use of OpenAI's ChatGPT (version GPT-5, OpenAI, San Francisco, CA, USA) to assist in language editing and stylistic refinement of the manuscript. All intellectual content, data interpretation, and conclusions were generated exclusively by the authors.

Conflicts of Interest: The authors declare no conflicts of interest.

Abbreviations

The following abbreviations are used in this manuscript:

CAD/CAM	Computer-Aided Design/Computer-Aided Manufacturing
PDL	Periodontal Ligament
PMMA	Polymethyl methacrylate
PTV	Periotest Value
FPD	Fixed Partial Denture
FEA	Finite Element Analysis
TCML	Thermocycling and Mechanical Loading

Appendix A

Appendix A.1. Literature Review Methodology

Appendix A.1.1. Search Strategy

A comprehensive literature search was conducted in April 2025 using MEDLINE via PubMed and Google Scholar. Google Scholar was searched to capture grey literature and studies not indexed in MEDLINE. Only studies published in English were considered, which may introduce a potential language bias. No review protocol was pre-registered in PROSPERO or OSF, and, consistent with the scoping review design, no formal quality assessment tool was applied, in line with its exploratory purpose. This review was conducted in accordance with the PRISMA Extension for Scoping Reviews (PRISMA-ScR) guidelines [68]. The objective was to identify studies focusing on the physical simulation of the PDL within in vitro research settings. Two independent reviewers (P.K. and J.K.) performed the search using the following Boolean string:

(periodontal ligament simulation OR PDL simulation OR artificial periodontal ligament OR artificial PDL OR artificial periodontium OR periodontium simulation) AND (prosthetic materials OR resin OR composite OR porcelain OR ceramic OR zirconia OR lithium disilicate OR enamel OR dentin) AND (chewing simulation OR chewing simulator OR fatigue OR wear OR mechanical testing OR mechanical properties OR fracture resistance OR fatigue strength).

Appendix A.1.2. Eligibility Criteria

Studies were included if they met the following criteria:

- Published between 2005 and 2025
- Written in English
- Peer-reviewed original research
- Use of physical models simulating the PDL

The 20-year publication window (2005–2025) was chosen to provide a balanced overview of contemporary research while excluding older studies no longer representative of current laboratory methods and materials.

Studies based exclusively on computational simulations (e.g., finite element analysis) or those that did not employ any form of PDL simulation were excluded.

Appendix A.1.3. Selection Process

Initial screening was conducted based on titles and abstracts. Articles meeting the inclusion criteria were subjected to full-text review. Additionally, backward citation tracking was performed to identify further relevant studies.

Study selection was carried out independently by two reviewers (P.K. and J.K.). Any disagreements regarding article inclusion were resolved through discussion with a third reviewer (A.M.), who assessed the full text prior to making the final decision. The study selection process is presented in Figure A1.

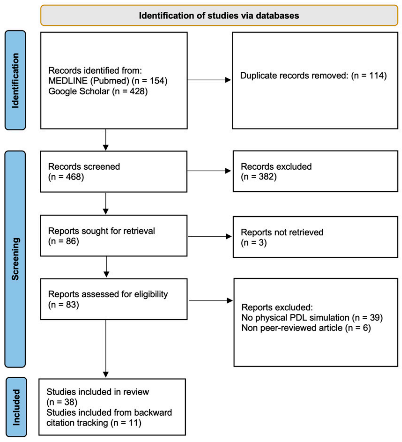


Figure A1. PRISMA 2020 flow diagram of the article selection process.

Appendix A.1.4. Data Extraction

The following parameters were extracted from each included study:

- Bibliographic details (authors, year of publication)
- Thickness of the simulated PDL layer
- Material used for PDL simulation
- Material of the abutment tooth
- Technique used for model fabrication
- Tested mechanical parameters
- Type of intervention tested
- Validation of the PDL model (if applicable)
- Assessment of the impact of PDL simulation on study outcomes
- Correlation of findings with in vivo data (if reported)

For clarity, validation procedures were classified as functional—when biomechanical behavior (mobility, damping, or load response) was experimentally verified—or geometric—when dimensional accuracy of the simulated PDL layer was confirmed using imaging or direct measurement. These distinctions are reflected in Appendix A, Table A1.

A data extraction template summarizing all recorded variables (e.g., PDL material, thickness, validation type, test parameters) is provided as Supplementary Table S2.

Appendix A.2

Table A1. Summary of included studies evaluating PDL simulation in in vitro models.

Study (Authors, Year)	Layer Thickness	PDL Simulation Material	Tooth Material	Technique for Artificial PDL	Experimental Model Validation	Parameters Tested	Treatment Type	Influence of PDL on Results	In Vivo Measurements Correlation
Aboushelib (2013) [28]	0.6 mm	Polyether impression material (Impregum; 3M ESPE, St. Paul, MN, USA)	Composite (Z250, 3M ESPE, St. Paul, MN, USA)	Polyether injection into the resin socket—technique not specified	Not specified	Cyclic loading resistance	Zirconia single crown	Lower peak stress in tested materials (lack of cone cracks typical for overloading in the testing environment)	Comparable survival rate
Alqarni et al. (2024) [48]	0.2 mm	3D printing resin—Flexible resin V2 (Formlabs Inc., Somerville, MA, USA)	Human teeth	PDL constructed using CAD/CAM process: root surface scanned and digitally offset 0.2 mm; PDL 3D-printed in flexible resin and fitted to the root, alveolar socket not mentioned	Not specified	Feasibility of digital workflow to fabricate PDL	None	Not specified	Not specified
AlZahrani & Richards (2018) [38]	0.25 mm	Latex rubber die-spacer (Rubber-Sep, Kerr Dental, Brea, CA, USA)—direct technique; light-body silicone impression material (Imprint 4, 3M ESPE, St. Paul, MN, USA)—wax technique	Human teeth	20 coats of paint-on latex rubber applied directly to the root (12 µm each) using a brush, tooth attached to a vertical rod with sticky wax to maintain position, embedded in acrylic resin (ProBase Cold, Ivoclar Vivadent, Schaan, Liechtenstein), comparison with the wax technique with teeth inserted with a silicone positioner	Geometric: Micro-CT—layer thickness measurement	Reproducibility and uniformity of PDL layer thickness (micro-CT analysis)	None	Not applicable	Not applicable
Berthold et al. (2011a) [61]	Not specified	Silicone (not specified) or silicone with rubber foam (not specified)	Stainless steel	Custom aluminum jaw model with alveolar sockets, silicone (not specified) or silicon with rubber foam (not specified) for increased mobility, apical screw for mobility adjustment	Functional: Universal testing machine—deflection measurement; Periotest	Tooth mobility: static (universal testing machine) and dynamic (Periotest)	Splint	Not applicable	Comparable damping capacity (Periotest)
Berthold et al. (2011b) [62]	Not specified	Silicone (not specified) or silicone with rubber foam (not specified)	Stainless steel	Custom aluminum jaw model with alveolar sockets, silicone (not specified) or silicon with rubber foam (not specified) for increased mobility, apical screw for mobility adjustment	Functional: Periotest	Different splint designs rigidity	Splint	Not applicable	Not specified
Boeckler et al. (2008) [49]	0.25 mm	Silicone impression material (Flexistone, Detax, Ettlingen, Germany)	Human teeth	Teeth coated with a wax sheet and embedded in an epoxy resin (Technovit 5000, Heraeus Kulzer, Hanau, Germany), wax removed, socket filled with silicone (Flexistone, Detax, Ettlingen, Germany), specimens reinserted manually	Functional: Periotest	Crowns marginal gap	Tooth-implant supported fixed partial denture	Not applicable	Not specified
Bömicke et al. (2017) [59]	Not specified	Heat-shrink tubing (Protect, Bahag AG, Mannheim, Germany), Light body polyvinylsiloxane impression material (Flexitime Correct Flow, Heraeus Kulzer GmbH, Hanau, Germany) in apical area	Cobalt chromium alloy	Metal abutments with roots covered with heat-shrink tubing and apically sealed with polyvinylsiloxane impression material; embedded in acrylic resin (Technovit 4071; Heraeus Kulzer GmbH, Hanau, Germany) in aluminum blocks, teeth inserted using silicone positioners	Not specified	Load-bearing capacity, fracture patterns under dynamic and static loading	Zirconia and metal-ceramic inlay-retained and wing-retained resin bonded fixed partial dentures	Not specified	Similar failure modes as in clinical conditions
Bruschi-Alonso et al. (2010) [50]	0.2–0.3 mm	Polyether impression material (Impregum; 3M ESPE St. Paul, MN, USA)	Bovine teeth	Roots dipped in melted wax, afterwards embedded in acrylic resin (Vipi Flash; Vipi Ind. Com., Pirassununguna, Brazil), wax removed, socket filled with polyether (Impregum), specimens reinserted manually	Not specified	Impact strength (fracture resistance at high speed impact)	Tooth fragment reattachment	Not specified	Not specified
Clausen et al. (2010) [40]	Not specified	Other elastomer/technical—gum resin (Anti-Rutsch-Lack, Wenko-Wenselaar GmbH, Hilden, Germany)	Human teeth	Roots coated directly with gum resin layer and embedded in acrylic resin (Technovit 4000, Heraeus Kulzer, Hanau,, Germany), roots secured in resin by 0.9 mm stainless steel bar in apical third of the root	Not specified	Survival under cyclic loading; fracture resistance	Ceramic overlay	Not specified	Not specified
Clavijo et al. (2009) [51]	0.2–0.3 mm	Polyether impression material (Impregum; 3M ESPE St. Paul, MN, USA)	Bovine teeth	Roots dipped in melted wax, afterwards embedded in polystyrene resin (Cristal, Araquímica, Araçariguama, Brazil), wax removed, socket was filled with polyether (Impregum), specimens inserted manually	Not specified	Root fracture resistance	Intraradicular post	Not specified	Comparable fracture resistance
Dulaimi & Al-Hashimi (2005) [52]	Not specified	Silicone impression material (DoriDent, Dr Hirschberg GmbH, Vienna, Austria)	Human teeth	Roots wrapped in lead foil embedded in acrylic resin (Medicus Cold Cure, DMP Ltd., Attiki, Greece), lead foil removed and silicone material painted on the root, specimens inserted in the socket manually	Not specified	Endodontic spreader load and penetration depth	Endodontic treatment	Not specified	Not specified

Table A1. *Cont.*

Study (Authors, Year)	Layer Thickness	PDL Simulation Material	Tooth Material	Technique for Artificial PDL	Experimental Model Validation	Parameters Tested	Treatment Type	Influence of PDL on Results	In Vivo Measurements Correlation
Fulde et al. (2025) [41]	0.2 mm	Other elastomer/technical—gum resin (Anti-Rutsch-Lack, Wenko-Wenselaar, Hilden, Germany)	Human teeth	Roots coated directly with 0.2 mm gum resin (dipping once and removing excess in apical portion), inserted into wax, 0.9 mm metal bar placed through the root to prevent rotation, specimens embedded in acrylic resin (Technovit 4000, Kulzer, Hanau, Germany), specimens reinserted using silicone positioner	Not specified	Survival after thermocycling and mechanical loading, fracture resistance under vertical load, failure mode, wear behaviour (chewing simulation)	Zirconia overlay	Not specified	Not specified
González-Lluch et al. (2016) [31]	0.2 mm	Polyethyl methacrylate denture relining material (Visco-gel, Dentsply Detrey GmbH, Konstanz, Germany)	Human teeth	Roots coated with Visco-gel layer using a brush, inserted into acrylic resin blocks (Vertex Self-Curing Liquid, Zeist, The Netherlands)	Not specified	Fracture resistance and failure modes: teeth with vs. without PDL simulation	Fiber post retained restoration, single crown/no crown	No significant difference in fracture load between PDL vs. no-PDL (p = 0.185)—median (538 N vs. 395 N)—limited number of specimens	Not specified
Hasna et al. (2022) [29]	Not specified	Red wax (not specified)	Human teeth	Roots coated with red wax (not specified), embedded in acrylic resin blocks (JET, Artigos Odontológicos Clássico Ltda., São Paulo, Brazil)—technique not specified	Not specified	Deflection, fracture resistance after artificial aging	Endodontic access restoration	Not specified	Not specified
Hayashi et al. (2006) [13]	0.2 mm	Polyvinylsiloxane impression material (Duplicone, Shofu, Kyoto, Japan)	Human teeth	Roots coated with polyvinylsiloxane, embedded in acrylic resin (Uni-Fast II, GC, Tokyo, Japan)—technique not specified	Not specified	Fracture resistance	Intraradicular post and crown	Without simulated PDL in the preliminary study, fracture resistance was approx. twice as high, acrylic resin in specimens without PDL acted as a ferrule	Not specified
Ille et al. (2023) [30]	Not specified	Polyvinylsiloxane impression material (Elite HD, Zhermack, Badia Polesine, Italy)	Human teeth	Teeth inserted in acrylic cylinder (not specified), apical part of the root embedded in polyvinylsiloxane low consistency impression material (not specified)—technique not specified	Not specified	Compressive strength	Ceramic/composite overlays	Not specified	Not specified
Jalalian & Mirzaei (2009) [33]	0.2 mm	Polyether impression material (Impregum, 3M ESPE, St. Paul, MN, USA)	Human teeth	Roots wrapped in 0.2 mm aluminum foil, embedded in acrylic resin; foil removed, sockets filled with Impregum, specimens repositioned manually	Not specified	Compressive strength	Intraradicular fiber post	Not specified	Not specified
Kolbeck et al. (2008) [18]	Not specified	Polyether impression material (Impregum, 3M ESPE, Seefeld, Germany)	Human teeth	Roots coated with polyether, embedded in resin blocks (Palapress Vario: Heraeus Kulzer, Hanau, Germany)—technique not specified	Not specified	Fracture resistance after chewing simulation	Tooth-tooth and implant-tooth fixed partial dentures	Not applicable	Not specified
Kosewski et al. (2023) [15]	0.85 mm	Polyether impression material (Impregum Penta; 3M ESPE, St. Paul, MN, USA)	Hybrid ceramic (Ambarino High Class, Creamed GmbH & Co., Marburg, Germany)	Teeth and bone constructed using CAD/CAM process: PMMA cubes with milled alveolar sockets digitally enlarged, teeth milled based on digital project, sockets filled with polyether material, specimens inserted in the socket with 3D-printed positioner	Functional: Periotest	Crown material wear	Single ceramic crown	Lower antagonist wear (0.172 mm vs. 0.220 mm), wear depth (0.098 mm vs. 0.186 mm), volume loss (0.107 mm^3 vs. 0.322 mm^3)	Not specified
Krummel et al. (2019) [42]	0.25 mm	Other elastomer/technical—gum resin (Anti-Rutsch-Lack, Wenko-Wenselaar GmbH, Hilden, Germany)	Human teeth	Roots coated directly with 0.25 mm gum resin layer and embedded in acrylic resin (Technovit 4000, Heraeus Kulzer, Hanau, Germany), roots secured in resin by 0.9 mm stainless steel bar in apical third of the root	Not specified	Survival rate after dynamic loading; fracture resistance; failure mode analysis	Ceramic overlay	Not specified	Not specified
Marchionatti et al. (2014) [32]	0.3 mm	Polyether impression material (Impregum Soft Medium Body, 3M ESPE, St. Paul, MN, USA) vs. silicone impression material (Express Medium Body, 3M ESPE, St. Paul, MN, USA)	Bovine teeth	Roots dipped in melted wax, afterwards embedded in acrylic resin (Dencrilay, Dencril, Caieiras, Brazil), wax removed, socket filled with simulation material, specimens reinserted manually	Not specified	Fiber posts bond strength, roots fracture resistance	Intraradicular fiber post	No effect	Not specified
Nawafleh et al. (2020) [17]	0.3 mm	Light body elastomeric impression material (3M ESPE, St. Paul, MN, USA)—not specified	Epoxy resin (Exakto-Form, Bredent, Senden, Germany)	Wax layer added to the roots, embedded in in acrylic resin, wax substituted with elastomeric impression material, specimens reinserted manually	Not specified	Fatigue survival and fracture resistance	Single zirconia crown	No effect	Not specified
Oliveira et al. (2022) [53]	Not specified	Polyether impression material (Impregum; 3M ESPE, St. Paul, MN, USA)	Polystyrene resin (Cristal, Araquímica, Araçariguama, Brazil)	Alveolus in acrylic model adapted manually, polyether injected in the socket, specimens reinserted manually	Not specified	Proximal contact force	Direct restoration	Not applicable	Not specified
Pişkin et al. (2008) [54]	Not specified	Light body silicone impression material (Speedex light body, Coltene/Whaledent, Altstätten, Switzerland)	Human teeth	Tooth wrapped in a single layer of aluminium foil, inserted in polystyrene resin, after aluminium foil removal, silicone material injected in the sockets, specimens reinserted manually	Not specified	Fracture resistance	Endodontic treatment	Not specified	Not specified

Table A1. *Cont.*

Study (Authors, Year)	Layer Thickness	PDL Simulation Material	Tooth Material	Technique for Artificial PDL	Experimental Model Validation	Parameters Tested	Treatment Type	Influence of PDL on Results	In Vivo Measurements Correlation
Preis et al. (2015) [55]	1 mm	Polyether impression material (Impregum, 3M Espe, Seefeld, Germany)	Human teeth	Roots dipped in melted wax, embedded in resin blocks (Palapress Vario, Kulzer, Hanau, Germany), wax removed, socket filled with polyether (Impregum), specimens reinserted manually	Not specified	Fracture resistance and marginal adaptation after thermocycling and mechanical loading under different cementation types	Monolithic zirconia-reinforced lithium silicate single crowns	Not specified	Not specified
Preis et al. (2018) [16]	1 mm	Polyether impression material (Impregum, 3M, Seefeld, Germany)	Human teeth	Roots dipped in melted wax, embedded in resin blocks (Palapress Vario, Kulzer, Hanau, Germany), wax removed, socket filled with polyether (Impregum), specimens reinserted manually	Not specified	Survival after thermocycling and mechanical loading, fracture resistance, effect of restoration type (tooth/implant), polishing vs. glazing	Lithium aluminosilicate single crowns	No effect	Not specified
Puschmann et al. (2009) [43]	0.2 mm	Latex rubber die-spacer (Erkoskin, Erkodent, Pfalzgrafenweiler, Germany)	Cobalt chromium alloy (Wironit; Bego, Bremen, Germany)	Roots of metal teeth coated directly with a 0.2 mm silicone layer; roots embedded in low-melting-point alloy (MCP 70) using a duplication form	Functional: Mobility checked with an electronical displacement transducer (W1T3; HBM, Darmstadt, Germany)—displacement of 50 µm under 20 N of horizontal load	Load-bearing capacity with and without fatigue loading; fracture mode analysis	Ceramic fixed partial denture	Not specified	Not specified
Rathke et al. (2022) [34]	0.25 mm	Other elastomer/technical—latex rubber milk (Suter Kunststoffe, Fraubrunnen, Switzerland)	Human teeth	Roots coated directly with one layer of latex rubber milk (Suter Kunststoffe, Switzerland) and embedded in acrylic resin (Technovit 4071, Heraeus Kulzer, Hanau, Germany)	Not specified	Root fracture resistance and crack formation after chewing simulation and static loading	Intraradicular glass fiber and metal posts	Not specified	Not specified
Ribeiro et al. (2023) [12]	0.3 mm	Polyether impression material (Impregum F, 3M Oral Care, St. Paul, MN, USA)	Bovine teeth	Roots coated with a 0.3 mm polyether layer and embedded in polystyrene resin (Cristal, Araquímica, Araçariguama, Brazil), technique not specified	Not specified	Survival rate and failure mode under cyclic loading	Intraradicular glass fiber post and direct restoration	Different results from studies without PDL simulation	Comparable differences between survival rates of teeth restored with fiber posts vs. no posts, and different fracture modes than in clinical studies
Rosentritt et al. (2006) [11]	1 mm	Polyether impression material (Impregum, 3M ESPE, Seefeld, Germany)	Liquid crystal polymer (LCP) and human teeth	Roots dipped in melted wax, afterwards embedded in PMMA resin (not specified), wax removed, socket filled with polyether (Impregum), specimens reinserted manually	Not specified	Fracture resistance of all-ceramic fixed partial dentures after thermocycling and mechanical loading	Ceramic fixed partial denture	Fracture resistance reduced by up to 70%; omission of PDL simulation leads to overestimation of fracture resistance	Not specified
Rosentritt et al. (2011) [10]	0.75 mm	Polyether impression matrial (Impregum, 3M ESPE, Seefeld, Germany)	Human teeth	Roots dipped in melted wax, embedded in PMMA (Palapress Vario, Kulzer, Hanau, Germany), wax removed, sockets filled with polyether (Impregum), specimens reinserted using repositioning cast	Functional/Geometric: Universal testing machine—deflection measurement; PDL thickness measured after slicing the specimens	Fracture resistance of three-unit fixed partial dentures after thermocycling and mechanical loading	Ceramic fixed partial denture	Fracture force reduced by 40–50% with PDL; no PDL led to overestimation; PDL presence seemed to be responsible for the ageing effect during TCML	Similar modes of failure as in clinical conditions
Rosentritt et al. (2020) [14]	1 mm	Polyether impression material (Impregum, 3M ESPE, Seefeld, Germany)	Human teeth	Roots dipped in melted wax, embedded in resin blocks (Palapress Vario, Kulzer, Hanau, Germany), wax removed, socket filled with polyether (Impregum), specimens reinserted manually	Not specified	Number and depth of wear traces	Zirconia/ceramic single crown	Crown on implants displayed increased antagonistic wear compared to teeth with PDL	Wear patterns similar to clinical findings
Sarafidou et al. (2012) [44]	Not specified	Latex rubber die-spacer (Erkoskin, Erkodent, Pfalzgrafenweiler, Germany)	Polyurethane resin (PUR, Alpha-Die-Top, Schütz Dental, Rosbach vor der Höhe, Germany)	Roots coated directly with latex layer (Erkoskin); embedded manually in polyurethane base	Not specified	Load-bearing capacity after artificial ageing (thermocycling and mechanical loading); fracture mode analysis	Fixed partial denture	not specified	not specified
Sasse et al. (2015) [45]	Not specified directly, reference to previous studies (0.2 mm)	Other elastomer/technical—gum resin (Anti-Rutsch-Lack, Wenko-Wenselaar GmbH, Hilden, Germany)	Human teeth	Roots coated directly with gum resin layer and embedded in acrylic resin (Technovit 4000, Heraeus Kulzer, Hanau, Germany), roots secured in resin by 0.9 mm stainless steel bar in apical third of the root	Not specified	Survival rate after dynamic loading; fracture resistance; failure mode analysis	Ceramic overlay	Not specified	Not specified

Table A1. *Cont.*

Study (Authors, Year)	Layer Thickness	PDL Simulation Material	Tooth Material	Technique for Artificial PDL	Experimental Model Validation	Parameters Tested	Treatment Type	Influence of PDL on Results	In Vivo Measurements Correlation
Sivieri-Araujo et al. (2015) [35]	0.25–0.3 mm	Polyether impression material (Impregum Soft, 3M ESPE AG, Seefeld, Germany)	Bovine teeth	Roots dipped in melted wax, embedded in polystyrene resin (not specified), wax removed, socket filled with polyether (Impregum), specimens reinserted manually	Not specified	Fracture resistance	Intraradicular post	Not specified	Not specified
Soares et al. (2005) [26]	0.2–0.3 mm	Polyether impression material (Impregum F, 3M-ESPE, Seefeld, Germany)/Polysulfide (Permlastic, Kerr, Brea, CA, USA)/Polyurethane elastomer (Ultra Flex, Solplas, Santo André, Brazil)	Bovine teeth	Roots dipped in melted wax, embedded in self-cured acrylic resin (Jet Clássico, Artigos Odontológicos Clássico Ltda., São Paulo, Brazil)/polystyrene resin (Cristal, Araquímica, Araçariguama, Brazil), wax removed, sockets filled with simulation material, specimens reinserted manually	Not specified	Fracture resistance and fracture modes	None	Presence of PDL simulation significantly altered fracture modes (more root fractures); absence of PDL caused concentration of fractures at coronal region	Fracture modes with PDL simulation similar to clinical fracture patterns
Soares et al. (2011) [56]	0.2–0.3 mm	Polyether impression material (Impregum F, 3M ESPE, St. Paul, MN, USA)	Human teeth	Roots dipped in melted wax, embedded in wax model of the mandible, wax from roots eliminated, wax mandible duplicated in polystyrene resin (Aerojet, Aerojet Fiberglass, São Paulo, Brazil), socket filled with polyether (Impregum), specimens reinserted manually	Not specified	Bone strain under different conditions of bone loss and periodontal splinting	Splint	Presence of artificial PDL allowed more realistic strain distribution	Strain values comparable to cadaveric bone measurements
Souza et al. (2014) [57]	0.2–0.3 mm	Polyether impression material (Impregum Soft, 3M ESPE, Seefeld, Germany)	Bovine teeth	Roots dipped in melted wax, afterwards embedded in polystyrene resin (Cristal, Araquímica, Araçariguama, Brazil), wax removed, socket filled with polyether (Impregum), specimens reinserted manually	Not specified	Fracture resistance	Endodontic irrigation	Not specified	Not specified
Sterzenbach et al. (2011) [1]	0.3 mm	Polyurethane (Anti-Rutschlack, Kaddi-Lack, Dortmund, Germany); Polyether (Impregum Penta, 3M ESPE, Seefeld, Germany); Polyvinylosiloxane (Mollosil, DETAX, Ettlingen, Germany)	Human teeth	Roots coated with 0.3 mm wax, embedded in acrylic resin (Technovit 4004), wax removed, sockets filled with simulation material; specimens reinserted manually	Functional: Periotest; universal testing machine—deflection measurement	Tooth deflection (mobility) under axial and perpendicular loading; damping capacity—Periotest values	None	Not applicable	Mobility similar to clinical conditions
Sterzenbach et al. (2012) [36]	0.1 mm	Other elastomer/technical—gum resin (Anti-Rutsch-Lack, Wenko-Wenselaar GmbH, Hilden, Germany)	Human teeth	Roots coated directly with 0.1 mm silicone (Anti-Rutsch-Lack) and embedded in autopolymerizing acrylic resin (Technovit 4000, Heraeus Kulzer, Hanau, Germany); long axis angled at 135° probably with an index—technique not specified directly, reference to previous study	Not specified	Maximum load capability after thermocycling and mechanical loading; fracture patterns depending on restorative stage	Post-endodontic restorations: fiber post vs. post and core vs. post, core, ceramic crown	Not specified	Not specified
Tanomaru-Filho et al. (2014) [58]	0.25 mm	Polyether impression material (Impregum Soft, 3M ESPE AG, Seefeld, Germany)	Bovine teeth	Roots dipped in melted wax, embedded in polystyrene resin (Cristal, Araquímica, Araçariguama, Brazil), wax removed, sockets filled with polyether (Impregum); specimens reinserted manually	Not specified	Fracture resistance	Intraradicular post/endodontic filling	not specified	Not specified
Waldecker et al. (2019) [9]	Not specified	Heat-shrink tubing (Protect, Bahag AG, Mannheim, Germany), low-viscosity polyvinylsiloxane impression material (Flexitime Correct Flow; Heraeus Kulzer GmbH, Hanau, Germany) in apical area	Cobalt-chromium alloy	Abutment roots covered with heat-shrink tubing sealed apically with polyvinyl siloxane; embedded in aluminum blocks with acrylic resin (Technovit 4071; Heraeus Kulzer GmbH, Hanau, Germany)	Functional: Model deflection measured under load of 100 N (technique not specified)	Validation of finite elements analysis model; stress distribution and fracture resistance under axial and oblique loading	Zirconia inlay-retained fixed partial denture	Absence of simulated resilience led to overestimation of fracture resistance by 50–95%	Good correlation with clinical behaviour
Weigl et al. (2018) [60]	1 mm	Polyether impression material (Impregum, 3M Oral Care, Seefeld, Germany)	Hybrid ceramic (Ambarino High Class, Creamed GmbH & Co., Marburg, Germany)	Roots dipped in wax, embedded in resin blocks (Palapress Vario, Heraeus-Kulzer, Hanau, Germany), sockets filled with polyether—technique not specified in detail, reference to previous studies	Not specified	Survival under thermocycling and mechanical loading, fracture resistance, fracture mode, influence of crown thickness and cement type	Zirconia single crown	Not specified	Not specified
Wong & Botelho (2014) [8]	Not specified	Orthodontic elastic rings (Z-pak Elastics, Ormco Corp., Brea, CA, USA)	Stainless steel	Two or more elastic orthodontic rings placed apically and circumferentially around the tooth analog, rings placed in grooves in the root surfaces, specimens reinserted in metal base holders	Functional: Universal testing machine—deflection measurement	Fatigue bond strength under cyclic loading	Fixed-fixed/cantilever resin-bonded fixed partial denture	Not applicable	More mobility of the roots than in the clinical situation (550 µm vs. 25–100 µm)
Yeslam et al. (2023) [46]	0.55 mm	Latex rubber die-spacer (Erkoskin, Erich Kopp GmbH, Pfalzgrafenweiler, Germany)	Die resin (Mirapont, Hager & Werken GmbH & Co. KG, Germany)	Roots painted directly with liquid latex (Erkoskin), embedded in die resin (Mirapont, Hager & Werken GmbH & Co. KG, Duisburg, Germany), manual insertion into the resin	Not specified	Fracture resistance	Hybrid ceramic single crown	Not specified	Fracture modes different than those in clinical conditions
Zhang et al. (2016) [69]	0.3 mm	Polyvinylsiloxane impression material (Imprint II, 3M ESPE, St. Paul, MN, USA)	Composite resin (Filtek P60, 3M ESPE, St. Paul, MN, USA)	Roots coated with 0.3 mm polyvinylsiloxane and embedded in PMMA support base—technique not specified	Not specified	Validation of finite elements analysis model; crack initiation and propagation under loading	Zirconia inlay and onlay fixed partial denture	Not specified	Not specified

Table A1. *Cont.*

Study (Authors, Year)	Layer Thickness	PDL Simulation Material	Tooth Material	Technique for Artificial PDL	Experimental Model Validation	Parameters Tested	Treatment Type	Influence of PDL on Results	In Vivo Measurements Correlation
Zhu et al. (2015) [4]	0.2–0.3 mm	Silicone impression material (Honigum-Mixstar Mono, DMG, Hamburg, Germany)	Human teeth	Roots dipped in melted wax, afterwards embedded in epoxy resin (Phoenix; Blue-star Manufacturing, Wuxi, China), wax removed, socket filled with silicone, specimens reinserted manually	Functional: Universal testing machine—compression testing—elastic modulus evaluation	Cyclic loading resistance	Vertical root fracture bonding	PDL simulation significantly increased fatigue resistance (PDL group: 150, 580 ± 21, 570 cycles vs. non-PDL group: 29, 2 [62] ±9, 473 cycles, $p < 0.001$)	Not specified
Zimmermann et al. (2020) [47]	0.1 mm	Polyvinylsiloxane impression material (PRESIDENT light body, Coltène AG, Altstätten, Switzerland)	Composite resin (BRILLIANT Crios CAD/CAM, Coltène AG, Altstätten, Switzerland)	Teeth and bone constructed using CAD/CAM process: teeth and alveolar sockets milled from composite (Brilliant) and (PEEK BreCAM.bioHPP—bredent medical GmbH; Senden, Germany), elastomeric material injected into the socket, specimen insertion not specified	Not specified	Fracture resistance after thermocycling and mechanical loading	Ceramic/composite fixed partial denture	Not specified	Not specified

Appendix A.3. Literature Review Results

A total of 582 records were identified by searching the Medline and Google Scholar electronic databases. After removal of the 114 duplicates, 468 records were screened based on the title and abstract, resulting in 86 reports sought for retrieval. Three records could not be retrieved because full-text versions were not accessible through institutional or open-access databases; therefore, 83 studies were included for full-text analysis. Of these, 45 studies were excluded after full-text review for the following reasons: lack of physical PDL simulation (n = 39) and non-peer-reviewed articles (n = 6). A total of 38 articles fulfilled all methodological requirements and were included in the final analysis. Additionally, citation tracking was performed, resulting in the identification of 11 additional relevant articles through the reference lists of the included studies, leading to a total of 49 studies included in the review.

The following data were extracted from the studies: material used for PDL simulation and its thickness, material used to simulate the tooth structure with details on the creation of the artificial PDL-tooth complex; methods of experimental model validation; parameters tested; and type of treatment examined. Furthermore, the influence of PDL simulation on the results and information about the correlation with in vivo measurements were also included. The results of the literature review are presented in Appendix A.2, in Table A1.

A concise summary of studies that directly evaluated the influence of PDL simulation on mechanical test outcomes is provided in Supplementary Table S3. The table includes only investigations in which the effect of PDL presence or absence was explicitly analyzed, listing the test type, PDL analog material, validation approach, and the direction of the observed effect. Studies that performed direct comparisons between models with and without simulated PDL are clearly highlighted, providing a transparent overview of the most methodologically relevant evidence available.

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Publikacja nr 2

Journal of
Functional Biomaterials

MDPI

Article

Effect of the Abutment Rigidity on the Wear Resistance of a Lithium Disilicate Glass Ceramic: An In Vitro Study

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Abstract: Lithium disilicate (LDS) glass ceramics are among the most common biomaterials in conservative dentistry and prosthodontics, and their wear behavior is of paramount clinical interest. An innovative in vitro model is presented, which employs CAD/CAM technology to simulate the periodontal ligament and alveolar bone. The model aims to evaluate the effect of the abutment rigidity on the wear resistance of the LDS glass ceramic. Two experimental groups (LDS restorations supported by dental implants, named LDS-on-Implant, or by hybrid ceramic tooth replicas with artificial periodontal ligament, named LDS-on-Tooth-Replica) and a control group (LDS-Cylinders) were compared. Fifteen samples (n = 15) were fabricated for each group and subjected to testing, with LDS antagonistic cusps opposing them over 120,000 cycles using a dual axis chewing simulator. Wear resistance was analyzed by measuring the vertical wear depth (mm) and the volume loss (mm^3) on each LDS sample, as well as the linear antagonist wear (mm) on LDS cusps. Mean values were calculated for LDS-Cylinders (0.186 mm, 0.322 mm^3, 0.220 mm, respectively), LDS-on-Implant (0.128 mm, 0.166 mm^3, 0.199 mm, respectively), and LDS-on-Tooth-Replica (0.098 mm, 0.107 mm^3, 0.172 mm, respectively) and compared using one-way-ANOVA and Tukey's tests. The level of significance was set at 0.05 in all tests. Wear facets were inspected under a scanning electron microscope. Data analysis revealed that abutment rigidity was able to significantly affect the wear pattern of LDS, which seems to be more intense on rigid implant-abutment supports compared to resilient teeth replicas with artificial periodontal ligament.

Keywords: dental materials; dental implants; ceramics; lithia disilicate; tooth wear; dental abutments; 3D printing; periodontium; periodontal ligament

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Citation: Kosewski, P.; De Angelis, F.; Sorrentino, E.; Mielczarek, A.; Buonvivere, M.; D'Arcangelo, C. Effect of the Abutment Rigidity on the Wear Resistance of a Lithium Disilicate Glass Ceramic: An In Vitro Study. *J. Funct. Biomater.* **2023**, *14*, 395. https://doi.org/10.3390/jfb14080395

Academic Editor: Dimitrios Dionysopoulos

Received: 29 June 2023
Revised: 20 July 2023
Accepted: 21 July 2023
Published: 25 July 2023

1. Introduction

Natural occlusion (i.e., the way in which the teeth of the upper and lower jaw contact each other) is a state of dynamic equilibrium and the wear of dental tissues is its inherent component [1]. Studies show that occlusion is self-adjusting through the process of wear and continuous tooth migration [2]. Under physiological conditions, tooth enamel wears off at an average rate of 30–35 µm per year [3]. Within certain limits, teeth wear is therefore essential for masticatory efficiency. Predisposing factors, however, are able to increase this physiological wear rate and gradually lead to an occlusal vertical dimension loss jeopardizing not only dental aesthetics, but the stomatognathic system functions as well [4–10].

Given the subtle boundary between physiological and pathological wear rate, accurate diagnosis is essential. While in the past this relied solely on the clinician's experience, nowadays it is possible to take advantage of intraoral scanners to perform 3D monitoring of the patients, thereby obtaining objective measurements (such as size and depth of the wear facets) of the patient's tooth wear over time, in order to determine whether treatment is necessary or not [11,12]. In cases of pathological wear, a clinical intervention with a minimally invasive approach becomes mandatory [13–15]. Modern restorative materials with favorable wear behavior, such as lithium disilicate (LDS) glass ceramic (luted with contemporary cements [16–18]), make it possible to work in an "additive way", reestablishing a proper occlusal vertical dimension without sacrificing sound tooth structure [19–21]. The periodontal ligament (PDL) is a layer of connective tissue that covers the root cement and forms the tooth's suspension apparatus. Inside the periodontium there is a dense network of blood vessels and fluid. Not only the connective tissue ligaments but also the fluid pressure in the periodontium and the blood pressure in the blood vessels are responsible for the mechanical properties of the periodontium. The complex functioning of the periodontium and its viscoelastic properties are responsible for enabling the dissipation of energy and absorption of forces exerted on the teeth [22]. Dental implants used to restore missing teeth do not have periodontal ligament nor mechanoreceptors and are not able to move in the bone, which means that the natural mechanisms of adapting to occlusion do not function in their case [23]. Crowns on implants made of ceramic materials or zirconium oxide are additionally characterized by greater hardness and a lower modulus of elasticity [24] as well as lower ability to wear than natural human enamel [25,26].

Dental implants used to restore missing teeth offer a highly predictable solution: meta-analyses show an average of 94.6% implant survival over an average follow-up of 13.4 years [27]. Nevertheless, complications of both biological and mechanical nature are relatively common. It is estimated that on average 24% of implants will develop peri-implantitis [28], whereas the incidence of mechanical failure is estimated to be 5.6% to 7.7% [29]. There are reports that overloading on implants may lead to an increased risk of failure in implant prosthetic treatment, contributing not only to mechanical complications, but also exacerbating the course of biological complications [30,31].

Due to the rigid anchorage of implants in the bone, they are subjected to a different type of load than natural teeth [32]. According to available research, the perception of pressure on implants is 4 to 20 times less sensitive than on natural teeth [33]. An experimental animal study by Cheng et al. comparing the biomechanics of teeth with natural dentition showed that tooth displacement under pressure is many times greater than with implants [22]. This study also showed that the energy dissipation capacity of the periodontium is multiple times greater than that of implants and, with a load of 300 g, a natural tooth was able to dissipate 50 times more energy [22]. The viscoelastic properties of the periodontium are able to absorb energy and transfer it to the surrounding bone, dissipating chewing forces. Implants, on the other hand, have a limited ability to dissipate energy [22]. Data shows that implant-supported prosthetic restorations are more likely to suffer from technical complications compared to crowns on natural teeth. Chippings or fractures of ceramic restorations are more common on implants than on natural teeth (8.8% vs. 2.9% over 5 years of observation) [34]. The restorative material choice, already crucial on natural teeth, becomes even more important in a system with reduced abutment mobility, such as the implant-supported crown. In these terms, LDS restorations, with their lower modulus of elasticity, could be a valid alternative to zirconia [35]. This is so true that LDS crowns are being increasingly used for single-unit implant-supported restorations [36–41].

Wear of materials inside the oral cavity is a process influenced by many biological, chemical and physical factors. Due to technological limitations and the presence of many factors affecting the behavior of prosthetic materials in the oral cavity, currently available clinical studies on the abrasion of dental materials or tissues are insufficiently precise [38,42–44]. Thus, in vitro tests are currently employed as an important tool for assessing the basic wear behavior of dental materials [45] and their effect on antagonistic materials. Several papers investigated the in vitro wear resistance of restorative materials opposing either human enamel antagonists or dedicated artificial abraders [25,46], underlining the effect of different antagonist materials on the wear rate of tested specimens. Furthermore, few studies analyzed the wear behavior of dental materials opposing themselves, which seems particularly interesting for simulating the clinical scenario of restored occlusal surfaces occluding in a full-mouth rehabilitation [26,47].

Although the complexity of tooth biomechanics and the variability of the oral cavity environment cannot be completely simulated, various methods of restoring the periodontal ligament have been described in the literature in order to take into account the resilience of PDL in mechanical tests [48] and bring the in vitro measurements closer to the clinical situation. The most commonly used approach is to create a layer of elastic material around the root of the tooth using silicone or polyether materials [49]. Previously described methods involved the process of manually soaking the tooth root in wax and then replacing it with a polyether material [50] or manually applying several layers of rubber on the tooth root [51]. Some authors, however, postulated that the production of periodontal spaces from wax could result in poor control of the thickness of the material simulating periodontium, and thus affect the mechanical properties of the models [51,52]. Since a standardized approach on the matter is still lacking, a large variety of methodologies for producing PDL simulations can still be observed in the available literature [48]. Currently, there are no conclusive data in the literature about the effect of different abutment rigidity on the wear of prosthetic materials. One pilot study by Rosentritt et al. described a quadrant model of the jaws including teeth with simulated periodontium and implants [38]. Models were subjected to wear analysis and the wear of the natural enamel on antagonistic teeth was calculated, but no data were obtained regarding wear depth of restorative materials supported by teeth or implants [38].

On the above basis, the aim of the present study was to evaluate the influence of the abutment rigidity on the wear resistance of lithium disilicate (LDS) restorations (supported by either rigid titanium dental implants or elastic hybrid ceramic tooth replicas) using an in vitro model simulating alveolar bone and periodontal ligament. The null hypothesis to be tested was that the abutment rigidity is not able to influence the wear resistance of lithium disilicate restorations.

2. Materials and Methods

The present study aimed at comparing two experimental groups (LDS-on-Implant and LDS-on-Tooth-Replica) and a control group (LDS-Cylinders).

The components of the experimental model were computer-designed using Solidworks 2016 SP 5.0 (Dassault Systémes SolidWorks Corporation, Vélizy-Villacoublay, France). The experimental models were developed and validated with the use of Periotest Classic device (Medizintechnik Gulden, Modautal, Germany)—a device used for measurement of mobility of natural teeth and implants, which uses percussion on the surface of the crown to measure the damping capacity, which is expressed as PTV (Periotest Value). PTV ranges from −8 to +50, where a lower PTV is correlated with lower mobility, and higher values with higher mobility of subject tested.

2.1. Sample Size Estimation

Sample size was estimated with Statistica 13.3 (StatSoft GmbH, Hamburg, Germany). Independent Sample t-Test was used. The base value for material wear of lithium disilicate and standard deviation of measurements was adopted from research by D'Arcangelo et al. [47]. The anticipated difference of 20% was used for calculations on the basis of a preliminary study by Rosentritt et al. [38] in which the depth of wear traces on prosthetic restorations opposing natural enamel and supported by resin teeth or implants was evaluated (516.8 ± 97.4 and 636.8 ± 187.1 µm, respectively). With statistical power of 0.8 and $\alpha = 0.05$, the required number of samples per group was estimated at 14.

2.2. Sample Fabrication

For the LDS-on-Implant group, fifteen PMMA blocks (20 × 12 × 12 mm) were milled through CAD/CAM procedures and used to simulate the alveolar bone. The implant bed was digitally designed and milled. Various diameters of the implant bed were evaluated with the use of Periotest Classic device (Medizintechnik Gulden) so that stability and damping capacity would reflect the mean stability of the implants in human maxilla (−2.0 PTV [53]). The diameter chosen for further manufacturing of experimental models was 3.2 mm. Fifteen 3.5 mm diameter (2.8 mm core diameter) cylindrical dental implants (SPI Element, Thommen Medical AG, Grenchen, Switzerland) were inserted in the blocks with the use of a 3D printed guide to accommodate implant insertion parallel to the long axis of the implant bed. After placing the prosthetic abutments (SPI Easy, Thommen Medical AG), monolithic lithium disilicate (IPS e.max CAD, Ivoclar Vivadent, Schaan, Liechtenstein) single crowns with a flat occlusal surface were milled (n = 15) and luted (Single Bond Universal; Relyx Ultimate, 3M ESPE, Maplewood, MN, USA) to the implant-abutment complex with adhesive protocol according to the manufacturer's instruction (Figure 1).

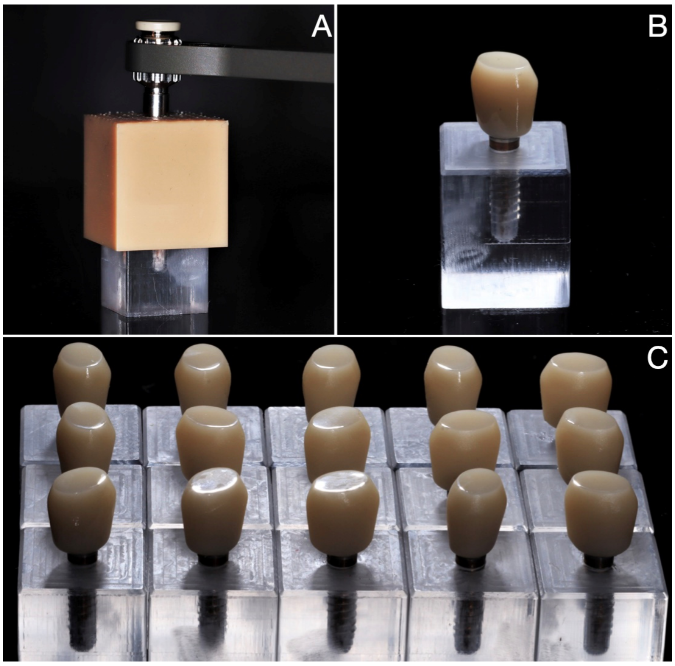

Figure 1. LDS-on-Implant samples. Cylindrical dental implants inserted parallel to the long axis of the implant bed in the blocks with the use of 3D printed guide (**A**). After placing the prosthetic abutments, monolithic lithium disilicate single crowns with a flat occlusal surface were adhesively luted to the implant-abutment complex (**B**). A total of 15 samples was fabricated (**C**).

For LDS-on-Tooth-Replica, a model of a tooth was designed digitally on the basis of mean dimensions of a second maxillary premolar [54]. The preparation for a full crown with a thickness reduction of 2 mm occlusally and 1.5 mm axially and a total occlusal convergence angle of 8° was designed and milled from a hybrid glass ceramic material (Ambarino High Class, Creamed GmbH & Co., Marburg, Germany) with mechanical properties comparable to human dentin (hardness compared to natural dentin was 710 MPa and 650 MPa, respectively, while modulus of elasticity was 10 GPa and 16.5 Gpa [55,56]). Cubic PMMA blocks with dimensions of 20 × 12 × 12 mm were milled to create a replica of the socket, leaving a space between the tooth root and the socket wall. According to the model adopted in the literature, a polyether material was used as a periodontal ligament imitation [48,50] (Impregum Penta, 3M ESPE). Based on tests using the Periotest Classic device (Medizintechnik Gulden), it was determined that in order to obtain a damping capacity equal to the average characteristics of maxillary teeth (2.5 PTV [53]) the thickness of periodontal space should be 0.85 mm. The polyether adhesive (3M ESPE) was applied on the surface of the roots and the socket. Tooth replicas were embedded inside PMMA blocks filled with polyether material using a 3D-printed positioner, which allowed for unambiguous positioning of the tooth replica in the model, leaving an established and homogeneous space around the root and in the apical area (around 0.85 mm). The positioner also allowed for evacuation of the polyether material excess. After full setting, the excessive polyether material was cut with a scalpel at the level of the top surface of the PMMA cube. Finally, fifteen monolithic lithium disilicate single crowns with a flat occlusal surface were milled (n = 15) and adhesively luted (Single Bond Universal; Relyx Ultimate, 3M ESPE) to each tooth replica (Figure 2).

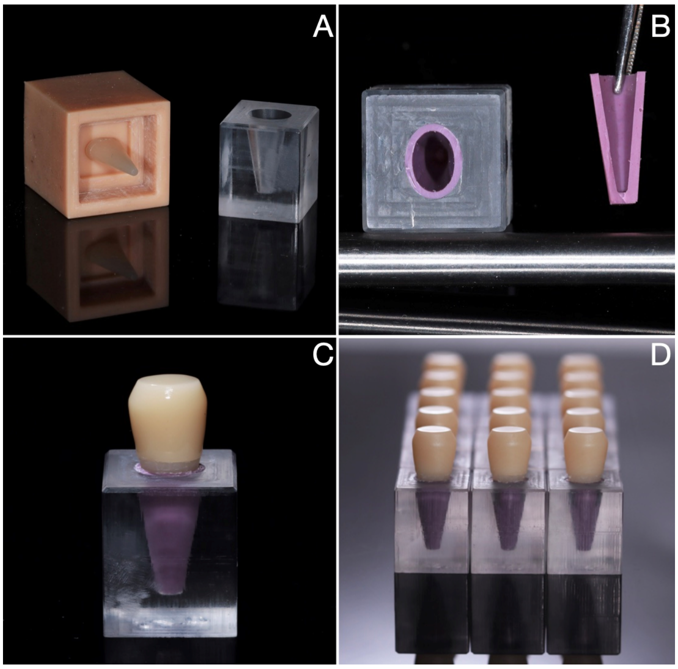

Figure 2. LDS-on-Tooth-Replica samples. Tooth replicas were embedded inside PMMA blocks filled with polyether material using a 3D printed positioner (**A**), which allowed for unambiguous positioning of the replica in the model, leaving an established and homogeneous space around the root (**B**). Monolithic lithium disilicate single crowns with a flat occlusal surface were adhesively luted to each tooth replica (**C**). A total of 15 samples was fabricated (**D**).

As for the control group (LDS-Cylinders), 15 cylindrical specimens (n = 15) with an 8 mm diameter and a 4 mm height were milled from lithium disilicate blocks with CAD/CAM procedures (Figure 3).

Figure 3. LDS-Cylinders sample.

Forty-five (n = 45) LDS conical cusps with a 2 mm diameter round tip were milled and used as antagonists in all groups (Figure 4) in accordance with the Ivoclar method and several in vitro studies [19,25,26,47,57].

Figure 4. Conical cusp used as antagonist.

All LDS crowns, LDS cylindrical specimens and LDS antagonistic cusps were carefully polished with medium and fine silicone polishers (Diapro, Eve Ernst Vetter GmbH, Keltern, Germany). No glaze was applied, because, according to the literature, pretreatment of LDS through grinding and polishing demonstrates more favourable wear characteristics than glazing [58,59].

Finally, models from LDS-on-Implant and LDS-on-Tooth-Replica groups were tested with Periotest Classic. Measurements were taken from mesial, distal, buccal and palatal directions and every measurement was repeated 3 times. The mean of 3 measurements was recorded as PT value for a tooth or implant model in measured direction. The mean PT values of 15 models are presented in the Table 1.

Table 1. Mean Periotest values of experimental models (on the scale from −8 to 50).

	LDS-on-Tooth-Replica		LDS-on-Implant	
Direction	**Mean (PTV)**	**SD**	**Mean (PTV)**	**SD**
Buccal	2.88	1.13	−1.38	0.84
Palatal	2.91	1.04	−1.71	0.88
Mesial	3.04	0.73	−1.69	0.51
Distal	3.14	0.75	−1.97	0.57
Overall	2.99	0.91	−1.69	0.73

2.3. Wear Testing

After being stored at 37 °C for 24 h, each sample of the three experimental groups was randomly paired with an antagonist cusp and placed in a chewing simulator with two axes (CS-4.2, SD Mechatronik GmbH, Feldkirchen-Westerham, Germany) following the methodology described in the literature [19,25,26,47,57]. Acrylic resin (VariDur 200, Buehler, IL, USA) was used to secure the samples inside the chambers and the antagonist cusps in their respective holders (Figure 5).

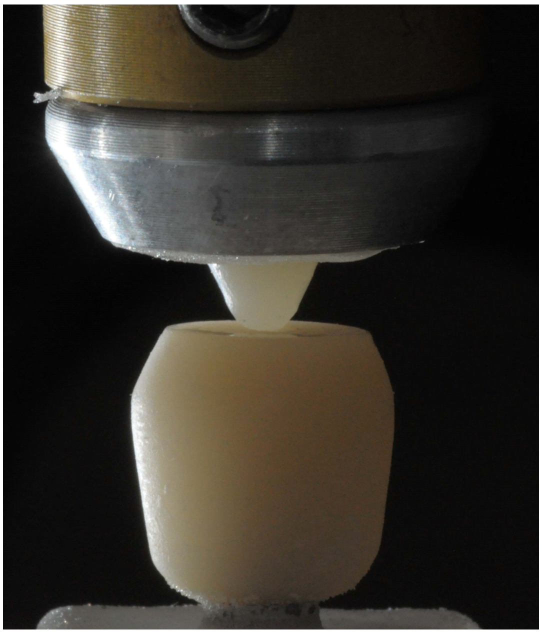

Figure 5. Cusp–sample pair placed in a two-axis chewing simulator.

Subsequently, all sample–cusp pairs underwent a two-body wear test, with parameters as listed in Table 2.

Table 2. Chewing simulator setting parameters.

Number of Cycles	120,000
Force	49 N
Height	3 mm
Lateral movement	−0.7 mm
Lowering speed	60 mm/s
Lifting speed	60 mm/s
Advanced speed	40 mm/s
Return speed	40 mm/s
Frequency	1.6 Hz

The specimens were subjected to 120,000 cycles while being submerged in artificial saliva throughout the entire test duration.

2.4. Data Analysis

After wear test, a quantitative surface assessment was carried out to evaluate the wear facets of the specimens in three dimensions. Three-dimensional meshes of each worn surface were obtained using a laboratory scanner (inEos X5, Dentsply Sirona, Charlotte, NC, USA) in STL (standard triangulation language) format. Subsequently, these meshes were converted to drawing interchange Format (DXF) and imported into computer-aided design software (AutoCAD 2009, Autodesk Inc., San Rafael, CA, USA). The unworn surface surrounding the wear facet was used as reference plane for wear depth (mm) and volume loss (mm^3) measurement. Cusps were measured both before and after the wear test to assess the detectable linear difference, referred to as antagonist wear (mm). Mean values and standard deviation for wear depth, volume loss and antagonist wear were calculated in each group. SPSS Statistics 24 (IBM Corp., Armonk, NY, USA) statistical software was employed to compare the mean values using three different analysis of variance (ANOVA) tests and the Tukey's method for multiple comparisons. Normality and homoscedasticity of the data set were respectively confirmed by means of Shapiro–Wilk test and Brown–Forsythe test.

2.5. SEM Wear Facet Analysis

After quantitative evaluations, specimens were gold-sputtered and examined under a scanning electron microscope (SEM) (EVO 50 XVP LaB6, Carl Zeiss SMT Ltd., Cambridge, UK). The surfaces were observed at a magnification of 250× to enable the evaluation of the wear facets on the examined specimens. The SEM was operated under the following settings: high vacuum (2107 Torr), a current output of 10 pA, an accelerating voltage of 10 kV, and a working distance of approximately 10 mm.

3. Results

Mean values and standard deviations recorded in each group for the three dependent variables tested (wear depth, volume loss and antagonist wear) are listed in Table 3 and graphically presented in Figure 6.

Table 3. Mean values (and standard deviations) recorded for antagonist vertical wear (mm), sample vertical wear depth (mm) and sample volumetric loss (mm^3) within the three experimental groups under investigation.

	Antagonist Wear (mm)	Wear Depth (mm)	Volume Loss (mm^3)
LDS-Cylinders	0.220 ^{a} (0.038)	0.186 ^{a} (0.032)	0.322 ^{a} (0.098)
LDS-on-Implant	0.199 ^{a,b} (0.038)	0.128 ^{b} (0.028)	0.166 ^{b} (0.033)
LDS-on-Tooth-Replica	0.172 ^{b} (0.055)	0.098 ^{c} (0.026)	0.107 ^{c} (0.045)

Same superscript letters indicate not statistically significant differences (reading vertically).

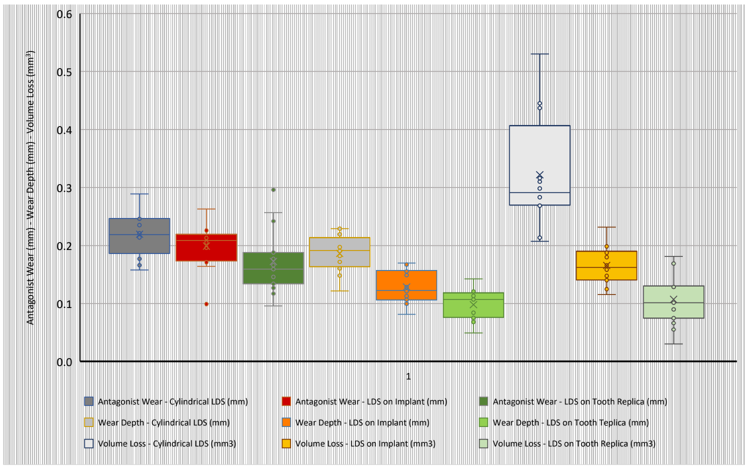

Figure 6. Box plot showing the wear behaviour (in terms of antagonist vertical wear, sample vertical wear depth and sample volumetric loss) of LDS (Lithium Disilicate) cylindrical specimens compared to LDS-implant-supported restorations and to LDS-restorations adhesively luted on tooth replicas.

The One-Way-ANOVA tests showed that the factor under investigation (abutment rigidity) was able to significantly influence all three variables. In detail, statistically significant differences ($p < 0.05$) were found among the experimental group for both wear depth and volume loss parameters, with the highest values showed by LDS-Cylinders (0.186 mm and 0.322 mm^3, respectively) and the lowest by LDS-on-Tooth-Replica (0.098 mm and 0.107 mm^3, respectively). As for antagonist wear, a statistically significant difference ($p < 0.05$) was found between cylindrical LDS (0.220 mm) and LDS-on-Tooth Replica (0.172 mm). The antagonist wear observed on LDS-on-Tooth-Replica was lower than what was observed on LDS-on-Implant (0.199 mm), although no statistically significant differences were detected for this comparison ($p > 0.05$).

Representative SEM images at 250× magnifications were collected for the LDS-on-Implant (Figure 7A), LDS-on-Tooth-Replica (Figure 7B), and LDS-Cylinders (Figure 7C) groups.

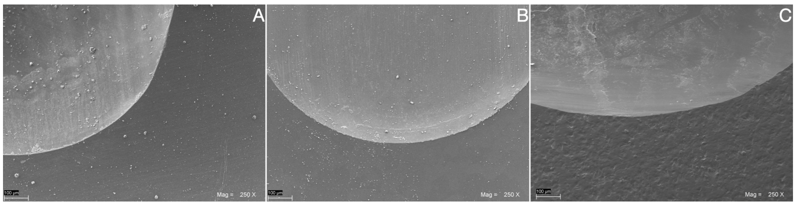

Figure 7. Representative SEM images at 250× magnifications collected for the LDS-on-Implant (**A**), LDS-on-Tooth-Replica (**B**), and LDS-Cylinders (**C**) groups.

A similar wear pattern with smooth scratches running the length of their wear track was evident in all groups. Microcracks were not detectable. Debris was noticeable along the surface, with no difference between worn and unworn areas.

4. Discussion

The null hypothesis tested in the present study had to be rejected. The abutment rigidity was a significant factor, able to affect the wear behaviour of LDS restorations. Chewing simulation showed statistically significant differences in both the wear depth and the volume loss variables among LDS-on-Tooth-Replica, LDS-on-Implant and LDS-Cylinders groups. The difference between LDS-on-Tooth-Replica and LDS-on-Implant was 24% in terms of wear depth and 35% in terms of volume loss. This difference indicates that the ability to dissipate the energy of the flexible tooth suspension apparatus may result in a reduced attrition on the prosthetic material itself. The greater energy absorbed by the implant-supported prosthetic restoration results in greater material wear. Data in the literature comparing the wear between implants and natural teeth are lacking, but biomechanical studies confirm that implants may dissipate energy less effectively than natural dentition during loading [22].

A pilot study by Stück et al. evaluating tooth wear through intraoral scans did not show statistically significant differences in wear of the LDS crown on the implant compared to the adjacent natural teeth, nor differences in wear of enamel of antagonistic teeth during the 24-month follow-up period. No comparison between the wear rates of the same prosthetic material supported either by teeth or by implants was performed. This study, as described by the authors themselves, was limited by the small number of cases and the large discrepancy of data related to the multitude of factors affecting tooth wear in the clinical setting [2]. A pilot in vitro study by Rosentritt et al. assessed that the type of abutment (implant or natural tooth) may affect the wear of the same prosthetic material, but the possibility of drawing conclusions from the obtained data was limited [38].

Our study showed significant differences in wear also when implant-supported LDS restorations were compared to cylindrical LDS samples directly embedded within the resin, without any supporting abutment. Wear on LDS-Cylinders was 45% greater in terms of linear depth and 94% greater in terms of volume loss than what was observed on LDS-on-Implant. This indicates some energy dissipation capacity even in the presence of a titanium implant/abutment support, probably also ascribable to the composite cement used for luting the prosthetic restoration [60].

Antagonist wear showed no statistically significant differences between LDS-Cylinders and LDS-on-Implant, and between LDS-on-Implant and LDS-on-Tooth-Replica. Significant differences in antagonist wear between LDS-Cylinders and LDS-on-Tooth-Replica (0.220 mm and 0.172 mm, respectively) were demonstrated. In our study, the effect of the system rigidity on the wear of the antagonistic cusps was less pronounced than that of the test samples. This could be explained, considering that, in all groups, all antagonistic cusps were rigidly fixed to the simulator arm in the same way, without any additional rigid/flexible support.

Two-body-wear, three-body wear, and toothbrushing tests have been described in the literature among the most common in vitro methodologies to investigate wear in ceramic restorative materials [61,62]. The toothbrushing test is used for assessing the abrasion wear of ceramic materials and examining the wear of the glaze or extrinsically stained layers. However, the impact on dental wear can vary significantly depending on various factors, such as the technique and timing of toothbrushing, the type of dentifrice and toothbrush employed, as well as the frequency and force of brushing [62]. The three-body wear test replicates clinical conditions to simulate occlusal wear through the introduction of a third abrasive body between the ceramic surface and the opposing surface simulating food bolus. Studies on composite and amalgam samples revealed that gradual change in the distance between the opposing substrates and even minor alterations of the abrasive-film features and thickness at the contact areas result in significant changes in wear rates and wear-rate ranking of the materials [61,62]. Two-body wear tests are suitable for predicting the wear behavior of dental materials [44] through the simulation of non-masticatory tooth movements such as swallowing, empty mastication, parafunctions and dynamic occlusion movements [63]. Moreover, a two-body wear test using standardized antagonists made out of the same material of the tested specimens allows simulation of the situation of restored occlusal surfaces occluding in a full-mouth rehabilitation following the treatment of a severely worn patient [20,26]. In the present paper, two-body wear test of LDS samples resulted in a SEM surface topography characterized by smooth wear tracks surrounded by debris. In dental ceramics, wear is strongly affected by surface roughness [64–66]. The repeated sliding contact on the ceramic surface leads to compressive stresses before movement, shear stresses at the contact interfaces, and tensile stresses at the trailing edge of the antagonist, thus resulting in fatigue wear and debris formation able to further increase wear [61,67]. Despite some controversies [68], several studies demonstrated greater wear for materials opposed to glazed ceramics than to polished ones [68–70]. Polished LDS showed a tribological behaviour close to human enamel [59,69] and should therefore be preferred to glazed LDS for single restorations [58].

According to Brosh et al., elastomeric dental impression materials perform similarly to periodontal ligament under loading and also in terms of immediate recovery when the load is not applied and thus are the material of choice in PDL simulations [71]. The presented methodology of making teeth replicas with periodontium using CAD/CAM technology showed high repeatability. The standard deviation of the measurements of damping capacity using the Periotest was 0.91 (on a scale of −8 to 50). Obtained Periotest values were comparable with those described for natural teeth and implants, and the appropriate difference in vibration damping between teeth replicas and implants was achieved [53]. The homogeneity of the obtained models made it possible to control the variables that could affect the tested parameters. The width of the periodontal gap used in the present model was around 0.85 mm, similar to a previously described in vitro model in which tooth mobility was directly measured to reflect mobility of natural dentition (0.075 ± 0.1 mm) [50]. These data indicate that simulating periodontal ligament with polyether material should take into account a larger periodontal gap width than that observed in natural tissues (0.15–0.38 mm) [72] to obtain adequate mobility and damping capacity of the in vitro models. The presented methodology made it possible to eliminate the previously described problem of obtaining a reproducible thickness of the silicone layer simulating PDL by manually soaking the root in wax [51,52], as well as reduced the time needed for model fabrication [50].

A limitation of the present approach was the use of solid PMMA blocks, which, according to the technical data, have a similar modulus of elasticity as human bone (approx. 3000 MPa) [73,74]. However, reproducing the trabecular and cortical bone layer with a material that would precisely emulate the properties of human bone might bring the presented model closer to the clinical situation.

In the present study, tooth replicas milled from hybrid ceramics with a similar modulus of elasticity and compression resistance as human dentine [75] were used instead of natural teeth. Despite the fact that the use of natural tissues would be the closest to clinical conditions, natural teeth are characterized by large discrepancies in shape, but also in mechanical properties, which results in increased inhomogeneity of the results [38]. The described experimental model is easy to reproduce and relatively time effective to perform, which allows for a wider application in the in vitro study of prosthetic materials.

5. Conclusions

Considering the limitations of the presented experimental model, it can be concluded that:

1. The rigidity of the abutment is able to significantly affect the wear pattern of lithium disilicate glass ceramics, which seems to be more intense in the presence of a rigid implant-abutment support than on restorations supported by more flexible teeth replicas and in the presence of periodontal ligament.
2. In vitro chewing simulation models that include a wider number of intra-oral parameters (alveolar bone, periodontal ligament, abutment rigidity/elasticity, etc.) might be closer to the actual clinical situation and seem to produce significantly different results from extremely simplified in vitro models.

Author Contributions: Conceptualization, P.K., F.D.A., A.M., M.B. and C.D.; Data curation, F.D.A., E.S., M.B. and C.D.; Formal analysis, P.K. and A.M.; Funding acquisition, P.K., F.D.A., A.M. and C.D.; Investigation, P.K. and A.M.; Methodology, P.K., F.D.A., E.S., A.M., M.B. and C.D.; Project administration, P.K., F.D.A., A.M. and C.D.; Resources, P.K., F.D.A., A.M. and C.D.; Software, P.K. and A.M.; Supervision, P.K., F.D.A., A.M. and C.D.; Validation, P.K., F.D.A., A.M. and M.B.; Visualization, P.K., E.S., A.M. and M.B.; Writing—original draft, P.K. and A.M.; Writing—review & editing, F.D.A., E.S., M.B. and C.D. All authors have read and agreed to the published version of the manuscript.

Funding: This research received no external funding.

Data Availability Statement: The data presented in this study are available on request from the corresponding author.

Acknowledgments: Special thanks to Grzegorz Palicki and C.Witt Dental for providing implants for the tests. We would also like to thank MaxLab dental laboratory and R.S. Team for their help with the CAD/CAM manufacturing of the study components. Last but not least, sincere thanks to Maciej Kalinowski for assisting with the digital design of the study.

Conflicts of Interest: The authors declare no conflict of interest.

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Omówienie wyników i dyskusja

Analiza piśmiennictwa dotyczącego fizycznej symulacji więzadła ozębnej w badaniach laboratoryjnych ujawniła znaczną heterogeniczność stosowanych metod oraz obecność istotnych ograniczeń metodologicznych. W przeważającej części proponowanych modeli *in vitro*, warstwa symulująca więzadło ozębnej była wytwarzana w sposób manualny, najczęściej z wykorzystaniem techniki traconego wosku lub ręcznego nakładania warstw materiałów elastomerowych. Zastosowane rozwiązania nie pozwalają na pełną kontrolę jednorodności oraz powtarzalności procesu. Brak możliwości precyzyjnej oceny grubości warstwy symulującej więzadło ozębnej skutkuje znaczną zmiennością parametrów biomechanicznych badanych próbek, nawet w obrębie tej samej serii eksperymentalnej[20-22].

Przegląd literatury ujawnia również braki w dokumentacji metodologicznej, dotyczące geometrii symulowanej warstwy więzadła ozębnej, jednorodności zastosowanego materiału oraz sposobu pozycjonowania korzenia względem modelu zębodołu. Ograniczenia te znacząco utrudniają rzetelną interpretację wyników, uniemożliwiają ich miarodajne porównanie i osłabiają wiarygodność wniosków dotyczących biomechanicznego funkcjonowania badanych układów[23-28].

Z analizy piśmiennictwa wynika ponadto, że znaczenie symulacji więzadła ozębnej w badaniach *in vitro* jest uzależnione od rodzaju badanego uzupełnienia protetycznego oraz charakteru testu mechanicznego. Najbardziej wyraźny wpływ obecności symulowanego PDL odnotowano w badaniach mostów protetycznych, zarówno konwencjonalnych, jak i adhezyjnych, w których brak elastycznego podparcia prowadził do przeszacowania wytrzymałości konstrukcji oraz do wzorców pęknięć odbiegających od obserwowanych w warunkach klinicznych[20,29,30]. W badaniach dotyczących wkładów koronowo-korzeniowych autorzy wskazywali, że zastosowanie warstwy imitującej więzadło ozębnej może modyfikować rozkład naprężeń w korzeniu i zmieniać typ powstających uszkodzeń. Zęby umocowane w elastycznej warstwie częściej podlegały złamaniom korzenia w porównaniu do sztywno umocowanych próbek[26,31]. Jednocześnie, w badaniach dotyczących odporności na złamanie pojedynczych koron nie wykazano istotnego wpływu obecności lub braku symulacji PDL[32,33].

Doniesienia dotyczące badań nad ścieralnością materiałów protetycznych są nieliczne. Dostępne wyniki sugerują, że zastosowanie symulacji PDL modyfikuje przebieg procesu zużycia materiału, prowadząc do mniejszego starcia próbek w porównaniu z modelami umocowanymi sztywno[34,35].

Ponadto wykazano, że uzyskiwana ruchomość zęba w modelach *in vitro* stanowi wypadkową zarówno wymiarów geometrycznych symulowanej przestrzeni więzadła ozębnej, jak i właściwości mechanicznych materiału zastosowanego do odtworzenia jej podatności. W konsekwencji, szerokość przestrzeni PDL w modelu eksperymentalnym odpowiadająca jej anatomicznym wymiarom (ok. 0.15-0.38 mm[36]) nie musi prowadzić do wiernego odtworzenia biomechaniki zęba. Użycie określonego materiału elastycznego oraz danej konfiguracji anatomicznej repliki zęba może prowadzić do powstania modelu o nadmiernej sztywności lub przeciwnie - zbyt dużej podatności[20,35]. Z tego względu, prawidłowe odwzorowanie warunków biomechanicznych wymaga kalibracji modelu i ustalenia niezbędnej szerokości PDL opartej na testach funkcjonalnych takich jak pomiar zdolności do tłumienia drgań czy pomiar ruchomości pod obciążeniem.

Wyniki badań nad nowym modelem *in vitro* symulującym więzadło ozębnej wykazały, że zastosowanie technologii CAD/CAM umożliwia precyzyjne zaprojektowanie i odtworzenie przestrzeni odpowiadającej PDL, co stanowi istotną różnicę w porównaniu do wcześniej opisywanych metod manualnych[19]. Walidacja funkcjonalna modelu, przeprowadzona z wykorzystaniem pomiarów urządzeniem Periotest, pozwoliła na uzyskanie wartości tłumienia drgań mieszczących się w zakresie raportowanym w literaturze dla zębów naturalnych[37]. Jednocześnie, niska zmienność wyników dla poszczególnych próbek wskazuje na wysoką powtarzalność właściwości biomechanicznych proponowanego modelu. Rezultaty te potwierdzają, że cyfrowo kontrolowany proces projektowania i wytwarzania pozwala na lepszą kontrolę geometrii warstwy PDL w porównaniu do metod opartych na procedurach manualnych, co ma szczególne znaczenie w badaniach porównawczych wymagających jednorodnych warunków eksperymentalnych.

Walidacja radiologiczna dostarczyła dodatkowych informacji dotyczących jednorodności warstwy symulującej ozębną. Zastosowanie technologii CAD/CAM umożliwiło zaprojektowanie modelu w sposób gwarantujący uzyskanie założonych parametrów geometrycznych. Wprowadzenie dedykowanych pozycjonerów pozwoliło na jednoznaczne i powtarzalne umiejscowienie repliki zęba w modelu. Takie rozwiązania znacząco ograniczyły wpływ czynników losowych typowych dla technik manualnych oraz zwiększyły powtarzalność otrzymywanych modeli.

Na podstawie przeglądu piśmiennictwa oraz badań własnych zaproponowano spójne podejście metodologiczne do projektowania i raportowania szczegółów modeli *in vitro* z symulacją więzadła ozębnej, które może stanowić punkt odniesienia dla dalszych badań nad właściwościami mechanicznymi materiałów i konstrukcji protetycznych.

Wyniki badania, w którym porównano ścieralność koron protetycznych wykonanych z dwukrzemianu litu, osadzonych na implantach stomatologicznych oraz na replikach zębów z symulowanym więzadłem ozębnej wskazują na istotne różnice w przebiegu procesu zużycia materiału w zależności od rodzaju filaru. Obserwowane różnice dotyczyły zarówno głębokości starcia, jak i ubytku objętościowego materiału (odpowiednio: 24% i 35%), co sugeruje, że sztywność filaru wpływa na sposób przenoszenia i absorpcji energii podczas cyklicznych obciążeń ceramiki. Brak elastycznego komponentu w konstrukcjach implantoprotetycznych prowadzi do zwiększonej koncentracji naprężeń w obrębie materiału protetycznego, co skutkuje jego bardziej nasilonym starciem. Obecność symulowanego PDL w modelu zęba sprzyja częściowemu rozpraszaniu energii oraz redukcji intensywności procesu ścierania.

Średnia głębokość starcia wyniosła 0,128 mm dla koron osadzonych na implantach oraz 0,098 mm dla koron opartych na replikach zębów z symulowanym więzadłem ozębnej, natomiast średni ubytek objętościowy materiału wynosił odpowiednio 0,166 mm^3 i 0,107 mm^3. Uzyskane wyniki wskazują, że wpływ rodzaju podparcia był bardziej wyraźny w odniesieniu do zużycia badanej korony niż ceramiki stanowiącej element przeciwstawny. Wartość starcia antagonisty była niższa w grupie z symulowanym więzadłem ozębnej (0,172 mm) niż w grupie implantologicznej (0,199 mm), jednak różnica ta nie osiągnęła istotności statystycznej. Porównanie z grupą kontrolną, w której próbki dwukrzemianu litu były osadzone bezpośrednio w żywicy, bez pośrednictwa filaru, wykazało jeszcze większe nasilenie zużycia materiału niż w grupie implantologicznej. W tej grupie głębokość starcia była większa o 45%, a ubytek objętościowy o 94% w porównaniu z koronami osadzonymi na implantach. Sugeruje to, że nawet w układzie implant–łącznik–korona występuje ograniczona zdolność do rozpraszania energii, prawdopodobnie częściowo związana z obecnością warstwy cementu kompozytowego czy z samą konstrukcją i umocowaniem implantu w podłożu. Analiza powierzchni zużycia w skaningowym mikroskopie elektronowym wykazała zbliżony charakter śladów starcia we wszystkich badanych grupach, z obecnością gładkich rys oraz bez widocznych mikropęknięć, co wskazuje, że obserwowane różnice dotyczyły przede wszystkim nasilenia procesu zużycia, a nie odmiennych mechanizmów uszkodzenia materiału.

Zastosowanie nowego modelu *in vitro* umożliwiło porównanie ścieralności koron protetycznych w warunkach różniących się wyłącznie charakterem podparcia filaru – sztywnym w przypadku implantu oraz elastycznym w przypadku zęba z symulowanym więzadłem ozębnej. Precyzyjnie kontrolowana geometria warstwy PDL oraz powtarzalność jej właściwości biomechanicznych pozwoliły na ograniczenie zmienności niezwiązanej z analizowanym czynnikiem oraz na jednoznaczną interpretację różnic w przebiegu zużycia materiału.

Przedstawione wyniki odnoszą się do warunków laboratoryjnych *in vitro* i opisują wpływ charakteru podparcia filaru protetycznego na ścieralność materiałów protetycznych w ramach przyjętego modelu badawczego. Pełna ocena znaczenia klinicznego obserwowanych zależności wymaga ich potwierdzenia w badaniach klinicznych. Należy jednak podkreślić, że w dostępnej literaturze brak jest badań klinicznych bezpośrednio porównujących ścieralność koron protetycznych osadzonych na implantach stomatologicznych z analogicznymi rekonstrukcjami opartymi na zębach naturalnych. W konsekwencji badania eksperymentalne prowadzone w kontrolowanych warunkach in vitro stanowią obecnie podstawowe źródło danych w tym zakresie.

Podsumowanie

Przeprowadzone badania wykazały, że dotychczas stosowane metody fizycznej symulacji więzadła ozębnej w badaniach in vitro są niejednorodne i obarczone istotnymi ograniczeniami metodologicznymi, co utrudnia porównywanie wyników. Opracowany model CAD/CAM umożliwił precyzyjne i powtarzalne odtworzenie warstwy symulującej PDL oraz uzyskanie właściwości biomechanicznych zbliżonych do zębów naturalnych. W badaniu ścieralności wykazano, że korony z dwukrzemianu litu osadzone na implantach zużywały się bardziej niż korony oparte na replikach zębów z symulowanym PDL, zarówno pod względem głębokości starcia, jak i ubytku objętościowego. Wyniki potwierdzają istotny wpływ charakteru i sztywności filaru protetycznego na przebieg procesu starcia materiału ceramicznego pod obciążeniem funkcjonalnym.

Wnioski

1. Symulacja więzadła ozębnej stanowi kluczowy element metodologiczny w badaniach in vitro oceniających właściwości mechaniczne konstrukcji protetycznych, w tym odporność na złamanie mostów protetycznych, wytrzymałość rekonstrukcji zębów z wkładami koronowo-korzeniowymi oraz ścieralność materiałów protetycznych.
2. Znaczna heterogeniczność dotychczas stosowanych metod fizycznej symulacji PDL, obejmująca m.in. brak standaryzacji geometrii oraz właściwości warstwy elastycznej, istotnie ogranicza porównywalność wyników badań mechanicznych prowadzonych w różnych ośrodkach badawczych.
3. Modele eksperymentalne uwzględniające symulację więzadła ozębnej powinny podlegać walidacji funkcjonalnej, niezależnie od weryfikacji geometrycznej. Etap ten jest niezbędny do potwierdzenia, że opracowany model możliwie wiernie odtwarza mechaniczne zachowanie naturalnego więzadła ozębnej.
4. Zastosowanie technologii CAD/CAM umożliwia precyzyjne i powtarzalne odtworzenie przestrzeni odpowiadającej więzadłu ozębnej w modelach in vitro. Zaprezentowany model eksperymentalny stanowi alternatywę dla dotychczasowych technik oraz może być adaptowany do różnych scenariuszy badawczych.
5. Większa sztywność filaru protetycznego istotnie wpływa na zwiększenie intensywności ścierania koron z dwukrzemianu litu w badaniach laboratoryjnych.
6. W warunkach badań in vitro rekonstrukcje protetyczne oparte na implantach stomatologicznych wykazują większą intensywność ścierania materiału ceramicznego w porównaniu z rekonstrukcjami wspartymi na replikach zębów z symulowanym więzadłem ozębnej.

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Oświadczenia współautorów

Warszawa, 02.04.26
(miejscowość, data)

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OŚWIADCZENIE

Jako współautor pracy pt. „*Simulation of the Periodontal Ligament in Dental Materials Research: A CAD/CAM-Based Method for PDL Modeling"* oświadczam, iż mój własny wkład merytoryczny w przygotowanie, przeprowadzenie i opracowanie badań oraz przedstawienie pracy w formie publikacji stanowi:
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OŚWIADCZENIE

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Warszawa, 02.02.2026
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prof. dr hab. n. med. Agnieszka Mielczarek
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Agnieszka Beata Mielczarek
Elektronicznie podpisany przez Agnieszka Beata Mielczarek
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*w szczególności udziału w przygotowaniu koncepcji, metodyki, wykonaniu badań, interpretacji wyników

(place, date)
Chieti (Italy) 02/04/2026

prof. Camillo D'Arcangelo

DECLARATION

As a co-author of the paper entitled *"Effect of the Abutment Rigidity on the Wear Resistance of a Lithium Disilicate Glass Ceramic: An In Vitro Study,"* I hereby declare that my own substantive contribution to the preparation, conduct, and analysis of the research, as well as to the presentation of the work in the form of a scientific publication, comprised:

Conceptualization of the study, data curation, development of the methodology, project administration, supervision, funding acquisition, provision of resources, and review and editing of the manuscript

My percentage contribution to the preparation of the publication is 10%

I hereby give my consent for the above-mentioned publication to be used as part of the doctoral dissertation of Przemysław Kosewski, DDS.

....
(signature of the declarant)

Chieti, 02 - 04 - 2026
(place, date)

prof. Francesco de Angelis

DECLARATION

As a co-author of the paper entitled *"Effect of the Abutment Rigidity on the Wear Resistance of a Lithium Disilicate Glass Ceramic: An In Vitro Study,"* I hereby declare that my own substantive contribution to the preparation, conduct, and analysis of the research, as well as to the presentation of the work in the form of a scientific publication, comprised:

Conceptualization of the study, development of the methodology, data curation, validation of the methodology, project administration, supervision, funding acquisition, provision of resources, and review and editing of the manuscript.

My percentage contribution to the preparation of the publication is 35%

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Francesco De Angelis
....
(signature of the declarant)

Chieti, 06/04/26

Matteo Buonvivere, DDS

DECLARATION

As a co-author of the paper entitled *"Effect of the Abutment Rigidity on the Wear Resistance of a Lithium Disilicate Glass Ceramic: An In Vitro Study,"* I hereby declare that my own substantive contribution to the preparation, conduct, and analysis of the research, as well as to the presentation of the work in the form of a scientific publication, comprised:

Conceptualization of the study, development of the methodology, data curation, validation of the methodology, visualization of the study results, and review and editing of the manuscript.

My percentage contribution to the preparation of the publication is 10%

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Matteo Buonvivere
....
(signature of the declarant)

CHIETI, 2 . 04 . 26
(place, date)

Edoardo Sorrentino, DDS

DECLARATION

As a co-author of the paper entitled *"Effect of the Abutment Rigidity on the Wear Resistance of a Lithium Disilicate Glass Ceramic: An In Vitro Study,"* I hereby declare that my own substantive contribution to the preparation, conduct, and analysis of the research, as well as to the presentation of the work in the form of a scientific publication, comprised:

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Edoardo Sorrentino ....
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