



Special Issue Reprint

Advanced Dental Materials

From Design to Application, Second Edition

Edited by
Tina Poklepović Peričić and Josip Kranjčić

mdpi.com/journal/materials



Advanced Dental Materials: From Design to Application, Second Edition

Advanced Dental Materials: From Design to Application, Second Edition

Guest Editors

Tina Poklepović Perićić

Josip Kranjčić



Basel • Beijing • Wuhan • Barcelona • Belgrade • Novi Sad • Cluj • Manchester

Guest Editors

Tina Poklepović Perićić
School of Medicine
University of Split
Split
Croatia

Josip Kranjčić
School of Dental Medicine
University of Zagreb
Zagreb
Croatia

Editorial Office

MDPI AG
Grosspeteranlage 5
4052 Basel, Switzerland

This is a reprint of the Special Issue, published open access by the journal *Materials* (ISSN 1996-1944), freely accessible at: <https://www.mdpi.com/journal/materials/special-issues/V98V1FS5HK>.

For citation purposes, cite each article independently as indicated on the article page online and as indicated below:

Lastname, A.A.; Lastname, B.B. Article Title. <i>Journal Name</i> Year , Volume Number, Page Range.
--

ISBN 978-3-7258-7581-8 (Hbk)

ISBN 978-3-7258-7582-5 (PDF)

<https://doi.org/10.3390/books978-3-7258-7582-5>

© 2026 by the authors. Articles in this reprint are Open Access and distributed under the Creative Commons Attribution (CC BY) license. The reprint as a whole is distributed by MDPI under the terms and conditions of the Creative Commons Attribution-NonCommercial-NoDerivs (CC BY-NC-ND) license (<https://creativecommons.org/licenses/by-nc-nd/4.0/>).

Contents

About the Editors	vii
-----------------------------	-----

Preface	ix
-------------------	----

Josip Kranjčić and Tina Poklepović Peričić Advanced Dental Materials: From Design to Application—Second Edition Reprinted from: <i>Materials</i> 2025 , <i>18</i> , 4902, https://doi.org/10.3390/ma18214902	1
---	---

Ragai-Edward Matta, Lara Berger, Oleksandr Sednyev, Dennis Bäuerle, Eva Maier, Werner Adler and Michael Taschner In Vivo Wear Analysis of Leucite-Reinforced Ceramic Inlays/Onlays After 14 Years Reprinted from: <i>Materials</i> 2025 , <i>18</i> , 3446, https://doi.org/10.3390/ma18153446	3
---	---

Lara Berger, Felix Förtsch, Ralf Robert Kretschmer, Oleksandr Sednyev, José Ignacio Zorzin, Manfred Wichmann and Ragai Edward Matta Influence of the Milling Strategy on the Marginal Fit of Chairside-Fabricated Lithium Disilicate Crowns Reprinted from: <i>Materials</i> 2025 , <i>18</i> , 2184, https://doi.org/10.3390/ma18102184	18
---	----

Alexander Strunz, Lara Berger, Anna Seidel, Johannes Ries, Werner Adler, Manfred Wichmann and Ragai Edward Matta The Impact of Three-Dimensional Printer Technology on the Accuracy of Dental Implant Models Reprinted from: <i>Materials</i> 2025 , <i>18</i> , 1902, https://doi.org/10.3390/ma18091902	31
--	----

Nicole Riberti, Michele Furlani, Emanuele Tognoli, Adriano Piattelli, Luca Comuzzi, Alessandra Giuliani and Andrea Gatto Biconometric Connections in Dental Implants: A Pilot Mechanical Study Reprinted from: <i>Materials</i> 2025 , <i>18</i> , 1415, https://doi.org/10.3390/ma18071415	45
--	----

Cosmina Raluca Fratila, Luis Óscar Alonso-Ezpeleta, Manuel Poveda-Saenz, Giovanni Giovannini, Ana Belén Lobo-Galindo, Javier Flores-Fraile and Álvaro Zubizarreta-Macho Accuracy Evaluation of Indirect Bonding Techniques for Clear Aligner Attachments Using 3D-Printed Models: An In Silico and Physical Model-Based Study Reprinted from: <i>Materials</i> 2025 , <i>18</i> , 780, https://doi.org/10.3390/ma18040780	58
--	----

Mariano Herrero-Climent, Fernando Sanchez-Lasheras, Jordi Martinez-Lopez, Javier Gil and Aritza Brizuela-Velasco Mechanical Behavior of PEEK and PMMA Graphene and Ti6Al4V Implant-Supported Frameworks: In Silico Study Reprinted from: <i>Materials</i> 2025 , <i>18</i> , 441, https://doi.org/10.3390/ma18020441	72
---	----

Mario Ceddia, Tea Romasco, Nilton De Bortoli, Jr., Bruno Freitas Mello, Adriano Piattelli, Eitan Mijiritsky, et al. Biomechanical Finite Element Analysis of Two Types of Short-Angled Implants Across Various Bone Classifications Reprinted from: <i>Materials</i> 2024 , <i>17</i> , 5680, https://doi.org/10.3390/ma17235680	97
---	----

Fabián Murillo-Gómez, José Roberto Hernández-Viquez,	
José Roberto Sauma-Montes de Oca, Cristina Vargas-Vargas, Natalia González-Vargas,	
José Roberto Vega-Baudrit and Daniel Chavarría-Bolaños	
Mechanical, Adhesive and Surface Properties of a Zirconia-Reinforced Lithium Silicate	
CAD/CAM Ceramic Exposed to Different Etching Protocols	
Reprinted from: <i>Materials</i> 2024 , <i>17</i> , 5039, https://doi.org/10.3390/ma17205039	116

About the Editors

Tina Poklepović Peričić

Tina Poklepović Peričić is a prosthodontist and an Assistant Professor at the University of Split School of Medicine, where she teaches research methodology and prosthodontics.

Since 2009, she has been actively involved in the voluntary work of Cochrane Croatia, first as a dental medicine coordinator and from 2015 to 2022 as a co-director of the Centre. Her interests include systematic reviews, the dissemination of evidence, and education. Since 2019, she has been a member of the international GRADE working group, which assesses the certainty of research evidence and develops clinical practice guidelines. She has authored various papers on evidence synthesis, clinical practice guidelines, and scientific methodology. She has also been a reviewer for several scientific journals and an Editorial Board Member for *BMC Medical Research Methodology*, *Systematic Reviews*, and *Frontiers in Medicine*. In addition to leading an ongoing EU project on fact-checking, she led the international Erasmus + project “Evidence Implementation in Clinical Practice” and has been coordinating the global initiative “Informed Health Choices” in Croatia, which deals with teaching children and the public about critical thinking regarding health issues. She was a clinical advisor to the Cochrane Clinical Answers and participated in the EVBRES (Evidence-Based Research) COST Action CA-17117 project on evidence-based research.

Josip Kranjčić

Josip Kranjčić is a prosthodontist and an Associate Professor at the University of Zagreb School of Dental Medicine.

As a student, he was the best in his year, earning five consecutive Dean’s Awards. Since 2011, he has been employed as an Assistant at the Department of Fixed Prosthodontics at the School of Dental Medicine in Zagreb. In 2014, he spent time at the Smithsonian Institution in Washington, D.C., USA, where he researched the characteristics of the temporomandibular joint in North American historical populations as part of his doctoral dissertation. That same year, he also participated in the “Young Prosthodontic Educators Workshop”, organized by the International College of Prosthodontists in Germany. He successfully defended his doctoral dissertation, titled “Research on Anthropometric Measures of Temporomandibular Joint’s Articular Eminence on Historical Samples”, in 2016. He also completed his specialization in prosthetic dentistry in 2016. In 2017, he was promoted to Assistant Professor, and in 2022, he was appointed Associate Professor at the Department of Fixed Prosthodontics, University of Zagreb School of Dental Medicine. As a specialist in prosthetic dentistry, he works at the University Hospital “Dubrava” in Zagreb. He has participated in scientific projects funded by the Croatian Science Foundation. He has published several scientific and professional papers in domestic and international journals and has actively participated in numerous international conferences. He is also a Guest Editor of the scientific journal *Materials*.

Preface

The field of dental materials has witnessed remarkable progress in recent years, with innovative designs, advanced technologies, and novel applications significantly transforming modern dental practice. This reprint, *Advanced Dental Materials: From Design to Application, Second Edition*, aims to provide a comprehensive and up-to-date overview of the latest developments in dental materials science, addressing fundamental principles, emerging technologies, and their translation into real-world clinical applications.

This volume is intended for dental professionals, researchers, and students seeking a deeper and more integrated understanding of the materials shaping the future of dentistry. By bringing together current research findings and practical insights, this Reprint highlights the critical connection between material design, performance, and clinical outcomes.

We hope that this work will serve not only as a valuable reference but also as a source of inspiration for further exploration and innovation in the field. Ultimately, it seeks to bridge the gap between material development and clinical use, supporting the continued advancement of evidence-based and patient-centred dental care.

Tina Poklepović Peričić and Josip Kranjčić

Guest Editors

Editorial

Advanced Dental Materials: From Design to Application—Second Edition

Josip Kranjčić ^{1,2,*} and Tina Poklepović Peričić ³

¹ Department of Prosthodontics, University Hospital Dubrava, Av. Gojka Šuška 6, 10000 Zagreb, Croatia

² Department of Fixed Prosthodontics, University of Zagreb School of Dental Medicine, Gundulićeva 5, 10000 Zagreb, Croatia

³ School of Medicine, University of Split, Šoltanska 2, 21000 Split, Croatia; tinapoklepovic@gmail.com

* Correspondence: kranjcic@sfzg.unizg.hr

“Advanced Dental Materials: From Design to Application—Second Edition” is the title of the second edition of a highly successful Special Issue in the journal *Materials*. When we were initially invited to consider the concept of a Special Issue as guest editors, one of our main ideas was to gather a comprehensive collection of the latest research in the field of dental materials and present it as a unified body of work. That vision has been successfully realized.

Following the unexpected success of the first edition—which culminated in the publication of a printed book under the same title and theme—we made the proud and bold decision to launch a second Special Issue. The dedicated effort and time invested by us, as well as by numerous scientists, clinicians, and esteemed colleagues from around the world (without whom none of this would have been possible), have quickly resulted in another successful volume. This second edition now stands as a coherent and well-defined scientific compilation.

I feel it is important to emphasize the need for the continuous improvement of existing dental materials on the market, as well as the development of new ones. In this context, the role of research in dental materials becomes even more significant, as it not only highlights the outcomes of material design and production but also points to their potential for therapeutic application.

In this spirit, I sincerely hope that this Special Issue will, at least to some extent, contribute to this important field—and, by extension, to the advancement of dentistry as a whole.

This Special Issue features a total of eight high-quality scientific papers published over the course of one year. More than 30 researchers and co-authors from around the world contributed to the publication of their scientific work.

The key themes on which this Special Issue is based include dental materials, ceramics, dental alloys, composite materials, CAD–CAM technology, dental implants, prosthodontics, fixed prosthodontics, removable dental prosthesis base materials, and restorative dentistry.

Matta R.E. et al. (1) contributed to this Special Issue with their article titled “In Vivo Wear Analysis of Leucite-Reinforced Ceramic Inlays/Onlays After 14 Years”; Berger L. et al. (2) contributed with their article titled “Influence of the Milling Strategy on the Marginal Fit of Chairside-Fabricated Lithium Disilicate Crowns”; Strunz A. et al. (3) contributed with their article titled “The Impact of Three-Dimensional Printer Technology on the Accuracy of Dental Implant Models”; Riberti N. et al. (4) contributed with their article titled “Biconometric Connections in Dental Implants: A Pilot Mechanical Study”; Fratila CR. et al. (5) contributed with their article titled “Accuracy Evaluation

of Indirect Bonding Techniques for Clear Aligner Attachments Using 3D-Printed Models: An In Silico and Physical Model-Based Study”; Herrero-Climent M. et al. (6) contributed with their article titled “Mechanical Behavior of PEEK and PMMA Graphene and Ti6Al4V Implant-Supported Frameworks: In Silico Study”; Ceddia M. et al. (7) contributed with their article titled “Biomechanical Finite Element Analysis of Two Types of Short-Angled Implants Across Various Bone Classifications”; Murillo-Gómez et al. (8) contributed with their article titled “Mechanical, Adhesive and Surface Properties of a Zirconia-Reinforced Lithium Silicate CAD/CAM Ceramic Exposed to Different Etching Protocols”. The research findings are presented through a series of tables, graphical illustrations, and images, which further enhance the readability and visual appeal of the articles.

This Special Issue—like the one before it—is rich in new scientific insights and serves as a vital link between research and clinical practice. In this way, the studies conducted and the results published herein gain their full meaning and relevance.

Once again, we extend our sincere gratitude to all contributors for their valuable participation in the creation of this Special Issue. Through their dedicated work, they have helped bring this project to life.

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

List of Contributions:

1. Matta, R.-E.; Berger, L.; Sednyev, O.; Bäuerle, D.; Maier, E.; Adler, W.; Taschner, M. In Vivo Wear Analysis of Leucite-Reinforced Ceramic Inlays/Onlays After 14 Years. *Materials* **2025**, *18*, 3446. <https://doi.org/10.3390/ma18153446>.
2. Berger, L.; Förtsch, F.; Kretschmer, R.R.; Sednyev, O.; Zorzin, J.I.; Wichmann, M.; Matta, R.E. Influence of the Milling Strategy on the Marginal Fit of Chairside-Fabricated Lithium Disilicate Crowns. *Materials* **2025**, *18*, 2184. <https://doi.org/10.3390/ma18102184>.
3. Strunz, A.; Berger, L.; Seidel, A.; Ries, J.; Adler, W.; Wichmann, M.; Matta, R.E. The Impact of Three-Dimensional Printer Technology on the Accuracy of Dental Implant Models. *Materials* **2025**, *18*, 1902. <https://doi.org/10.3390/ma18091902>.
4. Riberti, N.; Furlani, M.; Tognoli, E.; Piattelli, A.; Comuzzi, L.; Giuliani, A.; Gatto, A. Biconometric Connections in Dental Implants: A Pilot Mechanical Study. *Materials* **2025**, *18*, 1415. <https://doi.org/10.3390/ma18071415>.
5. Raluca Fratila, C.; Alonso-Ezpeleta, L.Ó.; Poveda-Saenz, M.; Giovannini, G.; Lobo-Galindo, A.B.; Flores-Fraile, J.; Zubizarreta-Macho, Á. Accuracy Evaluation of Indirect Bonding Techniques for Clear Aligner Attachments Using 3D-Printed Models: An In Silico and Physical Model-Based Study. *Materials* **2025**, *18*, 780. <https://doi.org/10.3390/ma18040780>.
6. Herrero-Climent, M.; Sanchez-Lasheras, F.; Martinez-Lopez, J.; Gil, J.; Brizuela-Velasco, A. Mechanical Behavior of PEEK and PMMA Graphene and Ti6Al4V Implant-Supported Frameworks: In Silico Study. *Materials* **2025**, *18*, 441. <https://doi.org/10.3390/ma18020441>.
7. Ceddia, M.; Romasco, T.; De Bortoli, N., Jr.; Mello, B.F.; Piattelli, A.; Mijiritsky, E.; Di Pietro, N.; Trentadue, B. Biomechanical Finite Element Analysis of Two Types of Short-Angled Implants Across Various Bone Classifications. *Materials* **2024**, *17*, 5680. <https://doi.org/10.3390/ma17235680>.
8. Murillo-Gómez, F.; Hernández-Viquez, J.R.; Sauma-Montes de Oca, J.R.; Vargas-Vargas, C.; González-Vargas, N.; Vega-Baudrit, J.R.; Chavarría-Bolaños, D. Mechanical, Adhesive and Surface Properties of a Zirconia-Reinforced Lithium Silicate CAD/CAM Ceramic Exposed to Different Etching Protocols. *Materials* **2024**, *17*, 5039. <https://doi.org/10.3390/ma17205039>.

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

Article

In Vivo Wear Analysis of Leucite-Reinforced Ceramic Inlays/Onlays After 14 Years

Ragai-Edward Matta ^{1,*}, Lara Berger ^{1,†}, Oleksandr Sednyev ¹, Dennis Bäuerle ², Eva Maier ², Werner Adler ³ and Michael Taschner ²

¹ Department of Prosthodontics, Erlangen University Hospital, Glueckstrasse 11, 91054 Erlangen, Germany; lara.berger@uk-erlangen.de (L.B.); oleksandr.sednyev@uk-erlangen.de (O.S.)

² Department of Operative Dentistry and Periodontology, Erlangen University Hospital, Glueckstrasse 11, 91054 Erlangen, Germany; baeuerle.dennis@hotmail.de (D.B.); eva.maier@uk-erlangen.de (E.M.); michael.taschner@uk-erlangen.de (M.T.)

³ Department of Medical Informatics, Biometry and Epidemiology, Friedrich-Alexander-University of Erlangen-Nuremberg, Waldstrasse 6, 91054 Erlangen, Germany; werner.adler@uk-erlangen.de

* Correspondence: ragai.matta@uk-erlangen.de; Tel.: +49-9131-8533604; Fax: +49-09131-8536781

† These authors contributed equally to this work.

Abstract

Material wear significantly impacts the clinical success and longevity of dental ceramic restorations. This in vivo study aimed to assess the wear behavior of IPS Empress[®] glass-ceramic inlays and onlays over 14 years, considering the influence of different antagonist materials. Fifty-four indirect restorations of 21 patients were available for comprehensive wear analysis, with complete follow-up data for up to 14 years. Three-dimensional measurements relied on digitized epoxy resin models produced immediately post-insertion (baseline) and subsequently at 2, 4, and 14 years. The occlusal region on the baseline model was delineated for comparative analysis. Three-dimensional superimpositions with models from subsequent time points were executed to assess wear in terms of average linear wear and volumetric loss. Statistical analyses were conducted in R (version 4.4.1), employing Mann–Whitney U tests (material comparisons) and Wilcoxon signed rank tests (time point comparisons), with a significance threshold of $p \leq 0.05$. During the entire study period, an increase in wear was observed at each assessment interval, gradually stabilizing over time. Significant differences in substance loss were found between the follow-up time points, both for mean (-0.536 ± 0.249 mm after 14a) and integrated distance ($-18,935 \pm 11,711$ mm³ after 14a). In addition, significantly higher wear was observed after 14 years with gold as antagonist compared to other materials ($p \leq 0.03$). The wear behavior of IPS Empress[®] ceramics demonstrates clinically acceptable long-term outcomes, with abrasion characteristics exhibiting stabilization over time.

Keywords: occlusal interactions; wear of glass-ceramic; wear behavior; all-ceramic restorations; ceramic inlays/onlays; long-term clinical investigation

1. Introduction

Indirect single-tooth ceramic restorations, such as inlays and onlays, offer advantages over direct composite restorations, when restoring larger defects [1–3]. These techniques have been shown to present good long-term survival rates and have proven to be reliable in everyday clinical practice [4,5].

In general, modern dental ceramics are known for their excellent esthetic properties, biocompatibility, and durability [6,7]. Glass-ceramics have color properties similar to

human enamel and are particularly translucent due to their high glass content; although, this can lead to an impairment of mechanical properties. These mechanical limitations are partially compensated by an increased crystalline content [8].

With the market launch of IPS Empress® from Ivoclar Vivadent AG (Schaan, Liechtenstein), an attempt was made to optimize mechanical properties by stabilizing the glass matrix through integration of leucite crystals and thus to develop a highly esthetic yet stable restorative material [9]. In this context, a review on longevity and clinical performance showed a survival rate of up to 91% for IPS Empress® inlays and onlays after seven years, which underlines the reliability and durability of this material [10].

However, long-term success is limited by factors such as secondary caries, adhesive failure, and fractures [11]. In addition, the experience of the dentist and the correct indication for the restoration of lost dental hard tissue are decisive influencing parameters for long-term outcomes [12].

In this respect, wear mechanisms are determined by a variety of influencing variables, including individual masticatory forces, the shape and size of tooth contact surfaces, and the surface texture, microstructure, and fracture strength of the material [13]. The hardness of the material proves to be less decisive, while the surface roughness and fracture toughness of the ceramic are of greater importance [14]. In principle, the occlusal wear properties of the antagonists should be comparable to ensure long-term functionality without structural deficits [15–17].

Concerning tribology, various primary mechanisms are described with regard to wear processes. These include chemical dissolutive transformations, in which material is removed by chemical reactions, as well as adhesive wear, which is caused by adhering material particles that are subsequently torn off at the contact surfaces. Other parameters include surface fatigue, which occurs when repeated stresses lead to microcracks and consequential failure of the material, or mechanical abrasion, when material is worn away by particles of variable hardness [18,19]. In this context, patient-specific factors, such as bruxism, forced tooth brushing, diet, or salivary flow, also play a central role [13,20,21]. Various systems are available for tribological investigations, with both two- and three-body wear testing systems frequently used to analyze the wear resistance of materials under realistic conditions [18,22,23]. To evaluate the results of these laboratory testing approaches, industrial light–optical scanners enable contactless and highly precise object registration in form of three-dimensional (3D) datasets, which can then be virtually aligned with each other and precisely superimposed or aligned using special software [24–26].

According to an *in vitro* study, leucite-reinforced glass-ceramics exhibit improved abrasion resistance [27]. Zirconia ceramics are characterized by lower wear rates compared to lithium disilicate ceramics, which consequently results in reduced material loss [28]. Compared to lithium disilicate ceramics, gold restorations show a higher abrasiveness *in vitro*, especially with IPS Empress® ceramics [29]. While there is a vast body of work regarding *in vitro* wear behavior of IPS Empress® [28], recent literature lacks conclusive evidence addressing the abrasion behavior of dental ceramics against various antagonist, particularly IPS Empress® in clinical context [30]. Therefore, laboratory studies conducted under defined and controlled conditions cannot simulate all the individual and complex factors of the chewing process [16,31]. This implies the need for long-term clinical studies to record the wear behavior under real conditions and thus evaluate the actual performance of different materials.

The aim of this *in vivo* study was, therefore, to quantitatively determine the occlusal wear rates of IPS Empress® glass-ceramic inlays and onlays in available patients over follow-up intervals of two, four, and fourteen years correlated with the respective antagonist

material in order to gain a comprehensive understanding of the long-term effects and clinical performance of this material.

2. Materials and Methods

2.1. Study Design

This prospective clinical study aimed to evaluate the wear behavior of leucite-reinforced glass-ceramic restorations using a light-optical 3D scanning technique by digitally comparing virtual patient models of the intraoral situations after 2, 4, and a total observation period of 14 years. The study project was approved in advance by the Ethics Committee of the University Hospital Erlangen (No. 2628) and conducted in accordance with EN 540 (Clinical investigation of medical devices in humans, European Committee for Standardization), the CONSORT declaration [32], and established protocols [33,34].

2.2. Subject Recruitment and Intervention

An experienced assistant professor treated 30 selected patients (average age 39.4 years; age ranging between 23–64 years; eleven men, nineteen women) at the Department of Conservative Dentistry and Periodontology at the University Hospital Erlangen according to defined inclusion criteria (Table 1 [35,36]). All participants gave their written informed consent, agreed to the study procedure, and committed to regular check-ups in the first 4 years after placement of the restoration; participation in the 14-year check-up was voluntary.

Table 1. Conditions of the study population.

Inclusion Criteria
- Symptom-free tooth for restoration
- Rubber dam application possible
- No further treatment needed for other posterior teeth
- Good oral hygiene
- No periodontal or pulpal disease
- Age 18–65
- No deep cavities in close proximity to the pulp

At the beginning of the study, a total of 83 IPS Empress® restorations were placed in 30 patients, consisting of 31 two-surface inlays, 39 MOD (mesial-occlusal-distal) inlays, and 13 onlays, which were distributed over 34 upper premolars, 14 upper molars, 13 lower premolars, and 22 lower molars.

Cavity preparations followed standardized protocols to ensure consistent quality and geometry, as described in previous publications [35,36]. Cavities close to the pulp were excluded to avoid adhesion reduction by calcium hydroxide. After preparation, two-phase polyvinylsiloxane impressions were taken (Dimension Garant H and Garant L, 3M, Seefeld, Germany), and chairside-fabricated temporary restorations (Luxatemp, DMG, Hamburg, Germany) were fixed with TempBond™ NE (Kerr, Herzogenrath, Germany). The leucite-reinforced glass-ceramic restorations were produced in a commercial dental laboratory according to manufacturer's instructions [37].

To ensure that the maximum cement gap width of 100 µm was not exceeded, a special dental probe was used (tip diameter 100 µm, tactile control). Before insertion, the thickness of the restorations was measured with a caliper (rapid caliper, accuracy 0.01 mm, Kroeplin, Schluechtern, Germany) to check a sufficient minimum layer thickness at the relevant points. Afterwards, the inlays and onlays were adhesively bonded according to the clinical instructions for the pre-treatment of tooth and restoration [36]. After removing excess insertion material and the rubber dam, occlusal contacts were adjusted with fine diamond

drills. Final polishing was performed with diamond and polishing strips interdentally (GC Dental Industrial Corp., Tokyo, Japan/3M), while the occlusal finish was achieved using felt discs (Dia-Finish E felt discs, Renfert, Hilzingen, Germany) and diamond polishing paste (Brinell, Renfert).

The follow-up examinations of the restorations were carried out at the following four points in time: after 2 weeks (=baseline) and after 2, 4 and 14 years. Double-mix impressions (Panasil, Kettenbach GmbH und Co. KG, Eschenburg, Germany) were taken, subsequently filled with epoxy resin (AlphaDie MF, Schütz Dental GmbH, Rosbach, Germany) and stored as precision models in the Dental Clinic 1—Conservative Dentistry and Periodontology—of the University Hospital Erlangen, Germany. In addition, the respective antagonists (enamel, composite, ceramic, gold) were recorded by means of photography (Figure 1).

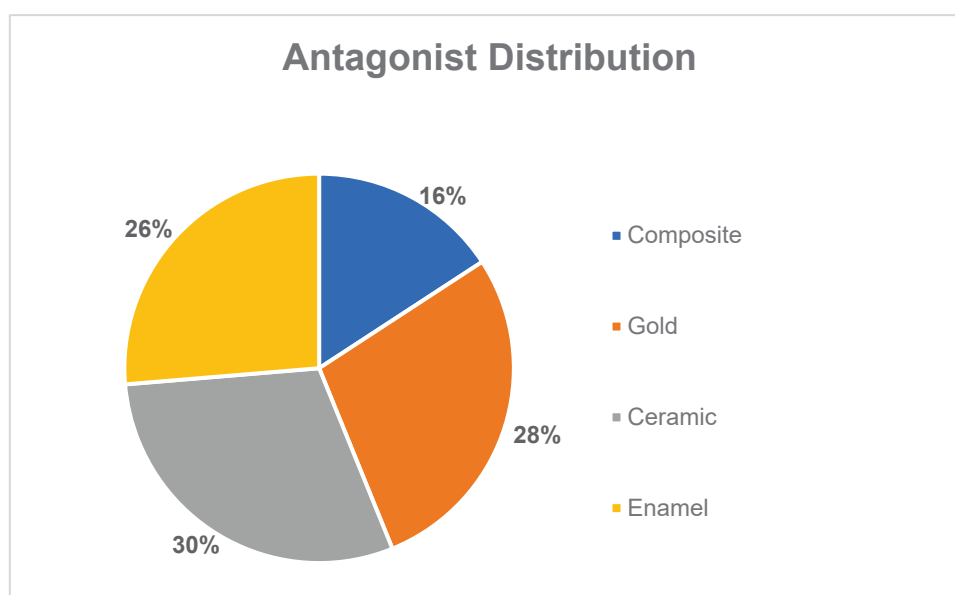


Figure 1. Pie chart of the antagonist distribution in percentage.

2.3. Three-Dimensional Data Acquisition and Volumetric Measurements

In preparation for the subsequent digitization processes, all physical patient models were equipped with ATOS 0.8 mm reference points (GOM GmbH, Braunschweig, Germany) to enable spatial orientation and thus linking areas of the individual measurements to a 3D overall representation of the object. In view of the subsequent evaluations, these were only attached to non-relevant areas, i.e., not to restorations and not to areas necessary for the virtual superimposition process. Furthermore, to improve the scan quality, matting was carried out using a titanium dioxide solution applied with an airbrush gun (RICH[®] AS-2, Fuso Seiki Co., Ltd., Tokyo, Japan), of which the minimum layer thickness can be considered negligible concerning measurements [38,39].

An industrial, high-precision non-contact scanner (ATOS So4 II, GOM GmbH) was used to digitize the models, as it has a smaller measurement error of less than 0.004 mm in the course of object registration compared to conventional dental laboratory scanners for the measurement volume used here [40]. The ATOS So4 II system was calibrated at the beginning of each measurement sessions in accordance with the manufacturer's instructions by a specifically manufacturer-trained employee of our faculty. All measurements were performed by a trained and experienced user. Based on the triangulation principle, precise light points were projected onto the object to be measured, reflected, and recorded by two light-sensitive sensors arranged at defined angles [41]. Several individual measurements per patient model from different perspectives resulted in a measurement

series of point clouds, which could be transformed into a common coordinate system using the reference points (GOM GmbH) attached in advance. In this way, four precise virtual 3D models were generated for each patient case in the common.stl (Surface/Standard Triangulation/Tessellation Language) format. Afterwards, the scanner's analysis software (GOM Inspect Professional 2018, GOM GmbH) was used to perform a digital data analysis by comparing (=matching) the occlusal surfaces of the restorations to be examined (=region of interest, ROI) between the models of the follow-up examinations (after 2, 4 and 14 years; =IST models) and the baseline model (after 2 weeks; =SOLL model) in order to evaluate the occlusal material loss of the restorations both numerically and graphically. In the course of the respective target–actual superimposition for the before-and-after comparisons at different time points, the model pairs were first roughly aligned manually by selecting prominent points in the central fissures or on the cusp tips. In a second step, the surfaces of the teeth adjacent to the restorations were selected, and an automated fine alignment was performed using a “best fit” algorithm.

This standardized procedure ultimately enabled highly precise recording of the occlusal volume changes in the form of color-coded distance maps to visualize local differences due to wear processes (Figure 2). In addition, a numerical calculation of the average surface loss (“mean distance” in [mm]) and the volume decrease (“integrated distance” in [mm³]) over time was carried out.

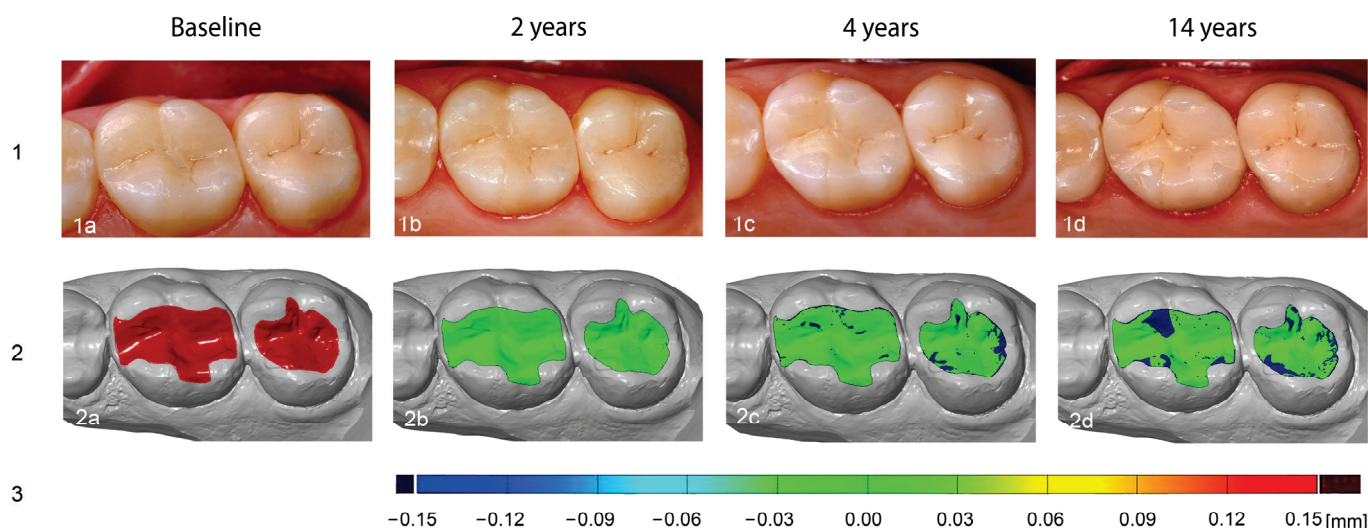


Figure 2. Clinical and virtual 3D visualization of occlusal wear over a 14-year period. Series (1a–1d): Clinical situations in an occlusal view. Series (2): Corresponding 3D surface models. (2a): Baseline model with defined region of interest (ROI). (2b–2d): Superimpositions of follow-up models with baseline. Color-coded distance maps illustrate material loss, with blue denoting wear and green indicating areas with no or almost no surface loss. Series (3): Color-coded scale bar to quantify surface changes.

2.4. Statistical Data Analysis

Statistical analyses to evaluate the wear on the IPS Empress® restorations in relation to the respective antagonist class were conducted using the R programming language (version 4.4.1) [42]. Mann–Whitney U tests (comparing different materials at the same time point) and Wilcoxon signed rank tests (comparing different time points within one material) were used to evaluate the data. The significance level was set at $p \leq 0.05$. An overview of the study procedure is shown in Figure 3.

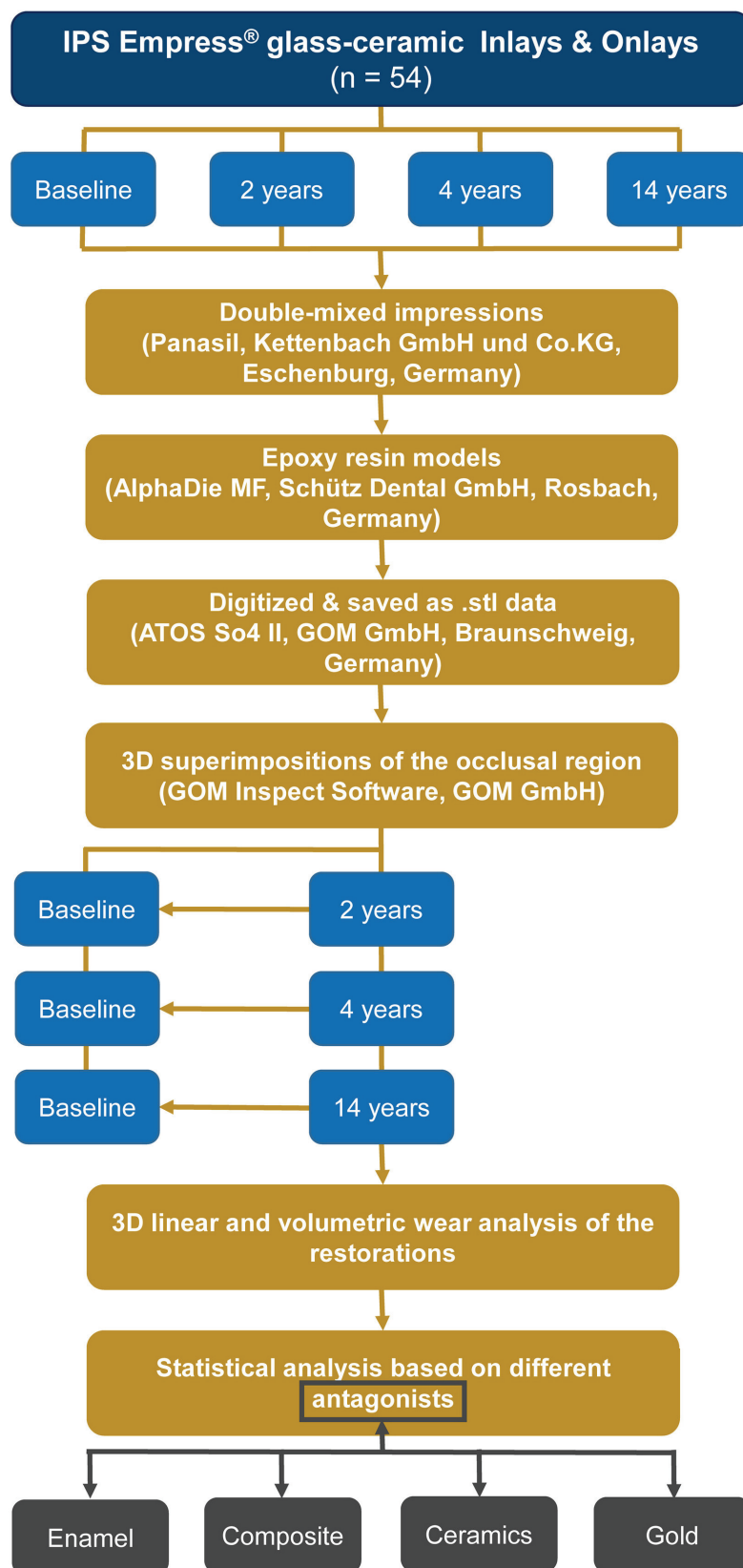


Figure 3. Flowchart illustrating the study design.

3. Results

3.1. Study Population

The investigations were based on digital patient models at the various follow-up times, with a total number of 83 IPS Empress® inlays/onlays (baseline, $n = 83$) in 30 patients at the beginning. Over time, the number decreased so that after 2 years, a total of 82 restorations (98.8%); after 4 years, 72 inlays and onlays (86.75%); and after a period of 14 years, 54 IPS Empress® glass-ceramic restorations (65%) were still included in the study. The reasons were failure (16.9%), change in residence, or death (18.1%), with a failure rate of 12% according to the Kaplan–Meier analysis [36]. During the last follow-up examination, the sample consisted of 21 patients, including 13 women and 8 men aged between 24 and 66 years (mean: 40 years).

3.2. Substance Loss Analysis Methodology

The results for the evaluation of the occlusal wear rates were derived exclusively from negative measured values up to a maximum limit of 0.00 mm, which in turn would not correspond to any substance loss. These negative values illustrate the extent of height loss of the restoration as a result of the effect of the corresponding antagonists.

3.3. Average Surface Loss

Over the course of the entire study period, the average surface loss of the IPS Empress® glass-ceramic inlays and onlays showed a progressive increase, which occurred independently of the antagonist material present. Two years after placement of the restorations, there was a total loss of -0.253 ± 0.185 mm; after four years, -0.380 ± 0.205 mm; and after fourteen years, -0.536 ± 0.249 mm. The comparison of the different follow-up times (2 vs. 4 years/2 vs. 14 years/4 vs. 14 years) within the same antagonist group (enamel, composite, ceramic, gold) provided consistently significant results ($p \leq 0.004$). In contrast, the comparison of the different antagonists with regard to abrasion on the ceramic restorations showed no significant differences after 2, 4, and 14 years (Figure 4, Table 2).

Table 2. p -values for comparisons of average surface loss (mean distance) and volume decrease (integrated distance) depending on different antagonists at one point in time and between different points in time within an antagonist material.

Mean Distance [mm]			
Material comparison	2 years	4 years	14 years
Enamel vs. Composite	0.27	0.421	0.743
Enamel vs. Ceramic	0.216	0.767	0.333
Enamel vs. Gold	0.412	1.000	0.709
Composite vs. Ceramic	0.955	0.777	0.777
Composite vs. Gold	0.858	0.421	0.411
Ceramic vs. Gold	0.752	0.922	0.202
Time point comparison	2 vs. 4 years	2 vs. 14 years	4 vs. 14 years
Enamel	<0.001	<0.001	0.001
Composite	0.004	0.004	0.004
Ceramic	<0.001	<0.001	<0.001
Gold	<0.001	<0.001	<0.001
Integrated Distance [mm ³]			
Material comparison	2 years	4 years	14 years
Enamel vs. Composite	0.194	0.411	0.421
Enamel vs. Ceramic	0.163	0.626	0.358
Enamel vs. Gold	0.806	0.233	0.026
Composite vs. Ceramic	0.978	0.760	0.760
Composite vs. Gold	0.482	0.108	0.025
Ceramic vs. Gold	0.304	0.140	0.030

Table 2. Cont.

Time point comparison	2 vs. 4 years	2 vs. 14 years	4 vs. 14 years
Enamel	0.002	<0.001	0.004
Composite	0.004	0.004	0.020
Ceramic	<0.001	<0.001	0.005
Gold	<0.001	<0.001	0.003

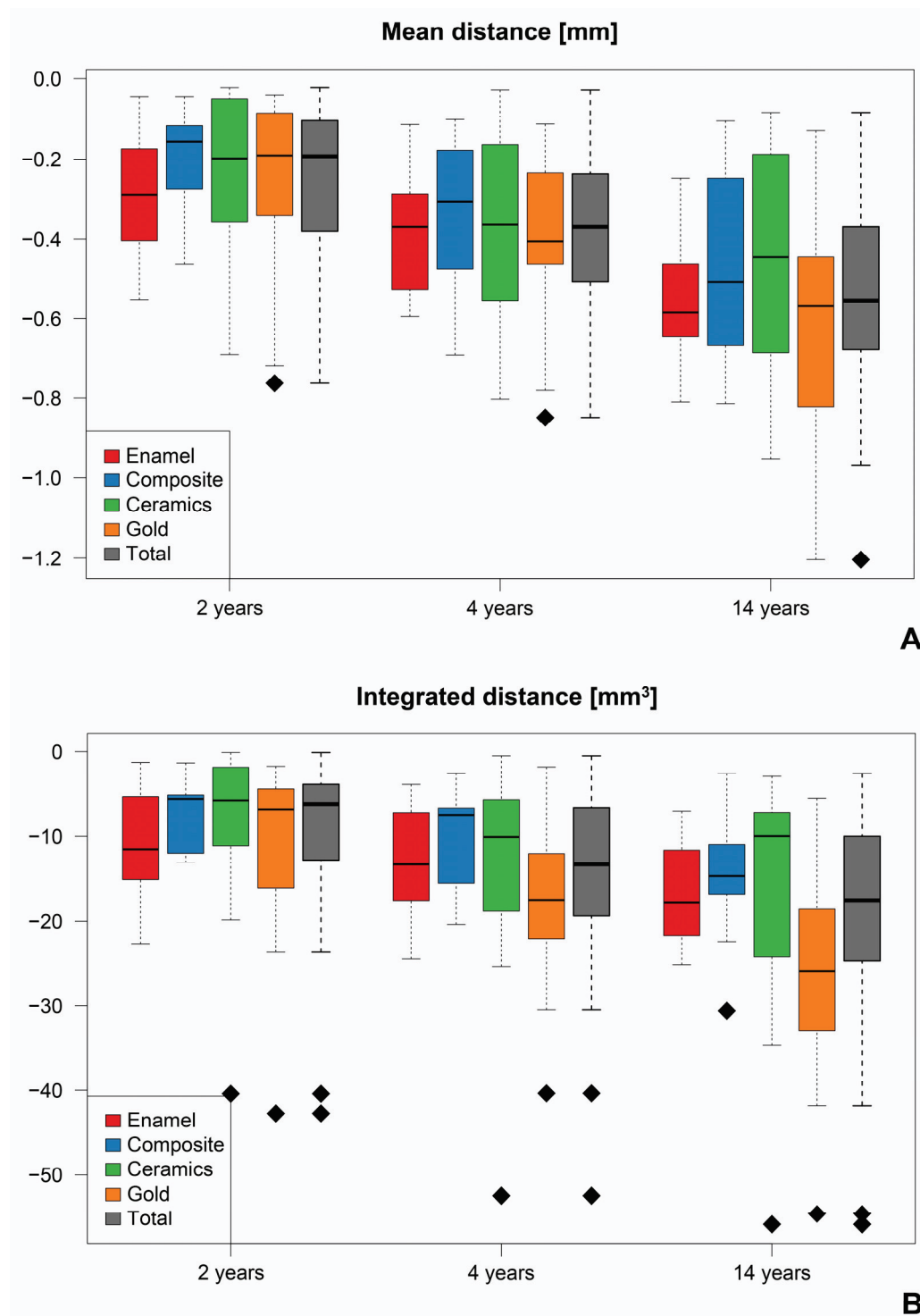


Figure 4. Boxplot diagrams depicting the occlusal wear of the ceramic restoration in terms of average surface loss (mean distance) (A) and volume decrease (integrate distance) (B) after 2, 4, and 14 years. The black rhombus symbols indicate outliers.

3.4. Volume Decrease

The occlusal volume of the inlays and onlays also decreased over time, resulting in a total volume loss of $-9981 \pm 8965 \text{ mm}^3$ after 2 years, $-14,207 \pm 9753 \text{ mm}^3$ after 4 years, and $-18,935 \pm 11,711 \text{ mm}^3$ after 14 years. The comparative analysis of the integrated distance between the time points 2 and 4 years, 2 and 14 years, and 4 and 14 years after final placement of the ceramic restorations showed consistently significant results for all antagonists, analogous to the average surface loss ($p \leq 0.02$). Furthermore, a significantly higher material loss was determined after 14 years if gold was present as an antagonist ($p < 0.03$) (Figure 4, Table 2).

4. Discussion

In this study, a 3D surface analysis was conducted to evaluate the abrasive properties of 54 IPS Empress® restorations over an observation period of 14 years in relation to different antagonists. In this respect, long-term clinical studies are essential in order to be able to adequately assess the suitability of dental ceramics for long-term use in patients and thus for clinical success [43]. The physical properties of the ceramic are of particular importance in order to withstand abrasive challenges in the oral cavity [44] and ultimately ensure satisfactory long-term performance for both the patient and the practitioner [10].

Many of the studies on abrasion behavior of leucite-reinforced glass-ceramic inlays found in the literature are in vitro studies that evaluate the wear of the antagonistic tooth and not of the restoration itself [45,46]. However, the quantitative comparison of in vitro and in vivo results shows a high correlation of the data and the associated relevance of laboratory studies for clinical performance [47]. Nevertheless, clinical studies are considered the most valuable method for determining wear behavior [17,48].

In the present analysis, the abrasion on the restorations of 13 female and 8 male test subjects aged 24 to 66 years was analyzed over the entire observation period and thus with a heterogeneous gender and age distribution. In this regard, Ohlmann et al. [49] showed in their study investigating the abrasive behavior of posterior crowns made of different materials using 3D laser scanning technology that neither the age nor the gender of the patient collective had a significant influence on wear.

Overall, the results obtained show that a continuous and significant increase in surface and volume loss occurred in IPS Empress® glass-ceramic inlays and onlays up to 14-years in vivo, irrespective of the antagonist material used. The comparison of the measured values for abrasion in the occlusal contact area between the individual examination intervals showed significant differences for both the mean distance and integrated distance compared to the individual antagonist groups, with the abrasion rate being more pronounced in the first few years and then decreasing ($p \leq 0.02$). Over the course of the 2-, 4-, and 14-year follow-up periods, no statistically significant differences in restoration wear were observed between the various antagonist materials, with the exception of a significantly greater volumetric material loss associated with gold antagonists at the 14-year recall ($p < 0.03$) (Figure 2, Table 2). These findings indicate that composite, ceramic, and enamel antagonists exhibit comparable long-term wear behavior when opposing IPS Empress® restorations. Accordingly, both ceramic and composite antagonists may be regarded as suitable and clinically favorable counterparts for IPS Empress® restorations in long-term applications. In view of the fact that natural and healthy enamel is subject to a continuous annual abrasion process of around 20 to 40 μm in the molar region even under physiological conditions [13,15], the loss of material on the occlusal surfaces of the restorations observed in this study after just two years can be considered appropriate. The FDI clinical criteria specify that restorations exhibiting wear within 80–120% of the enamel wear are deemed harmonious, while wear substantially exceeding the natural enamel wear—exceeding

300 µm occlusally—is regarded as problematic [50]. Thus, if a restoration's wear mimics enamel wear in both its rate and pattern, it is considered clinically acceptable even if some volume loss or subtle changes in contour become apparent over time. The fact that abrasion processes occur on ceramic restorations after a short time has already been confirmed *in vivo* by Aladağ et al. [48] using 3D data overlays. An average volumetric abrasion of $0.27 \pm 0.16 \text{ mm}^3$ was evaluated on lithium disilicate molar crowns after only six months. However, it must be noted that the hardness of a material is not the only determining factor for wear. The masticatory loads and the associated wear processes in the oral cavity represent a complex, multifactorial phenomenon that is influenced by numerous individual factors [51]. In the study by Aladağ et al. [48], patient-specific factors that influence the complex intraoral abrasion process and the associated material wear [13,15] were not documented, as is the case in the present study, which should be seen as a limitation. The variability of these factors, such as chewing pattern, saliva composition, and individual dietary habits, could affect the interpretation and transferability of the results. However, the relationship between maximum biting force and the wear rates of enamel and all-ceramic crowns, for example, was investigated in a clinical study, but no significant correlation was found [51]. Furthermore, other authors point out that the additional recording of food consumption over such a long observation period, which in the present study would cover more than 10% of a person's average lifespan, is not considered realistic [34].

Krejci et al. [52] analyzed the abrasion of MOD inlays made of four different ceramic materials on a total of 24 extracted, caries-free molars using chewing simulation. Palatal cusps of human maxillary first molars of similar shape and dimensions, which showed no existing abrasion, were used as antagonists. The investigations were carried out in a computer-controlled chewing simulator with six chambers, which simulated real chewing processes and *in vivo* load over five years. The samples were subjected to several factors, including chemical degradation (75% ethanol at 37 °C), mechanical toothbrush abrasion (2 N), cyclic masticatory loads (max. 49 N at 1.7 Hz), and thermal cycles in water with alternating temperatures of 5 °C and 55 °C. A material loss of $-21.8 \pm 8.8 \text{ µm}$ was measured in the occlusal contact point area for polished IPS Empress® restorations. In contrast, significantly higher values in the tenths of a millimeter range were recorded in the present study after only two years. This can be explained by the fact that the present study was conducted under real *in vivo* conditions, whereas Krejci et al. [52] conducted their study *in vitro* with only six test objects per group and enamel as the only antagonist, which limits the comparability of the results. In addition, not only molars but also premolars were included in the analyses in the current study. However, it has already been shown that there is no direct correlation between the loss of substance and the specific location of the inlays in the oral cavity [34].

In contrast, Krämer et al. [34] conducted a clinical study to evaluate the wear phenomena of 17 ceramic inlays on premolars and molars made of IPS Empress® against enamel antagonists over a period of 8 years. The average occlusal wear of the ceramic inlays analyzed using the 3D evaluation method was 78 µm after 4 years and 116 µm after 8 years, with the difference between the two time points being statistically significant. While the measured values are lower compared to the results of the present study, but tend to be comparable, the data of the current analysis also show significant differences between the examination intervals. This indicates progressive wear over time; although, this appears to be stabilizing.

In this regard, Lambrechts et al. [15] reported a tendency towards higher wear rates in the first few years (initial phase), which subsequently stabilize as the occlusal environment approaches a dynamic equilibrium (steady-state phase), provided no further restorative measures are taken. Consequently, the wear process shows a non-linear progression, as the

initially small contact areas can lead to high pressure concentrations, which in turn cause an increased wear rate [52,53]. This is consistent with the results of the study conducted, as after only 2 years, depending on the antagonist group, around 40–60% of the total material loss observed after 14 years occurred.

In addition, significantly higher wear was observed on the ceramic restorations after 14 years when gold was used as an antagonist ($p \leq 0.03$). This contrasts with the observations of a different *in vitro* study, which showed that gold in particular, and less so ceramic, exhibited accelerated wear when in contact with each other. However, the authors also describe a “functional polishing” that occurs on ceramic surfaces at the beginning of a masticatory load and removes the rough surface layers. Thus, progressive polishing of the ceramic leads to a gradual decrease in the abrasion rate. However, when gold was present as an antagonist, little evidence of these polishing mechanisms of the ceramic surface was documented, and instead, rapid wear of the gold was observed. Furthermore, gold particles were observed to transfer to the ceramic surfaces over time [53]. These microparticles could act as abrasives and thus lead to increased abrasion processes, as the surface roughness is directly related to the abrasion rate [29,53,54].

In contrast to the gold antagonists, the IPS Empress® inlays and onlays showed no significant differences in abrasion behavior compared to composite, ceramic, and natural enamel at the individual follow-up times. This indicates a comparable wear resistance and confirms the compatibility of these material classes (with the exception of gold) for long-term clinical use.

Overall, there are only a few *in vivo* studies in the literature that prospectively document the vertical dimensional loss of ceramic restorations, particularly inlays and onlays, over a longer period of time [34,55], which limits the comprehensive comparability of the results. The focus of this analysis was on the abrasion of ceramic restorations as a function of different antagonists, which has not yet been investigated in this form.

Although mechanical profilometry was previously described as a reliable method for determining surface changes [56], the present study is based on a modern digital 3D laser scanning method (ATOS So4 II, GOM GmbH) of replica models. The choice of measurement method significantly influences the accuracy and reliability of the results [17,57]. The ATOS system used generates high-precision and non-contact digitally recorded measurement data that can be analyzed quantitatively and qualitatively reproducibly in both two- and three-dimensional form [58]. In general, the use of 3D measurement techniques for the visualization of surface changes has become established in scientific research, as the smallest changes can be recorded with the highest precision and attention to detail [17,34,48,51]. Double-mix impressions, which were taken at each examination time and poured with epoxy resin, served as the basis for the measurements carried out. Both materials have been shown to be characterized by high detail accuracy, dimensional stability, and storage stability, thus enabling the precise reproduction of the intraoral structures [59,60]. Alternatively, plaster could have been used for model fabrication. However, the detail reproduction would have been limited to 20 microns with Class IV plaster compared to epoxy resin, which reveals structural details down to one micron [60]. Another option would have been digital impression taking using intraoral scanning, which has been shown to offer comparable accuracy to conventional methods and eliminates the need for physical model fabrication but was not available at the beginning of the present study. However, data capture is limited to visible areas, which can be a challenge in hard-to-reach regions. In addition, blood, saliva, and other patient-specific factors can affect image quality, while precision may be lower than with conventional techniques in large dental arches [61]. Furthermore, a recent *in vitro* study showed that an industrial precision scanner from

GOM GmbH generates the highest precision in object registration compared to laboratory scanners, photogrammetry devices, and intraoral scanners [62].

Overall, it should be mentioned that the analysis regarding the effect of different antagonist materials was not the focus of the study design when the present investigation was initiated, hence a control group is missing. At that point in time, the influence of different antagonists on wear of restorations was not yet a focus in the scientific literature. The lack of patient-specific factors and the comparably small sample size present further limitations within the scope of this study. Nevertheless, our data reflect real clinical conditions, in which a large number of different antagonist materials naturally occur.

Since the wear of dental materials is subject to numerous influencing factors and mechanisms, further data from long-term clinical studies are essential for a sound understanding of the relationships between wear rate and relevant parameters [47]. Future research should use a standardized method to investigate wear, such as the 3D analysis used in this study, in order to improve the comparability of the data to the already complex mechanisms of material wear. In addition, research should be conducted into potential selection tools based on artificial intelligence (AI) that take antagonists into account.

5. Conclusions

Conclusively, the evaluated results indicate that IPS Empress® ceramic restorations exhibit adequate wear resistance according to the FDI, which is stabilized due to superficial polishing processes, and can, therefore, be recommended for long-term clinical use.

Concerning the long-term success of restorative materials, such as the glass-ceramic inlays and onlays examined in this study, it is important to consider comparable wear properties of the opposing antagonists, to ensure the long-term integrity and functionality of dental restorations.

Author Contributions: Conceptualization, R.-E.M. and L.B.; methodology, R.-E.M., D.B. and M.T.; software, L.B. and R.-E.M.; validation, L.B. and E.M.; formal and statistical analysis, W.A.; investigation, D.B.; resources, M.T. and R.-E.M.; data curation, D.B.; writing—original draft preparation, L.B. and O.S.; writing—review and editing, L.B. and O.S.; visualization, E.M. and L.B.; supervision, L.B., R.-E.M. and M.T.; project administration, R.-E.M. and M.T.; funding acquisition, R.-E.M. and M.T. All authors have read and agreed to the published version of the manuscript.

Funding: This research was supported up to the 4-year recall by 3M Espe (Seefeld, Germany). E.M. was supported by the Interdisciplinary Center for Clinical Research at the University Hospital of the University of Erlangen, Nuremberg (Junior Project J103).

Institutional Review Board Statement: The study was conducted in accordance with the Declaration of Helsinki and approved by the Ethics Committee of the University Hospital Erlangen (No. 2628, 13 May 2002).

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study. Written informed consent has been obtained from the patients to publish this paper.

Data Availability Statement: The raw data supporting the conclusions of this article will be made available by the authors on request.

Acknowledgments: Parts of the results of the present study were incorporated into Dennis Bäuerle's doctoral thesis at Friedrich Alexander University Erlangen, Nuremberg.

Conflicts of Interest: The authors declare that this study received funding from 3M Espe (Seefeld, Germany). The funder was not involved in the study design, collection, analysis, interpretation of data, the writing of this article or the decision to submit it for publication.

Abbreviations

The following abbreviations are used in this manuscript:

3D	three-dimensional
EN	European norm
MOD	Mesial–occlusal–distal
ROI	region of interest
CONSORT	Consolidated Standards of Reporting Trials

References

1. Meyer, A., Jr.; Cardoso, L.C.; Araujo, E.; Baratieri, L.N. Ceramic inlays and onlays: Clinical procedures for predictable results. *J. Esthet. Restor. Dent.* **2003**, *15*, 338–351; discussion 352. [CrossRef] [PubMed]
2. Hopp, C.D.; Land, M.F. Considerations for ceramic inlays in posterior teeth: A review. *Clin. Cosmet. Investig. Dent.* **2013**, *5*, 21–32. [CrossRef] [PubMed]
3. Al-Fouzan, A.F.; Tashkandi, E.A. Volumetric measurements of removed tooth structure associated with various preparation designs. *Int. J. Prosthodont.* **2013**, *26*, 545–548. [CrossRef] [PubMed]
4. Collares, K.; Corrêa, M.B.; Laske, M.; Kramer, E.; Reiss, B.; Moraes, R.R.; Huysmans, M.C.; Opdam, N.J. A practice-based research network on the survival of ceramic inlay/onlay restorations. *Dent. Mater.* **2016**, *32*, 687–694. [CrossRef] [PubMed]
5. Forna, N.C.; Uleanu, R.; Lungu, I.; Dimofte, A.-R. Dental ceramics: Advantages and benefits. *Practice* **2024**, *4*, 7.
6. Contrepolis, M.; Soenen, A.; Bartala, M.; Laviolle, O. Marginal adaptation of ceramic crowns: A systematic review. *J. Prosthet. Dent.* **2013**, *110*, 447–454.e410. [CrossRef] [PubMed]
7. Kelly, J.R.; Benetti, P. Ceramic materials in dentistry: Historical evolution and current practice. *Aust. Dent. J.* **2011**, *56* (Suppl. 1), 84–96. [CrossRef] [PubMed]
8. Sen, N.; Us, Y.O. Mechanical and optical properties of monolithic CAD-CAM restorative materials. *J. Prosthet. Dent.* **2018**, *119*, 593–599. [CrossRef] [PubMed]
9. Albakry, M.; Guazzato, M.; Swain, M.V. Influence of hot pressing on the microstructure and fracture toughness of two pressable dental glass-ceramics. *J. Biomed. Mater. Res. B Appl. Biomater.* **2004**, *71*, 99–107. [CrossRef] [PubMed]
10. El-Mowafy, O.; Brochu, J.F. Longevity and clinical performance of IPS-Empress ceramic restorations—A literature review. *J. Can. Dent. Assoc.* **2002**, *68*, 233–237. [PubMed]
11. Abdulrahman, S.; Von See Mahm, C.; Talabani, R.; Abdulateef, D. Evaluation of the clinical success of four different types of lithium disilicate ceramic restorations: A retrospective study. *BMC Oral Health* **2021**, *21*, 625. [CrossRef] [PubMed]
12. Manhart, J.; Chen, H.; Hamm, G.; Hickel, R. Buonocore Memorial Lecture. Review of the clinical survival of direct and indirect restorations in posterior teeth of the permanent dentition. *Oper. Dent.* **2004**, *29*, 481–508. [PubMed]
13. Yip, K.H.; Smales, R.J.; Kaidonis, J.A. Differential wear of teeth and restorative materials: Clinical implications. *Int. J. Prosthodont.* **2004**, *17*, 350–356. [PubMed]
14. Sripetchdanond, J.; Leevailoj, C. Wear of human enamel opposing monolithic zirconia, glass ceramic, and composite resin: An in vitro study. *J. Prosthet. Dent.* **2014**, *112*, 1141–1150. [CrossRef] [PubMed]
15. Lambrechts, P.; Braem, M.; Vuylsteke-Wauters, M.; Vanherle, G. Quantitative in vivo wear of human enamel. *J. Dent. Res.* **1989**, *68*, 1752–1754. [CrossRef] [PubMed]
16. Dupriez, N.D.; von Koeckritz, A.K.; Kunzelmann, K.H. A comparative study of sliding wear of nonmetallic dental restorative materials with emphasis on micromechanical wear mechanisms. *J. Biomed. Mater. Res. B Appl. Biomater.* **2015**, *103*, 925–934. [CrossRef] [PubMed]
17. de Carvalho Ramos, N.; Augusto, M.G.; Alves, L.M.M.; Kleverlaan, C.J.; Dal Piva, A.M.d.O. Wear of dental ceramics. *Braz. Dent. Sci.* **2023**, *26*, e3638. [CrossRef]
18. Osiewicz, M.A.; Werner, A.; Roeters, F.J.M.; Kleverlaan, C.J. Wear of direct resin composites and teeth: Considerations for oral rehabilitation. *Eur. J. Oral Sci.* **2019**, *127*, 156–161. [CrossRef] [PubMed]
19. Schlueter, N.; Amaechi, B.T.; Bartlett, D.; Buzalaf, M.A.R.; Carvalho, T.S.; Ganss, C.; Hara, A.T.; Huysmans, M.; Lussi, A.; Moazzez, R.; et al. Terminology of Erosive Tooth Wear: Consensus Report of a Workshop Organized by the ORCA and the Cariology Research Group of the IADR. *Caries Res.* **2020**, *54*, 2–6. [CrossRef] [PubMed]
20. Lobbezoo, F.; Ahlberg, J.; Raphael, K.G.; Wetselaar, P.; Glaros, A.G.; Kato, T.; Santiago, V.; Winocur, E.; De Laat, A.; De Leeuw, R.; et al. International consensus on the assessment of bruxism: Report of a work in progress. *J. Oral Rehabil.* **2018**, *45*, 837–844. [CrossRef] [PubMed]
21. Dahl, B.L.; Carlsson, G.E.; Ekfeldt, A. Occlusal wear of teeth and restorative materials. A review of classification, etiology, mechanisms of wear, and some aspects of restorative procedures. *Acta Odontol. Scand.* **1993**, *51*, 299–311. [CrossRef] [PubMed]

22. Maier, E.; Grottschreiber, C.; Knepper, I.; Opdam, N.; Petschelt, A.; Loomans, B.; Lohbauer, U. Evaluation of wear behavior of dental restorative materials against zirconia in vitro. *Dent. Mater.* **2022**, *38*, 778–788. [CrossRef] [PubMed]
23. Someya, T.; Kasahara, M.; Takemoto, S.; Hattori, M. The Wear Behavior of Glass-Ceramic CAD/CAM Blocks against Bovine Enamel. *Materials* **2023**, *16*, 6839. [CrossRef] [PubMed]
24. Holst, S.; Karl, M.; Wichmann, M.; Matta, R.E. A technique for in vitro fit assessment of multi-unit screw-retained implant restorations: Application of a triple-scan protocol. *J. Dent. Biomech.* **2012**, *3*, 1758736012452181. [CrossRef] [PubMed]
25. Jeon, J.H.; Kim, H.Y.; Kim, J.H.; Kim, W.C. Accuracy of 3D white light scanning of abutment teeth impressions: Evaluation of trueness and precision. *J. Adv. Prosthodont.* **2014**, *6*, 468–473. [CrossRef] [PubMed]
26. Jeon, J.H.; Choi, B.Y.; Kim, C.M.; Kim, J.H.; Kim, H.Y.; Kim, W.C. Three-dimensional evaluation of the repeatability of scanned conventional impressions of prepared teeth generated with white- and blue-light scanners. *J. Prosthet. Dent.* **2015**, *114*, 549–553. [CrossRef] [PubMed]
27. Theocharopoulos, A.; Chen, X.; Hill, R.; Cattell, M.J. Reduced wear of enamel with novel fine and nano-scale leucite glass-ceramics. *J. Dent.* **2013**, *41*, 561–568. [CrossRef] [PubMed]
28. Cherian, J.; Jayakumar, R.; James, J.; Thomas, V.; Sramadathil, S.; Sasi, A.K. A comparative evaluation of enamel wear against different surface finished ceramics: An in vitro study. *Cureus* **2023**, *15*, e44689. [CrossRef] [PubMed]
29. Elmaria, A.; Goldstein, G.; Vijayaraghavan, T.; Legeros, R.Z.; Hittelman, E.L. An evaluation of wear when enamel is opposed by various ceramic materials and gold. *J. Prosthet. Dent.* **2006**, *96*, 345–353. [CrossRef] [PubMed]
30. Etman, M. Wear properties of dental ceramics. In *Non-Metallic Biomaterials for Tooth Repair and Replacement*; Elsevier: Amsterdam, The Netherlands, 2013; pp. 161–193.
31. Heintze, S.D.; Cavalleri, A.; Forjanic, M.; Zellweger, G.; Rousson, V. Wear of ceramic and antagonist—a systematic evaluation of influencing factors in vitro. *Dent. Mater.* **2008**, *24*, 433–449. [CrossRef] [PubMed]
32. Needleman, I.; Worthington, H.; Moher, D.; Schulz, K.; Altman, D.G. Improving the completeness and transparency of reports of randomized trials in oral health: The CONSORT statement. *Am. J. Dent.* **2008**, *21*, 7. [PubMed]
33. Frankenberger, R.; Taschner, M.; Garcia-Godoy, F.; Petschelt, A.; Krämer, N. Leucite-reinforced glass ceramic inlays and onlays after 12 years. *J. Adhes. Dent.* **2008**, *10*, 393–398. [PubMed]
34. Krämer, N.; Kunzelmann, K.-H.; Taschner, M.; Mehl, A.; Garcia-Godoy, F.; Frankenberger, R. Antagonist enamel wears more than ceramic inlays. *J. Dent. Res.* **2006**, *85*, 1097–1100. [CrossRef] [PubMed]
35. Taschner, M.; Frankenberger, R.; Garcia-Godoy, F. IPS Empress inlays luted with a self-adhesive resin cement after 1 year. *Am. J. Dent.* **2009**, *22*, 55. [PubMed]
36. Taschner, M.; Stirnweiss, A.; Frankenberger, R.; Kramer, N.; Galler, K.M.; Maier, E. Fourteen years clinical evaluation of leucite-reinforced ceramic inlays luted using two different adhesion strategies. *J. Dent.* **2022**, *123*, 104210. [CrossRef] [PubMed]
37. Beham, G. *IPS Empress: Eine Neue Keramik Technologie*; ZWR: Stuttgart, Germany, 1991; Volume 100.
38. Vlaar, S.T.; van der Zel, J.M. Accuracy of dental digitizers. *Int. Dent. J.* **2006**, *56*, 301–309. [CrossRef] [PubMed]
39. Matta, R.E.; von Wilmsky, C.; Neuhuber, W.; Lell, M.; Neukam, F.W.; Adler, W.; Wichmann, M.; Bergauer, B. The impact of different cone beam computed tomography and multi-slice computed tomography scan parameters on virtual three-dimensional model accuracy using a highly precise ex vivo evaluation method. *J. Craniomaxillofac. Surg.* **2016**, *44*, 632–636. [CrossRef] [PubMed]
40. Holst, S.; Karl, M.; Wichmann, M.; Matta, R.E. A new triple-scan protocol for 3D fit assessment of dental restorations. *Quintessence Int.* **2011**, *42*, 651–657. [PubMed]
41. Zimmermann, M.; Mehl, A.; Mörmann, W.H.; Reich, S. Intraoral scanning systems—A current overview. *Int. J. Comput. Dent.* **2015**, *18*, 101–129. [PubMed]
42. R Core Team. *R: A Language and Environment for Statistical Computing*; R Foundation for Statistical Computing: Vienna, Austria, 2024. Available online: <https://www.R-project.org/> (accessed on 30 May 2024).
43. Malament, K.A.; Natto, Z.S.; Thompson, V.; Rekow, D.; Eckert, S.; Weber, H.P. Ten-year survival of pressed, acid-etched e.max lithium disilicate monolithic and bilayered complete-coverage restorations: Performance and outcomes as a function of tooth position and age. *J. Prosthet. Dent.* **2019**, *121*, 782–790. [CrossRef] [PubMed]
44. Dondani, J.R.; Pardeshi, V.; Gangurde, A.; Shaikh, A.; Mahule, A.; Deval, P. Comparative Evaluation of Wear of Natural Enamel Antagonist Against Glazed Monolithic Zirconia Crowns and Polished Monolithic Zirconia Crowns: An In Vivo Study. *Int. J. Prosthodont.* **2023**, *36*, 273–281. [CrossRef] [PubMed]
45. Almejrad, L.; Almansour, A.; Bartlett, D.; Austin, R. CAD/CAM leucite-reinforced glass-ceramic for simulation of attrition in human enamel in vitro. *Dent. Mater.* **2024**, *40*, 173–178. [CrossRef] [PubMed]
46. Murbay, S.; Yeung, S.K.W.; Yip, C.Y.; Pow, E.H.N. Assessing Enamel Wear of Monolithic Ceramics with Micro-CT and Intra-oral Scanner. *Int. Dent. J.* **2023**, *73*, 496–502. [CrossRef] [PubMed]
47. Krejci, I.; Lutz, F. In-vitro test results of the evaluation of dental restoration systems. Correlation with in-vivo results. *Schweiz. Monatsschr. Zahnmed.* **1990**, *100*, 1445–1449. [PubMed]

48. Aladağ, A.; Oğuz, D.; Çömlekoğlu, M.E.; Akan, E. In vivo wear determination of novel CAD/CAM ceramic crowns by using 3D alignment. *J. Adv. Prosthodont.* **2019**, *11*, 120–127. [CrossRef] [PubMed]
49. Ohlmann, B.; Trame, J.P.; Dreyhaupt, J.; Gabbert, O.; Koob, A.; Rammelsberg, P. Wear of posterior metal-free polymer crowns after 2 years. *J. Oral Rehabil.* **2008**, *35*, 782–788. [CrossRef] [PubMed]
50. Hickel, R.; Peschke, A.; Tyas, M.; Mjör, I.; Bayne, S.; Peters, M.; Hiller, K.-A.; Randall, R.; Vanherle, G.; Heintze, S.D. FDI World Dental Federation: Clinical criteria for the evaluation of direct and indirect restorations—Update and clinical examples. *Clin. Oral Investig.* **2010**, *14*, 349–366. [CrossRef] [PubMed]
51. Suputtamongkol, K.; Anusavice, K.J.; Suchatlampong, C.; Sithiamnuai, P.; Tulapornchai, C. Clinical performance and wear characteristics of veneered lithia-disilicate-based ceramic crowns. *Dent. Mater.* **2008**, *24*, 667–673. [CrossRef] [PubMed]
52. Krejci, I.; Lutz, F.; Reimer, M.; Heinzmann, J.L. Wear of ceramic inlays, their enamel antagonists, and luting cements. *J. Prosthet. Dent.* **1993**, *69*, 425–430. [CrossRef] [PubMed]
53. Monasky, G.E.; Taylor, D.F. Studies on the wear of porcelain, enamel, and gold. *J. Prosthet. Dent.* **1971**, *25*, 299–306. [CrossRef] [PubMed]
54. Lee, A.; Swain, M.; He, L.; Lyons, K. Wear behavior of human enamel against lithium disilicate glass ceramic and type III gold. *J. Prosthet. Dent.* **2014**, *112*, 1399–1405. [CrossRef] [PubMed]
55. Zurek, A.D.; Alfaro, M.F.; Wee, A.G.; Yuan, J.C.; Barao, V.A.; Mathew, M.T.; Sukotjo, C. Wear Characteristics and Volume Loss of CAD/CAM Ceramic Materials. *J. Prosthodont.* **2019**, *28*, e510–e518. [CrossRef] [PubMed]
56. Krejci, I.; Lutz, F.; Reimer, M. Wear of CAD/CAM ceramic inlays: Restorations, opposing cusps, and luting cements. *Quintessence Int.* **1994**, *25*, 199–207. [PubMed]
57. Nawafleh, N.A.; Mack, F.; Evans, J.; Mackay, J.; Hatamleh, M.M. Accuracy and reliability of methods to measure marginal adaptation of crowns and FDPs: A literature review. *J. Prosthodont.* **2013**, *22*, 419–428. [CrossRef] [PubMed]
58. Mendřický, R. Determination of measurement accuracy of optical 3D scanners. *MM Sci. J.* **2016**, *2016*, 1565–1572. [CrossRef]
59. Bajoghli, F.; Sabouhi, M.; Nosouhian, S.; Davoudi, A.; Behnamnia, Z. Comparing the Accuracy of Three Different Impression Materials in Making Duplicate Dies. *J. Int. Oral Health* **2015**, *7*, 12–16. [PubMed]
60. Gujjarlupudi, M.C.; Reddy, S.V.; Madineni, P.K.; Ealla, K.K.; Nunna, V.N.; Manne, S.D. Comparative evaluation of few physical properties of epoxy resin, resin-modified gypsum and conventional type IV gypsum die materials: An in vitro study. *J. Contemp. Dent. Pract.* **2012**, *13*, 48–54. [CrossRef] [PubMed]
61. Ahlholm, P.; Sipilä, K.; Vallittu, P.; Jakonen, M.; Kotiranta, U. Digital Versus Conventional Impressions in Fixed Prosthodontics: A Review. *J. Prosthodont.* **2018**, *27*, 35–41. [CrossRef] [PubMed]
62. Pinto, R.J.; Casado, S.A.; Chmielewski, K.; Caramês, J.M.; Marques, D.S. Accuracy of different digital acquisition methods in complete arch implant-supported prostheses: An in vitro study. *J. Prosthet. Dent.* **2024**, *132*, 172–177. [CrossRef] [PubMed]

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

Article

Influence of the Milling Strategy on the Marginal Fit of Chairside-Fabricated Lithium Disilicate Crowns

Lara Berger ¹, Felix Förtsch ¹, Ralf Robert Kretschmer ¹, Oleksandr Sednyev ¹, José Ignacio Zorzin ², Manfred Wichmann ¹ and Ragai Edward Matta ^{1,*}

¹ Department of Prosthodontics, University Hospital Erlangen, Glückstrasse 11, 91054 Erlangen, Germany; lara.berger@uk-erlangen.de (L.B.); felix.foertsch@uk-erlangen.de (F.F.); ralf_kretschmer@t-online.de (R.R.K.); oleksandr.sednyev@uk-erlangen.de (O.S.); claudia.ehrhardt@uk-erlangen.de (M.W.)

² Department of Operative Dentistry and Periodontology, University Hospital Erlangen, Glückstrasse 11, 91054 Erlangen, Germany; jose.zorzin@fau.de

* Correspondence: ragai.matta@uk-erlangen.de; Tel.: +49-9131-8533604

Abstract: Utilizing dental CAD/CAM systems, chairside treatments allow patients to be treated within one day without the need for an intermediate dental laboratory. With this procedure, it is also essential to ensure a sufficient marginal fit of the restoration. Therefore, this study investigates the influence of the milling strategy on the marginal fit of CAD/CAM-fabricated lithium disilicate crowns. For this purpose, 10 crowns were fabricated from each of the materials IPS e.max CAD and Celtra Duo using the fastest and finest available milling strategy. The accuracy of the marginal fit of the 40 crowns was examined using the industrial scanner ATOS Triple Scan and the associated software using the triple scan protocol. In a comparison of the milling processes, the restorations fabricated using the finest process always exhibited a better marginal fit, with mean deviations of 87 μm and 146 μm for IPS e.max CAD and 111 μm and 118 μm for Celtra Duo. The difference was only statistically significant for the crowns made of IPS e.max CAD ($p = 0.008$). All mean deviations determined were below the clinically acceptable marginal gap values (150 μm). Both materials can be used for chairside-fabricated crowns; however, choosing the faster milling strategy leads to higher marginal discrepancies.

Keywords: CAD/CAM; marginal fit; milling accuracy; lithium disilicate

1. Introduction

Progressive digitalization has revolutionized dentistry significantly since the 1980s, particularly through the introduction of dental CAD/CAM systems (Computer-Aided Design/Computer-Aided Manufacturing) [1]. These computer-aided technologies enable the precise and efficient processing of a wide range of materials, including ceramics, metals, composites, and plastics, with a particular focus on prosthetic applications [2]. In combination with intraoral scanners, these systems enable so-called chairside treatments, in which patients can be treated within one day without the involvement of a dental laboratory. Recent surveys show that in economically strong countries, more than a quarter of dentists already use this procedure, which points to its numerous advantages [3–5]. These include significant time and cost savings while maintaining or even improving the quality of the restoration, increased comfort for both patients and practitioners, and the overall increased efficiency of the entire workflow [6,7]. These developments set new standards in dental care and highlight the potential of digital technologies in modern dentistry [2]. The increasing use of chairside systems is closely linked to constant innovations in the field of dental

ceramics, which have become indispensable as dental restorative materials due to their mechanical and esthetic properties and excellent biocompatibility. Dental ceramics can be divided into two main groups: oxide ceramics and glass ceramics. Oxide ceramics are characterized by very good strength properties but often involve esthetic compromises due to the opacities typical of the material. In contrast, glass-ceramics enable the fabrication of highly esthetic restorations, as they exhibit outstanding translucency with adequate mechanical properties, which is the main advantage of this material class [8].

Among glass-ceramics, lithium disilicate-based materials in particular have established themselves, achieving a flexural strength of 350 MPa to 450 MPa and thus exhibiting significantly increased mechanical properties compared to leucite-reinforced ceramics (80 to 150 MPa) or feldspar ceramics (100 to 160 MPa) [8,9]. The structural properties of this material class have considerably expanded the range of indications for glass-ceramics, so that monolithic lithium disilicate blocks are now not only used for single-tooth crowns, but can also be used for the fabrication of fixed partial dentures (FPDs) in the anterior region and for three-unit FPDs with premolar involvement [10–12]. However, the strength is not yet sufficient for longer-span restorations in the posterior region, which is why the optimization of the mechanical properties of glass-ceramics is a focus of current research [10,13,14]. A key approach to increasing the strength characteristics of glass-ceramics is to further improve crack resistance, as crack propagation is one of the main causes of failure in glass-ceramic restorations. This is where the class of zirconia-reinforced lithium disilicate ceramics comes in, as zirconia undergoes a volume increase of 3–5% when subjected to extreme forces due to a phase transformation from the monoclinic to the tetragonal structure. This increase in volume generates a compression pressure that can inhibit crack propagation and thus contribute significantly to the stability of the ceramic [13]. Recent reviews have shown that these materials are a good alternative to conventional lithium disilicate ceramics. When comparing the mechanical properties of the two material classes, however, the data situation is ambiguous, so further research is still required to classify the mechanical properties more precisely [15,16].

In addition to the material properties, various clinical quality characteristics have a direct influence on the long-term stability of dental restorations, including the accuracy of fit of the crown on the tooth [17]. A seamless transition between the crown and tooth would be ideal, but this cannot be achieved in practice due to technical limitations in the manufacturing process and the need for a spacer for the luting material between the tooth and crown. This inevitably results in a narrow gap, which is referred to as the marginal gap. This should be kept as small as possible, as too large a discrepancy promotes the accumulation of plaque and colonization by cariogenic microorganisms, which in turn increases the risk of secondary caries [18,19]. In addition, over-contoured or protruding crown margins can irritate the gingiva and lead to irreversible periodontal damage in the long term [20,21]. In the scientific literature, the information on a clinically acceptable range of the marginal gap value varies, with the upper limit typically being between 100 μm and 120 μm [22–24]. Marginal gaps of less than 120 μm can be realized realistically with common CAD/CAM systems [25,26].

In addition to conventional two-dimensional methods such as the silicone replica technique or the cross-section method, various digital three-dimensional methods are now available for measuring the marginal gap, which enable an even more precise and comprehensive examination of the accuracy of fit. In addition to optical coherence tomography and micro-computed tomography, these methods also include the triple scan method described by Holst, Matta et al., which was used in this study [27,28].

Another decisive factor for the longevity and mechanical stability of ceramic restorations is the surface quality. Industrially manufactured monolithic ceramic blocks, which

can be processed using the CAD/CAM technique, are characterized by a high degree of material homogeneity [29]. Nevertheless, the milling process during fabrication can create fine microcracks both on the surface and inside the material, which can act as potential fracture initiation points and compromise the mechanical integrity of the restoration [30,31]. Fractographic analysis of clinically failed all-ceramic restorations has shown that such defects are often the starting points for restoration failures [32,33].

The quality of the milling process is influenced by various factors, including the size, shape, and abrasiveness of the burs, the number of axes of the milling machine, the production mode, and the selected milling strategy [31,34,35]. A recent review has shown that five-axis milling machines deliver more precise results in terms of accuracy of fit than machines with three or four axes. In addition, the wear of the burs, regardless of their geometry, affects the accuracy of the manufacturing process [34]. The milling strategy also plays a decisive role: depending on the material used and the manufacturer's recommendations, at least one fast and one fine milling strategy should be implemented. While fast milling strategies are associated with a time saving of around ten minutes, the question arises as to whether this reduces the precision of the restoration [34,36].

The aim of the present *in vitro* study was, therefore, to investigate the influence of different milling strategies on the marginal fit of CAD/CAM-fabricated monolithic lithium disilicate crowns made of IPS e.max CAD and Celtra Duo. The null hypotheses were that neither the milling strategy nor the selected restorative material had an influence on the accuracy of fit.

2. Materials and Methods

To conduct this study, 40 composite tooth dies (Grandio Disc, VOCO GmbH, Cuxhaven, Germany) were fabricated, which were based on an ideally prepared maxillary canine with a chamfer preparation. To subsequently design and fabricate identical crowns, the software of the CEREC system (Dentsply Sirona, Charlotte, NC, USA) required information about the positions of the tooth die and the adjacent teeth. For this reason, a scan model was created into which the fabricated tooth dies could be reproducibly integrated and removed for digital impression taking. The dies and the corresponding scan model were matted with titan oxide spray (CEREC Optispray, Dentsply Sirona) to enable precise and fast digital capture [37] and then digitized with the CEREC Omnicam (Dentsply Sirona) by an experienced clinician.

The greatest possible equivalence of the crowns was crucial for the comparability of the individual crowns when examining the accuracy of fit. To ensure this, the crown shape was calculated using the factory settings (Table 1) stored by the manufacturer for the respective material in the CEREC SW 4.4.0 software (Dentsply Sirona) and the “biogeneric” function. It was checked whether the minimum thickness was met at all points before proceeding to crown fabrication. The materials IPS e.max CAD LT A2 C14 (Ivoclar Vivadent, Schaan, Liechtenstein) and Celtra Duo LT A2 C14 (Dentsply Sirona) were each processed with the CEREC MC XL (Dentsply Sirona) CNC unit for the fabrication of 20 crowns. For both materials, 10 crowns were fabricated using the fastest and finest milling strategy available: for the IPS e.max CAD material (Ivoclar Vivadent), the milling strategies were fast (Group A) and extra-fine (Group B), and for Celtra Duo (Dentsply Sirona), fine (Group C) and extra-fine (Group D). The mean manufacturing times for each group are listed in Table 2. For the crowns made of IPS e.max CAD (Ivoclar Vivadent), a crystallization firing was conducted at 850 °C for 30 min in the CEREC Speedfire (Dentsply Sirona) ceramic furnace in accordance with the manufacturer's instructions to achieve the final hardness of the material required for insertion. According to the manufacturer, this step was not necessary for the crowns made of Celtra Duo (Dentsply Sirona).

Table 1. Factory settings in the CEREC software for IPS e.max CAD (Ivoclar Vivadent) and Celtra Duo (Dentsply Sirona).

Parameter	Default Value
Spacer (radial)	120 μm
Spacer (occlusal)	120 μm
Proximal Contact Strength	25 μm
Occlusal Contact Strength	25 μm
Dynamic Contact Strength	25 μm
Margin Ramp Angle	60°
Minimal Thickness (radial)	1000 μm
Minimal Thickness (occlusal)	1500 μm
Margin Thickness	50 μm

Table 2. Mean manufacturing times for all groups.

Material	Milling Strategy	Mean Manufacturing Time	Group
IPS e.max CAD	fast	8 min	A
	extrafine	22 min	B
Celtra Duo	fine	14 min	C
	extrafine	22 min	D

The total of 40 crowns and 40 dies were digitized for the following analyses, both individually and in combination with the high-resolution, light-optical ATOS Triple Scan (Carl Zeiss GOM Metrology GmbH, Braunschweig, Germany), whose average measurement error during object registration is only 3 μm [28]. According to the scan protocol from the pilot study by Holst, Matta et al. [28], each scan series consisted of three individual scans: the crown alone, the die alone, and the respective crown on the corresponding die. For the scan of the crown on the corresponding die, the crown was reproducibly fixed with constant pressure in a special device using a pin in the central part of the crown. The captured data were transferred directly to a connected computer in the ATOS Professional 2016 software (Carl Zeiss GOM Metrology GmbH) to be able to combine the individual scans into one object and subsequently perform the matching of the scans and the analytical evaluation.

The marginal fit was analyzed both three-dimensionally using a marginal surface analysis and two-dimensionally in the form of a virtual cross-sectional analysis. For the three-dimensional marginal surface analysis, an area between the preparation margin and an offset curve located 1 cm further into the lumen was defined. This area was compared for each die with the surface of the corresponding inside of the crown, and the fit was calculated based on distance measurements. The mean deviation of the marginal measuring area was determined as the arithmetic mean of all measured orthogonal distances from the die surface to the inside of the crown (“Mean” in μm).

Within the two-dimensional cross-sectional analysis, which was also carried out using the software ATOS Professional (Carl Zeiss GOM Metrology GmbH), the linear discrepancies between the crown margin and the preparation margin were recorded with the help of pre-programmed analysis scripts. The program generated 20 radial sections at 18° intervals, starting from the center of the crown or die. For each section, three distances were calculated according to the classification by Holmes et al. [38]: the “z” distance along the vertical z-axis of the coordinate system (vertical marginal discrepancy), the “n” distance in the horizontal plane (horizontal marginal discrepancy), and the “xyz” distance as the

absolute linear distance between the crown margin and the preparation margin (absolute marginal discrepancy), as illustrated in Figure 1.

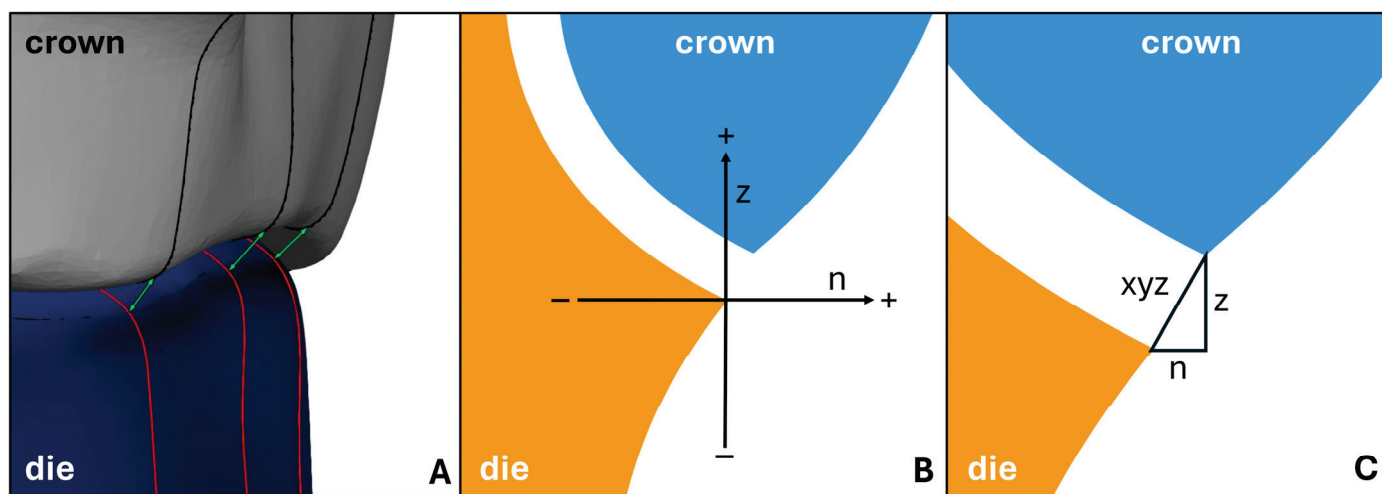


Figure 1. Two-dimensional cross-sectional analysis: (A) Visualization of the sections (red/black) and the absolute marginal discrepancy (green). (B) Illustration of the vectors of the horizontal (n) and vertical (z) marginal discrepancy. (C) Visualization of the measurement distances after Holmes [38].

A Wilcoxon signed rank test was performed for the statistical analysis of the acquired data using the statistical software SPSS (version 28.0.3, IBM Corporation, Armonk, NY, USA), with the significance level being set at 0.05. A graphical overview of the study design can be found in Figure 2.

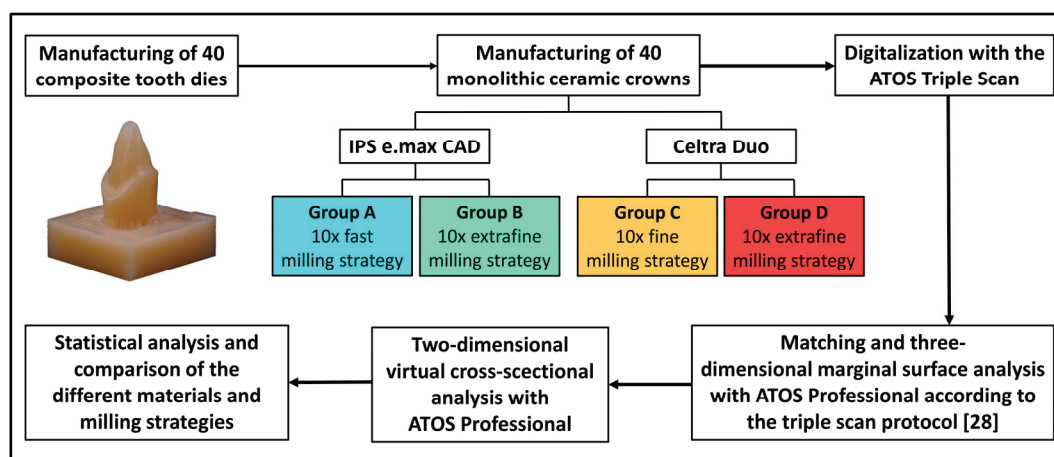


Figure 2. Overview of the study design.

3. Results

3.1. Three-Dimensional Marginal Surface Analysis

In the marginal surface analysis, smaller discrepancies were found for both materials when selecting the faster milling strategy (Table 3, Figure 3). However, these results were only statistically significant for the crowns made of Celtra Duo (Dentsply Sirona) (Table 4). In the material comparison, the extra-fine milled crowns made of IPS e.max CAD showed a significantly higher precision with a 16 μm lower mean deviation.

Table 3. Descriptive statistics of the three-dimensional marginal fit analysis: mean deviation (Mean) with standard deviation (SD) and the minimum (Min) and maximum (Max) deviation in μm for all groups.

Group	Mean [μm]	SD [μm]	Min [μm]	Max [μm]
A	63	11.2	50	91
B	66	14.2	38	83
C	55	3.3	48	59
D	82	10.6	65	96

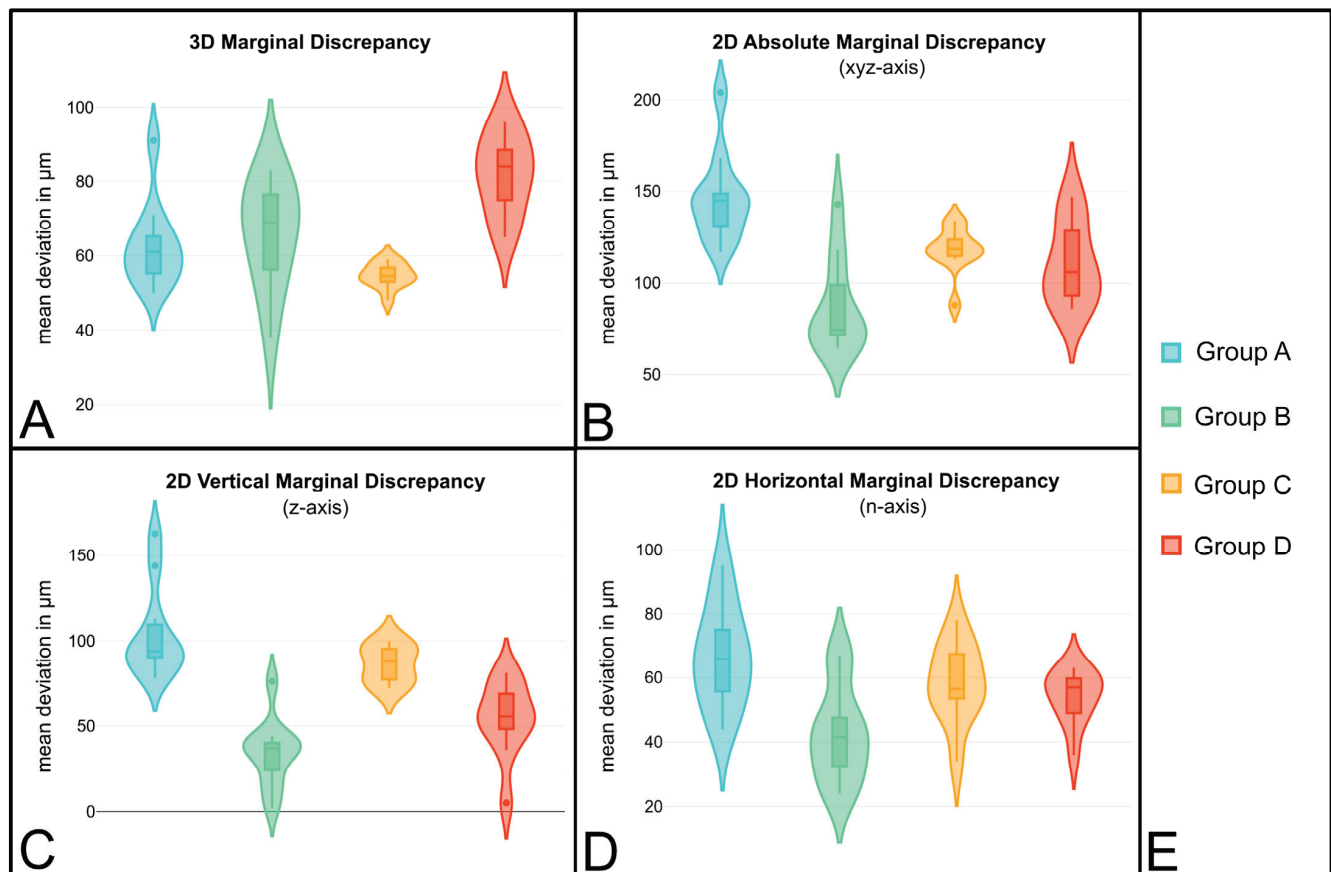


Figure 3. Violin plots showing the mean distance in μm for all groups (E) for the three-dimensional marginal discrepancy (A) and the two-dimensional absolute marginal discrepancy (B), as well as the vertical (C) and horizontal (D) marginal discrepancies.

Table 4. *p* values for all statistical comparisons as part of the three-dimensional marginal fit analysis.

Variable	Group 1	Group 2	<i>p</i> Value
Milling strategy	A	B	0.374
	C	D	0.005
Material	B	D	0.015

3.2. Two-Dimensional Virtual Cross-Sectional Analysis

In the two-dimensional virtual cross-sectional analysis, the deviations in relation to the vertical (z-axis), horizontal (n-axis), and absolute (xyz-axis) marginal discrepancy were examined and compared. For both materials, the restorations fabricated using the fastest possible milling strategy showed a higher deviation in all three axes (Table 5, Figure 3).

Table 5. Descriptive statistics of the two-dimensional virtual cross-sectional analysis: mean deviation (Mean) with standard deviation (SD) and the minimum (Min) and maximum (Max) deviation in μm for all groups.

Group	Axis	Mean [μm]	SD [μm]	Min [μm]	Max [μm]
A	xyz	146	24.8	117	204
	z	105	27.9	78	163
	n	67	16	44	95
B	xyz	87	26	65	143
	z	19	36.5	−76	44
	n	43	15	24	67
C	xyz	118	12.8	88	134
	z	86	10.8	72	100
	n	59	12	34	78
D	xyz	111	21.9	86	147
	z	54	22	5	81
	n	54	8	36	63

When the absolute marginal discrepancy was examined, only the comparison of the crowns made of IPS e.max CAD with a difference in the mean deviations of 59 μm was statistically significant (Table 6); for the crowns made of Celtra Duo (Dentsply Sirona), the difference was only 7 μm . In the material comparison, the restorations made of IPS e.max CAD (Ivoclar Vivadent) showed significantly higher accuracy, with the mean deviation in the xyz axis being 24 μm lower.

Table 6. *p* values for all statistical comparisons as part of the two-dimensional virtual cross-sectional analysis.

Axis	Variable	Group 1	Group 2	<i>p</i> Value
xyz	Milling strategy	A	B	0.008
		C	D	0.26
	Material	B	D	0.022
z	Milling strategy	A	B	0.007
		C	D	0.005
	Material	B	D	0.005
n	Milling strategy	A	B	0.022
		C	D	0.332
	Material	B	D	0.066

The same results were obtained for the vertical and horizontal marginal discrepancy. All comparisons were statistically significant in the z-axis, but only the comparison of the milling strategies for crowns made of IPS e.max CAD (Ivoclar Vivadent) was significant in the n-axis.

4. Discussion

In the scientific examination of the accuracy of fit of crowns, there are various methodological approaches that are characterized by individual influencing factors that signifi-

cantly determine the accuracy and clinical relevance of the analyses [39–42]. A key aspect is the distinction between *in vivo* patient studies and *in vitro* laboratory studies. In *in vivo* studies, the fabrication of crowns requires customized tooth preparations that vary from patient to patient, potentially limiting the comparability of data. In addition, patient-specific challenges during impression-taking can affect the accuracy of measurements [43]. In contrast, *in vitro* analyses allow the fabrication of impressions and the corresponding crowns under standardized conditions, which simplifies a differentiated evaluation of the accuracy of fit. This is also reflected in the fact that a large proportion of the investigations into the accuracy of fit of crowns are conducted as *in vitro* studies [44].

The selected restorative material and the manufacturing process also affect the accuracy of fit. Nowadays, the dental market offers a wide range of products for monolithic glass-ceramic restorations manufactured using the CAD/CAM workflow, so it is necessary to conduct studies to verify the material properties stated by the manufacturer and thus to find the best possible materials for the clinical application [8,45]. Since both IPS e.max CAD (Ivoclar Vivadent) and Celtra Duo (Dentsply Sirona) are frequently used in chairside procedures in everyday clinical practice [46], these materials were to be investigated in the present study.

The choice of measurement method is a crucial factor in the significance of the study, whereby Nawafleh et al. recommend a combined measurement method for analyzing the marginal fit in a review [44]. The triple-scan protocol introduced by Holst, Matta et al. [28] enables the generation of a large number of measurements per crown during three-dimensional surface analysis, which significantly increases the informative value of the fit analysis. In addition, the data generated by the scanning protocol can be used to perform a further virtual two-dimensional cross-sectional analysis, which eliminates potential sources of error in the production of replicas and sections. Both measurement methods complement each other; critical areas with unacceptable margins are more easily identified in the two-dimensional analysis, while the three-dimensional surface analysis, due to the high number of measured values, may hide potentially unacceptable deviations within acceptable average values. Conversely, large deviations in the individual measurements of the two-dimensional analysis can make a precise assessment of the fit more difficult. Accordingly, it can be assumed that the combined measurement methodology can ensure a sufficiently high level of validity for the study [40,47,48].

The first null hypothesis was that the choice of milling strategy has no influence on the marginal fit of the restoration. This hypothesis must be rejected since the crowns milled in the fastest possible milling process showed higher marginal discrepancies in the two-dimensional cross-sectional analysis. When the absolute marginal discrepancy (xyz-axis) was considered, all mean deviations determined were below the clinically acceptable value of 120 μm [22,24], except for the fast-milled crowns made of IPS e.max CAD. Other authors consider margins of up to 150 μm to be sufficient [49]. However, the maximum deviation in group A significantly exceeded this value at 204 μm . For all other groups, the maximum deviation remained below 150 μm . The differences between the various milling strategies were statistically significant only for the crowns made of IPS e.max CAD, with a difference in the mean deviation of 59 μm ($p = 0.008$); for the crowns made of Celtra Duo (Dentsply Sirona), the difference was only 7 μm ($p = 0.26$).

The results of this study, with a mean deviation of 87 and 146 μm for IPS e.max CAD (Ivoclar Vivadent) and 111 and 118 μm for Celtra Duo (Dentsply Sirona), are in the same range as the results of similar comparative studies [39,49–53]. For IPS e.max CAD (Ivoclar Vivadent), values between 55 μm [39] and 146 μm [53] are found in the literature, while for Celtra Duo (Dentsply Sirona), the range includes values from 46 μm [52] to 131 μm [49]. These considerable differences can be attributed to different manufacturing parameters in

crown production and differences in the measurement methodology. However, the chosen manufacturing strategy was not mentioned in any of the comparative studies, making it difficult to critically evaluate the differences identified in this work.

When the individual axes were examined, the crowns manufactured using the extra-fine grinding process always showed less deviation, although the comparison in the horizontal axis for the crowns made of Celtra Duo (Dentsply Sirona) was not significant ($p = 0.332$). There is also hardly any comparative data available here; some results for IPS e.max CAD (Ivoclar Vivadent) can be found for the vertical discrepancy alone. In a study by Neves et al., this was $39 \pm 9 \mu\text{m}$ [54]; Mostafa et al. found a value of $33 \pm 20 \mu\text{m}$ in their study [55]. Both results are within the range of values in this study of $19 \pm 37 \mu\text{m}$ and $105 \pm 28 \mu\text{m}$, respectively. However, no information on the milling strategy used can be found in the studies, either.

When comparing the milling strategies as part of the three-dimensional marginal surface analysis, the crowns produced in the respective fastest possible process proved to be more precise, whereby the difference with a mean deviation of $27 \mu\text{m}$ was only significant for the restorations made of Celtra Duo (Dentsply Sirona) ($p = 0.005$). For the crowns made of IPS e.max CAD (Ivoclar Vivadent), the results were very close to each other, with deviations of $63 \pm 11 \mu\text{m}$ and $66 \pm 14 \mu\text{m}$, respectively ($p = 0.374$). This three-dimensional marginal surface analysis considers a 1 cm wide area starting from the preparation margin, which is referred to as the “internal marginal gap” according to Holmes et al. [38]. Within this area, over 10,000 individual measurement sections are defined, from which the mean value is calculated. In contrast, the two-dimensional virtual cross-sectional analysis evaluates the absolute marginal discrepancy, i.e., the actual distance between the cervical restoration margin and the preparation margin. In addition, the vertical and horizontal marginal discrepancies can be viewed in isolation by vector determination. Due to these facts, this two-dimensional analysis method can be considered more relevant with regard to the marginal fit of the restoration.

The second null hypothesis was that the selected restorative material has no influence on the marginal fit. This hypothesis must also be rejected since the restorations made of Celtra Duo (Dentsply Sirona) showed significantly higher marginal discrepancies in both analysis methods. In the three-dimensional analysis, the difference in mean deviations was $16 \mu\text{m}$ ($p = 0.015$), and in the two-dimensional analysis, it was $24 \mu\text{m}$ ($p = 0.022$) for the absolute marginal discrepancy. In a recent study, Elsayed et al. also found a higher marginal fit for IPS e.max CAD (Ivoclar Vivadent) compared to Celtra Duo (Dentsply Sirona) [39], but studies with opposite results can also be found [51,52]. This inconsistent data situation can be attributed to different study designs and the associated challenges in the comparability of the measurement methods. It is also possible that these differences are influenced by other factors, such as the thickness of the titanium oxide powder layer or the positioning of the crowns on the dies [56].

Despite the informative results of this study, its limitations must also be considered. For one thing, the selected crown shape could have influenced the milling precision since the milling machine may have had more difficulty reaching some areas depending on the geometry of the restoration. For another, although the two materials examined are frequently used, they represent only a small part of the product range of dental glass-ceramics [10].

Regarding future research approaches, the investigation of CAD/CAM-fabricated glass-ceramic crowns in relation to the possible influence of the crown material and the manufacturing strategy could be extended by considering additional parameters in addition to the accuracy of fit. These include surface quality, wear behavior, and survival rates, which could be determined using chewing simulators. Such an extension of the analysis

would enable well-founded conclusions to be drawn that would allow a differentiated recommendation for clinical application. Given the increasing use of monolithic single-tooth crowns made of different composite or zirconia blocks in clinical practice, it would also be useful to examine these materials in more detail in the context of the present work. This could provide valuable insights for practical application and help to optimize treatment outcomes.

5. Conclusions

Considering the limitations of this in vitro study, the following conclusions can be drawn:

- Choosing the faster milling strategy leads to higher marginal discrepancies, regardless of the material used
- When fabricating crowns from IPS e.max CAD (Ivoclar Vivadent), choosing the fast milling strategy can result in clinically unacceptable marginal gap values
- The smallest marginal discrepancies were achieved with the material IPS e.max CAD when the extra-fine milling strategy was selected
- Both materials investigated are suitable for crown fabrication using the chairside technique if the appropriate milling strategy is applied

When using lithium disilicate ceramics chairside, it is important to consider whether the time saved by choosing the faster milling strategy justifies the risk of clinically unacceptable marginal gap values.

Author Contributions: Data curation, R.R.K.; Formal analysis, R.E.M.; Investigation, R.R.K.; Methodology, L.B. and R.E.M.; Project administration, J.I.Z., M.W. and R.E.M.; Resources, M.W.; Software, O.S.; Supervision, J.I.Z., M.W. and R.E.M.; Validation, L.B.; Visualization, F.F. and O.S.; Writing—original draft, L.B. and F.F.; Writing—review and editing, R.E.M. All authors have read and agreed to the published version of the manuscript.

Funding: The authors would like to thank Ivoclar (Ivoclar Vivadent GmbH) and Dentsply Sirona for kindly providing the materials used in this study.

Institutional Review Board Statement: Not applicable.

Informed Consent Statement: Not applicable.

Data Availability Statement: The underlying data are available from the corresponding author upon reasonable request.

Acknowledgments: Parts of the results of the present study were used in Ralf Robert Kretschmer's doctoral thesis.

Conflicts of Interest: The authors declare no conflicts of interest. The funders had no role in the design of the study; in the collection, analyses, or interpretation of the data; in the writing of the manuscript; or in the decision to publish the results.

References

1. Miyazaki, T.; Hotta, Y.; Kunii, J.; Kuriyama, S.; Tamaki, Y. A review of dental CAD/CAM: Current status and future perspectives from 20 years of experience. *Dent. Mater. J.* **2009**, *28*, 44–56. [CrossRef] [PubMed]
2. Suganna, M.; Kausher, H.; Tarek Ahmed, S.; Sultan Alharbi, H.; Faraj Alsubaie, B.; Ds, A.; Haleem, S.; Meer Rownaq Ali, A.B. Contemporary Evidence of CAD-CAM in Dentistry: A Systematic Review. *Cureus* **2022**, *14*, e31687. [CrossRef] [PubMed]
3. Krastev, T.; Payer, M.; Krastev, Z.; Cardelles, J.F.P.; Vegh, A.; Banyai, D.; Geczi, Z.; Vegh, D. The Utilisation of CAD/CAM Technology Amongst Austrian Dentists: A Pilot Study. *Int. Dent. J.* **2023**, *73*, 430–434. [CrossRef]
4. Muhetaer, A.; Yang, H.Y.; Huang, C. Application of Chairside CAD/CAM and Its Influencing Factors among Chinese Dental Practitioners: A Crosssectional Study. *Chin. J. Dent. Res.* **2024**, *27*, 253–262. [CrossRef] [PubMed]

5. Nassani, M.Z.; Ibraheem, S.; Shamsy, E.; Darwish, M.; Faden, A.; Kujan, O. A Survey of Dentists' Perception of Chair-Side CAD/CAM Technology. *Healthcare* **2021**, *9*, 68. [CrossRef]
6. Abdullah, A.O.; Tsi trou, E.A.; Pollington, S. Comparative in vitro evaluation of CAD/CAM vs conventional provisional crowns. *J. Appl. Oral Sci.* **2016**, *24*, 258–263. [CrossRef]
7. Aswal, G.S.; Rawat, R.; Dwivedi, D.; Prabhakar, N.; Kumar, V. Clinical Outcomes of CAD/CAM (Lithium disilicate and Zirconia) Based and Conventional Full Crowns and Fixed Partial Dentures: A Systematic Review and Meta-Analysis. *Cureus* **2023**, *15*, e37888. [CrossRef]
8. Li, R.W.; Chow, T.W.; Matinlinna, J.P. Ceramic dental biomaterials and CAD/CAM technology: State of the art. *J. Prosthodont. Res.* **2014**, *58*, 208–216. [CrossRef]
9. Pereira, R.M.; Ribas, R.G.; Montanheiro, T.; Schatkoski, V.M.; Rodrigues, K.F.; Kito, L.T.; Kobo, L.K.; Campos, T.M.B.; Bonfante, E.A.; Gierthmuehlen, P.C.; et al. An engineering perspective of ceramics applied in dental reconstructions. *J. Appl. Oral Sci.* **2023**, *31*, e20220421. [CrossRef]
10. Fu, L.; Engqvist, H.; Xia, W. Glass-Ceramics in Dentistry: A Review. *Materials* **2020**, *13*, 1049. [CrossRef]
11. Reich, S.; Endres, L.; Weber, C.; Wiedhahn, K.; Neumann, P.; Schneider, O.; Rafai, N.; Wolfart, S. Three-unit CAD/CAM-generated lithium disilicate FDPs after a mean observation time of 46 months. *Clin. Oral Investig.* **2014**, *18*, 2171–2178. [CrossRef]
12. Zarone, F.; Di Mauro, M.I.; Ausiello, P.; Ruggiero, G.; Sorrentino, R. Current status on lithium disilicate and zirconia: A narrative review. *BMC Oral Health* **2019**, *19*, 134. [CrossRef] [PubMed]
13. Liu, X.; Yao, X.; Zhang, R.; Sun, L.; Zhang, Z.; Zhao, Y.; Zhang, T.; Yan, J.; Zhang, Y.; Wu, X.; et al. Recent advances in glass-ceramics: Performance and toughening mechanisms in restorative dentistry. *J. Biomed. Mater. Res. B Appl. Biomater.* **2024**, *112*, e35334. [CrossRef]
14. Teichmann, M.; Göckler, F.; Weber, V.; Yildirim, M.; Wolfart, S.; Edelhoff, D. Ten-year survival and complication rates of lithium-disilicate (Empress 2) tooth-supported crowns, implant-supported crowns, and fixed dental prostheses. *J. Dent.* **2017**, *56*, 65–77. [CrossRef] [PubMed]
15. Manziuc, M.; Kui, A.; Chisnoiu, A.; Labunet, A.; Negucioiu, M.; Ispas, A.; Buduru, S. Zirconia-Reinforced Lithium Silicate Ceramic in Digital Dentistry: A Comprehensive Literature Review of Our Current Understanding. *Medicina* **2023**, *59*, 2135. [CrossRef]
16. Zarone, F.; Ruggiero, G.; Leone, R.; Breschi, L.; Leuci, S.; Sorrentino, R. Zirconia-reinforced lithium silicate (ZLS) mechanical and biological properties: A literature review. *J. Dent.* **2021**, *109*, 103661. [CrossRef]
17. Larson, T.D. The clinical significance of marginal fit. *Northwest Dent.* **2012**, *91*, 22–29. [PubMed]
18. Rossetti, P.H.; do Valle, A.L.; de Carvalho, R.M.; De Goes, M.F.; Pegoraro, L.F. Correlation between margin fit and microleakage in complete crowns cemented with three luting agents. *J. Appl. Oral Sci.* **2008**, *16*, 64–69. [CrossRef]
19. Zoellner, A.; Brägger, U.; Fellmann, V.; Gaengler, P. Correlation between clinical scoring of secondary caries at crown margins and histologically assessed extent of the lesions. *Int. J. Prosthodont.* **2000**, *13*, 453–459.
20. Knoernschild, K.L.; Campbell, S.D. Periodontal tissue responses after insertion of artificial crowns and fixed partial dentures. *J. Prosthet. Dent.* **2000**, *84*, 492–498. [CrossRef]
21. Srimanepong, V.; Heboyan, A.; Zafar, M.S.; Khurshid, Z.; Marya, A.; Fernandes, G.V.O.; Rokaya, D. Fixed Prosthetic Restorations and Periodontal Health: A Narrative Review. *J. Funct. Biomater.* **2022**, *13*, 15. [CrossRef]
22. Karataşlı, O.; Kursoğlu, P.; Capa, N.; Kazazoğlu, E. Comparison of the marginal fit of different coping materials and designs produced by computer aided manufacturing systems. *Dent. Mater. J.* **2011**, *30*, 97–102. [CrossRef]
23. Reich, S.; Gozdowski, S.; Trentzsch, L.; Frankenberger, R.; Lohbauer, U. Marginal fit of heat-pressed vs. CAD/CAM processed all-ceramic onlays using a milling unit prototype. *Oper. Dent.* **2008**, *33*, 644–650. [CrossRef]
24. Tan, P.L.; Gratton, D.G.; Diaz-Arnold, A.M.; Holmes, D.C. An in vitro comparison of vertical marginal gaps of CAD/CAM titanium and conventional cast restorations. *J. Prosthodont.* **2008**, *17*, 378–383. [CrossRef] [PubMed]
25. Kale, E.; Cilli, M.; Özçelik, T.B.; Yilmaz, B. Marginal fit of CAD-CAM monolithic zirconia crowns fabricated by using cone beam computed tomography scans. *J. Prosthet. Dent.* **2020**, *123*, 731–737. [CrossRef] [PubMed]
26. Naert, I.; Van der Donck, A.; Beckers, L. Precision of fit and clinical evaluation of all-ceramic full restorations followed between 0.5 and 5 years. *J. Oral Rehabil.* **2005**, *32*, 51–57. [CrossRef] [PubMed]
27. Ayres, A.P.; Cuschieri, L.A.; Bianchi, D.M.; Pradiés, G.; Côrtes, A.R.G. Advantages and drawbacks of different methods to measure marginal gaps in fixed dental prostheses: A scoping review. *J. Dent.* **2024**, *151*, 105400. [CrossRef]
28. Holst, S.; Karl, M.; Wichmann, M.; Matta, R.-E.T. A new triple-scan protocol for 3D fit assessment of dental restorations. *Quintessence Int.* **2011**, *42*, 651–657.
29. Zhang, Y.; Lawn, B.R. Evaluating dental zirconia. *Dent. Mater.* **2019**, *35*, 15–23. [CrossRef]
30. Fraga, S.; Amaral, M.; Bottino, M.A.; Valandro, L.F.; Kleverlaan, C.J.; May, L.G. Impact of machining on the flexural fatigue strength of glass and polycrystalline CAD/CAM ceramics. *Dent. Mater.* **2017**, *33*, 1286–1297. [CrossRef]

31. Romanyk, D.L.; Martinez, Y.T.; Veldhuis, S.; Rae, N.; Guo, Y.; Sirovica, S.; Fleming, G.J.P.; Addison, O. Strength-limiting damage in lithium silicate glass-ceramics associated with CAD-CAM. *Dent. Mater.* **2019**, *35*, 98–104. [CrossRef] [PubMed]
32. Quinn, J.B.; Quinn, G.D.; Kelly, J.R.; Scherrer, S.S. Fractographic analyses of three ceramic whole crown restoration failures. *Dent. Mater.* **2005**, *21*, 920–929. [CrossRef]
33. Thompson, J.; Anusavice, K.; Naman, A.; Morris, H. Fracture surface characterization of clinically failed all-ceramic crowns. *J. Dent. Res.* **1994**, *73*, 1824–1832. [CrossRef] [PubMed]
34. Oliveira, A.R.; Ziglioli, N.U.; Marocho, S.M.S.; Satterthwaite, J.; Borba, M. Effect of the CAD/CAM Milling Protocol on the Fracture Behavior of Zirconia Monolithic Crowns. *Materials* **2024**, *17*, 2981. [CrossRef]
35. Pilecco, R.O.; Machry, R.V.; Baldi, A.; Tribst, J.P.M.; Sarkis-Onofre, R.; Valandro, L.F.; Kleverlaan, C.J.; Scotti, N.; Pereira, G.K.R. Influence of CAD-CAM milling strategies on the outcome of indirect restorations: A scoping review. *J. Prosthet. Dent.* **2024**, *131*, 811.e1–811.e10. [CrossRef] [PubMed]
36. Lubauer, J.; Belli, R.; Schünemann, F.H.; Matta, R.E.; Wichmann, M.; Wartzack, S.; Völkl, H.; Petschelt, A.; Lohbauer, U. Inner marginal strength of CAD/CAM materials is not affected by machining protocol. *Biomater. Investig. Dent.* **2021**, *8*, 119–128. [CrossRef]
37. Oh, H.-S.; Lim, Y.-J.; Kim, B.; Kim, M.-J.; Kwon, H.-B.; Baek, Y.-W. Influence of Scanning-Aid Materials on the Accuracy and Time Efficiency of Intraoral Scanners for Full-Arch Digital Scanning: An In Vitro Study. *Materials* **2021**, *14*, 2340. [CrossRef]
38. Holmes, J.R.; Bayne, S.C.; Holland, G.A.; Sulik, W.D. Considerations in measurement of marginal fit. *J. Prosthet. Dent.* **1989**, *62*, 405–408. [CrossRef]
39. Elsayed, S.M.; Emam, Z.N.; Abu-Nawareg, M.; Zidan, A.Z.; Elsis, H.A.; Abuelroos, E.M.; Fansa, H.A.; Shokier, H.M.R.; Elbanna, K.A. Marginal gap distance and cyclic fatigue loading for different all-ceramic endocrowns. *Eur. Rev. Med. Pharmacol. Sci.* **2023**, *27*, 879–887. [CrossRef]
40. Matta, R.E.; Schmitt, J.; Wichmann, M.; Holst, S. Circumferential fit assessment of CAD/CAM single crowns—a pilot investigation on a new virtual analytical protocol. *Quintessence Int.* **2012**, *43*, 801–809.
41. Rizonaki, M.; Jacquet, W.; Bottenberg, P.; Depla, L.; Boone, M.; De Coster, P.J. Evaluation of marginal and internal fit of lithium disilicate CAD-CAM crowns with different finish lines by using a micro-CT technique. *J. Prosthet. Dent.* **2022**, *127*, 890–898. [CrossRef]
42. Zimmermann, M.; Valcanaia, A.; Neiva, G.; Mehl, A.; Fasbinder, D. Influence of Different CAM Strategies on the Fit of Partial Crown Restorations: A Digital Three-dimensional Evaluation. *Oper. Dent.* **2018**, *43*, 530–538. [CrossRef] [PubMed]
43. Abduo, J.; Elseyoufi, M. Accuracy of Intraoral Scanners: A Systematic Review of Influencing Factors. *Eur. J. Prosthodont. Restor. Dent.* **2018**, *26*, 101–121. [CrossRef] [PubMed]
44. Nawafleh, N.A.; Mack, F.; Evans, J.; Mackay, J.; Hatamleh, M.M. Accuracy and reliability of methods to measure marginal adaptation of crowns and FDPs: A literature review. *J. Prosthodont.* **2013**, *22*, 419–428. [CrossRef] [PubMed]
45. Spitznagel, F.A.; Boldt, J.; Gierthmuehlen, P.C. CAD/CAM Ceramic Restorative Materials for Natural Teeth. *J. Dent. Res.* **2018**, *97*, 1082–1091. [CrossRef]
46. Marchesi, G.; Camurri Piloni, A.; Nicolin, V.; Turco, G.; Di Lenarda, R. Chairside CAD/CAM Materials: Current Trends of Clinical Uses. *Biology* **2021**, *10*, 1170. [CrossRef]
47. Berger, L.; Matta, R.E.; Weiß, C.M.; Adler, W.; Wichmann, M.; Zorzin, J.I. Effect of Luting Materials on the Accuracy of Fit of Zirconia Copings: A Non-Destructive Digital Analysis Method. *Materials* **2024**, *17*, 2130. [CrossRef]
48. Kim, H.; Hong, S.J.; Choi, Y.; Lee, Y.; Won, H.; Kim, H.S.; Pae, A. Assessment of the fit of lithium disilicate crowns at various locations fabricated by three different methods using the triple-scan protocol. *Int. J. Comput. Dent.* **2023**, *26*, 37–47. [CrossRef]
49. Zimmermann, M.; Valcanaia, A.; Neiva, G.; Mehl, A.; Fasbinder, D. Three-Dimensional Digital Evaluation of the Fit of Endocrowns Fabricated from Different CAD/CAM Materials. *J. Prosthodont.* **2019**, *28*, e504–e509. [CrossRef]
50. Anadioti, E.; Aquilino, S.A.; Gratton, D.G.; Holloway, J.A.; Denry, I.; Thomas, G.W.; Qian, F. 3D and 2D marginal fit of pressed and CAD/CAM lithium disilicate crowns made from digital and conventional impressions. *J. Prosthodont.* **2014**, *23*, 610–617. [CrossRef]
51. Kang, S.Y.; Yu, J.M.; Lee, J.S.; Park, K.S.; Lee, S.Y. Evaluation of the Milling Accuracy of Zirconia-Reinforced Lithium Silicate Crowns Fabricated Using the Dental Medical Device System: A Three-Dimensional Analysis. *Materials* **2020**, *13*, 4680. [CrossRef] [PubMed]
52. Saloum, A.; Abdullah, M. *Marginal Gap and Internal Fit of CEREC e.Max and Celtra Duo Crowns*; University of British Columbia: Vancouver, Canada, 2019.
53. Schaefer, O.; Decker, M.; Wittstock, F.; Kuepper, H.; Guentsch, A. Impact of digital impression techniques on the adaption of ceramic partial crowns in vitro. *J. Dent.* **2014**, *42*, 677–683. [CrossRef] [PubMed]
54. Neves, F.D.; Prado, C.J.; Prudente, M.S.; Carneiro, T.A.; Zancopé, K.; Davi, L.R.; Mendonça, G.; Cooper, L.F.; Soares, C.J. Micro-computed tomography evaluation of marginal fit of lithium disilicate crowns fabricated by using chairside CAD/CAM systems or the heat-pressing technique. *J. Prosthet. Dent.* **2014**, *112*, 1134–1140. [CrossRef] [PubMed]

55. Mostafa, N.Z.; Ruse, N.D.; Ford, N.L.; Carvalho, R.M.; Wyatt, C.C.L. Marginal Fit of Lithium Disilicate Crowns Fabricated Using Conventional and Digital Methodology: A Three-Dimensional Analysis. *J. Prosthodont.* **2018**, *27*, 145–152. [CrossRef]
56. Prudente, M.S.; Davi, L.R.; Nabbout, K.O.; Prado, C.J.; Pereira, L.M.; Zancopé, K.; Neves, F.D. Influence of scanner, powder application, and adjustments on CAD-CAM crown misfit. *J. Prosthet. Dent.* **2018**, *119*, 377–383. [CrossRef]

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

Article

The Impact of Three-Dimensional Printer Technology on the Accuracy of Dental Implant Models

Alexander Strunz ¹, Lara Berger ¹, Anna Seidel ¹, Johannes Ries ¹, Werner Adler ², Manfred Wichmann ¹ and Ragai Edward Matta ^{1,*}

¹ Department of Prosthodontics, University Hospital Erlangen, Glückstrasse 11, 91054 Erlangen, Germany; alexander.strunz@uk-erlangen.de (A.S.); lara.berger@uk-erlangen.de (L.B.); anna.seidel@uk-erlangen.de (A.S.); johannes.ries@uk-erlangen.de (J.R.); claudia.ehrhardt@uk-erlangen.de (M.W.)

² Institute of Medical Informatics, Biometry and Epidemiology (IMBE), Friedrich-Alexander-University Erlangen-Nuremberg, Waldstrasse 6, 91054 Erlangen, Germany; werner.adler@uk-erlangen.de

* Correspondence: ragai.matta@uk-erlangen.de; Tel.: +49-9131-8533604

Abstract: This study examines the impact of different 3D printing technologies on the accuracy of implant positions in printed dental models, a crucial factor in implant-supported prosthetics. A standardized titanium model with three bone-level implants was scanned using an industrial scanner to create a virtual reference model. Ten intraoral scans of the same model were performed, and the generated STL files were used to design physical models printed with three different 3D printers: two utilizing digital light processing (DLP) technology and one employing stereolithography (SLA) ($n = 30$). The printed models were then rescanned, and deviations from the reference STL file were analyzed. Results showed that the SLA printer exhibited the highest deviations (0.26 ± 0.17 mm), whereas the DLP printers demonstrated greater accuracy, with one DLP system (0.07 ± 0.02 mm) performing slightly better than the other (0.12 ± 0.13 mm). The SLA printer exhibited the most significant errors in the vestibulo-oral and occlusal-apical directions. The findings suggest that DLP printers offer superior precision for implant-supported restorations in digital workflows. Clinically, the choice of 3D printing technology significantly impacts model accuracy, emphasizing the importance of selecting the appropriate printer based on the required precision.

Keywords: computer-aided design/computer-aided manufacturing (CAD/CAM); digital workflow accuracy; implant dentistry; 3D printing in dentistry; additive manufacturing dentistry

1. Introduction

The treatment of patients with implant-supported or implant-retained solutions plays an increasingly important role in everyday clinical practice [1]. Dental implants offer advantages in terms of esthetics as well as tooth preservation and function. They are increasingly being used not only to replace individual lost teeth, but also to treat edentulous patients [2]. Implants are a valuable alternative to mucosa-supported full dentures in such situations [3].

After successful implant placement and a variable healing phase, the prosthetic restoration is fitted, involving impression taking, restoration fabrication, and placement on the patient [4]. Implant prosthetic restorations can follow conventional or digital workflows, which should not be viewed as strictly separate.

In the conventional workflow, open and closed impressions are used, with studies showing that the open approach is more precise [5,6]. After fabrication of plaster models from patient impressions, the superstructure can be constructed either conventionally or digitally after digitization [7].

Scanbody geometries are then superimposed on the virtual 3D impression to register the exact implant position [1,8,9]. Studies on intraoral scanners indicate that their precision is clinically acceptable for individual implants or short-span bridges, but not yet sufficient for long-span restorations, prompting the investigation of scanning aids for such cases [10–13]. The digital design of the superstructure is directly based on intraoral scan data, minimizing errors by reducing additional transmissions and manual steps [8].

However, in most cases, a physical model is required to check and finalize the machine-made superstructure. On the one hand, manual work steps such as individual veneering can only be carried out on a model. On the other hand, restorations fabricated in the laboratory are classified as medical devices according to European law and must be checked by the dental technician for criteria such as gap-free and rocking-free fit on the model, correct clearance, positioning, or proximal contact verification before delivery to the dentist [14–16].

As the digital workflow becomes increasingly established, there is a need to transfer the information obtained during intraoral scanning into an exact physical model. In addition to subtractive methods, in which the models are milled from various plastics using milling machines, additive methods are also available [17,18].

Various 3D printing systems have increasingly found their way into everyday dental practice and dental technology in recent years. Initially used primarily for the fabrication of models, the technology now enables the precise manufacturing of provisional or definitive prostheses, bridges, or crowns [19,20]. Additionally, 3D printing is employed in the precise planning of surgical procedures, such as the creation of drill guides for implant placements [21,22]. Frequently used technologies include digital light processing (DLP) and stereolithography (SLA) methods [17,19,20]. The main advantages over subtractive production methods are their material efficiency and speed, which also result in lower costs [21,22]. Studies investigating the accuracy of 3D-printed models show that clinically acceptable results are achieved [23,24]. At the same time, numerous factors were identified that significantly influence the results of an additively manufactured model. In addition to the printer technology, the specific printer model, and the material used, parameters such as build angle or layer thickness can also have a significant impact on quality [19,25,26].

Stereolithography (SLA) was the first 3D printing process and served as the foundation for all subsequent technologies. It remains widely used today. During the printing process, the build platform is submerged in a reservoir filled with printer resin until it stops just before reaching the bottom. A UV laser is then used to polymerize the first layer of the object onto the build platform through the translucent window of the reservoir. The build platform subsequently moves upward allowing fresh resin to flow beneath it before returning to the viewing window, where the next layer is polymerized [27–29]. Digital light processing (DLP) follows a similar principle to SLA, with the key distinction being the light source used for photopolymerization. Instead of a laser, DLP employs a semiconductor plate with microscopically small mirrors, known as a digital micromirror device, which exposes the entire cross-section of the object at once [19,28,29].

Ultimately, many processes in additive manufacturing with 3D printers are specially tailored to the printer model being used. For example, the resolution of different printers varies and material properties such as the viscosity of the printer resin or the wavelength and light intensity required for curing must be compatible between the printer and its resin [19]. While both mechanical and biological material properties are of central importance for applications such as drilling templates or temporary and permanent restorations,

it is above all the precision of the material that is decisive for models that are used purely in the dental laboratory [24–26]. Table 1 shows additional information on the printer resins used and their recommended processing parameters.

Table 1. More details on the printer resins and their recommended processing parameters.

	Resin 1 IMPRIMO® LC Model Beige (DLP—ASIGA MAX UV)	Resin 2 PRO RESIN, MODEL X (DLP—Straumann P30)	Resin 3 Formlabs Dental Model Resin (SLA—Form 3+)
composition	alkoxylated bisphenol-A dimethacrylate, phosphine oxide, stabilizers, dyes, and pigments	UDMA, diacrylate, acrylic resin, phosphine oxide	bismethacrylate, methacrylate monomers, ethyl phenylphosphinate
layer height	50 µm	50 µm	50 µm
build angle	0°	0°	0°
mean printing time	37 min 32 s	45 min 12 s	1 h 50 min 8 s
printing temperature	30.0 °C	35.0 °C	35.0 °C
wavelength	385 nm	385 nm	405 nm
light intensity	11.7 $\frac{\text{mW}}{\text{cm}^2}$	no available information	250 mW Laser

While the influence of individual factors on the quality of printed models has been well researched [25–28], the extent to which individual printers and their technologies, together with the recommended materials are suitable for the production of high-precision implant models has been little investigated to date. The aim of this study was therefore to compare the achievable accuracy of different validated systems for the fabrication of implant models. The data presented in the study by Revilla-León et al. [24], which examined the accuracy of 3D-printed compared to conventionally produced implant models, has already demonstrated that 3D-printed models can be used in a comparable manner to the conventional workflow. Building on this, the novelty of the present study lies in the comparative analysis of different self-contained workflows for a comprehensive evaluation of these systems and their suitability for the production of implant models. To the best of our knowledge, there is no available information regarding the positioning accuracy of laboratory analogs in comparison to a reference for the printers examined in this study, using the printer resins employed. The influence of the printer model and its technology was investigated using the parameters, materials, and post-processing specified by the manufacturer of each printer.

The null hypothesis is that the printer model and its technology have no influence on the position of the implants in the printed implant model.

2. Materials and Methods

The aim of this in vitro study was to examine the trueness of the implant position in printed models as a function of the printer technology used. The accuracy was investigated as a function of three different 3D printers using a total of two different technologies. The ASIGA Max UV (ASIGA, Sydney, Australia) and the Straumann P30+ (2) printer (Rapidshape, Heimsheim, Germany) were examined as representatives of DLP technology, while the SLA printer Form 3+ (3) (Formlabs GmbH, Berlin, Germany) was compared.

As a reference model, an edentulous maxilla was milled from titanium using a CNC milling machine (NobelProcera, Nobel Biocare Services AG, Zürich, Switzerland). The reference model represented the left half of the upper jaw. The total volume was 7269 mm³ with a surface area of 4422 mm². Three bone-level implants (TiZr, ø 4.8 mm RC, SLA 12 mm, Roxolid, Loxim; Straumann AG, Basel, Switzerland) were then placed in the milled

titanium model and fixed using a laser. The implants in the anterior and premolar regions were placed at a right angle to the model base and the implant in the posterior region was placed at an angle of approximately 15° to the distal (see Figure 1a). Scanbodies from Medentika (L-Series L1410; Medentika GmbH—Straumann Group, Hügelsheim, Germany) were selected to record the implant position according to a digital workflow (see Figure 1b).

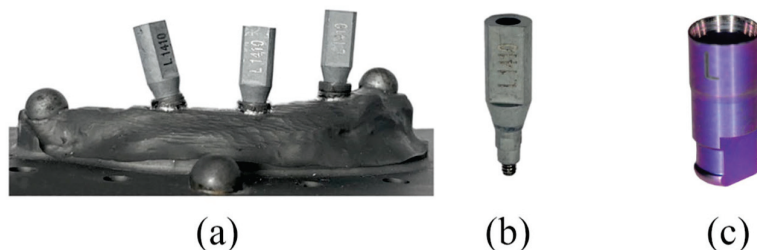


Figure 1. (a) Medentika scanbodies screwed onto reference titanium model; (b) Medentika Scanbody L-Series L1410; (c) laboratory analog from Medentika (L-Series L81; Straumann GmbH, Freiburg im Breisgau, Germany).

In order to ensure a non-reflective, matte surface, the titanium model was sandblasted with aluminum oxide abrasive, leaving out the implants. To increase the accuracy of the individual scan alignments, the model was then covered with measurement markers (Reference markers 0.8 mm; GOM GmbH, Braunschweig, Germany). The titanium model with manually screwed-in scanbodies was digitized using the ATOS So4 II industrial scanner (ATOS, GOM GmbH, Braunschweig, Germany) to generate a Standard Tessellation Language (STL) reference file [30].

Using the Primescan intraoral scanner (Dentsply Sirona, Bensheim, Germany; Primescan AC; Connect SW 5.2), 10 scans of the same model were then taken. To simulate as realistic a patient scenario as possible, the scanbodies were screwed in manually before each scan. The generated intraoral scan STL files were imported into the CAD software 3Shape (3Shape Dental Manager, Copenhagen K, Denmark, Version 2.21.1.0) and models were constructed. The implant position was aligned using a 3-point overlay process. As in other studies, this method showed better results than automatic single-point alignment [30]. The implant-analog-holder offset (IAH), which refers to the gap between the laboratory analog and the printed model, was set to 0.06 mm, as other studies have shown that the highest positional accuracy is achieved for IAH offsets between 0.04 and 0.06 mm [31].

The STL model data ($n = 10$) were then exported and converted into 10 physical models each ($n = 30$) using three different 3D printers. The first of the 3D printers with DLP technology to be tested was the ASIGA MAX UV (group A) (ASIGA, Sydney, Australia). After the nesting process with the associated nesting software ASIGA Composer (Version 1.2.12), 10 models were produced with the printer resin IMPRIMO[®] LC Model Beige (SCHEUDENTAL GmbH, Iserlohn, Germany) ($n = 10$). In addition, the DLP printer Straumann P30+ (group B) (Rapidshape, Heimsheim, Germany) was examined, using the nesting software AUTODESK NETFABB (Autodesk, Version 2022.0) and the printer resin PRO RESIN, MODEL X (Straumann GmbH, Freiburg im Breisgau, Germany). The third system examined was the SLA printer Form 3+ (group C) (Formlabs GmbH, Berlin, Germany) with the nesting software PreForm (Version 3.33.1) and the printer resin Formlabs Dental Model Resin (Formlabs GmbH, Berlin, Germany). The selection of materials and parameters for the printing process was based on the manufacturers' recommended specifications for the 3D printers. However, all models were nested with their base parallel to the build platform, and a layer thickness of 50 μm was maintained (see Figure 2a). Table 1 provides more detailed information on the materials used and their corresponding processing parameters.

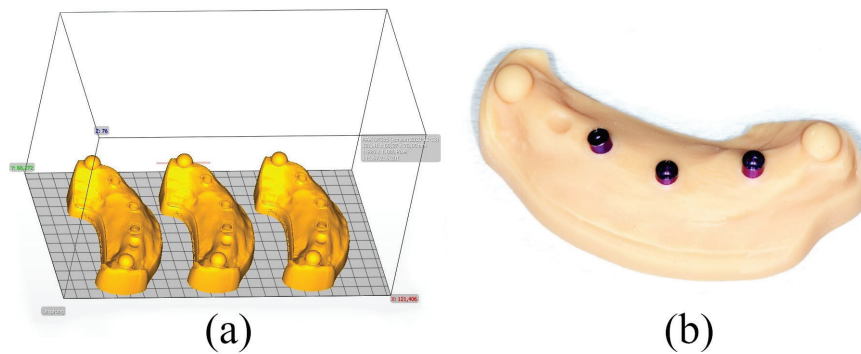


Figure 2. (a) Nesting of digital implant models using ASIGA Composer nesting software as an example; (b) printed model after post-processing and with laboratory analogs placed.

The post-processing of the printed models ($n = 30$) was also carried out according to the manufacturer's specifications using the washing and light-curing devices associated with the printers. All models printed with DLP technology were post-processed with the Straumann P-Wash and P-Cure device (Straumann GmbH, Freiburg im Breisgau, Germany). The parameters stored by the manufacturer were selected for the corresponding materials. The P-Wash unit uses two separate containers of isopropanol as a solvent, one for an initial rough cleaning and a second pass for final cleaning. Post-curing in the P-Cure unit takes place under vacuum. For the SLA-fabricated models, cleaning was performed using the Form Wash (Formlabs GmbH, Berlin, Germany) cleaning device. The models were washed in isopropanol for 10 min before being post-cured for 30 min in the Form Cure light-curing unit.

After post-processing, the printed models were fitted with laboratory analogs from Medentika (L-Series L81; Straumann GmbH, Freiburg im Breisgau, Germany) (see Figures 1c and 2b).

The models were then coated using an airbrush gun and a solution of Rutile Titanium White in ethanol (GOM GmbH, Braunschweig, Germany) to avoid reflections and covered with reference markers. The scanbodies were then screwed into the laboratory analogs of the printed models and the models were digitized using the ATOS So4 II industrial scanner (ATOS, GOM GmbH, Braunschweig, Germany) ($n = 30$). The digitization of the printed models was performed for all groups within 2 days after their production in order to minimize dimensional changes due to longer storage periods.

The accuracy of the implant positions in the printed models was analyzed by superimposing the resulting STL data with the reference STL file in the ATOS Professional software (v2018; GOM GmbH, Braunschweig, Germany) using a best-fit alignment of the model surface while omitting the scanbody surface (see Figure 3b–d). This step is essential to prevent alignment with the scanbody, as this would be undesirable at this stage. A best-fit alignment that includes the scanbody surface would distort the results, leading to significantly reduced deviations.

In order to compare the implant position with the reference STL file, measurement points were determined on the scanbodies of the reference STL and the individual STL data sets. For this purpose, a fitting cylinder with maximum contact to the scanbody surface and a maximum congruent plane perpendicular to the fitting cylinder was constructed. The measurement point for each implant is defined by the intersection of the cylinder axis with the vertical plane at the top of each scanbody, resulting in three measuring points for each model. Deviations from these reference points were plotted in a 3D coordinate system, with the X-axis representing the mesio-distal orientation, the Y-axis representing the oro-vestibular orientation, and the Z-axis representing the cranial-caudal orientation (see Figure 3a). In addition to the deviations in the individual planes, the Euclidean distance

dXYZ was determined as the 3D deviation resulting from the vector of the displacement relative to the three spatial planes. The deviations from the reference STL were given as maximum, average, and minimum deviation in mm (see Table 1). This method can be classified as suitable, as it has also been used successfully in other studies [6,30,32,33].

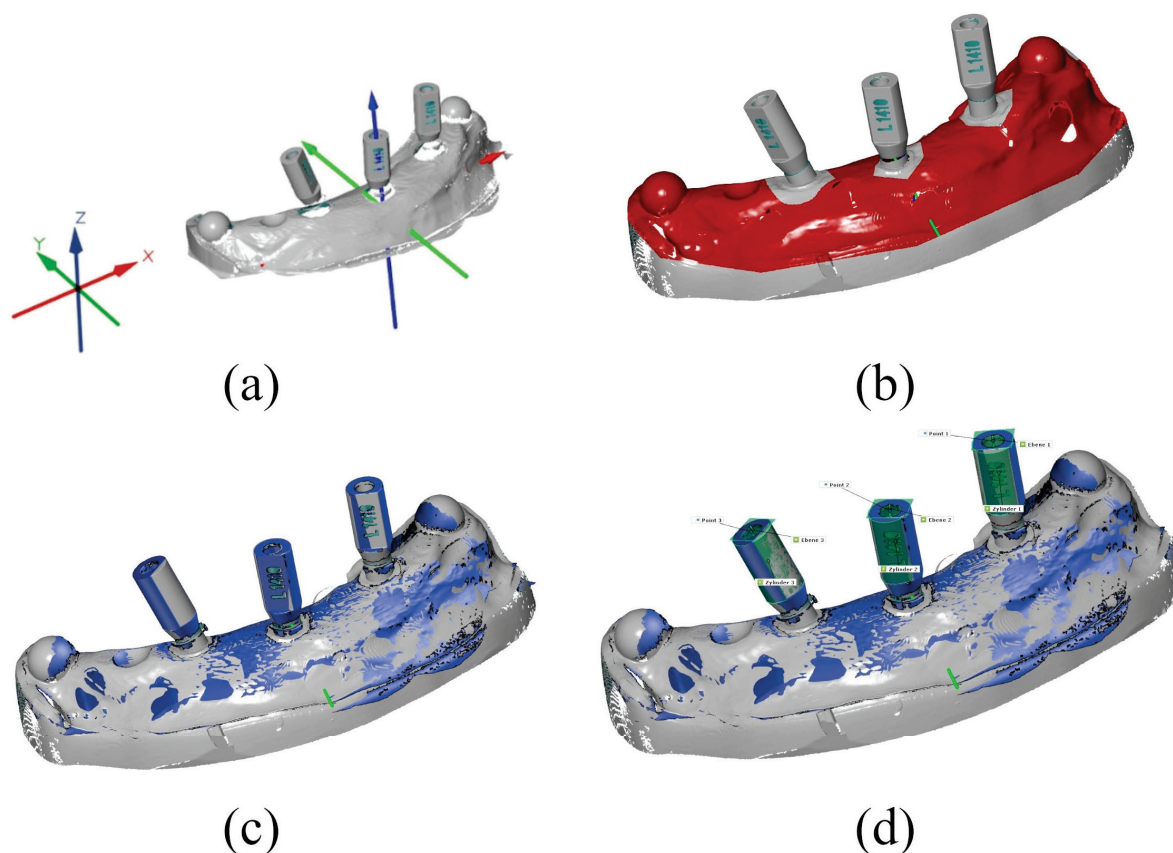


Figure 3. (a) The 3D coordinate system used, where the X-axis represents the mesio-distal orientation, the Y-axis represents the oro-vestibular orientation, and the Z-axis represents the cranial-caudal orientation. (b–d) Examples of the virtual calculation of 3D deviations in the ATOS Professional Software: (b) surface selection of the digitized printed models for the matching process with the reference STL data set, (c) overlaying of the STL data with the reference STL file, and (d) determination of the deviation between the STL of a printed model and the reference STL using a fitting cylinder and fitting plane.

The data were categorized into three groups: group A includes measurements from models printed with ASIGA MAX UV, group B contains those from Straumann P30+, and group C consists of models produced with Form 3+. Figure 4 provides an overview of the study design.

Statistical analysis was performed with the software R (Version 3.1.1 2014-07-11, R Core Team, R Foundation for Statistical Computing, Vienna, Austria). The Mann–Whitney U test was used to compare the printers with each other. This test is particularly suitable if the data are not normally distributed or correspond to ordinal scales, as it makes no assumptions about the distribution of the data. The significance level was set at $p < 0.05$, which means that the probability of observing an outcome that is at least as extreme as the observed outcome must be less than 5% for a true null finding to be considered statistically significant. This strict threshold ($p < 0.05$) was chosen to ensure high precision in the results and to minimize the risk of 1st type errors (falsely rejecting the null hypothesis).

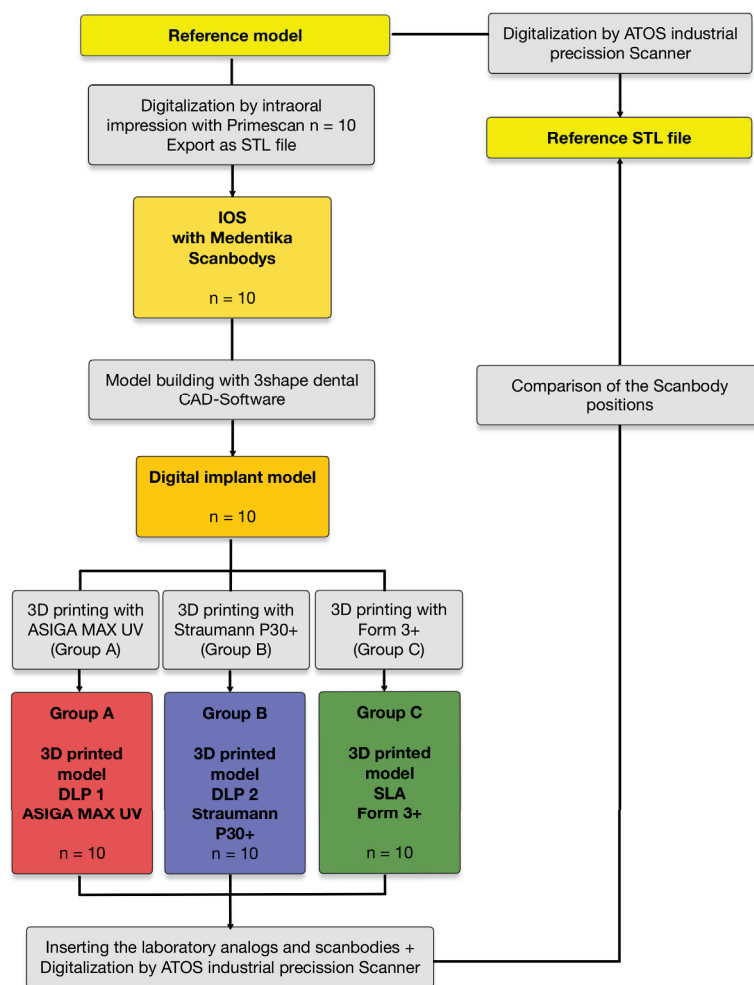


Figure 4. A representation of the study procedure with the individual study groups.

3. Results

Overall, the data obtained showed that the implant models produced by the SLA printer (Form 3+/Formlabs/SLA) (group C) exhibited significantly greater discrepancies (0.26 ± 0.17 mm) in the overall three-dimensional deviation of the implant positions, represented by the Euclidean distance dXYZ, compared to the reference, than the models produced by the two printers using DLP technology ($p < 0.001$) (see Table 2).

The two DLP printers again showed comparable values to each other, whereby the Straumann P30+ printer (0.07 ± 0.02 mm) (group B) achieved significantly higher accuracy in the overall deviation compared to ASIGA MAX UV (0.12 ± 0.13 mm) (group A) ($p = 0.006$) (see Figure 5). The collected data show a significant standard deviation, particularly in the Z-axis measurements of the SLA printer Form 3+. A lower standard deviation is observed in the measurements of the DLP printer ASIGA MAX UV, with the lowest standard deviations found in the DLP printer Straumann P30+. Our research has shown that the repeatability of the examined Straumann P30 DLP printer can be classified as very good. No outliers were observed, unlike the case with the DLP printer ASIGA MAX UV, and to an even greater extent with the SLA printer Form 3+. Figures 5 and 6 illustrate these outliers as circles on the corresponding measured values.

Table 2. Summary of the mean distance with standard deviation (SD) and the minimum (Min) or maximum (Max) distance in mm between the STLs of the respective groups A, B, and C and the corresponding reference files for the X-, Y-, and Z-axis as well as the Euclidean distance dXYZ.

Group			Mean	SD	Min	Max
Group A	ASIGA MAX UV DLP 1	dXYZ	0.12 mm	0.13 mm	0.03 mm	0.79 mm
		x	0.06 mm	0.05 mm	0.00 mm	0.21 mm
		y	0.05 mm	0.03 mm	0.01 mm	0.11 mm
		z	0.06 mm	0.14 mm	0.01 mm	0.78 mm
Group B	Straumann P30+ DLP 2	dXYZ	0.07 mm	0.02 mm	0.03 mm	0.10 mm
		x	0.03 mm	0.02 mm	0.00 mm	0.07 mm
		y	0.04 mm	0.02 mm	0.00 mm	0.08 mm
		z	0.03 mm	0.01 mm	0.00 mm	0.05 mm
Group C	Form 3+ SLA	dXYZ	0.26 mm	0.17 mm	0.09 mm	0.93 mm
		x	0.05 mm	0.05 mm	0.00 mm	0.18 mm
		y	0.17 mm	0.06 mm	0.08 mm	0.37 mm
		z	0.14 mm	0.20 mm	0.01 mm	0.91 mm

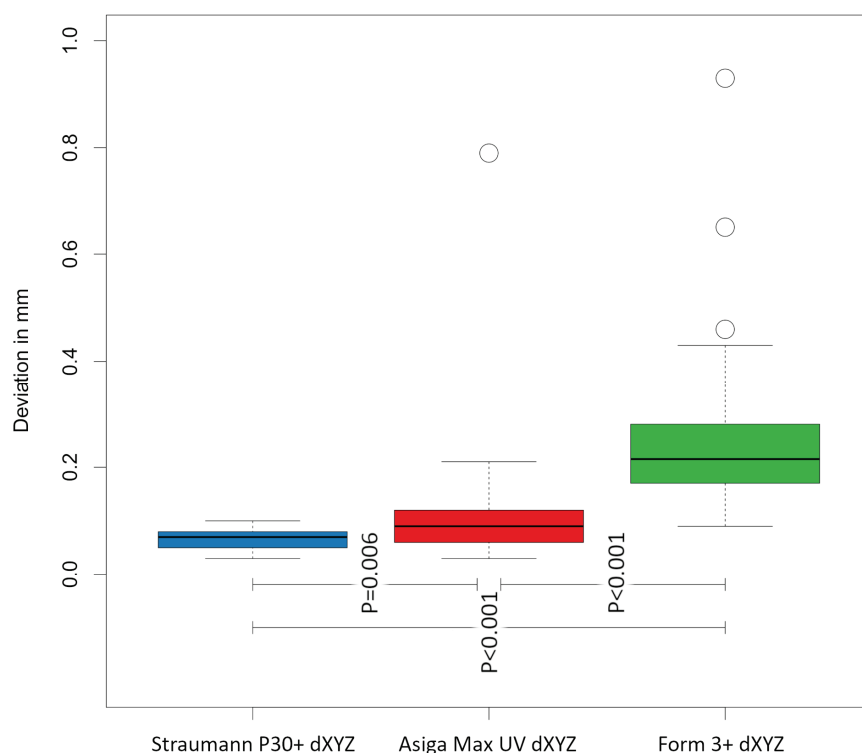


Figure 5. Statistical comparison of groups A and B, B and C, and A and C, with corresponding *p*-values for Euclidean distance dXYZ. The white circles represent the outliers.

When looking at the deviations in the individual spatial planes, the models manufactured using SLA technology (group C) show a greater deviation in the Y-axis (0.17 ± 0.06 mm) and Z-axis (0.14 ± 0.20 mm) compared to the X-axis (0.05 ± 0.05 mm) (see Figure 6).

No significant difference was found with regard to the implant position in the anterior, premolar, or posterior region.

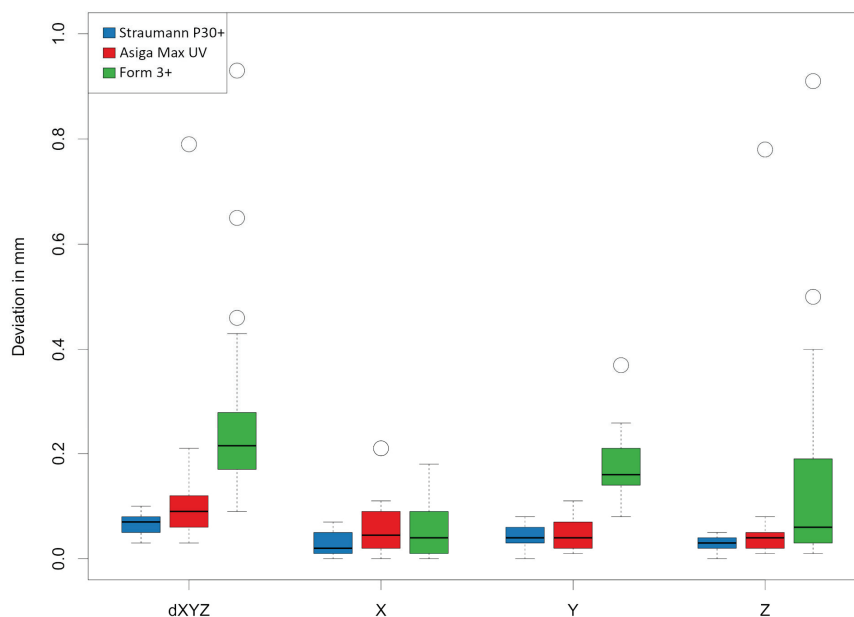


Figure 6. Boxplots for the comparison of groups A, B, and C for the Euclidean distance dXYZ, as well as the individual spatial planes X-, Y-, and Z-axis. The white circles represent the outliers.

4. Discussion

The digital workflow is increasingly being used in the treatment of patients with fixed dentures on implants. The focus is on saving time for both practitioners and patients, reducing costs and material usage, and minimizing sources of error [7,8,14].

In addition to the long-established subtractive methods of manufacturing dentures using CNC milling, additive manufacturing methods using 3D printers are becoming more relevant. The production of models is a frequent application of this technology [19–22].

If models are designed and fabricated purely based on an intraoral scan, there is no physical working basis for the manual post-processing and subsequent final inspection of implant restorations fabricated in the CAD/CAM workflow [6–8]. This makes it necessary to transfer the information obtained from the intraoral scan as accurately as possible into a physical model [14].

For this study, three different printers were selected for the production of implant models. Two printers using DLP technology were compared with each other as well as with a printer using SLA technology. The high-precision industrial scanner ATOS So4 II was used for this study both to generate a reference STL and for the 3D evaluation of the printed models, which also allows potential errors in the intraoral scan to be detected [6,32,33]. To minimize other factors influencing the printed models, the same parameters were applied to all models and the manufacturer's validated printer resin was used.

Numerous studies have examined the quality of printed models [27–29,34–38]. Various factors were found to influence the accuracy of such models. In addition to the printing technology examined in this study, the selected printing parameters, such as the build angle, the layer thickness, placement on the build platform, and printing temperature, also play a role [25,27,34,35]. The procurement of the printing resin and the post-processing after the printing process also have an influence on the quality [36]. The influence of different printers and their technology on the positioning accuracy of the implant analogs in the model is still poorly understood. In this context, the aim of this study was to determine whether clinically acceptable implant models can be produced with the selected printers and which systems produce the best results with the processes specified by the manufacturer.

To test the null hypothesis, the deviation of the implant position was compared by superimposing the scanbody geometry in the printed model with the reference model, with regard to the Euclidean distance and along the x, y, and z axes. The x and y axes together ensure two-dimensional movement in the print plane and thus generate the horizontal outline of the model to be printed. The resolution of the printer plays a decisive role here. In the case of DLP technology, this depends on the pixel density of the projector. In this context, a smaller print area also means a higher resolution. With SLA technology, this resolution is theoretically infinitely variable, as the laser has no fixed pixels. In practice, the resolution is limited by the dot size of the laser and its movement via the mirror system [29]. Figure 7 shows a schematic representation of the different operating modes in relation to their resolution. The z-axis adds the third dimension by changing the height of the print. The precision of the movement of the building platform and the curing depth of the light source used for polymerization may play a decisive role here. The larger deviations in the z-axis shown in our study for the SLA printer compared to the two DLP printers could indicate a lower precision in this movement.

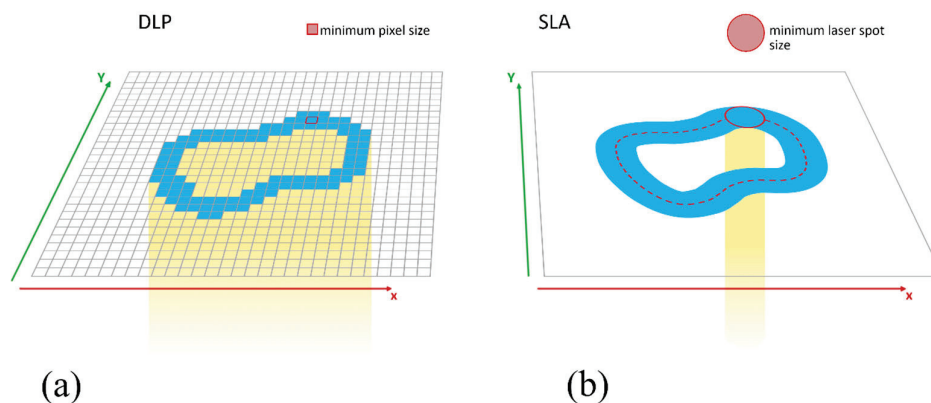


Figure 7. A comparison of the functionality of DLP versus SLA in relation to the x/y-axis and the resulting resolution. (a) In the DLP process, the entire layer is exposed simultaneously. The resolution is determined by the pixel size. (b) In the SLA process, the layer is drawn by a laser dot. The resolution depends on the diameter of the laser spot.

Overall, the null hypothesis was rejected, as there were statistically significant deviations in the implant positions in the printed models depending on the printer used. The lower deviations in Euclidean distance achieved by the two DLP printers (0.07 ± 0.02 mm for Straumann P30 group B and 0.12 ± 0.13 mm for ASIGA MAX UV group A) can be compared with values from similar studies [27,28,37]. In a study in which surgical guides were produced using DLP printers (D20 II; Rapid Shape) and SLA printers (Form 2; Formlabs), Vinzenz Le et al. found that the accuracy of the DLP-produced guides was higher [28]. Elisa Caussin et al. explained in a systematic review that for small print volumes, DLP printers enable higher accuracy than SLA printers, as the resolution of the DLP technology's projector is higher on a smaller area than can be achieved by the laser of an SLA printer [29]. At $85 \mu\text{m}$, the diameter of the laser spot on the SLA Form 3+ printer examined in this study is larger than the individual pixels of the DLP printers examined. According to the manufacturer, these have a resolution of $62 \mu\text{m}$ (ASIGA MAX UV) and $34 \mu\text{m}$ (Straumann P30). The resolution of the printers examined in the x and y planes thus follows the same gradation as the results measured in this study with regard to the accuracy achieved. A direct correlation with the resolution of the printer used is therefore very likely.

A study that examined two of the printers investigated here with regard to the accuracy of a printed maxillary dental arch also found a higher trueness for the DLP printer ASIGA Max UV compared to the SLA printer Form 3+. The simultaneous and therefore more even

polymerization in each layer was cited as a possible reason for this [27]. In a systematic review from 2020, deviations of 59–150 μm were classified as clinically acceptable for implant prosthetic restorations on printed models [38]. In the present study, all measured deviations of the DLP printer Straumann P30 (group B) showed values below 150 μm , so that the printer can be assessed as suitable for the production of implant models. On average, the measured values of the DLP printer ASIGA Max UV (group A) were also below 150 μm , although a deviation above 150 μm was measured at 3 of the 30 measuring points. With a maximum deviation of 150 μm , the models of the SLA printer (group C) can be rated as clinically unacceptable, which showed significantly larger deviations of 0.26 ± 0.17 mm both in individual measurements and on average. The largest deviations were found in the Y (0.17 ± 0.06 mm) and Z axes (0.14 ± 0.20 mm). In a previous study in which the positional accuracy of laboratory analogs in models produced with different printer technologies was also investigated, a larger overall deviation was found for the SLA printer compared to the DLP printer investigated, whereby higher accuracy in the Z-axis was found for the SLA printer than for the DLP printer [24]. However, the comparability of these data with the present study is limited by the different printers and printing parameters used.

The various factors already identified, such as the printer resin and the post-processing used, which make it difficult to draw clear conclusions about the sole dependence of trueness on the printer used and its technology, can be cited as a limitation of this study. Furthermore, all data collected in this study were based on intraoral scans in an idealized in vitro situation. It is to be expected that patient-specific factors in the clinical setting generate additional errors that could accumulate in the further workflow. As a potentially relevant parameter in the design of the models in the CAD software, the set implant-analog-holder (IAH) should be considered, which could influence the positioning of the analogs in the printed model. In the present work, a spacer of 0.060 mm was selected, as this proved best results in preceding studies and was the recommendation of the model design software [31]. The extent to which different values influence the position of the analog should be investigated further. In order to produce the most accurate results possible, a calibration was carried out for each printer before the printing process according to the manufacturer's specifications. A reduced calibration interval or irregular maintenance of the 3D printers could also have a negative impact on the trueness of the printed models.

Future studies should extend the results obtained here with other printers and materials and add clinical parameters such as fit, occlusal, and proximal contacts of implant prosthetic restorations fabricated with printed models. In addition, in view of the large number of different laboratory analogs for printed models, the question arises as to what influence their technology and geometry have on the positional accuracy in the printed model. While the results presented here focus on the dental technology aspect of the digital implant workflow, it still allows conclusions to be drawn about, for example, the suitability of different printer technologies for the production of surgical guides. It is recommended that future studies investigate the extent to which sufficiently precise surgical guides can be produced with different systems and which manufacturing parameters have an influence on accuracy.

5. Conclusions

Within the limitations of this in vitro study, the following conclusions were drawn:

1. The technology of the printer used has a significant influence on the accuracy of 3D-printed implant models ($p < 0.001$).
2. The two DLP printers examined show mostly higher trueness, (0.07 ± 0.02 mm) and (0.12 ± 0.13 mm), respectively, than the SLA printer examined (0.26 ± 0.17 mm).

3. The DLP printer P30 shows very good reproducibility in its results and can be recommended for the fabrication of implant models.

Based on this study, the printer used should be carefully selected depending on the accuracy required for the respective indication. For the production of implant models, a high-precision dental DLP printer is recommended.

It is to be expected that newer generations of 3D printers will increase their accuracy. Larger build platforms and a trend towards the automation of individual work steps, such as separation from the build platform, will make production more efficient and further reduce potential sources of error. The obtained results also provide insights into the suitability of the employed printer technologies for additive manufacturing of other materials, such as ceramics, within a digital workflow. High-precision 3D printers could be utilized for the fabrication of patient-specific ceramic implant bodies, thus expanding the range of applications for 3D printing in implant dentistry.

Author Contributions: Conceptualization, R.E.M. and A.S. (Alexander Strunz); methodology, R.E.M. and A.S. (Alexander Strunz); software, L.B. and A.S. (Anna Seidel); validation, M.W. and J.R.; formal and statistical analysis, W.A.; investigation, R.E.M. and A.S. (Alexander Strunz); data curation, A.S. (Alexander Strunz); writing—original draft preparation, A.S. (Alexander Strunz); writing—review and editing, A.S. (Alexander Strunz), L.B. and R.E.M.; visualization, A.S. (Alexander Strunz); supervision, R.E.M. and M.W.; project administration, R.E.M.; funding acquisition, R.E.M. 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 original contributions presented in this study are included in the article. Further inquiries can be directed to the corresponding author.

Acknowledgments: This work was performed by Strunz in fulfillment of the requirements for obtaining the degree “Dr. med. dent.” at Friedrich Alexander University Erlangen Nuremberg. We would also like to thank the Straumann Group, in particular Medentika and Michael Wagner, for providing the laboratory analogues and scanbodies. The provision of these materials had no influence on the contents of this study.

Conflicts of Interest: The authors declare no conflicts of interest. The funders had no role in the design of the study; in the collection, analyses, or interpretation of the data; in the writing of the manuscript; or in the decision to publish the results.

References

1. Buser, D.; Sennerby, L.; De Bruyn, H. Modern implant dentistry based on osseointegration: 50 years of progress, current trends and open questions. *Periodontology 2000* **2017**, *73*, 7–21. [CrossRef] [PubMed]
2. Hebel, K.; Gajjar, R.; Hofstede, T. Single-tooth replacement: Bridge vs. implant-supported restoration. *J. Can. Dent. Assoc.* **2000**, *66*, 435–438.
3. Lee, D.J.; Saponaro, P.C. Management of Edentulous Patients. *Dent. Clin. N. Am.* **2019**, *63*, 249–261.
4. Sailer, I.; Karasan, D.; Todorovic, A.; Ligoutsikou, M.; Pjetursson, B.E. Prosthetic failures in dental implant therapy. *Periodontology 2000* **2022**, *88*, 130–144. [CrossRef] [PubMed]
5. Papaspyridakos, P.; Chen, C.J.; Gallucci, G.O.; Doukoudakis, A.; Weber, H.P.; Chronopoulos, V. Accuracy of implant impressions for partially and completely edentulous patients: A systematic review. *Int. J. Oral Maxillofac. Implants* **2014**, *29*, 836–845.
6. Matta, R.E.; Adler, W.; Wichmann, M.; Heckmann, S.M. Accuracy of impression scanning compared with stone casts of implant impressions. *J. Prosthet. Dent.* **2017**, *117*, 507–512. [CrossRef]
7. De Angelis, P.; Manicone, P.F.; De Angelis, S.; Grippaudo, C.; Gasparini, G.; Liguori, M.G.; Camodeca, F.; Piccirillo, G.B.; Desantis, V.; D’Amato, G.; et al. Patient and Operator Centered Outcomes in Implant Dentistry: Comparison between Fully Digital and Conventional Workflow for Single Crown and Three-Unit Fixed-Bridge. *Materials* **2020**, *13*, 2781. [CrossRef]

8. Siqueira, R.; Galli, M.; Chen, Z.; Mendonça, G.; Meirelles, L.; Wang, H.L.; Chan, H.L. Intraoral scanning reduces procedure time and improves patient comfort in fixed prosthodontics and implant dentistry: A systematic review. *Clin. Oral. Investig.* **2021**, *25*, 6517–6531.
9. Richert, R.; Goujat, A.; Venet, L.; Viguie, G.; Viennot, S.; Robinson, P.; Farges, J.C.; Fages, M.; Ducret, M. Intraoral Scanner Technologies: A Review to Make a Successful Impression. *J. Healthc. Eng.* **2017**, *2017*, 8427595.
10. Marques, S.; Ribeiro, P.; Falcão, C.; Lemos, B.F.; Ríos-Carrasco, B.; Ríos-Santos, J.V.; Herrero-Climent, M. Digital Impressions in Implant Dentistry: A Literature Review. *Int. J. Environ. Res. Public Health* **2021**, *18*, 1020. [CrossRef]
11. Andriessen, F.S.; Rijkens, D.R.; van der Meer, W.J.; Wismeijer, D.W. Applicability and accuracy of an intraoral scanner for scanning multiple implants in edentulous mandibles: A pilot study. *J. Prosthet. Dent.* **2014**, *111*, 186–194. [PubMed]
12. Kihara, H.; Hatakeyama, W.; Komine, F.; Takafuji, K.; Takahashi, T.; Yokota, J.; Oriso, K.; Kondo, H. Accuracy and practicality of intraoral scanner in dentistry: A literature review. *J. Prosthodont. Res.* **2020**, *64*, 109–113. [PubMed]
13. Kernen, F.R.; Recca, M.; Vach, K.; Nahles, S.; Nelson, K.; Flügge, T.V. In vitro scanning accuracy using different aids for multiple implants in the edentulous arch. *Clin. Oral. Implants Res.* **2022**, *33*, 1010–1020. [CrossRef] [PubMed]
14. Graf, T.; Güth, J.F.; Diegritz, C.; Liebermann, A.; Schweiger, J.; Schubert, O. Efficiency of occlusal and interproximal adjustments in CAD-CAM manufactured single implant crowns—Cast-free vs. 3D printed cast-based. *J. Adv. Prosthodont.* **2021**, *13*, 351–360.
15. Buda, M.; Bratos, M.; Sorensen, J.A. Accuracy of 3-dimensional computer-aided manufactured single-tooth implant definitive casts. *J. Prosthet. Dent.* **2018**, *120*, 913–918.
16. Mata-Mata, S.J.; Donmez, M.B.; Meirelles, L.; Johnston, W.M.; Yilmaz, B. Influence of digital implant analog design on the positional trueness of an analog in additively manufactured models: An in-vitro study. *Clin. Implant Dent. Relat. Res.* **2022**, *24*, 821–830.
17. Rungrojwittayakul, O.; Kan, J.Y.; Shiozaki, K.; Swamidass, R.S.; Goodacre, B.J.; Goodacre, C.J.; Lozada, J.L. Accuracy of 3D Printed Models Created by Two Technologies of Printers with Different Designs of Model Base. *J. Prosthodont.* **2020**, *29*, 124–128.
18. Alshawaf, B.; Weber, H.P.; Finkelman, M.; El Rafie, K.; Kudara, Y.; Papaspyridakos, P. Accuracy of printed casts generated from digital implant impressions versus stone casts from conventional implant impressions: A comparative in vitro study. *Clin. Oral. Implants Res.* **2018**, *29*, 835–842.
19. Revilla-León, M.; Özcan, M. Additive Manufacturing Technologies Used for Processing Polymers: Current Status and Potential Application in Prosthetic Dentistry. *J. Prosthodont.* **2019**, *28*, 146–158.
20. Schweiger, J.; Edelhoff, D.; Güth, J.-F. 3D Printing in Digital Prosthetic Dentistry: An Overview of Recent Developments in Additive Manufacturing. *J. Clin. Med.* **2021**, *10*, 2010. [CrossRef]
21. Jockusch, J.; Özcan, M. Additive manufacturing of dental polymers: An overview on processes, materials and applications. *Dent. Mater. J.* **2020**, *39*, 345–354. [PubMed]
22. van Noort, R. The future of dental devices is digital. *Dent. Mater.* **2012**, *28*, 3–12. [PubMed]
23. Jeong, Y.G.; Lee, W.S.; Lee, K.B. Accuracy evaluation of dental models manufactured by CAD/CAM milling method and 3D printing method. *J. Adv. Prosthodont.* **2018**, *10*, 245–251. [PubMed]
24. Revilla-León, M.; Gonzalez-Martín, Ó.; Pérez López, J.; Sánchez-Rubio, J.L.; Özcan, M. Position Accuracy of Implant Analogs on 3D Printed Polymer versus Conventional Dental Stone Casts Measured Using a Coordinate Measuring Machine. *J. Prosthodont.* **2018**, *27*, 560–567.
25. Ide, Y.; Nayar, S.; Logan, H.; Gallagher, B.; Wolfaardt, J. The effect of the angle of acuteness of additive manufactured models and the direction of printing on the dimensional fidelity: Clinical implications. *Odontology* **2017**, *105*, 108–115.
26. Puebla, K.; Arcaute, K.; Quintana, R.; Wicker, R.B. Effects of environmental conditions, aging, and build orientations on the mechanical properties of ASTM type I specimens manufactured via stereolithography. *Rapid Prototyp. J.* **2012**, *18*, 374–388.
27. Demirel, M.; Diken Türksayar, A.A.; Donmez, M.B.; Yilmaz, B. Effect of 3D printing technology and print orientation on the trueness of additively manufactured definitive casts with different tooth preparations. *J. Dent.* **2024**, *148*, 105244. [CrossRef]
28. Le, V.; Keßler, A.; Folwaczny, M. Influence of DLP and SLA printer technology on the accuracy of surgical guides for implant dentistry in free-end situations. *Int. J. Comput. Dent.* **2023**, *26*, 217–226.
29. Caussin, E.; Moussally, C.; Le Goff, S.; Fasham, T.; Troizier-Cheyne, M.; Tapie, L.; Dursun, E.; Attal, J.P.; François, P. Vat Photopolymerization 3D Printing in Dentistry: A Comprehensive Review of Actual Popular Technologies. *Materials* **2024**, *17*, 950. [CrossRef]
30. Kropfeld, J.; Berger, L.; Adler, W.; Schulz, K.L.; Motel, C.; Wichmann, M.; Matta, R.E. Impact of Scanbody Geometry and CAD Software on Determining 3D Implant Position. *Dent. J.* **2024**, *12*, 94. [CrossRef]
31. Jin, G.; Shin, S.H.; Shim, J.S.; Lee, K.W.; Kim, J.E. Accuracy of 3D printed models and implant-analog positions according to the implant-analog-holder offset, inner structure, and printing layer thickness: An in-vitro study. *J. Dent.* **2022**, *125*, 104268. [CrossRef] [PubMed]
32. Matta, R.E.; Bergauer, B.; Adler, W.; Wichmann, M.; Nickenig, H.J. The impact of the fabrication method on the three-dimensional accuracy of an implant surgery template. *J. Cranio-Maxillofac. Surg.* **2017**, *45*, 804–808. [CrossRef] [PubMed]

33. Motel, C.; Kirchner, E.; Adler, W.; Wichmann, M.; Matta, R.E. Impact of Different Scan Bodies and Scan Strategies on the Accuracy of Digital Implant Impressions Assessed with an Intraoral Scanner: An In Vitro Study. *J. Prosthodont.* **2020**, *29*, 309–314. [CrossRef] [PubMed]
34. Alharbi, N.; Osman, R.; Wismeijer, D. Effects of build direction on the mechanical properties of 3D-printed complete coverage interim dental restorations. *J. Prosthet. Dent.* **2016**, *115*, 760–767. [CrossRef]
35. Park, G.S.; Kim, S.K.; Heo, S.J.; Koak, J.Y.; Seo, D.G. Effects of Printing Parameters on the Fit of Implant-Supported 3D Printing Resin Prosthetics. *Materials* **2019**, *12*, 2533. [CrossRef]
36. Kirby, S.; Pesun, I.; Nowakowski, A.; França, R. Effect of Different Post-Curing Methods on the Degree of Conversion of 3D-Printed Resin for Models in Dentistry. *Polymers* **2024**, *16*, 549. [CrossRef]
37. Chen, Y.; Li, H.; Zhai, Z.; Nakano, T.; Ishigaki, S. Impact of internal design on the accuracy of 3-dimensionally printed casts fabricated by stereolithography and digital light processing technology. *J. Prosthet. Dent.* **2023**, *130*, 381.e1–381.e7. [CrossRef]
38. Etemad-Shahidi, Y.; Qallandar, O.B.; Evenden, J.; Alifui-Segbaya, F.; Ahmed, K.E. Accuracy of 3-Dimensionally Printed Full-Arch Dental Models: A Systematic Review. *J. Clin. Med.* **2020**, *9*, 3357. [CrossRef]

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

Article

Biconometric Connections in Dental Implants: A Pilot Mechanical Study

Nicole Riberti ¹, Michele Furlani ², Emanuele Tognoli ³, Adriano Piattelli ⁴, Luca Comuzzi ⁵, Alessandra Giuliani ^{6,*} and Andrea Gatto ³

¹ Neuroscience, Imaging and Clinical Sciences Department, University of Chieti-Pescara, 66100 Chieti, Italy; nicole.riberti@unich.it

² Biomedical Science and Public Health Department, Polytechnic University of Marche, 60131 Ancona, Italy; m.furlani@staff.univpm.it

³ Department of Engineering “Enzo Ferrari”, University of Modena and Reggio Emilia, 41125 Modena, Italy; emanuele.tognoli@unimore.it (E.T.); agatto@unimore.it (A.G.)

⁴ School of Dentistry, Saint Camillus International University for Health Sciences, 00131 Rome, Italy; adriano.piattelli@unicamillus.org

⁵ Independent Researcher, 31020 San Vendemiano, TV, Italy; luca.comuzzi@gmail.com

⁶ Odontostomatologic and Specialized Clinical Sciences Department, Polytechnic University of Marche, 60131 Ancona, Italy

* Correspondence: a.giuliani@univpm.it

Abstract: Background: In dental implants, micro-gaps at the fixation–abutment interface can cause peri-implantitis and/or loosening or loss of the fixation screw; therefore, three-dimensional imaging is widely used to examine different types of connections. In the present study, we focus on the analysis on biconometric connections to detect and (possibly) measure the presence of micro-gaps in the as-positioned state and after repeated loading and unloading. Methods: Seven biconometric dental implants were characterized using micro-computed tomography (micro-CT). In two specimens (group 1), the cap was inserted, and only the apical portion was imaged, to evaluate the cap–abutment connection; in the remaining five specimens (group 2), the fixture–abutment connection was analyzed. Two implants in group 2 were also subjected to load tests to verify whether stresses could induce the formation of micro-gaps as a consequence of preload loss. Results: Micro-CT analysis showed the absence of micro-gaps greater than 10 µm in both cap–abutment and abutment–fixture connections. This was verified, in the fixture–abutment connection, even after mechanical loading and unloading. The results were reproducible in all the investigated samples in the different experimental conditions. Conclusions: In the human force range during chewing, the conical connection showed a high level of resistance to micro-gap formation at the implant–abutment interface. The absence of micro-gaps, as demonstrated here, provides encouraging preliminary data regarding the stability of the biconometric connections, which will be further verified in follow-up studies on a larger sample size.

Keywords: dental implants; conometric connection; mechanical testing; micro-computed tomography; micro-gaps

1. Introduction

A mismatch between fixture and abutment in dental implants may result in two main implications: biological complications, such as peri-implantitis [1,2], and mechanical complications, such as loosening or loss of the fixation screw and, consequently, fracture

of the prosthetic superstructure [3]. Therefore, a good fixture–abutment fit, together with the choice of proper materials/coatings for implant and crown manufacturing [4,5], can determine the success of the implant and good osseointegration [6].

In this context, the availability of a peri-implant mucosa of adequate size and organization certainly plays a primary role, as it allows for a correct continuum between epithelial and connective tissue, avoiding bone resorption [7,8]. Furthermore, the implant–abutment interface also controls the degree and location of peri-implant inflammatory cell accumulation, which can contribute to the extent of implant-associated alveolar bone loss [9]. Indeed, if the soft tissue size is not adequate, bone resorption occurs. Relatedly, changes in bone ridge level appear to be dependent on the location of the micro-gap [10,11].

To date, the two most-used techniques for prosthetics on endo-osseous implants are screwed and cemented ones. Screwing the prostheses onto the implants has the advantage of an easier application of the prosthesis and equally easier removal, but it has issues related to the access holes for the screwdriver, which must always be closed for esthetic reasons. This step requires the use of thin porcelain, with a risk of chipping, and it could represent an access route for bacteria that can contaminate the inner portion of the implant [12]. Conversely, the cementation of prostheses has the primary advantage of a lack of screw holes, enhancing esthetics; however, the principal disadvantage is the challenge of removing the restoration once cemented, compounded by the cement frequently accumulating beneath the gum, which is difficult to remove and serves as a prime reservoir for plaque and inflammation, often resulting in “pericementitis”. As previously noted, in cemented prostheses, the abutment is also affixed beneath the gingiva, leading to the aforementioned issues [13].

Currently, a conometric connection represents an alternative with a different rationale, as it does not involve the use of either screwing or cementation. The conometric prosthesis utilizes a tapered abutment affixed to the fixture via a Cone–Morse connection, positioned in a platform-switching manner beneath the gingival and bone profile. This is implemented during the implant surgery using the “One Abutment One Time” technique, which regulates all variables concerning the infiltration of organic fluids and tissue debris at the interface. The abutment, uniform across all implant diameters and lengths, is distinguished solely by its height/angle adjustment, to accommodate tissue thickness for superficialization and to correct disparallelisms between implants. The benefits include the absence of holes, the elimination of cement, and straightforward restoration removal. Indeed, an optimal healthy tissue appearance has been clinically observed, and proved to be sustained throughout time [14,15]. In this context, a bi-abutment connection—called “biconometry”—represents a novelty; it is made of three components: a fixture and abutment, coupled via a Cone–Morse connection, and a metal or poly-ether-ether ketone (PEEK) cap, coupled by friction to the abutment via a 4° conometric connection.

According to Schmitt et al. (2014) [16], the conical connection provides an ideal fit between the implant and the abutment, resulting in isolation of the internal part of the implant from the oral cavity. In this way, microleakage and the risk of rapid loosening are reduced. Furthermore, the load of the conical connection is transferred along the two constructions, ensuring an even distribution of forces and also providing greater stability from a mechanical point of view [16,17]. Moreover, in a literature review on different implant connections, Vinhas et al. (2020) showed that Cone–Morse connections are less prone to mechanical complications, such as fractures and screw loss [18].

Despite the success demonstrated by conical connections in clinical practice, the incidental presence of micro-gaps continues to be repeatedly investigated within different frameworks and at different resolutions. Among these, worthy of mention is a study that used synchrotron radiation to analyze, during compression tests up to 100 N, the

mechanical behavior in the coupling area of conical implants, also attempting to detect micro-movements [19]. Another study recorded, by means of a high-resolution and high-speed X-ray camera, a chewing simulator with dynamic and static loading up to 200 N on twenty implants, some with conical implant–abutment connections and others with flat connections; the conical connections did not show micro-gaps, while some micro-movements were found for the flat connections, both in static and dynamic loading conditions [20]. Other studies showed that crestal bone alterations around titanium implants are considerably influenced by potential movements between implants and abutments, but not by micro-gap size [21]. Some other groups investigated preload loss and bacterial penetration at the implant–abutment interface of conical and internal–external hexagon connection systems exposed to temperature cycling and mechanical wear, in order to determine the influence of implant geometry on bacterial penetration and load deformations as a result of testing different models [22–25].

In this context, micro-computed tomography (micro-CT) has allowed for studying the effectiveness of the seal between the implant and abutment in different settings [26–29]; indeed, it was recently demonstrated by synchrotron-based micro-CT that adequate and well-structured connective tissue surrounded the abutment in a conometric system implanted in a patient. Specifically, the implant’s effectiveness was supported by a quantitative assessment of the connective tissue achieved by artificial intelligence tools, featuring interlaced longitudinal and transverse collagen bundles [30].

The present study aims to detect and measure, by micro-CT, the possible presence of micro-gaps in biconometric connections, before and after application of a series of two successive load tests, the first up to 150 N and the second up to 300 N. Indeed, beyond the above-mentioned successful clinical results shared in the reference literature, this study is necessary to start a systematic investigation into the mechanical performance of biconometric connections, which have so far been marginally treated in the literature.

2. Materials and Methods

2.1. Experiment Workflow

Tests to detect the possible presence of micro-gaps were carried out by micro-CT analysis.

In the first phase of the study, a total of seven ($n = 7$) implant samples were acquired, each of them screwed in plexiglass in the basal part (Figure 1). In two ($n = 2$) dental implants (group 1), the cap was inserted; here, only the apical portion was acquired, to evaluate the junction with the underlying abutment. In the remaining five ($n = 5$) samples (group 2), the connection portion between the fixture and the abutment was analyzed.

In the second phase of the study, two ($n = 2$) implants of group 2 were also subjected to mechanical tests. In order to reproduce in-service conditions, these samples were screwed in plexiglass at an angle of inclination of 60° relative to the horizontal plane (Figure 1). A compression test was performed twice on each sample: the first test was conducted up to a load of 150 N, while the second was conducted up to 300 N. After each load test, a micro-CT scan was performed to detect and eventually measure the presence of micro-gaps at the fixture/abutment connection post-compression up to 150 N and post-compression up to 300 N, as a result of the relative displacement generated by the load cycle.

A table summarizing the groups and sample size is reported in Figure 1b.

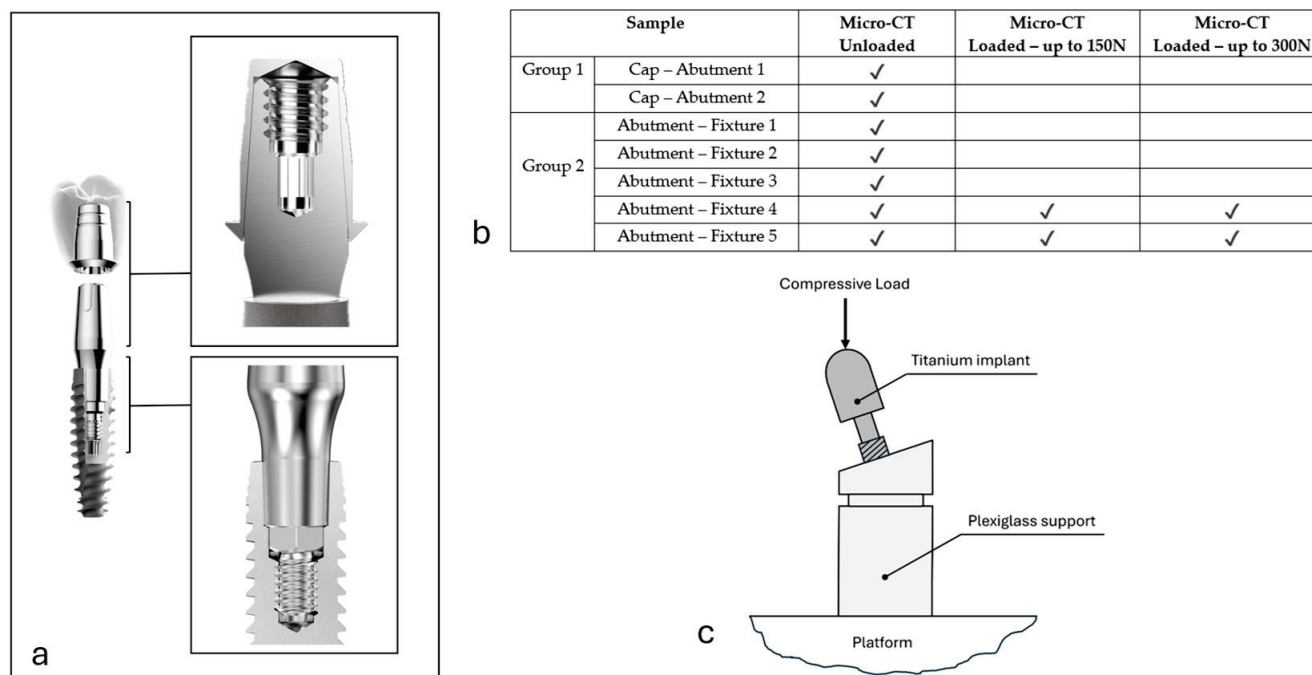


Figure 1. (a) Biconometric connection made of three components: fixture, abutment, and cap; (b) table summarizing groups and sample size; (c) scheme of titanium implant screwed in plexiglass support and subjected to external compressive load. Arrow indicates direction in which load was applied.

2.2. Implant Manufacturing

The investigated biconometric dental implants (AoN Implants Srl, Grisignano di Zocco (VI)—Italy) were endo-osseous screws designed for dental surgery; they can be implanted in previously prepared sites of edentulous areas, according to specific operative sequences.

A biconometric connection (Figure 1a) is made up of three components: (1) the implant (or fixture); (2) the abutment, which is coupled to the implant via a Cone–Morse connection; (3) the cap, coupled by friction to the abutment via a 4° conometric connection.

The material used for the implant production was cold-drawn Gr. 4 Titanium; it was selected for its strength and because it is not contaminated by other elements. The machining process was carried out through numerically controlled machinery, nominally maintaining control down to 1 µm details. During production, the produced components were checked by the systems VICI Vision (Vici & C S.p.A., Santarcangelo di Romagna (RN), Italy) and Hexagon OPTIV M (Hexagon Metrology Spa, Grugliasco (TO), Italy). The body of the implant was structured with a two-main loop so that, with each revolution, the screw advanced twice as much as a classic pitch implant, without creating stress in the recipient bone.

Afterwards, in the post-production checks, thorough cleaning was performed to completely degrease the parts with an ultrasonic metal washer (ILSA-MC Srl, San Pietro in Casale (BO), Italy); moreover, each component was checked for quality by visual inspection and centesimal calipers, micrometers, pin calipers, and internal and external thread calipers. Subsequently, a final cleaning process was performed, and the product was bagged in medical paper.

The abutment, made of Gr. 5 Titanium, was washed and checked as described for the implant; furthermore, the prosthetic cap, also made of Gr. 5 Titanium, was checked with the previously described instruments.

The abutment–implant coupling was performed by fixing the screw at 30 Ncm, as specified in the manufacturer’s instructions, while the cap–abutment coupling was per-

formed using a calibrated instrument by applying a force of 30 Ncm axially to the head of the cap, as previously reported [31].

2.3. X-Ray Micro-CT Analysis

The Metro-Tom 1500 ZEISS (Carl Zeiss Industrielle Messtechnik GmbH, Oberkochen, Germany) was used for three-dimensional (3D) micro-CT analysis, setting the X-ray source with a voltage at 200 kV and a current at 40 μ A. The pixel size was set to 7.93 μ m. An aluminum filter, 1.5 mm thick, was introduced to select photons in the high-energy range; finally, two hours of scanning time were spent for each specimen, with a total of 2050 projections acquired per sample. The image reconstruction was automatically performed by the Metrotom os (vers. 3.0) software.

2.4. Image Segmentation, Thickness Map, and Image Registration

All images were elaborated with the Dragonfly software (Dragonfly, vers. 2022.2, Comet Technologies Canada Inc., Montreal, QC, Canada; software available at <https://dragonfly.comet.tech/>, accessed on 21 March 2025) [32]. The software was used in the image processing phase; it allowed for a search for micro-gaps at the cap–abutment and fixture–abutment connections, segmenting the different components of the samples and creating thickness maps capable of detecting any internal micro-gap caused by relative misalignment after the compression tests. The software has the capability to calculate and create color-coded meshes that display the reference values of the local thickness between boundary points, which is useful for visualization and analysis. The thickness was determined by measuring the maximum diameter of an imaginary sphere that could be inserted inside the boundary walls.

Registration between the 3D reconstructions performed after the 150 N load and the one after the 300 N load was performed using the Dragonfly Image Registration tool. This software module can automatically register datasets by applying the scaling, rotations, and translations required to match features between two datasets. Two matching processes are made available by the software for registering datasets: the Mutual Information and the SSD (sum of squared differences). These methods are based on different concepts to quantify the degree of similarity between images and to apply the required linear transformations. The results may differ significantly: the performance of each algorithm is also dependent on the input data and the mode (Basic or Advanced) selected for registration. After several trials, we chose the SSD method in the Advanced mode.

2.5. Mechanical Compression Tests

The compression tests were performed on an INSTRON 5567 machine (Instron, Norwood, MA, USA) equipped with a load cell with a maximum load capacity of 1 kN. The speed of the crosshead was kept constant at a value of 5 mm/min during the test. Both samples were tested twice, with a first compression test up to a maximum load of 150 N, and a second compression test up to a maximum load of 300 N. At the end of each test, the sample was unloaded immediately after reaching the maximum load. The load was applied directly to the upper part of the implant, in a direction perpendicular to the base of the sample (Figure 1c). The curves of the tests were determined as the compressive displacement (X-axis) and compressive load (Y-axis) for both tested samples.

3. Results

3.1. Micro-CT Analysis

Figure 2 shows the different portions acquired by micro-CT before the mechanical compression tests. In the top panels (Figure 2b), the connection of the cap to the abutment is

imaged, while the implant–abutment connection is observed in the bottom panel (Figure 2c). The portions rendered in yellow or red represent the structural voids inside the sample, respectively, in the apical portion or in the implant–abutment junction.

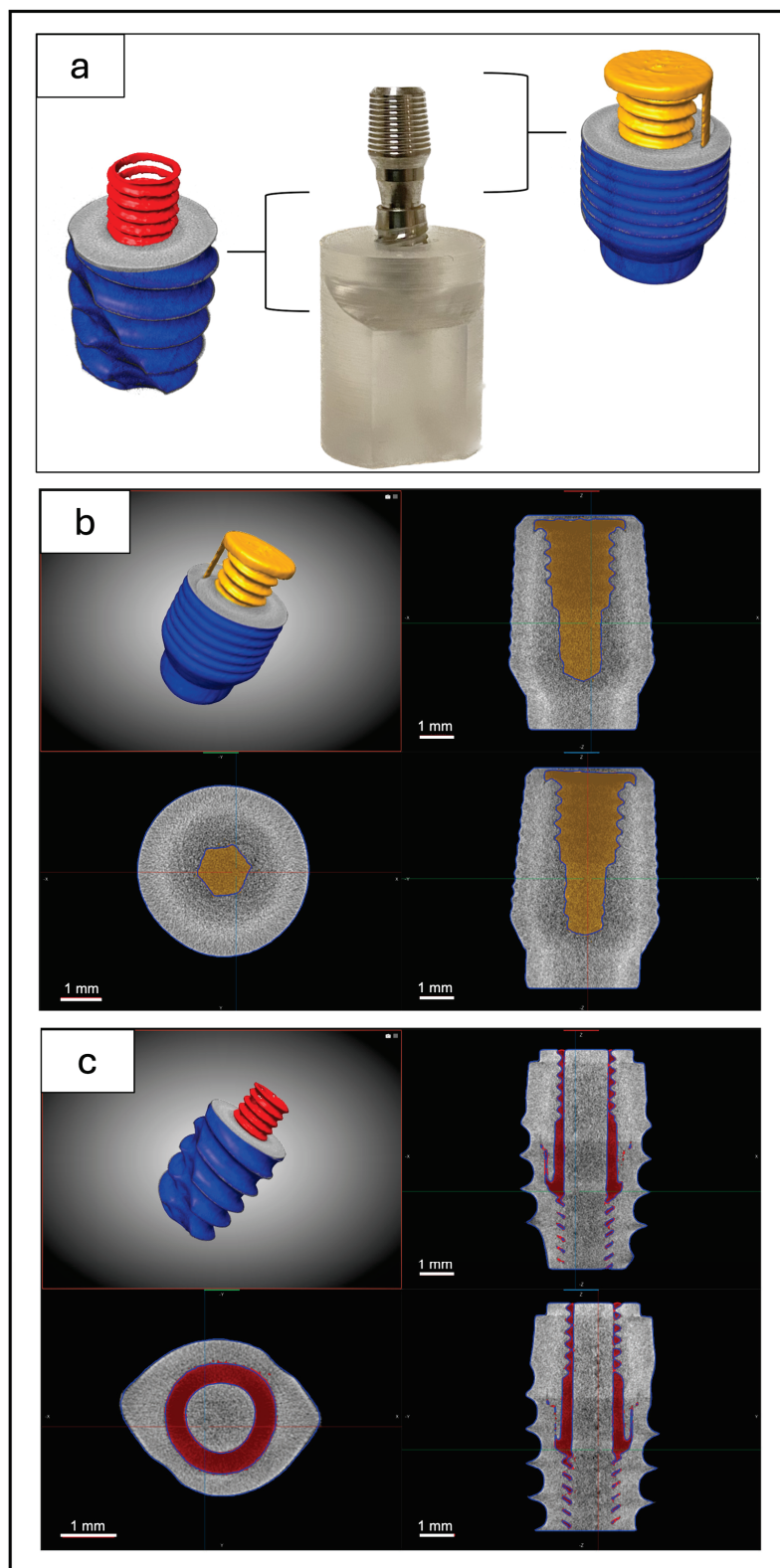


Figure 2. (a) Micro-CT 3D reconstructions: different portions of representative samples imaged in static conditions after preload at 30 Ncm; (b) apical portion: cap–abutment connection; (c) implant–abutment connection. Portions rendered in yellow or red represent structural voids inside samples, respectively, in apical portion or in implant–abutment junction.

The sharp difference in gray tones between the background (i.e., voids) and the titanium material allowed for segmentation of the brightness histograms using the thresholding method. From the highlighted voids and according to the thickness analysis performed, there was no evidence of micro-gaps either between the implant and abutment or between the abutment and cap. Therefore, at the present experimental resolution, the presence of micro-gaps of size $>10\text{ }\mu\text{m}$, after the preload at 30 Ncm, can be excluded in both cap–abutment and abutment–fixture connections.

All micro-CT analyses performed on the samples before compressive loading excluded the presence of micro-gaps both in the two samples of group 1, referring to the study of the cap–abutment connection, and in the five samples of group 2, referring to the study of the fixture–abutment connection.

As a second experimental step, the 3D scans were repeated on two samples of group 2, both after compression up to a 150 N load, and after compression up to a 300 N load. The vertical compression on the implant did not produce evident signs of permanent micro-gaps due to a loss of preload or misalignment. In fact, no micro-gaps were recorded at the implant–abutment connection in either of the two samples, neither after reaching the load of 150 N, nor after loading up to 300 N. To further confirm this result, a registration was performed between the 3D reconstructions after the 150 N load and those after the 300 N load using the Image Registration Dragonfly's feature-based registration workflows.

The registration outputs gave no evident sign of decoupling at the fixture–abutment connection, as shown by the white arrows in Figure 3. In particular, no micro-gaps were found at the fixture–abutment connection for both the samples after the two compression processes, as displayed in the magnifications of Figure 3.

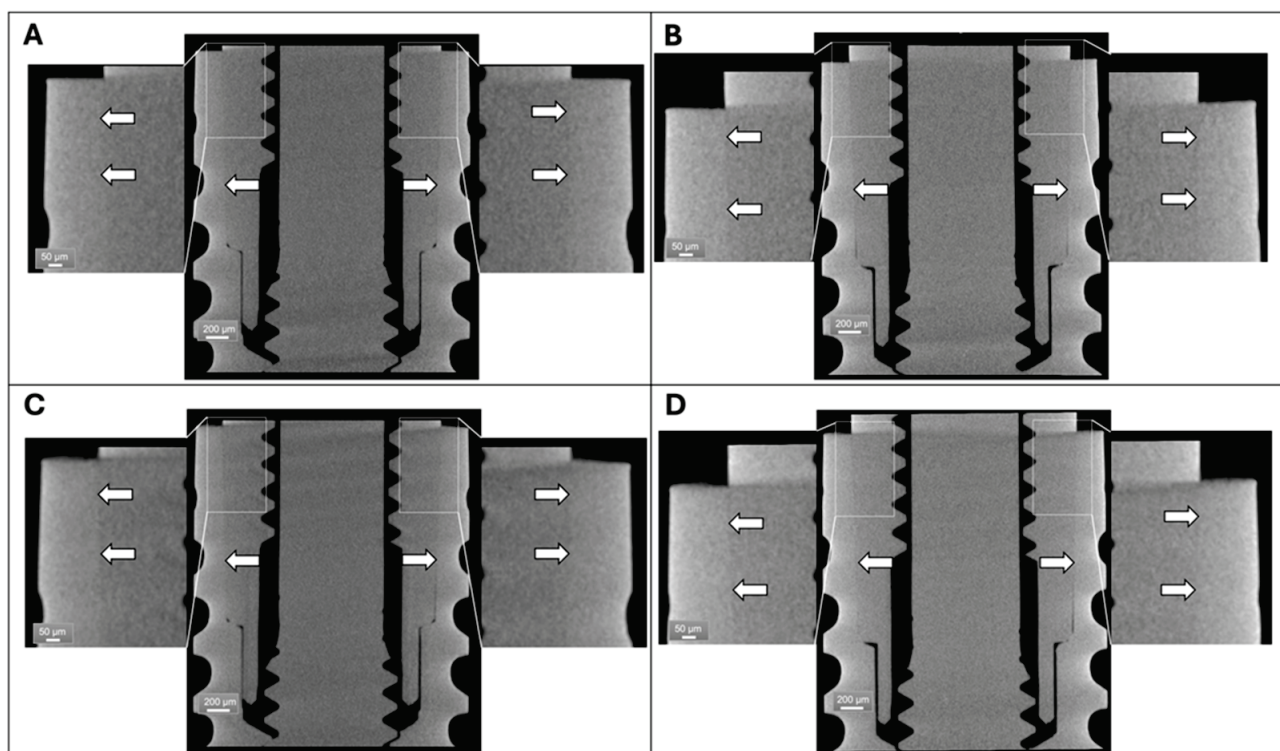


Figure 3. Micro-CT reconstruction in sagittal section of (A) Sample 1, compressed up to 150 N; (B) Sample 2, compressed up to 150 N; (C) Sample 1, compressed up to 300 N; and (D) Sample 2, compressed up to 300 N. White arrows: abutment – fixture interface.

3.2. Mechanical Tests

The load–displacement compression curves were registered for each test performed (Figure 4). The curves reveal, in both samples, that the tests performed up to a maximum load of 300 N have a greater slope than the previous tests carried out—on the same samples—up to a maximum load of 150 N. In addition, for both applied loads, a deviation can be observed between the curves of the two samples tested, greater than the possible uncertainty of the machine. This phenomenon can be explained by the fact that the preparation of the samples was a manual process, which could have led to slight differences between each sample. Finally, in all the tests performed, the components did not reach their maximum load bearing ability and remained structurally intact, with no visible cracks or damage after the load had been removed.

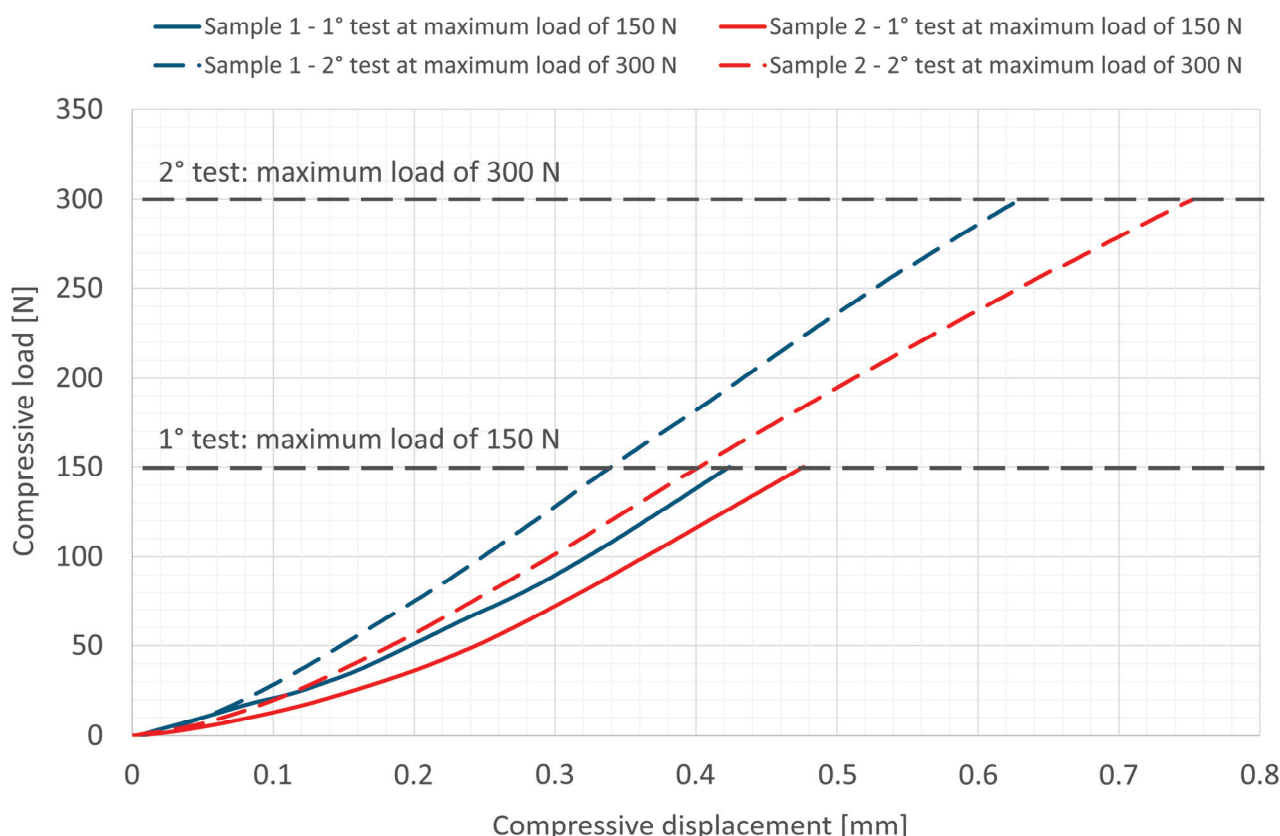


Figure 4. Load–displacement compression curves. Curves are shown for each sample tested, up to maximum load of 150 N (full lines) and 300 N (dotted lines), respectively.

4. Discussion

In this study, the focus is on the two different connections present within a biconometric dental implant: the fixture–abutment connection and the prosthetic cap–abutment connection.

Several clinical studies on this prosthetic technique have been performed [14,15,33–38]; all of these authors reported extremely satisfactory clinical data with the use of this technique, with follow-up ranging from 1 up to 5 years. Two microbiological studies have been conducted to evaluate bacterial infiltration at the level of the caps [39,40], in which no bacterial infiltration was found.

Other studies have reported that the functional load transmission to the fixture is influenced by the abutment connection area’s design [41–44]. Implants with different internal connections (for example, hexagons and octagons) were subjected to cyclic load tests, and it was shown that connections approximating circular shapes are less prone to failure [44]. Furthermore, the materials and production processes of the implant can

influence the presence of micro-gaps and marginal bone loss [45,46]. In particular, implants with different mismatching distances and different prosthetic abutment heights may have different effects with regard to bacterial infiltration and structural fracture of the implant.

For this reason, in the present study, innovative biconometric dental implants, characterized by the selection of Gr. 4 Titanium, with processes controlled at each step at the micrometric level of detail [47], were analyzed using the micro-CT imaging technique to verify the stability of the connections after compressive loading.

Although mechanical tests showed, in both specimens, a steeper slope of the load/displacement curve in the second test, i.e., up to a maximum load of 300 N, compared to tests performed up to a maximum load of 150 N, both specimens showed no decoupling (micro-gaps), i.e., loosening of connection, after the removal of the weight. Moreover, as the components did not reach the compressive load limit, they remained structurally intact, with no fractures, obvious cracks, or damage.

Furthermore, micro-CT allowed for imaging of the samples in three dimensions, achieving evaluation of the internal connections and interfaces between the different components in both of the groups. The acquisition pixel size was set to 7.93 μm , which means that micro-gaps with dimensions $<10 \mu\text{m}$ were not detectable. However, this level of resolution is still reliable for evaluating the quality of the connections, since the micro-gaps that determine an actual instability of the coupling during chewing are in this dimensional range, ultimately producing two effects: the first is biological, creating and amplifying the inflammation of the soft tissues and favoring bacterial invasion; the second is structural, causing possible microfractures of the implant crown, of the implant itself, or of the peri-implant bone.

Acquisitions before use, in the as-implanted condition, were performed to evaluate the quality of the connections just after the preload of the connections at 30 Ncm, while acquisitions following the mechanical tests allowed for documentation of the quality of the connections after a use phase. During the virtual dissection of the 3D images of the samples, no micro-gaps were observed in the apical regions, considered the most critical due to their greater exposure to the apex of the implant. Consequently, the presence of micro-gaps with dimensions $>10 \mu\text{m}$ was excluded. Such good performances are most likely related to the connection concept, which facilitates greater contact between the implant and the abutment, thus mitigating abutment instability (flexion instability), tissue inflammation, bacterial infiltration, and potential implant failure.

The results of the present study support and justify the successful clinical performance of these biconometric implants, already shown in the literature [30,33,35]. Indeed, many surgical and prosthetic benefits for the surgeon, prosthetist, and patient have been documented. For instance, the surgical benefit of sub-crestal implant placement at 1–2 mm was related to the preservation of the implant within the bone envelope, creating additional vertical space for the prosthetist, while also providing a mechanically stable connection that inhibits inflammatory infiltration and bacterial colonization [39,40]. The prosthetist utilizes tapered abutments and hoods to leverage the coupling between components, eliminating the need for cement and screws, and hence enhancing precision while minimizing bulk and ensuring adequate vertical space; this is nearly always achievable. The patient experiences definite outcomes, achieved rapidly; during dental hygiene sessions, plaque removal is easily accomplished at will, and chairside seating is expedited. Additional benefits encompass the utilization of minimal prosthetic components, hence substantially reducing implementation costs and streamlining all operations.

Our observations support the outcome of a previous study, where different connections were compared; indeed, conical connections resulted in reduced microleakage at the implant–abutment interface compared to parallel connections [24]. Moreover, conical

connections appeared to have better performance, exhibiting less peri-implant bone loss; thus, crestal bone levels are more effectively preserved [48].

However, despite the good results achieved, this pilot study presents some limitations that will be partially overcome in the follow-up studies.

The resolution of the CT measurements limits the study to gaps larger than 10 μm , but this threshold should be evaluated against the dimensional accuracy expected to characterize the specific application. In particular, the study involves experimental conditions that replicate a clinical setting where the dimensional deviations are affected by the following: (i) the typical manufacturing tolerance; (ii) the typical positioning tolerance during the surgical phase; (iii) the thermal deformation due to the temperature in the mouth (which is different from the test temperature); and (iv) the different stiffness and consequent differential deformation of the implant fixture and the biological tissues (bone, connective tissue, etc.). All of these factors add up in a chain of tolerances to an expected deviation of more than 10 μm .

Moreover, in our study, bone was modeled with plexiglass. The different stiffness measured during the two loading phases and observed in Figure 4 could very likely be attributable to the plexiglass undergoing changes as a result of loading. This effect will be the subject of follow-up studies to evaluate only the mechanical response of the plexiglass alone, to decouple it from the response of the entire construct.

Other limitations of this study include the limited sample size and loading steps. Indeed, the force generated during routine chewing of foods such as carrots or meat is about 70–150 N. The maximum chewing force in some people can reach up to 500–700 N [49]. Therefore, we selected two values within the range and spaced by 150 N; however, we plan to add two steps, namely 450 N and 600 N, in follow-up studies. Furthermore, regarding the resolution capacity of the tomographic instrumentation, it will be useful to use tomographs capable of reaching higher resolutions, in order to identify and measure even smaller gaps at the fixture/abutment interface.

5. Conclusions

Within the limits of the small sample size, our investigation revealed that, within the range of human force during chewing, the conical connection showed high resistance against the creation of micro-gaps at the implant–abutment and abutment–prosthesis interfaces. The absence of micro-gaps in biconometric implants, at micrometric resolution, as demonstrated here, provides encouraging preliminary data regarding the stability of the biconometric connections, which will be further verified in follow-up studies on a larger sample size.

Author Contributions: Conceptualization, N.R.; data curation, E.T., M.F., A.P. and L.C.; formal analysis, A.G. (Alessandra Giuliani) and A.G. (Andrea Gatto); investigation, N.R., E.T. and M.F.; methodology, N.R., E.T. and M.F.; project administration, A.P. and L.C.; resources, A.G. (Alessandra Giuliani), A.G. (Andrea Gatto), A.P. and L.C.; software, N.R. and M.F.; supervision, A.G. (Alessandra Giuliani) and A.P.; visualization, A.P. and L.C.; writing—original draft, N.R., E.T. and A.G. (Alessandra Giuliani); writing—review and editing, M.F., A.P., A.G. (Alessandra Giuliani), A.G. (Andrea Gatto), and L.C. 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 original contributions presented in this study are included in the article. Further inquiries can be directed to the corresponding author.

Acknowledgments: The authors acknowledge AoN Implants Srl for providing all the dental implants studied in this investigation. The authors do not have any financial interests, either directly or indirectly, in the products or information mentioned or elaborated within this paper. Nicole Riberti has a doctorate fellowship (code n. DOT1353500-1) in the framework of PON RI 2014/2020, Action IV. 4 “Doctorates and research contracts on innovation issues”, DM 1061/2021, funded by the Ministry of University and Research (MUR), Italy, FSE REACT-EU.

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

Abbreviations

The following abbreviations are used in this manuscript:

Micro-CT Micro-computed tomography

References

1. Fernandes, G.V.d.O.; Martins, B.G.d.S.; Fraile, J.F. Revisiting Peri-Implant Diseases in Order to Rethink the Future of Compromised Dental Implants: Considerations, Perspectives, Treatment, and Prognosis. *Dent. Med. Probl.* **2024**, *61*, 637–640. [CrossRef] [PubMed]
2. Lo Bianco, L.; Montevecchi, M.; Ostanello, M.; Checchi, V. Recognition and Treatment of Peri-Implant Mucositis: Do We Have the Right Perception? A Structured Review. *Dent. Med. Probl.* **2021**, *58*, 545–554. [CrossRef] [PubMed]
3. Katsoulis, J.; Takeichi, T.; Sol Gaviria, A.; Peter, L.; Katsoulis, K. Misfit of Implant Prostheses and Its Impact on Clinical Outcomes. Definition, Assessment and a Systematic Review of the Literature. *Eur. J. Oral Implantol.* **2017**, *10* (Suppl. S1), 121–138.
4. Abarno, S.; Gehrke, A.; Dedavid, B.; Gehrke, S. Stress Distribution around Dental Implants, Generated by Six Different Ceramic Materials for Unitary Restoration: An Experimental Photoelastic Study. *Dent. Med. Probl.* **2021**, *58*, 453–461. [CrossRef]
5. Biju, D.; Arumugam, P.; Kannan, S.; Kumar Yadalam, P.; Ronsivalle, V.; Cicciù, M.; Minervini, G. Development, Characterization, and Biocompatibility and Corrosion Analyses of a Silver-Decorated Graphene Oxide and Chitosan Surface Coating for Titanium Dental Implants: A Preliminary Report. *Dent. Med. Probl.* **2024**, *61*, 627–632. [CrossRef]
6. Krebs, M.; Schmenger, K.; Neumann, K.; Weigl, P.; Moser, W.; Nentwig, G. Long-Term Evaluation of ANKYLOS® Dental Implants, Part I: 20-Year Life Table Analysis of a Longitudinal Study of More Than 12,500 Implants. *Clin. Implant. Dent. Relat. Res.* **2015**, *17*, e275–e286. [CrossRef] [PubMed]
7. Araujo, M.G.; Lindhe, J. Peri-implant Health. *J. Periodontol.* **2018**, *89*, S230–S236. [CrossRef]
8. Arai, Y.; Takashima, M.; Matsuzaki, N.; Takada, S. Marginal Bone Loss in Dental Implants: A Literature Review of Risk Factors and Treatment Strategies for Prevention. *J. Prosthodont. Res.* **2024**, *69*, 12–20. [CrossRef]
9. Broggini, N.; McManus, L.M.; Hermann, J.S.; Medina, R.; Schenk, R.K.; Buser, D.; Cochran, D.L. Peri-Implant Inflammation Defined by the Implant-Abutment Interface. *J. Dent. Res.* **2006**, *85*, 473–478. [CrossRef]
10. Piattelli, A.; Vrespa, G.; Petrone, G.; Iezzi, G.; Annibali, S.; Scarano, A. Role of the Microgap Between Implant and Abutment: A Retrospective Histologic Evaluation in Monkeys. *J. Periodontol.* **2003**, *74*, 346–352. [CrossRef]
11. Molinero-Mourelle, P.; Cascos-Sanchez, R.; Yilmaz, B.; Lam, W.Y.H.; Pow, E.H.N.; Del Río Highsmith, J.; Gómez-Polo, M. Effect of Fabrication Technique on the Microgap of CAD/CAM Cobalt–Chrome and Zirconia Abutments on a Conical Connection Implant: An In Vitro Study. *Materials* **2021**, *14*, 2348. [CrossRef] [PubMed]
12. Ma, S.; Fenton, A. Screw- Versus Cement-Retained Implant Prostheses: A Systematic Review of Prosthodontic Maintenance and Complications. *Int. J. Prosthodont.* **2015**, *28*, 127–145. [CrossRef] [PubMed]
13. Gómez-Polo, M.; Ortega, R.; Gómez-Polo, C.; Celemin, A.; Del Río Highsmith, J. Factors Affecting the Decision to Use Cemented or Screw-Retained Fixed Implant-Supported Prostheses: A Critical Review. *Int. J. Prosthodont.* **2018**, *31*, 43–54. [CrossRef]
14. Degidi, M.; Nardi, D.; Sighinolfi, G.; Degidi, D. The Conometric Concept for the Definitive Rehabilitation of a Single Posterior Implant by Using a Conical Indexed Abutment: A Technique. *J. Prosthet. Dent.* **2020**, *123*, 576–579. [CrossRef]
15. Degidi, M.; Nardi, D.; Sighinolfi, G.; Degidi, D.; Piattelli, A. The Conometric Concept: Definitive Rehabilitation of the Single Posterior Implant Using the Acuris Abutment: A Preliminary Report. *Clin. Oral Implants Res.* **2020**, *31*, 223. [CrossRef]
16. Schmitt, C.M.; Nogueira-Filho, G.; Tenenbaum, H.C.; Lai, J.Y.; Brito, C.; Döring, H.; Nonhoff, J. Performance of Conical Abutment (Morse Taper) Connection Implants: A Systematic Review. *J. Biomed. Mater. Res. A* **2014**, *102*, 552–574. [CrossRef]
17. Arshad, M.; Mahgoli, H.; Payaminia, L. Effect of Repeated Screw Joint Closing and Opening Cycles and Cyclic Loading on Abutment Screw Removal Torque and Screw Thread Morphology: Scanning Electron Microscopy Evaluation. *Int. J. Oral Maxillofac. Implants* **2018**, *33*, 31–40. [CrossRef]
18. Vinhas, A.S.; Aroso, C.; Salazar, F.; López-Jarana, P.; Ríos-Santos, J.V.; Herrero-Climent, M. Review of the Mechanical Behavior of Different Implant–Abutment Connections. *Int. J. Environ. Res. Public Health* **2020**, *17*, 8685. [CrossRef]

19. Rack, A.; Rack, T.; Stiller, M.; Riesemeier, H.; Zabler, S.; Nelson, K. In Vitro Synchrotron-Based Radiography of Micro-Gap Formation at the Implant-Abutment Interface of Two-Piece Dental Implants. *J. Synchrotron Radiat.* **2010**, *17*, 289–294. [CrossRef]
20. Zipprich, H.; Weigl, P.; Ratka, C.; Lange, B.; Lauer, H.C. The Micromechanical Behavior of Implant-Abutment Connections under a Dynamic Load Protocol. *Clin. Implant. Dent. Relat. Res.* **2018**, *20*, 814–823. [CrossRef]
21. Hermann, J.S.; Schoolfield, J.D.; Schenk, R.K.; Buser, D.; Cochran, D.L. Influence of the Size of the Microgap on Crestal Bone Changes Around Titanium Implants. A Histometric Evaluation of Unloaded Non-Submerged Implants in the Canine Mandible. *J. Periodontol.* **2001**, *72*, 1372–1383. [CrossRef]
22. Ricomini Filho, A.P.; Fernandes, F.S.d.F.; Straioto, F.G.; Silva, W.J.d.; Del Bel Cury, A.A. Preload Loss and Bacterial Penetration on Different Implant-Abutment Connection Systems. *Braz. Dent. J.* **2010**, *21*, 123–129. [CrossRef] [PubMed]
23. Piermatti, J.; Yousef, H.; Luke, A.; Mahevich, R.; Weiner, S. An In Vitro Analysis of Implant Screw Torque Loss With External Hex and Internal Connection Implant Systems. *Implant. Dent.* **2006**, *15*, 427–435. [CrossRef] [PubMed]
24. Tsuruta, K.; Ayukawa, Y.; Matsuzaki, T.; Kihara, M.; Koyano, K. The Influence of Implant–Abutment Connection on the Screw Loosening and Microleakage. *Int. J. Implant. Dent.* **2018**, *4*, 11. [CrossRef]
25. Tripodi, D.; Vantaggiato, G.; Scarano, A.; Perrotti, V.; Piattelli, A.; Iezzi, G.; D’Ercole, S. An In Vitro Investigation Concerning the Bacterial Leakage at Implants With Internal Hexagon and Morse Taper Implant-Abutment Connections. *Implant. Dent.* **2012**, *21*, 335–339. [CrossRef]
26. Kowalski, J.; Puszczkarz, A.K.; Radwanski, M.; Sokolowski, J.; Cichomski, M.; Bourgi, R.; Hardan, L.; Sauro, S.; Lukomska-Szymanska, M. Micro-CT Evaluation of Microgaps at Implant–Abutment Connection. *Materials* **2023**, *16*, 4491. [CrossRef]
27. Keklikoglou, K.; Arvanitidis, C.; Chatzigeorgiou, G.; Chatzinikolaou, E.; Karagiannidis, E.; Koletsa, T.; Magoulas, A.; Makris, K.; Mavrothalassitis, G.; Papanagnou, E.-D.; et al. Micro-CT for Biological and Biomedical Studies: A Comparison of Imaging Techniques. *J. Imaging* **2021**, *7*, 172. [CrossRef]
28. Sacker, T.N.; Trentin, M.S.; dos Santos, T.M.P.; Lopes, R.T.; Rivaldo, E.G. In Vitro Microtomography Evaluation of the Implant-Abutment Interface—Gap Microtomography Evaluation/ Avaliação Da Microtomografia in Vitro Da Interface Implante-Pilar—Avaliação Da Microtomografia de Gap. *Braz. J. Dev.* **2021**, *7*, 57552–57565. [CrossRef]
29. Radwanski, M.; Leski, M.; Puszczkarz, A.K.; Sokolowski, J.; Hardan, L.; Bourgi, R.; Sauro, S.; Lukomska-Szymanska, M. A Micro-CT Analysis of Initial and Long-Term Pores Volume and Porosity of Bioactive Endodontic Sealers. *Biomedicines* **2022**, *10*, 2403. [CrossRef]
30. Riberti, N.; D’Amico, E.; Pierfelice, T.V.; Furlani, M.; Giuliani, A.; Piattelli, A.; Iezzi, G.; Comuzzi, L. Case Report of a Dental Implant with Conometric Abutment–Prosthetic Cap Connection: Advanced High-Resolution Imaging and Peri-Implant Connective Tissue Performance. *Clin. Pract.* **2024**, *14*, 556–569. [CrossRef]
31. Ceddia, M.; Comuzzi, L.; Di Pietro, N.; Romasco, T.; Specchiulli, A.; Piattelli, A.; Trentadue, B. Finite Element Analysis (FEA) for the Evaluation of Retention in a Conometric Connection for Implant and Prosthesis. *Osteology* **2023**, *3*, 140–156. [CrossRef]
32. Makovetsky, R.; Piche, N.; Marsh, M. Dragonfly as a Platform for Easy Image-Based Deep Learning Applications. *Microsc. Microanal.* **2018**, *24*, 532–533. [CrossRef]
33. Lupi, S.M.; Todaro, C.; De Martis, D.; Blasi, P.; Rodriguez y Baena, R.; Storelli, S. The Conometric Connection for the Implant-Supported Fixed Prosthesis: A Narrative Review. *Prosthesis* **2022**, *4*, 458–467. [CrossRef]
34. Degidi, M.; Nardi, D.; Sighinolfi, G.; Piattelli, A. The Conometric Concept: Definitive Fixed Lithium Disilicate Restorations Supported by Conical Abutments. *J. Prosthodont.* **2018**, *27*, 605–610. [CrossRef]
35. Lupi, S.M.; De Martis, D.; Todaro, C.; Isola, G.; Beretta, M.; Rodriguez y Baena, R. Conometric Connection for Implant-Supported Crowns: A Prospective Clinical Cohort Study. *J. Clin. Med.* **2023**, *12*, 7647. [CrossRef]
36. Sharma, V.; Paliwal, J.; Bhansali, S.; Meena, K.; Gupta, N.; Dadarwal, A. Evaluation of Clinical and Radiographic Outcome of Friction Fit Conical Abutment System in Implant-Supported Dental Prostheses: An in Vivo Study. *J. Indian Prosthodont. Soc.* **2022**, *22*, 29. [CrossRef]
37. Bressan, E.; Venezzes, A.; Magaz, V.; Lops, D.; Ghensi, P. Fixed Conometric Retention with CAD/CAM Conic Coupling Abutments and Prefabricated Syncone Caps: A Case Series. *Int. J. Periodontics Restor. Dent.* **2018**, *38*, 277–280. [CrossRef]
38. Bressan, E.; Sbricoli, L.; Guazzo, R.; Bambace, M.; Lops, D.; Tomasi, C. Five-Year Prospective Study on Conometric Retention for Complete Fixed Prostheses. *Int. J. Oral Implantol.* **2019**, *12*, 105–113.
39. Gehrke, P.; Burg, S.; Peters, U.; Beikler, T.; Fischer, C.; Rupp, F.; Schweizer, E.; Weigl, P.; Sader, R.; Smeets, R.; et al. Bacterial Translocation and Microgap Formation at a Novel Conical Indexed Implant Abutment System for Single Crowns. *Clin. Oral Investig.* **2022**, *26*, 1375–1389. [CrossRef]
40. Gehrke, P.; Hartjen, P.; Smeets, R.; Gosau, M.; Peters, U.; Beikler, T.; Fischer, C.; Stolzer, C.; Geis-Gerstorfer, J.; Weigl, P.; et al. Marginal Adaptation and Microbial Leakage at Conometric Prosthetic Connections for Implant-Supported Single Crowns: An In Vitro Investigation. *Int. J. Mol. Sci.* **2021**, *22*, 881. [CrossRef]
41. Murmura, G.; Di Iorio, D.; Cicchetti, A.R.; Sinjari, B.; Caputi, S. In Vitro Analysis of Resistance to Cyclic Load and Preload Distribution of Two Implant/ Abutment Screwed Connections. *J. Oral Implantol.* **2013**, *39*, 293–301. [CrossRef] [PubMed]

42. Cashman, P.M.; Schneider, R.L.; Schneider, G.B.; Stanford, C.M.; Clancy, J.M.; Qian, F. In Vitro Analysis of Post-Fatigue Reverse-Torque Values at the Dental Abutment/Implant Interface for a Unitarian Abutment Design. *J. Prosthodont.* **2011**, *20*, 503–509. [CrossRef] [PubMed]
43. Rocha Bernardes, S.; Chiarello de Mattos, M.d.G.; Hobkirk, J.; Faria Ribeiro, R. Loss of Preload in Screwed Implant Joints as a Function of Time and Tightening/Untightening Sequences. *Int. J. Oral Maxillofac. Implant.* **2014**, *29*, 89–96. [CrossRef]
44. Shin, H.-M.; Huh, J.-B.; Yun, M.-J.; Jeon, Y.-C.; Chang, B.M.; Jeong, C.-M. Influence of the Implant-Abutment Connection Design and Diameter on the Screw Joint Stability. *J. Adv. Prosthodont.* **2014**, *6*, 126–132. [CrossRef]
45. Galindo-Moreno, P.; León-Cano, A.; Monje, A.; Ortega-Oller, I.; O'Valle, F.; Catena, A. Abutment Height Influences the Effect of Platform Switching on Peri-implant Marginal Bone Loss. *Clin. Oral Implant. Res.* **2016**, *27*, 167–173. [CrossRef]
46. Laleman, I.; Lambert, F.; Gahlert, M.; Bacevic, M.; Woelfler, H.; Roehling, S. The Effect of Different Abutment Materials on Peri-implant Tissues—A Systematic Review and Meta-analysis. *Clin. Oral Implant. Res.* **2023**, *34*, 125–142. [CrossRef]
47. Deporter, D. Dental Implant Design and Optimal Treatment Outcomes. *Int. J. Periodontics Restor. Dent.* **2009**, *29*, 625–633.
48. Caricasulo, R.; Malchiodi, L.; Ghensi, P.; Fantozzi, G.; Cucchi, A. The Influence of Implant-abutment Connection to Peri-implant Bone Loss: A Systematic Review and Meta-analysis. *Clin. Implant. Dent. Relat. Res.* **2018**, *20*, 653–664. [CrossRef]
49. Zhang, Y.-R.; Du, W.; Zhou, X.-D.; Yu, H.-Y. Review of Research on the Mechanical Properties of the Human Tooth. *Int. J. Oral Sci.* **2014**, *6*, 61–69. [CrossRef]

Disclaimer/Publisher's Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

Article

Accuracy Evaluation of Indirect Bonding Techniques for Clear Aligner Attachments Using 3D-Printed Models: An In Silico and Physical Model-Based Study

Cosmina Raluca Fratila ¹, Luis Óscar Alonso-Ezpeleta ², Manuel Poveda-Saenz ³, Giovanni Giovannini ¹, Ana Belén Lobo-Galindo ⁴, Javier Flores-Fraile ^{4,*} and Álvaro Zubizarreta-Macho ^{1,4}

¹ Faculty of Dentistry, Alfonso X el Sabio University, 28691 Madrid, Spain; cfrat@myuax.com (C.R.F.); giovanni@uax.es (G.G.); amacho@uax.es (Á.Z.-M.)

² Endodontics Section, Department of Stomatology, School of Dentistry, University of Zaragoza, 50009 Zaragoza, Spain; lalopezp@unizar.es

³ Faculty of Medicine, CEU-San Pablo University, 28039 Madrid, Spain; manuelpoveda1@gmail.com

⁴ Department of Surgery, Faculty of Medicine, University of Salamanca, 37008 Salamanca, Spain; alobogal@hotmail.com

* Correspondence: j.flores@usal.es

Abstract: An inaccurate bonding procedure of the attachments related to clear aligner systems could influence the predictability of tooth movement. The aim of this study was to compare the positioning reliability of horizontal and vertical orthodontic clear aligner attachments. **Materials and Methods:** A total of 70 horizontal and 70 vertical orthodontic clear aligner attachments were bonded to five upper and five lower experimental anatomically based acrylic resin models with 14 clinical crowns each. The experimental anatomically based acrylic resin models were randomly distributed to the following study groups: Group A—horizontal orthodontic clear aligner attachments ($n = 70$) (HORIZONTAL) and Group B—vertical orthodontic clear aligner attachments ($n = 70$) (VERTICAL). Afterward, the orthodontic clear aligner attachments were digitally planned using orthodontic planning software, and orthodontic templates were manufactured by thermoforming on 3D-printed models in trilayer glycol-modified polyethylene terephthalate. Both horizontal and vertical orthodontic clear aligner attachments were put through an intra-oral scan to obtain a postoperative digital image, and PAPver, PCPver, PMVver, AUver, Alver, PPMhor, PPDhor, PMVhor, AMhor and ADhor cephalometric parameters were analyzed using a *t*-test or a non-parametric Mann–Whitney–Wilcoxon test. **Results:** The results showed that all cephalometric parameters showed statistically significant differences ($p < 0.05$) between the accuracy of the indirect bonding technique for horizontal and vertical orthodontic clear aligner attachments, except for the PAPver ($p = 0.6079$) and PMVhor ($p = 0.5001$) cephalometric parameters. **Conclusions:** The vertical orthodontic clear aligner attachments are more accurate than the horizontal orthodontic clear aligner attachments through the indirect bonding technique.

Keywords: accuracy; attachments; orthodontics; clear aligners; digital; composite resin

1. Introduction

Patient demand for aesthetic orthodontic treatments is continually increasing in adults and pediatric patients [1]; specifically, gender influences the perception of the aesthetics of the smile [2], with women being more critical than men and generating the most demands for orthodontic treatment. Moreover, Saccomanno et al. reported that 87.94% of patients are

female and 72.57% of them are required to improve their aesthetics, while only 54.84% of male patients state this reason for consultation [3]. Therefore, dentists must have knowledge of the different types of aesthetic orthodontics to make a correct diagnosis and properly execute the treatment plan [4].

Furthermore, Baron reported on three types of orthodontic appliances with a high rate of aesthetic acceptance by patients: thermoplastic aligners, lingual orthodontics and aesthetic braces [5]; however, conventional multibracket appliance techniques do not have high acceptance by patients due to insufficient aesthetics, hygiene problems and quality of life (pain, discomfort when chewing, mouth injuries, etc.) [6,7].

A systematic review published in 2020 verified that fixed orthodontic treatment can have a negative impact on quality of life 24 h after placement of the appliance and up to three months after the start of treatment [8].

Thermoplastic aligners are more discreet and comfortable to wear and can be removed for eating and brushing teeth. Alami et al. evaluated in a descriptive cross-sectional study the perception and satisfaction of patients treated with transparent aligners, and the main reason for choosing this type of orthodontics was its “invisibility” (89.7%) [9]. The possibility of having correct oral hygiene at home is essential for periodontal patients, making this type of orthodontics their main choice if the duration of treatment is longer than 18 months [10]. Likewise, the risk of developing stain lesions, white spots or dental cavities is reduced; therefore, it is a recommended orthodontic treatment for patients with a high risk of cavities. Moreover, Abay et al. analyzed the prevalence of white spots during orthodontic treatment and observed that the number of patients who developed them after treatment with aligners (41.18%) was lower than who underwent fixed orthodontic treatment with self-ligating brackets (63.64%) and conventional brackets (52.94%) [11].

For all these reasons, interest in clear aligners has grown [7]. The treatment consists of wearing transparent thermoplastic splints individualized to each patient for a minimum of 20 h a day, except for eating and drinking [12]. In addition, computer-aided design is combined with three-dimensional virtual model correction software to organize the dental movements and correct malocclusion, obtaining a simulation of the results of the treatment before starting [13,14]. Currently, clear aligners can be used for a wide variety of complex malocclusions using various auxiliary resources such as attachments, mini-screws and elastics [7,15,16].

Moreover, attachments have undergone a great evolution over the years, since at the beginning they only had an ellipsoidal shape (horizontal for active intrusion movements and vertical for retention objectives). Later, rectangular attachments were introduced, which are distinguished by their shape: horizontal, vertical and beveled. In particular, beveled attachments have developed into beveled attachments tailored to the requirement for dental mobility. These attachments serve the same purpose as traditional ones, but they are positioned along the tooth's vertical axis. Teardrop-shaped attachments are used to direct the canines' derotation in situations of several rotations or when the correction is generally larger than two degrees. They are always customized based on the form, length and width of the teeth. Later additions include the “power ridge”, which is utilized to achieve proper vertical control of the incisor axis in numerous motions and enhance the correction of torque larger than three degrees. Similarly, to address deep bites, “bite ramps” were added as a horizontal accessory to the palatal surface of the upper front teeth; they were only inserted buccal in cross-bite situations.

In every situation, the accuracy with which the attachments are cemented and the composite material utilized have a significant impact on how well the aligner, attachment and tooth interact [17].

The nature of composite resin (flowable or conventional) influences the adhesion of attachments to the tooth and also to the shape accuracy. Likewise, the thermoplastic template used to reproduce the shape of the attachments is a fundamental factor for the precision and predictability of future tooth movement [18,19]. There are several risk factors for early failure of attachments and/or changes in their shape and, with it, a loss of retention and predictable movement of the aligners, both related to the operator and the patient. Regarding factors related to the operator, the choice of an appropriate composite with ideal properties represents a relevant aspect for the maintenance of the attachment [20]; hence, flowable composites or orthodontic bonding composites with the greatest amount of filler possible and a rigid transfer template for greater precision and strength of the composite bond are used [21]. In general, patient-related causes of attachment loss are wearing aligners for less than 18 h a day, the use of aligner seats or “chewies” and unilateral chewing. This represented 56.25% of the early loss of the attachment. However, tooth-related reasons resulting in attachment loss were 6.73% [22]. Likewise, it must be considered that the attachment can wear out. Li Q et al. demonstrated that the wear volume of the attachment gradually increases as the treatment time increases. The average wear volume was 0.813 mm^3 after 8 months of treatment [23].

Currently, the scientific literature does not provide information related to the effect of the geometrical design of clear aligner attachments on the positioning reliability of clear aligner attachments, since an inaccurate positioning with respect to planning can lead to an unpredictable orthodontic treatment, undesirable complications and longer-lasting treatments. The present study provides practical implications with direct clinical application that could influence the predictability of tooth movement, making the clinical results of orthodontic treatment with clear aligners more precise.

The aim of this work was to compare the positioning reliability of horizontal and vertical orthodontic clear aligner attachments placed through indirect bonding technique, with a null hypothesis (H_0) stating that no differences were shown between the positioning reliability of horizontal and vertical orthodontic clear aligner attachments placed through indirect bonding technique.

2. Materials and Methods

2.1. Study Design

A total of 70 horizontal and 70 vertical orthodontic clear aligner attachments (Spectra ST flow, Dentsply, Ballaigues, Switzerland) were bonded on 5 upper and 5 lower experimental anatomically based acrylic resin models (Ref. 20-8130-128, EpoxiCure®, Buehler, Lake Bluff, IL, USA) with 14 clinical crowns each, applying a photo-polymerized composite resin cement (Transbond™ XT, 3M ESPE™, Saint Paul, MN, USA) in the middle of the buccal surface of the crown, followed by 20 s of etching the buccal surface with 37% orthophosphoric acid (VOCOCID, VOCO GmbH, Cuxhaven, Germany) and 20 s of applying a photo-polymerized resin adhesive primer (Unitek Transbond™ XT, 3M ESPE™, Saint Paul, MN, USA) to replicate a clinical setting. The research was carried out at the Dental Centre of Innovation and Advanced Specialties at Alfonso X El Sabio University (Madrid, Spain), between September 2023 and March 2024. A randomized controlled study was conducted in accordance with the principles defined in the German Ethics Committee’s statement for the use of organic tissues in medical research (Zentrale Ethikkommission, 2003). Sample size determination was determined for a total of 140 samples (70 per group), which would provide 95% power in determining statistically significant differences in variables analyzed among the 2 groups (effect size: 0.70, $\alpha = 5\%$).

2.2. Experimental Procedure

The following research groups received the experimental anatomically based acrylic resin models at random (Epidat 4.1, Galicia, Spain): Group A, horizontal orthodontic clear aligner attachments ($n = 70$) (HORIZONTAL); and Group B, vertical orthodontic clear aligner attachments ($n = 70$) (VERTICAL). A 3D printing system (ProJet® 6000. 3D Systems®, Rock Hill, SC, USA) was used to construct the upper and lower experimental anatomically based acrylic resin models, which were designed using 2D/3D computer-aided design/computer-aided engineering (CAD/CAE) (Midas FX+®, Milton Keynes, UK). Afterward, the orthodontic clear aligner attachment placement was digitally planned using orthodontic planning software (QuickSmile®, <https://quicksmile.es/> (access on 27 November 2024), Madrid, Spain) (Figure 1).

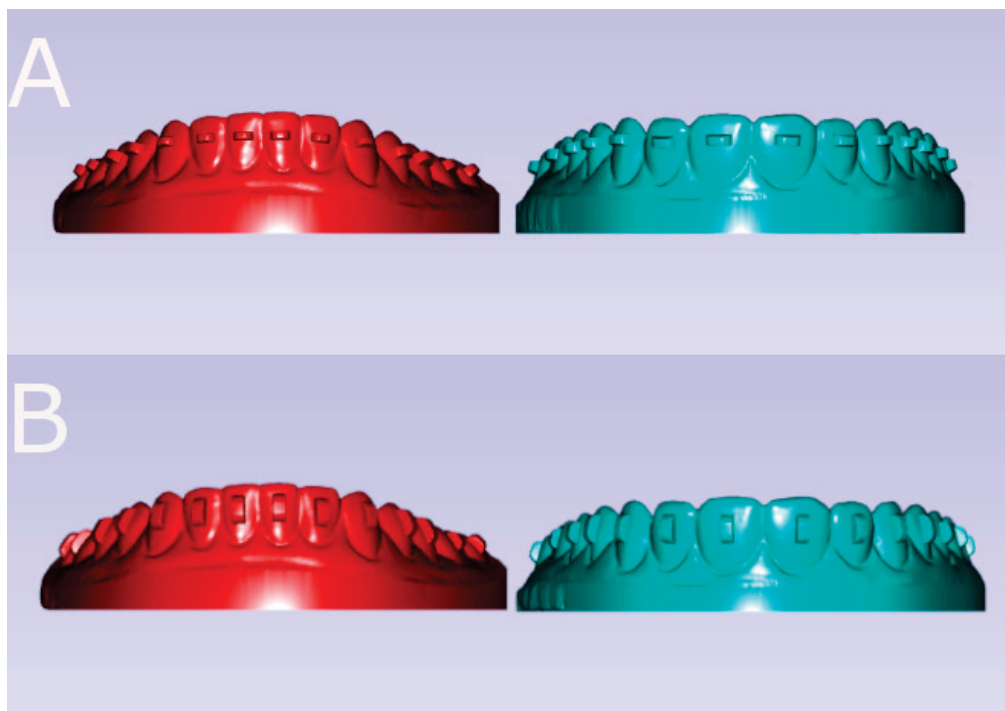


Figure 1. (A) Frontal view of the lower (red) and upper (green) experimental anatomically based acrylic resin models with the horizontal orthodontic clear aligner attachments planning and (B) frontal view of the lower (red) and upper (green) experimental anatomically based acrylic resin models with the vertical orthodontic clear aligner attachments planning.

Then, digital orthodontic templates were also designed using orthodontic planning software (QuickSmile®, Madrid, Spain). Subsequently, the orthodontic templates were manufactured by thermoforming on 3D-printed models (BioStar, Scheu Dental, Madrid, Spain) in 0.76 mm trilayer glycol modified polyethylene terephthalate (PET-G), following the manufacturer's recommendations.

Subsequently, both horizontal and vertical orthodontic clear aligner attachments were performed by flowable composite resin (Spectra ST flow, Dentsply, Ballaigues, Switzerland) on both upper (from tooth 1.7 to 2.7) and lower (from tooth 3.7 to 4.7) acrylic resin models based on experimental anatomy placed in the middle of the clinical crown's buccal surface without etching or resin glue application. In order to obtain an accurate standard tessellation language (STL) digital file, all experimental anatomically based acrylic resin models were then subjected to a postoperative digital impression using an intra-oral scan (iTero Element 5D Plus, Align Technology, Santa Clara, CA, USA) using 3D in-motion video

imaging technology. In accordance with the manufacturer's instructions, the occlusal plane was scanned first, and then the buccal and palatal surfaces, to capture the image.

2.3. Alignment Procedure

Subsequently, the STL digital files were imported into a reverse-engineering morphometric software program (3D Geomagic Capture Wrap, 3D Systems®, Rock Hill, SC, USA), where the teeth were individually segmented to facilitate the accurate alignment procedure between the STL digital files of the virtually planned positions of the digitalized horizontal and vertical orthodontic clear aligner attachments (QuickSmile®, Madrid, Spain) and the postoperative STL digital files of the same attachments bonded to the experimental anatomically based acrylic resin models, following the guidelines established by Brandão et al. [24]. The alignment procedure utilized the palatal surfaces of the anterior teeth and the occlusal and palatal surfaces of the posterior teeth, employing the best-fit algorithm (Figure 2).

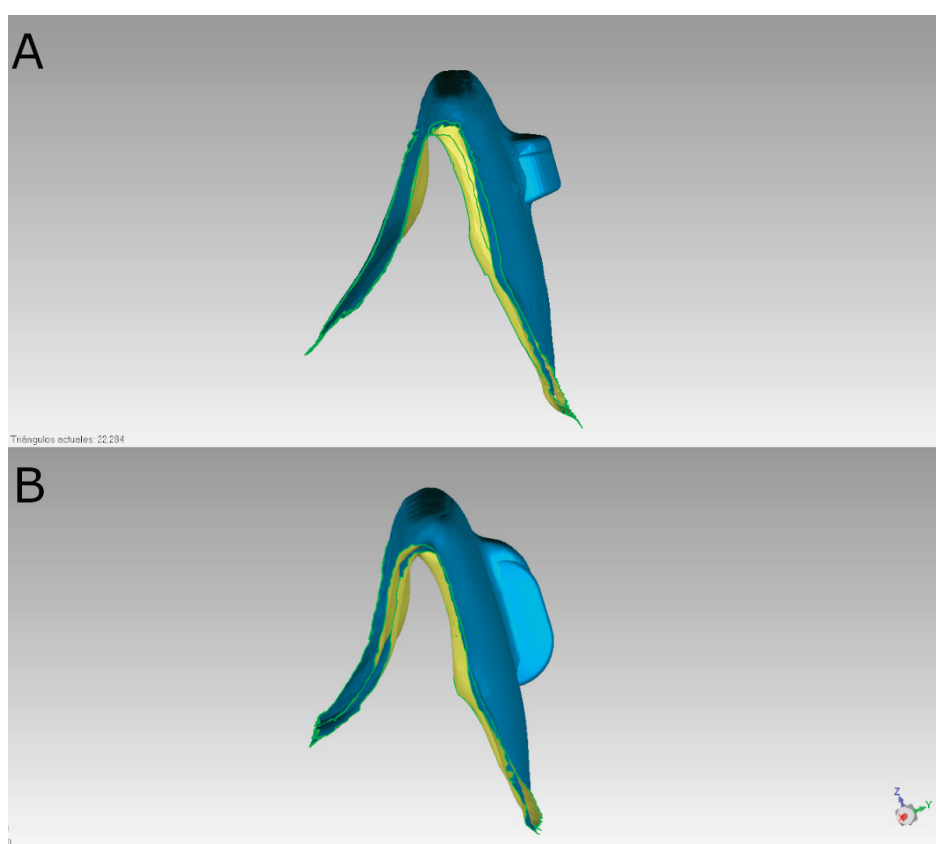


Figure 2. (A) Lateral views of the alignment procedure of the preoperative and postoperative STL digital files of horizontal and (B) vertical orthodontic clear aligner attachments in the buccal surface of tooth 1.1.

2.4. Measurement Procedure

The following characteristics were measured on both the vertical and the horizontal cut planes following the alignment process (Figure 3).

The vertical cutting plane's variables include the following:

PAPver: The greatest difference between the two apico-palatal attachment points (point-apico-palatal-vertical) of the physical and virtual models.

PCPver: point-coronal-palatal-vertical, the greatest difference between the two most coronal-palatal points of the attachments of the virtual and physical models.

PMVver: The greatest difference between the virtual and physical models' attachments' two middle vestibular points (point–middle vestibular–vertical)

AUver: The angle (angle–upper vertical) between the virtual and physical models' upper horizontal and vertical attachment profiles.

ALver: The angle (angle–lower vertical) between the virtual and physical models' lower horizontal and vertical attachment profiles.

Horizontal cutting plane variables include:

PPMhor: The greatest difference between the virtual and physical models' two most mesial–palatal attachment points (point–palatal–mesial–horizontal).

PPDhor: The greatest difference between the virtual and physical models' two furthest disto–palatal attachment points (point–palatal–distal–horizontal).

PMVhor: Point–middle–vestibular–horizontal, the greatest difference between the two middle vestibular attachment points of the physical and virtual models.

AMhor: The angle–mesial–horizontal angle that forms between the virtual and physical models' upper approximal attachment profiles.

ADhor: The angle (angle–distal–horizontal) established between the virtual and physical models' approximate lower approximation attachment profiles.

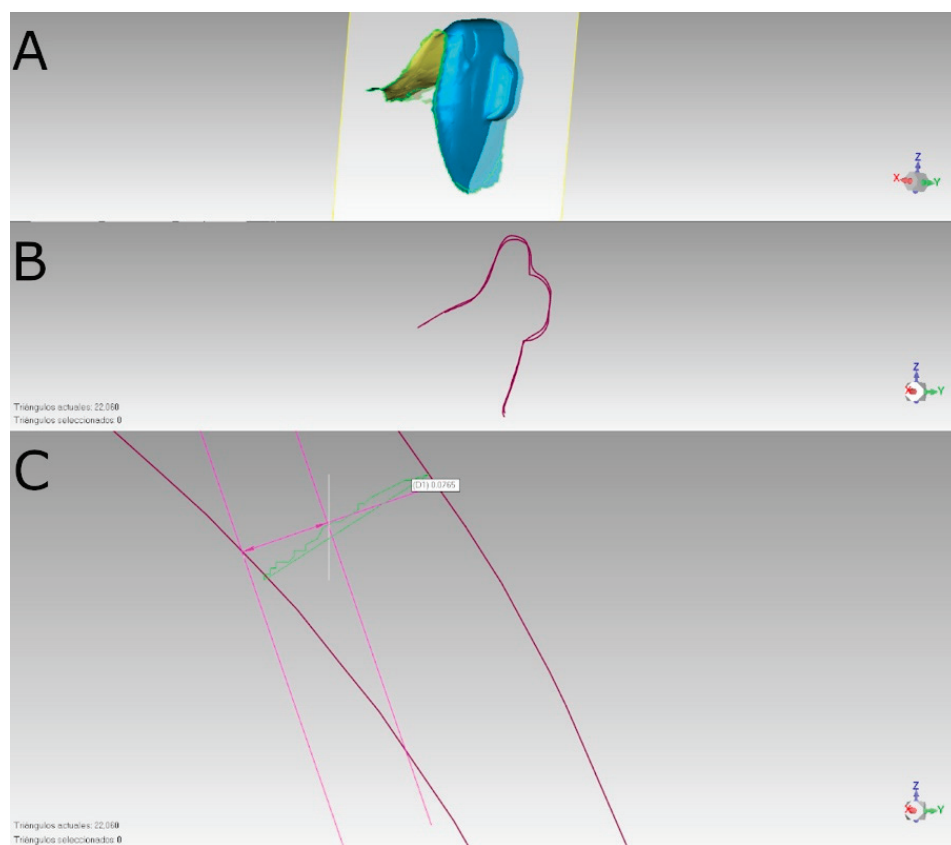


Figure 3. (A) Vertical plane crossing the aligned STL digital files of the vertical orthodontic clear aligner attachments in the buccal surface of tooth 1.1; (B) transversal plane of the aligned STL digital files of the vertical orthodontic clear aligner attachments in the buccal surface of tooth 1.1 and (C) Measurement procedure.

2.5. Statistical Analysis

A table has been obtained with the summary statistics for each of the response variables based on the group: number of observations, mean, median, standard deviation and the minimum and maximum values. They have been represented graphically using box plots.

In addition, the means between groups were compared using a *t*-test (PCPver, PPMhor, PMVhor and ADhor) or a non-parametric Mann–Whitney–Wilcoxon test (PAPver, PMVver, AUver, ALver, PPDhor and AMhor) according to application criteria. Normality tests were carried out using the Shapiro–Wilk test. Statistical analysis was performed with the software: SAS v9.4, SAS Institute Inc., Cary, NC, USA. The statistical decisions have been made taking the value 0.05 as the level of significance.

3. Results

The mean, median and SD values for the PAPver linear measurement (mm) between the STL digital files of both horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4A.

Table 1. Descriptive statistics of PAPver linear measurement (mm), PCPver linear measurement (mm), PMVver linear measurement (mm), AUver angle measurement (°), ALver angle measurement (°), PPMhor linear measurement (mm), PPDhor linear measurement (mm), PMVhor linear measurement (mm), AMhor angle measurement (°) and ADhor angle measurement (°) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments.

PAPver Linear Measurement (mm)						
Study Group	<i>n</i>	Mean	Median	SD	Minimum	Maximum
Horizontal	70	0.330	0.325	0.021	0.310	0.380
Vertical	70	0.325	0.320	0.022	0.310	0.370
PCPver linear measurement (mm)						
Horizontal	70	0.326	0.320	0.021	0.310	0.380
Vertical	70	0.267	0.270	0.016	0.230	0.280
PMVver linear measurement (mm)						
Horizontal	70	0.621	0.620	0.007	0.610	0.630
Vertical	70	0.573	0.575	0.008	0.560	0.580
AUver angle measurement (°)						
Horizontal	70	32.290	32.300	0.425	31.600	33.200
Vertical	70	30.740	30.700	0.190	30.400	31.100
ALver angle measurement (°)						
Horizontal	70	19.418	19.360	0.206	19.310	20.000
Vertical	70	17.827	17.825	0.040	17.750	17.900
PPMhor linear measurement (mm)						
Horizontal	70	0.361	0.360	0.021	0.320	0.400
Vertical	70	0.321	0.320	0.009	0.310	0.330
PPDhor linear measurement (mm)						
Horizontal	70	0.326	0.330	0.005	0.320	0.330
Vertical	70	0.288	0.290	0.010	0.270	0.310
PMVhor linear measurement (mm)						
Horizontal	70	0.270	0.270	0.016	0.250	0.300
Vertical	70	0.264	0.265	0.023	0.230	0.300
AMhor angle measurement (°)						
Horizontal	70	13.292	13.290	0.025	13.260	13.340
Vertical	70	12.454	12.390	0.204	12.320	13.020
ADhor angle measurement (°)						
Horizontal	70	22.520	22.515	0.084	22.390	22.690
Vertical	70	21.013	21.020	0.062	20.890	21.100

SD: standard deviation.

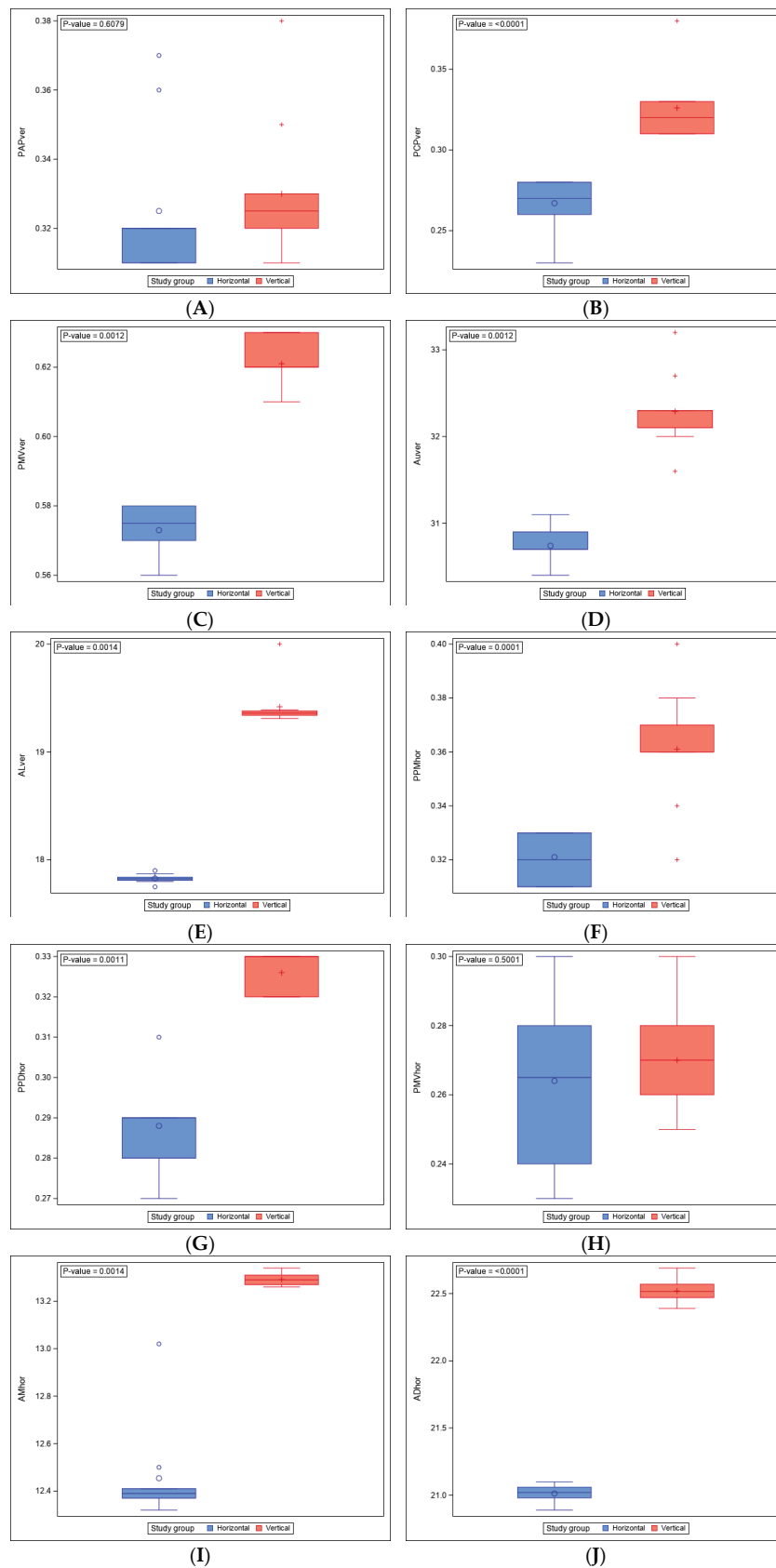


Figure 4. Box plot of the (A) PAPver linear measurement (mm), (B) PCPver linear measurement (mm), (C) PMVver linear measurement (mm), (D) AUver angle measurement (°), (E) ALver angle measurement (°), (F) PPMhor linear measurement (mm), (G) PPDhor linear measurement (mm), (H) PMVhor linear measurement (mm), (I) AMhor angle measurement (°) and (J) ADhor angle measurement (°) of the horizontal and vertical orthodontic clear aligner attachments. Horizontal line represents the median and “+,” “o” represents the mean.

A non-parametric Mann–Whitney–Wilcoxon test did not show statistically significant differences between the PAPver linear measurement (mm) of horizontal and vertical orthodontic clear aligner attachments ($p = 0.6079$) (Figure 4A).

The mean, median and SD values for the PCPver linear measurement (mm) between the STL digital files of both horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4B.

Student's *t*-test showed statistically significant differences between the PCPver linear measurement (mm) of the horizontal and vertical orthodontic clear aligner attachments ($p < 0.0001$) (Figure 4B).

The mean, median and SD values for the PMVver linear measurement (mm) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4C.

The non-parametric Mann–Whitney–Wilcoxon test showed statistically significant differences between the PMVver linear measurement (mm) of horizontal and vertical orthodontic clear aligner attachments ($p = 0.0012$) (Figure 4C).

The mean, median and SD values for the AUver angle measurement ($^{\circ}$) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4D.

The non-parametric Mann–Whitney–Wilcoxon test showed statistically significant differences between the AUver angle measurement (mm) of the horizontal and vertical orthodontic clear aligner attachments ($p = 0.0012$) (Figure 4D).

The mean, median and SD values for the ALver angle measurement ($^{\circ}$) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4E.

The non-parametric Mann–Whitney–Wilcoxon test showed statistically significant differences between the ALver angle measurement (mm) of horizontal and vertical orthodontic clear aligner attachments ($p = 0.0014$) (Figure 4E).

The mean, median and SD values for the PPMhor linear measurement (mm) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4F.

Student's *t*-test showed statistically significant differences between the PPMhor linear measurement (mm) of the horizontal and vertical orthodontic clear aligner attachments ($p = 0.0001$) (Figure 4F).

The mean, median and SD values for the PPDhor linear measurement (mm) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4G.

The non-parametric Mann–Whitney–Wilcoxon test showed statistically significant differences between the PPDhor linear measurement (mm) of the horizontal and vertical orthodontic clear aligner attachments ($p = 0.0011$) (Figure 4G).

The mean, median and SD values for the PMVhor linear measurement (mm) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4H.

Student's *t*-test did not show statistically significant differences between the PMVhor linear measurement (mm) of the horizontal and vertical orthodontic clear aligner attachments ($p = 0.5001$) (Figure 4H).

The mean, median and SD values for the AMhor angle measurement ($^{\circ}$) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4I.

The non-parametric Mann–Whitney–Wilcoxon test showed statistically significant differences between the AMhor angle measurement (mm) of the horizontal and vertical orthodontic clear aligner attachments ($p = 0.0014$) (Figure 4I).

The mean, median and SD values for the ADhor angle measurement ($^{\circ}$) between the STL digital files of both the horizontal and vertical orthodontic clear aligner attachments are displayed in Table 1 and Figure 4J.

Student's *t*-test showed statistically significant differences between the ADhor angle measurement (mm) of the horizontal and vertical orthodontic clear aligner attachments ($p < 0.0001$) (Figure 4J).

The results of this study indicate that the most of the variables of the descriptive statistics between STL digital files of both the horizontal and vertical orthodontic clear aligner attachments show significant differences between virtual and in vivo precision of attachments reproduction; however, PAPver and PMVhor linear measurements (mm) of the horizontal and vertical orthodontic clear aligner attachments did not show significant differences. These differences may be due to slight changes experienced in the attachments after removal of the attachment cementing splint; that is, greater resistance and greater traction may be experienced when removing a retainer with horizontal attachments than one with vertical attachments, due to the smaller base size.

4. Discussion

The results obtained in the present study refuted the null hypothesis (H_0), which stated that no differences would be shown between the positioning reliability of horizontal and vertical orthodontic clear aligner attachments placed through indirect bonding.

The accuracy of the attachments is one of the key components of clear aligner therapy to obtain more precision on tooth movement [25].

Clear aligners have revolutionized orthodontic therapy, offering an esthetic treatment in addition to correct dental malocclusion in patients of all ages [26,27]. Technological developments in aligner materials and techniques, especially 3D technology for planning dental movements, allow for the tooth movements of the teeth along the three planes of space more predictably. The type of movements with clear aligners are the following: (a) traslational movements: molar distalization, displacement of teeth to extraction sites and vestibulo-lingual/mesio-distal tooth displacement; (b) extrusion movement; (c) intrusion movement; (d) tipping movement: uncontrolled tipping and controlled tipping; torque and rotation movement [28]. Clear aligners are driven by two systems to obtain tooth movements. The first is a displacement-driven system which favors less complicated movements, like tipping and control of minor rotations, because the thermoplastic aligner designed by software is placed at the final location of the tooth, allowing the tooth to move or displace until it aligns with the aligner [29]. Buccolingual inclination of upper and lower incisors and produce satisfied clinical results [30]. The second is a force-driven system which operates on biomechanical principles. The amount and type of force applied depend on the shape of the aligners, with each tooth receiving a specific magnitude and type of force as determinate by software [29]. Clear aligners were not effective enough to produce adequate occlusal contacts, controlling tooth torque and retention [30].

Castroflorio et al. [18]. report that tipping and torque movements are the most difficult to control with aligners, and the loss of information increases moving toward the distal portion of the aligner. For most tooth angulation and inclination (mesio-distal tipping and bucco-lingual inclination), the worst conditions are revealed in the first and second molars on both arches.

Treatment duration, accuracy, attachability, cost, accessibility and biocompatibility are some of the issues that clear aligners must deal with [25–29]. The length of treatment

depends on the degree of malocclusion and patient compliance [26]. Different levels of therapy accuracy have been found in clinical investigations, indicating the need for more investigation. In aligner treatment results, prediction models have demonstrated an accuracy of around 78% [25,30,31]. Two crucial elements need to be taken into account in order to enhance aligner retention and system performance: the use of power ridges and attachments, as well as the careful selection of aligner material [32,33]. In contrast to the virtual setup, attachments are essential to clear aligner procedures because they provide torque, rotate teeth and affect treatment results [31,34,35]. Shape attachments determine how an aligner tooth moves, and any disparities might result in a crown-pushing force [36]. The mechanical characteristics of the aligner surface are improved by improving composite attachments, which increases aligner efficacy [37]. Issues such as heat deformation, material expansion and shrinkage might result in thickness variations that impact the aligner's final result and prevent the proper occlusion [38,39].

Aligners now have better mechanical qualities, biocompatibility and overall quality thanks to ongoing advancements in material development. Because of the release of monomers during thermoforming, certain thermoplastic polymers used in aligners may have cytotoxic consequences. ISO 20795-2 [26,40] specifies the mechanical and physical characteristics of dental resins used in orthodontic devices in order to guarantee their quality and safety. As a result, aligners are now less intrusive and more affordable and produce results that are on par with those of other available treatments [25].

Definitely, clear aligners represent a paradigm shift in modern orthodontics, providing a more convenient solution for malocclusion than conventional fixed orthodontic treatment because they facilitate good oral hygiene maintenance, are esthetic and comfortable and grant the patient satisfaction of occlusion and functionality [28]; however, Meto et al. reported microbial plaque formation on orthodontic clear aligners and recommended a copper–calcium hydroxide-based compound to remove microbial plaque formation [41].

This study used horizontal and vertical shape attachment, flowable composite and PET-G transfer templates, which are considered gold standards in the study of attachment reproduction [16], and the sample size is greater than that of the few other studies in the literature evaluated. However, the present investigation has limitations related to the *in vitro* experimental design of the study.

Previous studies have been conducted to assess the influence of different variables on the accuracy of orthodontic clear aligner attachment placement. Topsakal et al. analyzed the accuracy of the ovoid, hemi-ellipsoid and vertical rectangular attachments produced using digital light-processing three-dimensional printing technologies with 25 µm, 75 µm and 125 µm layer thickness and concluded that the layer thickness of the 3D printer is a crucial factor in determining attachment accuracy [25]; moreover, Fiorillo et al. retrospectively evaluated the accuracy of dental rotational movements using clear aligners with different attachment configurations and stated that attachment configurations to clear aligners improve rotational accuracy, but not significantly; specially optimized attachments showed the highest median accuracy (70%), followed by rectangular (65%) [42]; in addition, Brandão et al. evaluated the 3D accuracy of attachment positioning using in-house templates made with either polyethylene terephthalate glycol (PETG) or ethylene-vinyl acetate (EVA) and either pressure or vacuum thermoforming machines and established that the inaccuracy of the attachment positioning was slight and the vacuum and pressure thermoplastification machines showed no difference in attachment positioning accuracy [24]; furthermore, Valeri et al. assessed the accuracy of the process of attachment bonding in aligner treatments using two types of transfer templates and two light-curing resin-based composites (Transbond™ XT Light Cure Paste Adhesive) usually used in orthodontics and stated that a resin-based composite with a rigid transfer template is always associated with significant accuracy

and minor dispersion [43], although Chen et al. concluded that the operation time of Z350XT Flowable and SonicFill was shorter than Z350XT composite resin [44], and D'Antò et al. stated that flowable composite resin cements were appropriate for attachment fabrication; moreover, the fidelity of attachment reproduction was similar among them, and the orthodontic composite showed more overflow when compared with the flowable one [45]. Nevertheless, the present scientific literature fails to directly compare the impact of the geometric design of the clear aligner attachments on the precision and dependability of their positioning.

The present study provides practical implications with direct clinical application that could influence the predictability of tooth movement, making the clinical results of orthodontic treatment with clear aligners more precise. However, the study also showed limitations, since the researchers considered only one density of composite resin cement and a clear aligner system in the study to standardize the sample; moreover, the present laboratory-based study used 3-D printed models with different physical properties than natural teeth models. Furthermore, the authors advocate the creation of randomized controlled clinical trials incorporating different densities of composite resin cements and a clear aligner system for further research.

5. Conclusions

The vertical orthodontic clear aligner attachments are more accurate than the horizontal orthodontic clear aligner attachments through the indirect bonding technique; therefore, the authors recommend selecting attachments whose positioning is more reliable with respect to planning; especially in cases requiring mesiodistal inclination, mesiodistal translation or intrusion tooth movements for lower incisors, leading to a more predictable orthodontic treatment, avoiding complications and longer lasting treatments.

Author Contributions: Conceptualization, L.Ó.A.-E. and M.P.-S.; methodology, Á.Z.-M. software, Á.Z.-M.; validation, A.B.L.-G. and J.F.-F. data curation, G.G. writing—original draft preparation, C.R.F.; writing—review and editing, J.F.-F.; supervision, Á.Z.-M. 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; the study did not involve human participants/data or human tissue.

Informed Consent Statement: Not applicable.

Data Availability Statement: The original contributions presented in this study are included in the article. Further inquiries can be directed to the corresponding author.

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

References

1. Paglia, L.; Marzo, G. Aligners, can my child use them too? *Eur. J. Paediatr. Dent.* **2023**, *24*, 259.
2. Bolas-Colvee, B.; Tarazona, B.; Paredes-Gallardo, V.; Arias-De Luxan, S. Relationship between perception of smile esthetics and orthodontic treatment in Spanish patients. *PLoS ONE* **2018**, *13*, e0201102. [CrossRef] [PubMed]
3. Saccomanno, S.; Saran, S.; Laganà, D.; Mastrapasqua, R.F.; Grippaudo, C. Motivation, Perception, and Behavior of the Adult Orthodontic Patient: A Survey Analysis. *BioMed Res. Int.* **2022**, *4*, 2754051. [CrossRef]
4. Alhasyimi, A.A.; Ayub, A.; Farmasyanti, C.A. Effectiveness of the Attachment Design and Thickness of Clear Aligners during Orthodontic Anterior Retraction: Finite Element Analysis. *Eur. J. Dent.* **2024**, *18*, 174–181. [CrossRef] [PubMed]
5. Baron, P. Les appareils orthodontiques invisibles et presque invisibles [Invisible and almost invisible orthodontic appliances]. *L'Orthodontie Française* **2014**, *85*, 59–91. [CrossRef] [PubMed]

6. Li, Q.; Du, Y.; Yang, K. Comparison of pain intensity and impacts on oral health-related quality of life between orthodontic patients treated with clear aligners and fixed appliances: A systematic review and meta-analysis. *BMC Oral Health* **2023**, *23*, 920. [CrossRef]
7. Mendes Ribeiro, S.M.; Aragón, M.L.S.C.; Espinosa, D.D.S.G.; Shibasaki, W.M.M.; Normando, D. Orthodontic aligners: Between passion and science. *Dent. Press J. Orthod.* **2024**, *28*, e23spe6. [CrossRef] [PubMed]
8. Zhang, B.; Huang, X.; Huo, S.; Zhang, C.; Zhao, S.; Cen, X.; Zhao, Z. Effect of clear aligners on oral health-related quality of life: A systematic review. *Orthod. Craniofacial Res.* **2020**, *23*, 363–370. [CrossRef] [PubMed]
9. Alami, S.; Sahim, S.; Hilal, F.; Essamlali, A.; El Quars, F. Perception and satisfaction of patients treated with orthodontic clear aligners. *Open Access Lib J.* **2022**, *9*, e9300. [CrossRef]
10. Chhibber, A.; Agarwal, S.; Yadav, S.; Kuo, C.-L.; Upadhyay, M. Which orthodontic appliance is best for oral hygiene? A randomized clinical trial. *Am. J. Orthod. Dentofac. Orthop.* **2018**, *153*, 175–183. [CrossRef] [PubMed]
11. Abay, F.; Buyuk, S.K.; Korkmaz, Y.N. Prevalence of white spot lesions during clear aligner therapy: A systematic review. *Australas. Orthod. J.* **2022**, *38*, 368–379. [CrossRef]
12. Galluccio, G. Is the use of clear aligners a real critical change in oral health prevention and treatment? *Clin. Ter.* **2021**, *172*, 113–115. [PubMed]
13. Zheng, M.; Liu, R.; Ni, Z.; Yu, Z. Efficiency, effectiveness and treatment stability of clear aligners: A systematic review and meta-analysis. *Orthod. Craniofacial Res.* **2017**, *20*, 127–133. [CrossRef]
14. AlMogbel, A. Clear Aligner Therapy: Up to date review article. *J. Orthod. Sci.* **2023**, *12*, 37. [CrossRef] [PubMed]
15. Ma, X.Q.; Xiang, F.; Fan, M.Y.; Song, Y.; Wang, X.H.; Zhang, L.; Qian, W.H. Clinical efficacy of the combination of miniscrew with clear aligner in controlling the roller coaster effect. *Shanghai Kou Qiang Yi Xue* **2022**, *31*, 193–197.
16. Bellocchio, A.M.; Portelli, M.; Ciraolo, L.; Ciano, E.; Militi, A.; Peditto, M.; Barbera, S.; Nucera, R. Evaluation of the Clinical Variables Affecting Attachment Reproduction Accuracy during Clear Aligner Therapy. *Materials* **2023**, *16*, 6811. [CrossRef] [PubMed]
17. Putrino, A.; Barbato, E.; Galluccio, G. Clear Aligners: Between Evolution and Efficiency—A Scoping Review. *Int. J. Environ. Res. Public Health* **2021**, *18*, 2870. [CrossRef]
18. Castrolforio, T.; Sedran, A.; Parrini, S.; Garino, F.; Reverdito, M.; Capuozzo, R.; Mutinelli, S.; Grybauskas, S.; Vaitiekūnas, M.; Deregibus, A. Predictability of orthodontic tooth movement with aligners: Effect of treatment design. *Prog. Orthod.* **2023**, *24*, 2. [CrossRef]
19. Nucera, R.; Dolci, C.; Bellocchio, A.M.; Costa, S.; Barbera, S.; Rustico, L.; Farronato, M.; Militi, A.; Portelli, M. Effects of Composite Attachments on Orthodontic Clear Aligners Therapy: A Systematic Review. *Materials* **2022**, *15*, 533. [CrossRef]
20. Gazzani, F.; Bellisario, D.; Quadrini, F.; Danesi, C.; Alberti, A.; Cozza, P.; Pavoni, C. Light-curing process for clear aligners' attachment reproduction: Comparison between two nanocomposites cured by the auxiliary of a new tool. *BMC Oral Health* **2022**, *22*, 376. [CrossRef]
21. Jedliński, M.; Mazur, M.; Greco, M.; Belfus, J.; Grocholewicz, K.; Janiszewska-Olszowska, J. Attachments for the Orthodontic Aligner Treatment-State of the Art-A Comprehensive Systematic Review. *Int. J. Environ. Res. Public Health* **2023**, *20*, 4481. [CrossRef] [PubMed]
22. Yaosen, C.; Mohamed, A.M.; Jinbo, W.; Ziwei, Z.; Al-Balaa, M.; Yan, Y. Risk Factors of Composite Attachment Loss in Orthodontic Patients During Orthodontic Clear Aligner Therapy: A Prospective Study. *BioMed Res. Int.* **2021**, *23*, 6620377. [CrossRef] [PubMed]
23. Li, Q.; Yang, K. Surface wear of attachments in patients during clear aligner therapy: A prospective clinical study. *Prog. Orthod.* **2024**, *25*, 7. [CrossRef]
24. Brandão, N.M.C.B.; Maia, R.M.; Gomes, V.M.; Resende, C.; Antunes, A.N.D.G.; Souki, B.Q. Bonding positional accuracy of attachments and marginal adaptation of in-house aligners—A quality improvement laboratory study. *Orthod. Craniofacial Res.* **2024**, *27* (Suppl. S2), 120–130. [CrossRef]
25. Topsakal, K.G.; Gökmen, Ş.; Yurdakurban, E.; Duran, G.S.; Görgülü, S. The effect of layer thickness on the accuracy of the different in-house clear aligner attachments. *Clin. Oral Investig.* **2023**, *27*, 5331–5341. [CrossRef]
26. Narongdej, P.; Hassanpour, M.; Alterman, N.; Rawlins-Buchanan, F.; Barjasteh, E. Advancements in Clear Aligner Fabrication: A Comprehensive Review of Direct-3D Printing Technologies. *Polymers* **2024**, *16*, 371. [CrossRef] [PubMed]
27. Aljabaa, A.H. Clear Aligner Therapy—Narrative Review. *J. Int. Oral Health* **2020**, *12*, S1–S4. [CrossRef]
28. Moutawakil, A. Biomechanics of aligners: Literature Review. *Adv. Dent. Oral Health* **2021**, *13*, 555872. [CrossRef]
29. Katib, H.S.; Hakami, A.M.; Albalawei, M.; Alhajri, S.A.; Alruwaily, M.S.; Almusallam, M.I.; Alqahtani, G.H. Stability and Success of Clear Aligners in Orthodontics: A Narrative Review. *Cureus* **2024**, *16*, e52038. [CrossRef]
30. Yu, X.; Li, G.; Zheng, Y.; Gao, J.; Fu, Y.; Wang, Q.; Huang, L.; Pan, X.; Ding, J. 'Invisible' orthodontics by polymeric 'clear' aligners molded on 3D-printed personalized dental models. *Regen. Biomater.* **2022**, *9*, rbac007. [CrossRef]

31. Izhar, A.; Singh, G.; Goyal, V.; Singh, R.; Gupta, N.; Pahuja, P. Comparative Assessment of Clinical and Predicted Treatment Outcomes of Clear Aligner Treatment: An in Vivo Study. *Turk. J. Orthod.* **2019**, *32*, 229–235. [CrossRef]
32. Upadhyay, M.; Arqub, S.A. Biomechanics of Clear Aligners: Hidden Truths & First Principles. *J. World Fed. Orthod.* **2022**, *11*, 12–21.
33. Lombardo, L.; Palone, M.; Carlucci, A.; Siciliani, G. Clear Aligner Hybrid Approach: A Case Report. *J. World Fed. Orthod.* **2020**, *9*, 32–43. [CrossRef] [PubMed]
34. Weir, T. Clear Aligners in Orthodontic Treatment. *Aust. Dent. J.* **2017**, *62*, 58–62. [CrossRef] [PubMed]
35. Elshazly, T.M.; Keilig, L.; Salvatori, D.; Chavanne, P.; Aldesoki, M.; Bourauel, C. Effect of Trimming Line Design and Edge Extension of Orthodontic Aligners on Force Transmission: An in Vitro Study. *J. Dent.* **2022**, *125*, 104276. [CrossRef] [PubMed]
36. Ammann, R.; Tanner, C.; Schulz, G.; Osmani, B.; Nalabothu, P.; Töpper, T.; Müller, B. Three-Dimensional Analysis of Aligner Gaps and Thickness Distributions, Using Hard X-ray Tomography with Micrometer Resolution. *J. Med. Imaging* **2022**, *9*, 031509. [CrossRef]
37. Ho, C.-T.; Huang, Y.-T.; Chao, C.-W.; Huang, T.-H.; Kao, C.-T. Effects of Different Aligner Materials and Attachments on Orthodontic Behavior. *J. Dent. Sci.* **2021**, *16*, 1001–1009. [CrossRef]
38. Hartshorne, J.; Wertheimer, M.B. Emerging Insights and New Developments in Clear Aligner Therapy: A Review of the Literature. *AJO-DO Clin. Companion* **2022**, *2*, 311–324. [CrossRef]
39. Panayi, N.C. Directly Printed Aligner: Aligning with the Future. *Turk. J. Orthod.* **2023**, *36*, 62–69. [CrossRef] [PubMed]
40. ISO 20795-2; Dentistry—Base polymers. Part 2: Orthodontic Base Polymers. International Organization for Standardization: Geneva, Switzerland, 2013.
41. Meto, A.; Colombari, B.; Castagnoli, A.; Sarti, M.; Denti, L.; Blasi, E. Efficacy of a Copper-Calcium-Hydroxide Solution in Reducing Microbial Plaque on Orthodontic Clear Aligners: A Case Report. *Eur. J. Dent.* **2019**, *13*, 478–484. [CrossRef]
42. Fiorillo, G.; Campobasso, A.; Croce, S.; Hussain, U.; Battista, G.; Lo Muzio, E.; Mandelli, G.; Ambrosi, A.; Gastaldi, G. Accuracy of clear aligners in the orthodontic rotational movement using different attachment configurations. *Orthod. Craniofacial Res.* **2024**, *27*, 996–1003. [CrossRef] [PubMed]
43. Valeri, C.; Aloisio, A.; Mummolo, S.; Quinzi, V. Performance of Rigid and Soft Transfer Templates Using Viscous and Fluid Resin-Based Composites in the Attachment Bonding Process of Clear Aligners. *Int. J. Dent.* **2022**, *12*, 1637594. [CrossRef]
44. Chen, W.; Qian, L.; Qian, Y.; Zhang, Z.; Wen, X. Comparative study of three composite materials in bonding attachments for clear aligners. *Orthod. Craniofacial Res.* **2021**, *24*, 520–527. [CrossRef]
45. D’Antò, V.; Muraglie, S.; Castellano, B.; Candida, E.; Sfondrini, M.F.; Scribante, A.; Grippaudo, C. Influence of Dental Composite Viscosity in Attachment Reproduction: An Experimental In Vitro Study. *Materials* **2019**, *12*, 4001. [CrossRef] [PubMed]

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

Article

Mechanical Behavior of PEEK and PMMA Graphene and Ti6Al4V Implant-Supported Frameworks: In Silico Study

Mariano Herrero-Climent ¹, Fernando Sanchez-Lasheras ^{2,3}, Jordi Martinez-Lopez ⁴, Javier Gil ^{5,*} and Aritza Brizuela-Velasco ^{6,*}

¹ Porto Dental Institute, Av. De Montevideo 810, 4150-518 Porto, Portugal; dr.herrero@herrerocliment.com

² University Institute of Space Sciences and Technologies of Asturias (ICTEA), University of Oviedo, 33004 Oviedo, Spain; fsanchez@tecniproject.com

³ Department of Mathematics, Faculty of Sciences, University of Oviedo, 33007 Oviedo, Spain

⁴ Soadco Research and Development Department, AD700 Escaldes, Andorra; j.martinez@soadco.com

⁵ Biomimetics Oral Biomaterials and Interfaces (BOBI), Department Ciencia e Ingeniería de Materiales, Escola d'Enginyeria Barcelona Est, Universitat Politècnica de Catalunya, c/Eduard Maristany 16, 08029 Barcelona, Spain

⁶ DENS-ia Research Group, Faculty of Health Sciences, Miguel de Cervantes European University, C/del Padre Julio Chevalier 2, 47012 Valladolid, Spain

* Correspondence: javier.gil.mur@upc.edu (J.G.); aritzabrizuela@hotmail.com (A.B.-V.)

Abstract: A comparative analysis has been carried out between three different dental materials suitable for the prostheses manufacturing. The analysis performed is based on the finite elements method (FEM) and was made to evaluate their performance under three different loading conditions. Three different materials were modeled with 3D CAD geometry, all of them suitable to be simulated by means of a linear elastic model. The materials employed were graphene polymethyl methacrylate (G-PMMA) with 0.25% of graphene, polyether ether ketone (PEEK), and Ti6Al4V. Three loading conditions have been defined: distal, medial, and central. In all cases under study, the load was applied progressively, 5 N by 5 N until a previously fixed threshold of 25 N was reached, which always ensures that work is carried out in the elastic zone. The behavior of G-PMMA and PEEK in the tests performed is similar. Regarding maximum deformations in the model, it has been found that deformations are higher in the G-PMMA models when compared to those made of PEEK. The highest values of maximum stress according to the von Mises criteria are achieved in models made of Ti6Al4V, followed by G-PMMA and PEEK. G-PMMA is more prone to plastic deformations compared to Ti6Al4V. However, due to its relatively higher stiffness compared to other common polymers, G-PMMA is able to withstand moderate stress levels before significant deformation occurs, placing it in the intermediate position between Ti6Al4V and PEEK in terms of stress capacity. It should be noted that there is also a difference in the results obtained depending on the applied load, whether distal, medial, or central, proving that, in all simulations, it is the distal test that offers the worst results in terms of presenting a higher value for both displacement and tension. The results obtained allow us to identify the advantages and limitations of each material in terms of structural strength, mechanical behavior, and adaptability to loading conditions that simulate realistic scenarios.

Keywords: PEEK; G-PMMA; Ti6Al4V; finite element simulation; implant-supported frameworks

1. Introduction

When rehabilitating an edentulous arch, one of the possibilities that provides the best solution, both from a functional and esthetic point of view, is an implant-retained fixed prosthesis. However, the recommendation regarding the number of necessary support implants is diverse and ranges from recommendations of using one implant for each lost tooth [1–3], to only four to rehabilitate up to twelve teeth in occlusion, with optimal survival results, even in post-extraction and immediate loading protocols [4,5].

However, it is evident that the use of a smaller number of implants entails more challenging biomechanical situations, such as the tendency to increase the length of the pontics and favor deflection of the prosthetic structure or to use distal extensions, decreasing the mechanical advantage by increasing power arm. The choice of a correct prosthetic superstructure could possibly compensate for this mechanical behavior that tends to overload.

The prosthetic suprastructure is the part of the dental prostheses that is retained by the support implants, either directly through their connection or through an intermediate abutment, and, in the case of a full arch rehabilitation, can splint all the support implants. A dental prosthesis supported by several implants results in a combined structure in which the distribution of the applied loads depends on the relative rigidity of the various members involved, as well as the geometry of their distribution [6,7]. An osseointegrated implant forms an almost intimate connection with the bone, so it can be expected that the response to any load will be elastic, which means that the deflection of the fixation will be proportional to the applied load. On the other hand, with a rigid metal prosthesis, it can be assumed that its behavior will be relatively rigid compared to the implants, whose section diameter is smaller. The combined structure will normally be so complex that the static equations alone are insufficient to determine the distribution of the load performance, and the deformations of the prosthesis, implants, and jaw must be taken into account to determine said distribution. Such a structure is considered statically indeterminate and the analysis must treat the prosthesis in general and its suprastructure in particular as a curved elastic beam, subjected to bending and torsion [8].

In essence, this mechanical response of the suprastructures, when faced with the application of loads that come from masticatory forces, can lead to technical complications, of which fractures of these suprastructures are less frequent (0.2 to 4.7%) and the fracturing of the esthetic coating materials is more frequent (3.2% to 25.5%) [9].

However, as mentioned above, when evaluating the properties and mechanical behavior of a prosthetic suprastructure, it is too simple to do so in relation to its low tendency to fracture. These structures will have a greater or lesser capacity for deformation upon application of load and will consume a greater or lesser amount of tension in this deformation so that the rest of the system (retention screws, abutments, implants, and supporting bone) will receive more or less mechanical demand. In this regard, certain studies in the literature can be found that describe different degrees of deformation of the peri-implant bone depending on the material from which the prosthesis is made [10]. The deformation of the superstructure may also influence the technical complications of the covering materials, such as feldspathic ceramics, disilicates, and acrylic resins, with weak bonds to the frameworks, often due to mere mechanical entrapment, which, by having different elastic properties and different coefficients of thermal expansion, can favor failure.

In general, the most commonly used material in the manufacturing of these suprastructures has been metal, especially cobalt chrome alloys, using lost wax casting techniques [11,12]. However, the advent of CAD/CAM technology applications has led to a revolution in the possibility of using other different materials, most of them processed using milling or sintering-machining methods. Among these, titanium and its alloys stand

out, such as grade V and zirconium oxide, as well as other polymeric materials that have advanced first as provisional rehabilitation materials but are now beginning to be used as definitive ones, although with reasonable doubts about its mechanical behavior [13–15]. Of the latter, Polyetheretherketone (PEEK) and polymethylmethacrylate (PMMA) stand out. PEKK biomaterials are polymeric elastic materials with good shock absorbance and fracture resistance and present ultra-high performance among all thermoplastic composites for excellent mechanical strength, chemical resistance, and high thermal stability (tensile strength 90–100 MPa, compressive strength similar and modulus of elasticity between 3.6 and 4.1 GPa) [16]. PMMA was initially used as a dental material in 1937, although it must be acknowledged that it has inferior mechanical properties (tensile strength 50–70 MPa, similar compression resistance, and modulus of elasticity between 2 and 3.1 GPa), when compared to PEEK [17–20]. In this sense, several strategies have been applied in order to increase the mechanical properties of PMMA polymer resins, almost all of which were based on the addition of inorganic fillers such as mesoporous silica nanoparticles (MSNs), titanium dioxide (TiO₂), carbon nanotubes (CNTs), and also nano-sized graphene oxides (nGO) [21].

Despite everything, the mechanical properties of polymeric materials are very far from those of metals or ceramics. For example, Ti6Al4V has a tensile strength of 900–1200 MPa, a similar compressive strength, and an elastic modulus of 110 GPa [22–27]. Therefore, although the use of this type of polymeric materials opens new expectations in oral rehabilitation, due to its simplicity of manufacturing, price, ability to adhere to coating materials, adjustment possibilities, etc., it still raises doubts regarding its biomechanic behavior.

In principle, the trend in prosthetic materials is to try to avoid metallic materials because of the corrosion and ion release problems they cause [28–30]. For this reason, the materials usually used are titanium or titanium alloys, especially Ti6Al4V, due to their good mechanical properties and electrochemical similarity with the metals of dental implants, which are generally made of commercially pure titanium [31]. Ceramic materials are difficult to form and also have the problem of brittleness. One of the most tenacious materials is zirconia partially stabilized with yttria, which, due to phase transformations produced by tension, improves its tenacity. However, these improvements in mechanical properties are not sufficient for the mechanical bending loads that often occur in the oral cavity. These ceramic materials are nevertheless excellent with respect to dental esthetics [32–34]. That is why polymeric materials such as PMMA or PEEK are being tried as they are thermoplastic materials with good mechanical properties that will not have creep due to cross-linking of their structures or improvements in their mechanical properties. This improvement in response to masticatory loads can be achieved by introducing graphene, as is the case in this work. However, the amount of graphene that can be incorporated into PMMA is limited due to the black coloration that it acquires in the PMMA as the graphene content increases [35,36].

For this reason, the motivation for this in vitro experimental study arises: to use a finite element analysis that allows for the studying of the tensions and deformations of these materials subject to loading in a full arch rehabilitation design on four implants, with the objective of comparing its results with those of a high-performance material, such as Ti6Al4V, and, based on that, being able to infer its clinical applications.

2. Materials and Methods

A 3D CAD model has been created. This model has an adaptive structured mesh in which the places where loads will be applied had especially been considered. As can be observed in Figure 1, this model has a total of 1,502,113 nodes that generate 1,024,036 elements, with an average quality of the element of 0.8116 [37], an aspect ratio [38] of 2.1113, and a

Jacobian value of 1.0067 [39]. Also, its average deformation (skewness) is 0.21742 [40–42]. Figure 1 shows the 3D meshed model employed for the present research and details of the mesh. Although a formal mesh convergence test was not conducted, the results obtained for the mesh parameters presented before confirmed that the mesh met the criteria for numerical stability and accuracy.

Three different materials were modeled with 3D CAD geometry, as described before, all of which were suitable to be simulated by means of a linear elastic model. The materials employed were graphene polymethyl methacrylate (G-PMMA), polyether ether ketone (PEEK), and titanium alloy ASTM grade V (Ti6Al4V). Their main properties are listed in Table 1. The three models under study not only differ in the materials from which they are made of, but the models of G-PMMA and PEEK also have internal Ti6Al4V inserts in the area where they would be fixed to the other components. In the simulation, the prostheses are constrained with fully fixed supports at their base. In the referred models, the interfaces between pieces are defined as bonded, assuming perfect adhesion without relative motion, and under this assumption, no frictional contacts were required.

Table 1. Properties of the materials employed in the present study [18,19,22–24,26]. G-PMMA: glycol-modified polymethyl methacrylate (0.25% of graphene), PEEK: polyether ether ketone Ti6Al4V: titanium alloy ASTM grade V.

Material	Property	Value	Units
G-PMMA	Density	1190	kg/m ³
	Young's modulus	$3.2 \cdot 10^9$	N/m ² (Pascal)
	Poisson's coefficient	0.35	
	Tensile yield strength	$6.5 \cdot 10^7$	Pascals
PEEK	Density	1310	kg/m ³
	Young's modulus	$4.2 \cdot 10^9$	N/m ² (Pascal)
	Poisson's coefficient	0.38	
	Tensile yield strength	$1.16 \cdot 10^8$	Pascal
Ti6Al4V	Density	4430	kg/m ³
	Young's modulus	$1.138 \cdot 10^{11}$	N/m ² (Pascal)
	Poisson's coefficient	0.342	
	Tensile yield strength	$8.8 \cdot 10^8$	Pascal

In this research, three loading conditions have been defined for each of the materials under consideration. The first one, with the model loaded onto one of its extremes, will be called distal; the one in which the load is applied between the first two inserts will be called medial; and, finally, the loading situation in which the load is in the center of the middle span will be called central. For these three load cases, control surfaces were generated. Please note that the generation of control surfaces is necessary because if the force were theoretically applied on a surface with a dimension of zero, the stress value would be infinite. Therefore, the use of control surfaces contributes to a more realistic model from a biomechanical point of view. Although each load is applied to a different model, Figure 2 represents the three loads applied to the same model for the reader be able to clearly see their location in the model and compare them.

In all cases under study, the load was applied progressively, 5 N by 5 N until a previously fixed threshold of 25 N was reached, which always ensures that work is carried out in the elastic zone for all materials. The value of 25 N was fixed considering that none of the materials would enter into the plasticity zone with such load value.

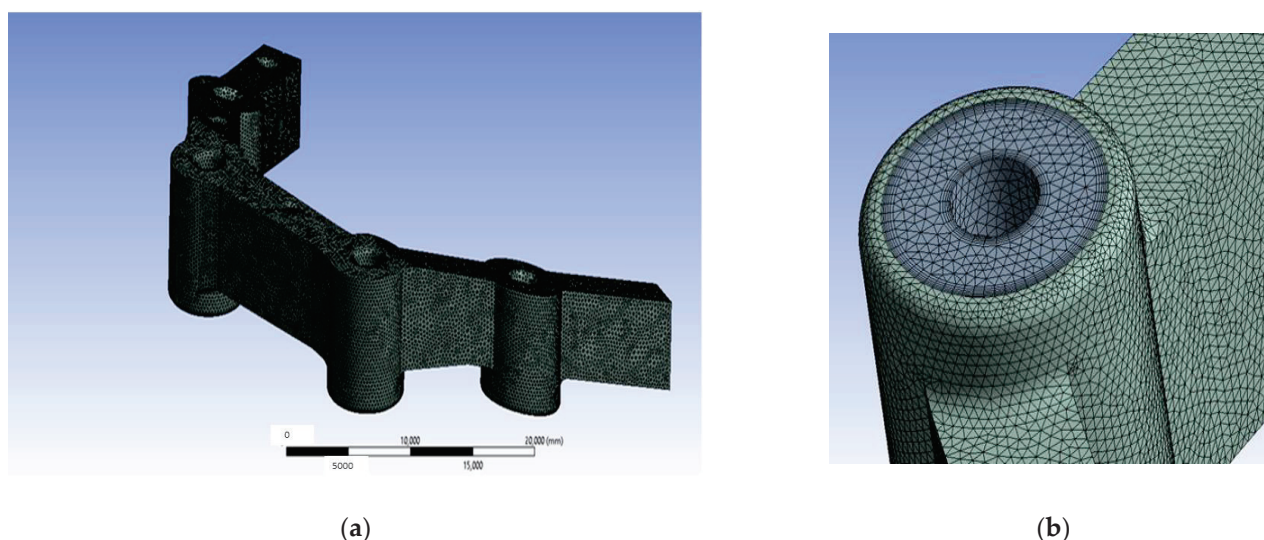


Figure 1. Finite elements model: (a) Meshed 3D view of the model employed for the present research; (b) detail of the mesh.

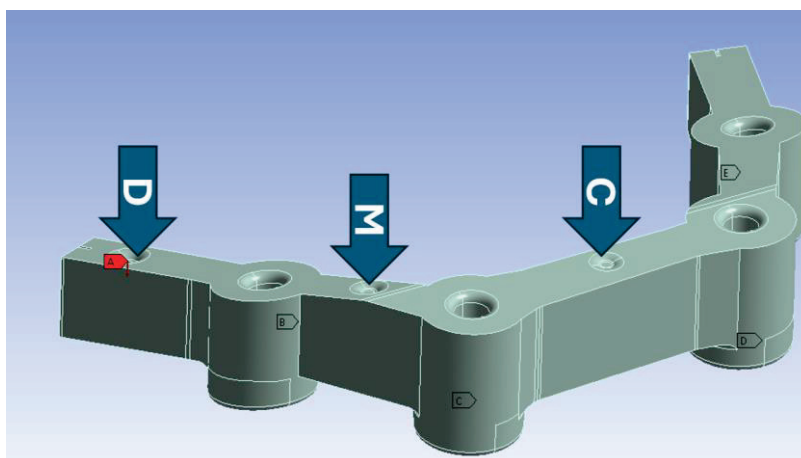


Figure 2. Load application points in the models under study (D: distal, M: medial, and C: central). Positions A, B, C, D and E are where dental implants are placed.

For the validation of the results obtained in the simulation by the finite element method, analyzed structures with the same dimensions and geometries have been elaborated with the three materials studied. Five samples have been elaborated for each of the materials and for each of the positions where the mechanical stress will be exerted (distal, medial, and central). Consequently, 45 structures were elaborated. Figure 3 shows one of the structures obtained.

The mechanical tests were performed using an MTS-Bionix servo-hydraulic testing machine (Bionix 358, MTS, Minneapolis, MN, USA). The tests performed are single-point bending tests, the same as those performed in the simulation. The load application rate was 5 N/s. Figure 4 shows specimens of the three materials prepared for testing. Five mechanical tests were performed for each material and each loading position. All results were expressed as mean and standard deviation except for the bacterial adhesion test results which were expressed as median and standard error. The comparative *t*-test (with the Excel software) was carried out between the different groups at 95%, which means that for values of $p < 0.05$, there are significant differences.

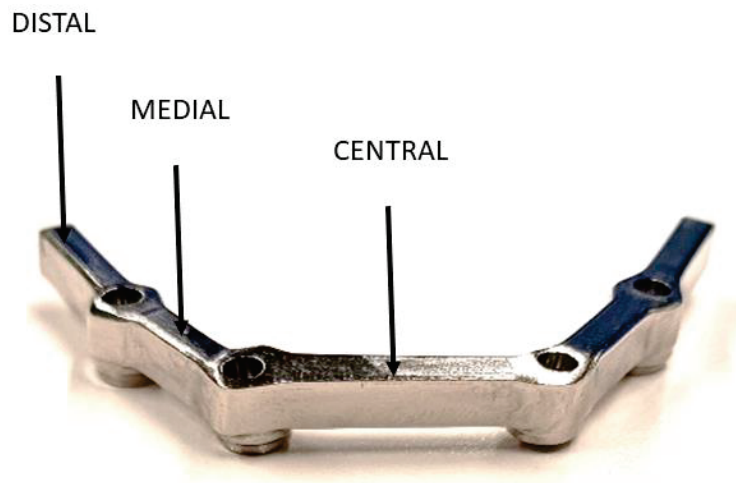


Figure 3. Prosthetic structure obtained for the mechanical tests.

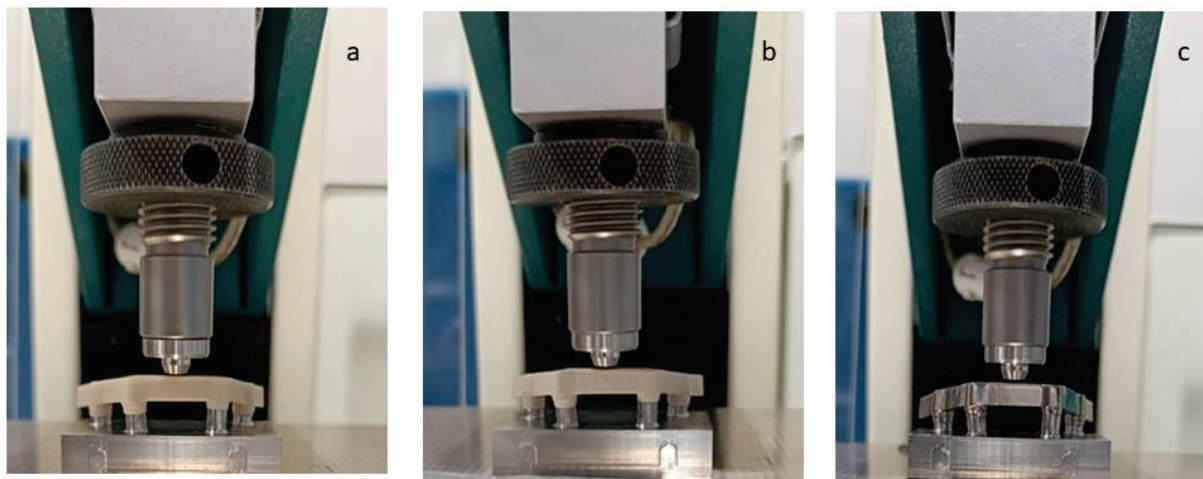


Figure 4. Flexural mechanical test (a) G-PMMA, (b) PEEK, (c) Ti6Al4V.

3. Results

This section presents the results obtained for all the models of each different load condition studied. The results are divided into subsections. Each subsection includes all the load conditions of a material while a final subsection presents graphs that summarize all the results obtained.

3.1. Results of the Model Made of G-PMMA for the Different Load Conditions Under Study

Figure 5 shows the deformation map in a mid-plane of the model that uses G-PMMA as a base material and has inserts of Ti6Al4V and presents the load located in the distal position. This median plane is obtained by joining the centers of the point of application of the load and the axis of the inserts. As can be observed, the largest deformations are shown in the area closer to the point where the load is applied. In the same way, Figure 6 shows the deformation map corresponding to the bushing of this model with the same load condition, most affected by the application of the load, and Figure 7 shows the stress map of the G-PMMA model with distal loading.

As can be seen in Figure 7, the maximum stress value obtained is 68.045 MPa. This maximum value is close to 65 MPa, which is the elastic limit of G-PMMA. Please note that this figure determined the magnitude of the force applied in all cases. Having obtained the general von Mises equivalent stress map, we proceeded to study the most loaded zone of the experiment, in other words, the area of the inserts. Furthermore, it is necessary to

point out that, although slices of the objects are shown by the plane indicated before to facilitate visualization, this plane does not necessarily contain the maximum equivalent stress, even if it contains the surface of application of the load or the embedment. This is because in complex geometries, like the one under study in the present research, it would be difficult to know a priori where the maximum load will be found and, also, in many cases, the plane in which such area is contained changes from one model to another.

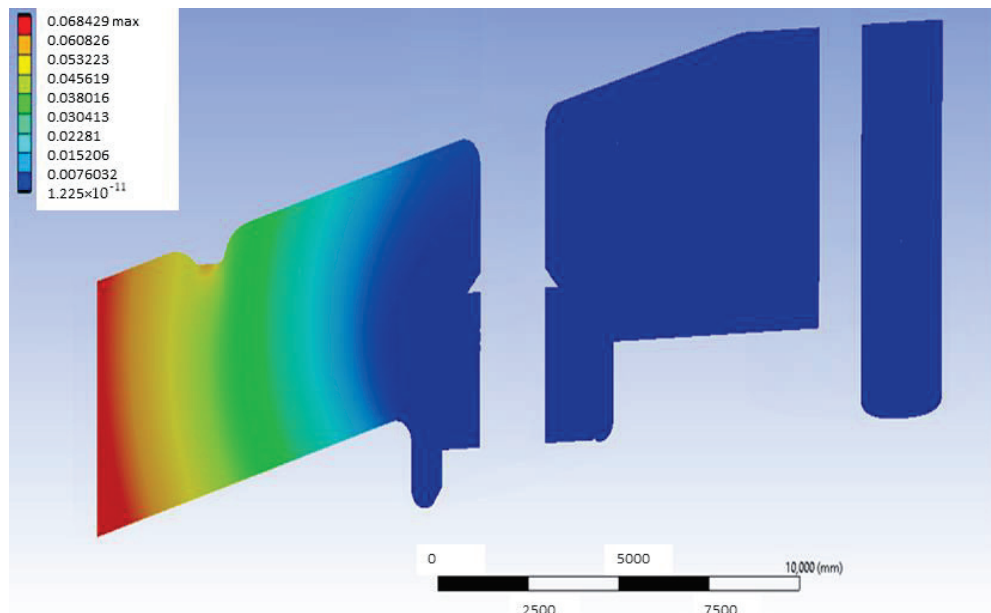


Figure 5. Mid-plane deformation map of the distally loaded G-PMMA model (units: mm).

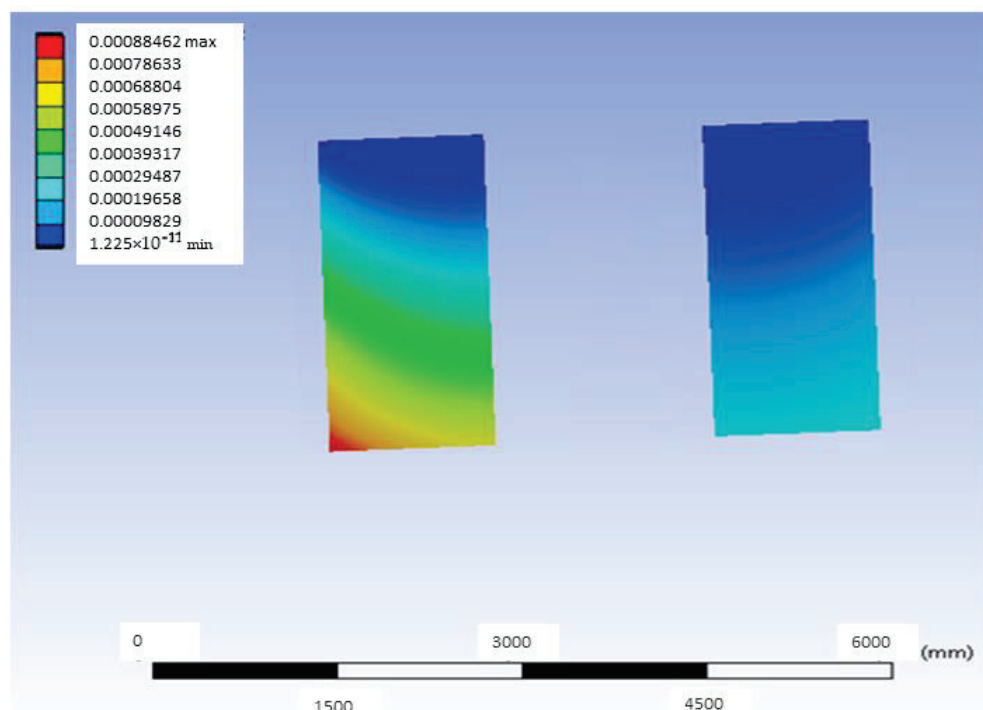


Figure 6. Deformation map of the Ti6Al4V bushing most affected by the application of the distal load in the G-PMMA model (units: mm).

The deformation presented in Figure 6 is complemented by Figure 8, which shows the stress map of the most loaded bushing of this model in distal load. In the same way as the figures shown before, in Figure 8, there is a concentration of stress in the upper left zone,

so that in real practice, it would be expected that the model would erode or deform, and the load in that zone in real cases would be somewhat lower than that of the theoretical model. In other words, it would be possible that, in practice, in the red zone, erosion or deformation would occur and this would reduce the real stress value.

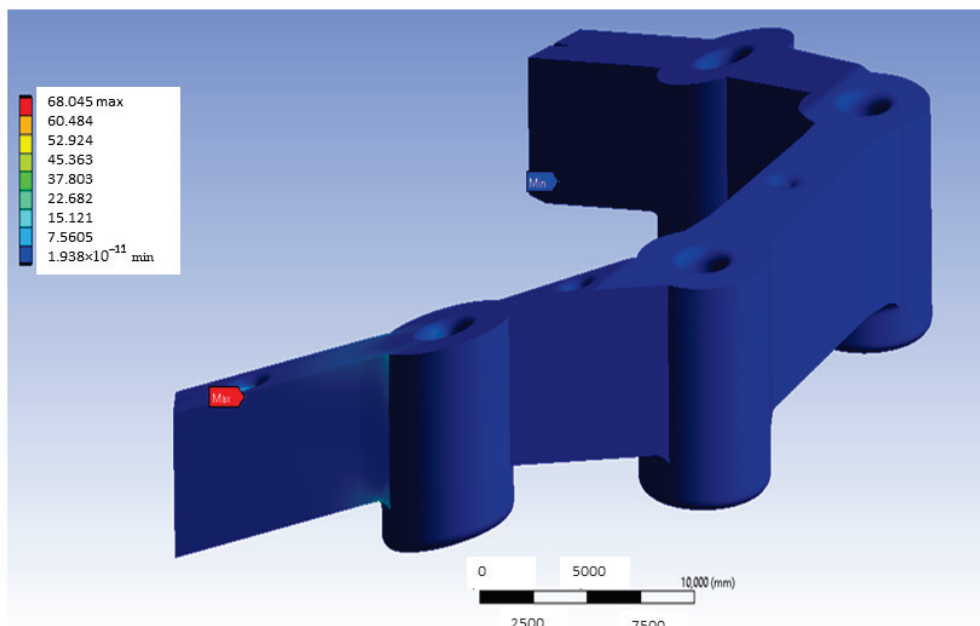


Figure 7. Stress map of the G-PMMA model with distal loading (units: MPa).

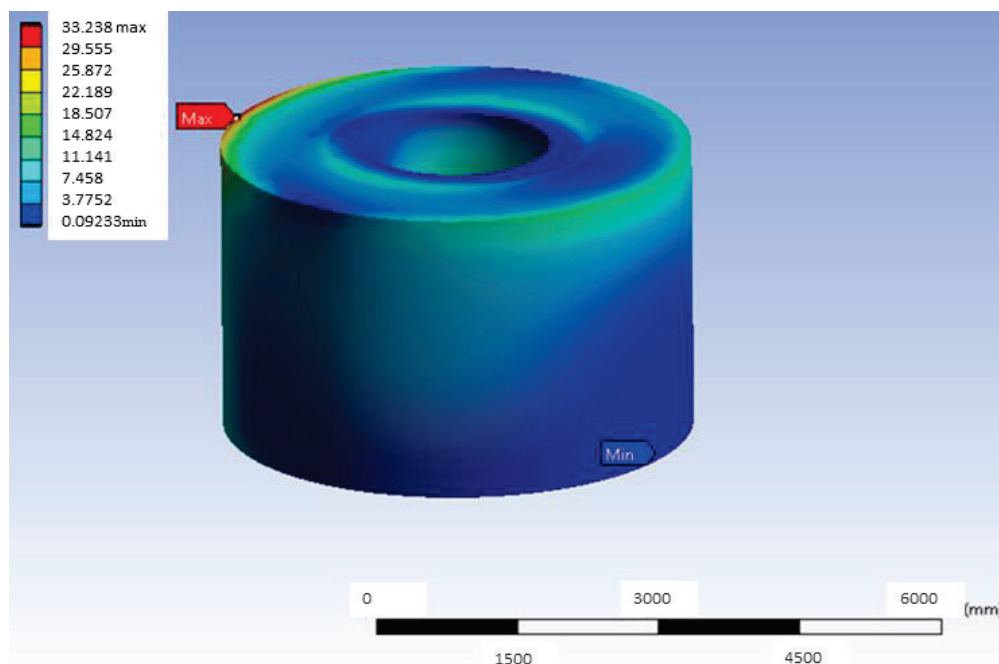


Figure 8. Stress map of the most loaded bushing of Ti6Al4V in the G-PMMA model with distal load.

The most important results obtained for the G-PMMA model with distal load applied are shown in Table 2, which presents the global displacement (minimum and maximum), the combined von Mises stress (minimum and maximum), and the maximum strain of the assembly that forms Model 1, while the data for the area of the inserts are presented in Table 2. Also, another table that presents the summarized results for the most loaded bushing of the same model are presented in Table 3. This scheme of results layout will be repeated throughout the nine models under study.

Table 2. Summary of results obtained for distal load G-PMMA model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	$2.45 \cdot 10^{-12}$	$1.37 \cdot 10^{-2}$	$3.88 \cdot 10^{-12}$	13.609	$4.27 \cdot 10^{-3}$
10 N	$4.90 \cdot 10^{-12}$	$2.74 \cdot 10^{-2}$	$7.75 \cdot 10^{-12}$	27.218	$8.55 \cdot 10^{-3}$
15 N	$7.35 \cdot 10^{-12}$	$4.11 \cdot 10^{-2}$	$1.16 \cdot 10^{-11}$	40.827	$1.28 \cdot 10^{-2}$
20 N	$9.80 \cdot 10^{-12}$	$5.47 \cdot 10^{-2}$	$1.55 \cdot 10^{-11}$	54.436	$1.71 \cdot 10^{-2}$
25 N	$1.23 \cdot 10^{-11}$	$8.85 \cdot 10^{-2}$	$1.94 \cdot 10^{-11}$	68.045	$2.13 \cdot 10^{-2}$

Table 3. Results obtained for the most loaded bushing in the distal load G-PMMA model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	$2.45 \cdot 10^{-12}$	$1.77 \cdot 10^{-4}$	$5.39 \cdot 10^{-8}$	6.6476
10 N	$4.90 \cdot 10^{-12}$	$3.54 \cdot 10^{-4}$	$1.08 \cdot 10^{-7}$	13.295
15 N	$7.35 \cdot 10^{-12}$	$5.31 \cdot 10^{-4}$	$1.62 \cdot 10^{-7}$	19.943
20 N	$9.80 \cdot 10^{-12}$	$7.08 \cdot 10^{-4}$	$2.15 \cdot 10^{-7}$	26.59
25 N	$1.23 \cdot 10^{-11}$	$8.85 \cdot 10^{-4}$	$2.69 \cdot 10^{-7}$	33.238

The same kind of analysis was performed for the G-PMMA models with loads in medial and central positions. In order to keep the present scientific article within a reasonable length, no figures of either of them are included in this paper, but tabulated summarized results are presented in Table 4 for the G-PMMA model with medial load and in Table 5 for its most loaded bushing, while Table 6 presents global results for the G-PMMA model with distal load, and Table 7 details the results of the most loaded bushing of the same model.

Table 4. Summary of results obtained for medial load G-PMMA model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	$3.10 \cdot 10^{-25}$	$3.14 \cdot 10^{-3}$	$3.85 \cdot 10^{-11}$	13,314	$4.17 \cdot 10^{-3}$
10 N	$6.20 \cdot 10^{-25}$	$6.29 \cdot 10^{-3}$	$7.70 \cdot 10^{-11}$	26,628	$8.35 \cdot 10^{-3}$
15 N	$9.30 \cdot 10^{-25}$	$9.43 \cdot 10^{-3}$	$1.16 \cdot 10^{-10}$	39,942	$1.25 \cdot 10^{-2}$
20 N	$1.24 \cdot 10^{-24}$	$1.26 \cdot 10^{-2}$	$1.54 \cdot 10^{-10}$	53,256	$1.67 \cdot 10^{-2}$
25 N	$1.55 \cdot 10^{-24}$	$1.57 \cdot 10^{-2}$	$1.93 \cdot 10^{-10}$	66,570	$2.09 \cdot 10^{-2}$

Table 5. Results obtained for the most loaded bushing in the medial load G-PMMA model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	$3.10 \cdot 10^{-25}$	$1.93 \cdot 10^{-5}$	$6.90 \cdot 10^{-7}$	2.8598
10 N	$6.20 \cdot 10^{-25}$	$3.85 \cdot 10^{-5}$	$1.38 \cdot 10^{-6}$	5.7196
15 N	$9.30 \cdot 10^{-25}$	$5.77 \cdot 10^{-5}$	$2.07 \cdot 10^{-6}$	8.5794
20 N	$1.24 \cdot 10^{-24}$	$7.70 \cdot 10^{-5}$	$2.76 \cdot 10^{-6}$	11.439
25 N	$1.55 \cdot 10^{-24}$	$9.62 \cdot 10^{-5}$	$3.45 \cdot 10^{-6}$	14.299

Table 6. Summary of results obtained for central load G-PMMA model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	0	$3.72 \cdot 10^{-3}$	$2.87 \cdot 10^{-9}$	13,140	$4.12 \cdot 10^{-3}$
10 N	0	$7.43 \cdot 10^{-3}$	$5.74 \cdot 10^{-9}$	26,280	$8.25 \cdot 10^{-3}$
15 N	0	$1.11 \cdot 10^{-2}$	$8.60 \cdot 10^{-9}$	39,421	$1.24 \cdot 10^{-2}$
20 N	0	$1.49 \cdot 10^{-2}$	$1.15 \cdot 10^{-8}$	52,561	$1.65 \cdot 10^{-2}$
25 N	0	$1.86 \cdot 10^{-2}$	$1.43 \cdot 10^{-8}$	65,701	$2.06 \cdot 10^{-2}$

Table 7. Results obtained for the most loaded bushing in the central load G-PMMA model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	0	$2.03 \cdot 10^{-5}$	$5.14 \cdot 10^{-5}$	3.4705
10 N	0	$4.05 \cdot 10^{-5}$	$1.03 \cdot 10^{-4}$	6.9409
15 N	0	$6.08 \cdot 10^{-5}$	$1.54 \cdot 10^{-4}$	10.411
20 N	0	$8.11 \cdot 10^{-5}$	$2.06 \cdot 10^{-4}$	13.882
25 N	0	$1.01 \cdot 10^{-4}$	$2.57 \cdot 10^{-4}$	17.352

3.2. Results of the Model Made of PEEK for the Different Load Conditions Under Study

The methodology employed for the analysis of the model made of PEEK was the same as the G-PMMA model. In this section, equivalent figures to those presented for G-PMMA in Section 3.1 are presented. Figure 9 shows the deformation map in a mid-plane of the PEEK model with Ti6Al4V inserts. This figure presents the load located in the distal position. The median plane was obtained with the same methodology as in the G-PMMA model. As can be seen in the referred figure, the deformation map in the plane is continuous, a fact that validates that the geometrical modeling has been correct since a smooth transition of the load between components is achieved. This fact is present in all the models under analysis in this research. It is important to consider that the load is applied vertically downwards, and that the model is embedded in the inserts, so that a state of triaxial stresses with a certain buckling component is produced.

Figure 10 shows the deformation map corresponding to the bushing of the PEEK material model with distal load most affected by the application of the referred load. Also, Figure 11 shows the stress map of the PEEK model with distal loading.

As can be seen in Figure 11, the maximum stress value obtained in the PEEK model with distal loading is 66.485 MPa. Having obtained the general von Mises equivalent stress map, the area of the inserts is considered the most critical for the present experiment; the graphical results of the most loaded bushing are presented in Figure 12. As with the G-PMMA model, there is a concentration of stress in the upper left zone, so that in real practice it would be expected that the model would erode or deform, and the load in that zone would be somewhat lower than that of theoretical model. In other words, it would be possible that in the red zone, erosion or deformation would occur and this would reduce the real stress.

As in the case of the G-PMMA material, for the PEEK models with the load in medial and central positions, studies equivalent to those carried out with the model with the load in the distal position were performed; therefore, analogous tables are presented. Table 8 presents a summary of the results obtained for PEEK model with distal load while Table 9 presents the results of the most loaded bushing in the same model.

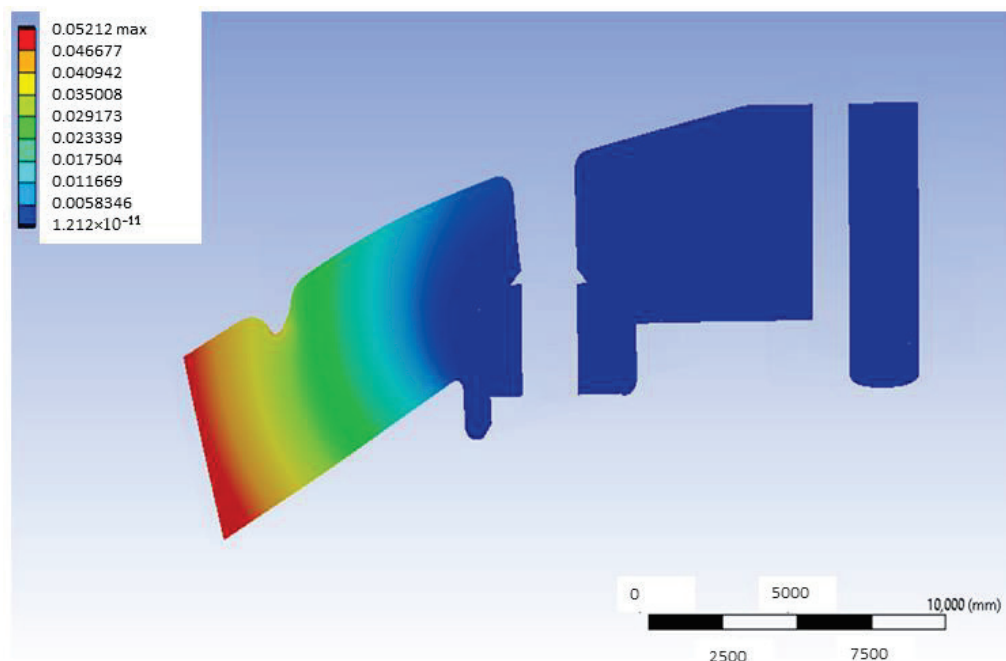


Figure 9. Mid-plane deformation map of the distally loaded PEEK model (units: mm).

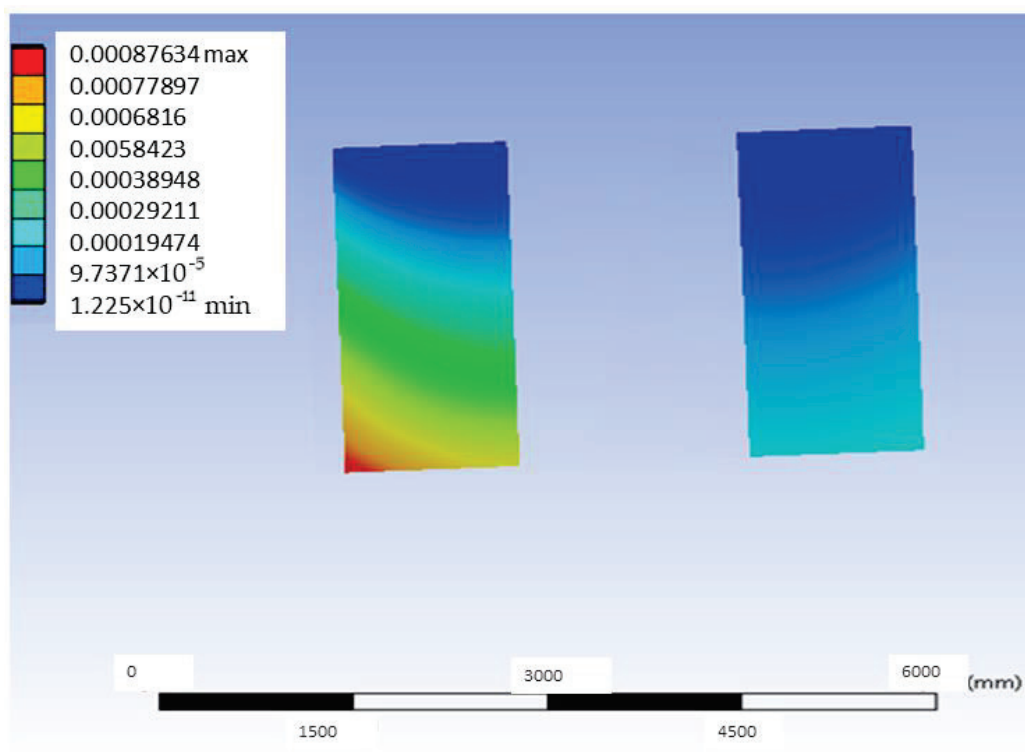


Figure 10. Deformation map of the Ti6Al4V bushing most affected by the application of the distal load in the PEEK model (units: mm).

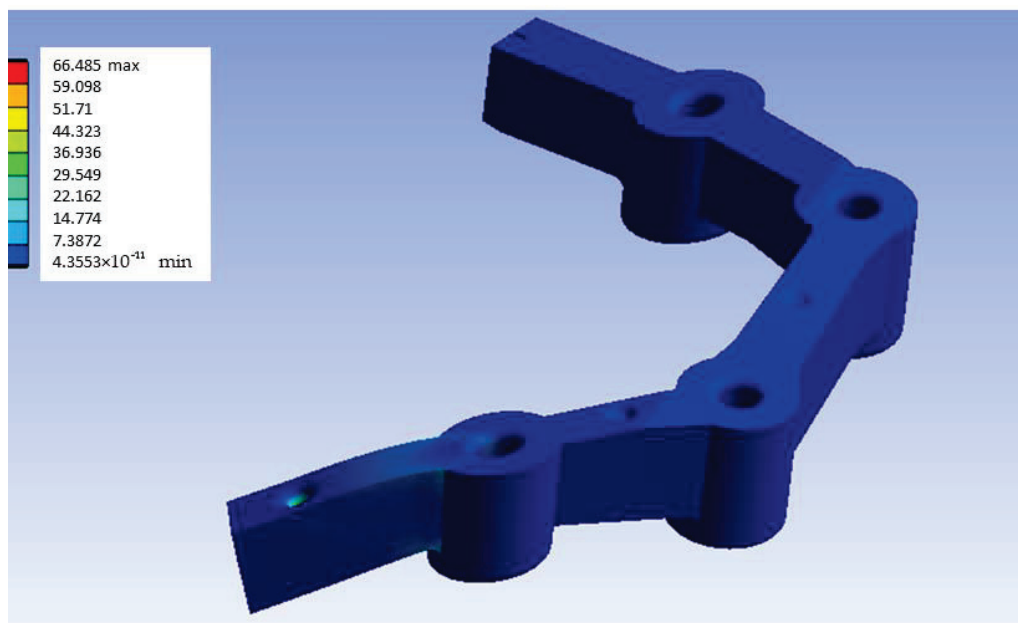


Figure 11. Stress map of the PEEK model with distal loading (units: MPa).

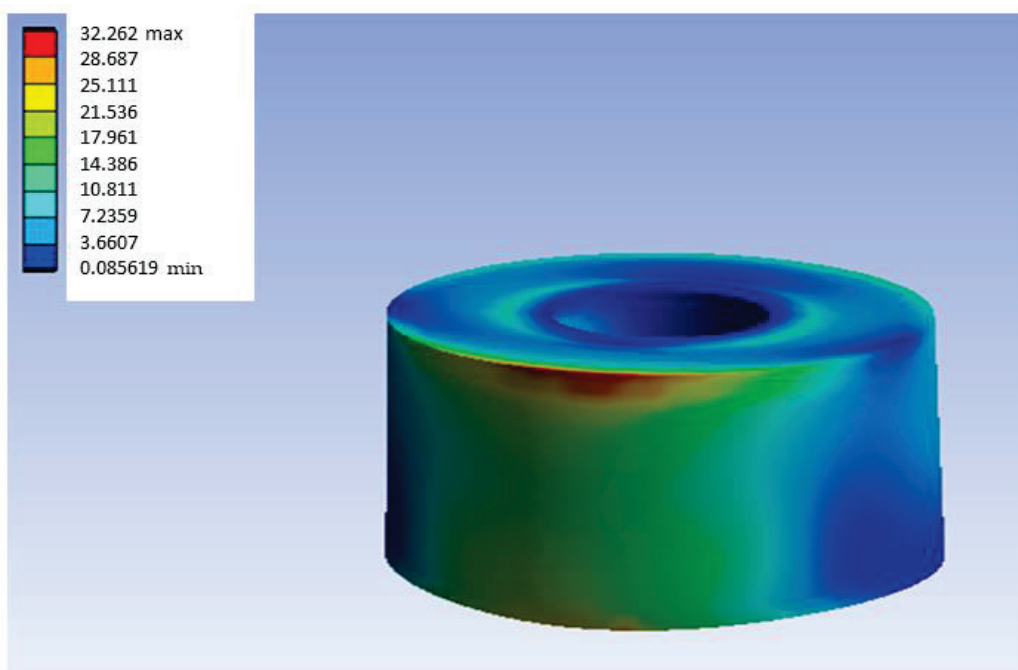


Figure 12. Stress map of the most loaded bushing of Ti6Al4V in the PEEK model with distal load.

Table 8. Summary of results obtained for distal load PEEK model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	$2.43 \cdot 10^{-12}$	$1.05 \cdot 10^{-2}$	$8.72 \cdot 10^{-12}$	13.297	$3.18 \cdot 10^{-3}$
10 N	$4.85 \cdot 10^{-12}$	$2.10 \cdot 10^{-2}$	$1.74 \cdot 10^{-11}$	26.594	$6.36 \cdot 10^{-3}$
15 N	$7.28 \cdot 10^{-12}$	$3.15 \cdot 10^{-2}$	$2.61 \cdot 10^{-11}$	39.891	$9.55 \cdot 10^{-3}$
20 N	$9.70 \cdot 10^{-12}$	$4.20 \cdot 10^{-2}$	$3.48 \cdot 10^{-11}$	53.188	$1.27 \cdot 10^{-2}$
25 N	$1.21 \cdot 10^{-11}$	$5.25 \cdot 10^{-2}$	$4.36 \cdot 10^{-11}$	66.485	$1.59 \cdot 10^{-2}$

Table 9. Results obtained for the most loaded bushing in the distal load PEEK model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	$2.43 \cdot 10^{-12}$	$1.75 \cdot 10^{-4}$	$9.26 \cdot 10^{-8}$	6.4523
10 N	$4.85 \cdot 10^{-12}$	$3.51 \cdot 10^{-4}$	$1.85 \cdot 10^{-7}$	12.905
15 N	$7.28 \cdot 10^{-12}$	$5.26 \cdot 10^{-4}$	$2.78 \cdot 10^{-7}$	19.357
20 N	$9.70 \cdot 10^{-12}$	$7.01 \cdot 10^{-4}$	$3.70 \cdot 10^{-7}$	25.809
25 N	$1.21 \cdot 10^{-11}$	$8.76 \cdot 10^{-4}$	$4.63 \cdot 10^{-7}$	32.262

In the following tables, equivalent results to those presented for PEEK models with medial and central loads are detailed. Thus, following the same system, Table 10 summarizes the results obtained for the PEEK model with medial load, while Table 11 shows the results obtained for the most loaded bushing in this model. In the same way, Tables 12 and 13 present the results obtained for the PEEK model with central load and the results of the most loaded bushing of such model, respectively.

Table 10. Summary of results obtained for medial load PEEK model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	$2.67 \cdot 10^{-13}$	$5.15 \cdot 10^{-4}$	$4.09 \cdot 10^{-11}$	13.045	$3.12 \cdot 10^{-3}$
10 N	$5.34 \cdot 10^{-13}$	$1.03 \cdot 10^{-3}$	$8.18 \cdot 10^{-11}$	26.090	$6.23 \cdot 10^{-3}$
15 N	$8.01 \cdot 10^{-13}$	$1.55 \cdot 10^{-3}$	$1.23 \cdot 10^{-10}$	39.135	$9.35 \cdot 10^{-3}$
20 N	$1.07 \cdot 10^{-12}$	$2.06 \cdot 10^{-3}$	$1.64 \cdot 10^{-10}$	52.180	$1.25 \cdot 10^{-2}$
25 N	$1.33 \cdot 10^{-12}$	$2.58 \cdot 10^{-3}$	$2.04 \cdot 10^{-10}$	65.225	$1.56 \cdot 10^{-2}$

Table 11. Results obtained for the most loaded bushing in the distal load PEEK model, being: Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	$2.67 \cdot 10^{-13}$	$1.08 \cdot 10^{-5}$	$8.30 \cdot 10^{-7}$	2.6418
10 N	$5.34 \cdot 10^{-13}$	$2.15 \cdot 10^{-5}$	$1.66 \cdot 10^{-6}$	5.2836
15 N	$8.01 \cdot 10^{-13}$	$3.23 \cdot 10^{-5}$	$2.49 \cdot 10^{-6}$	7.9254
20 N	$1.07 \cdot 10^{-12}$	$4.31 \cdot 10^{-5}$	$3.32 \cdot 10^{-6}$	10.567
25 N	$1.33 \cdot 10^{-12}$	$5.39 \cdot 10^{-5}$	$4.15 \cdot 10^{-6}$	13.209

Table 12. Summary of results obtained for central load PEEK model, being: Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	0	$2.79 \cdot 10^{-3}$	$2.83 \cdot 10^{-9}$	12.881	$3.08 \cdot 10^{-3}$
10 N	0	$5.58 \cdot 10^{-3}$	$5.65 \cdot 10^{-9}$	25.761	$6.16 \cdot 10^{-3}$
15 N	0	$8.37 \cdot 10^{-3}$	$8.48 \cdot 10^{-9}$	38.642	$9.24 \cdot 10^{-3}$
20 N	0	$1.12 \cdot 10^{-2}$	$1.13 \cdot 10^{-8}$	51.523	$1.23 \cdot 10^{-2}$
25 N	0	$1.39 \cdot 10^{-2}$	$1.41 \cdot 10^{-8}$	64.404	$1.54 \cdot 10^{-2}$

Table 13. Results obtained for the most loaded bushing in the central load PEEK model, being: Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	0	$1.93 \cdot 10^{-5}$	$4.86 \cdot 10^{-5}$	3.1576
10 N	0	$3.86 \cdot 10^{-5}$	$9.71 \cdot 10^{-5}$	6.3152
15 N	0	$5.79 \cdot 10^{-5}$	$1.46 \cdot 10^{-4}$	9.4727
20 N	0	$7.71 \cdot 10^{-5}$	$1.94 \cdot 10^{-4}$	12.63
25 N	0	$9.64 \cdot 10^{-5}$	$2.43 \cdot 10^{-4}$	15.788

3.3. Results of the Model Made of Ti6Al4V for the Different Load Conditions Under Study

The methodology employed for the analysis of the model made of Ti6Al4V was the same as for the other two materials. Figure 13 shows the deformation map in the mid-plane of the Ti6Al4V model with distal load. This figure presents the load located in the distal position. This figure is of special relevance, given that continuity can be observed between the area of the bushing, and the rest of the model, which, in this case, is made of the same material, so continuity was expected. The figure serves to verify compliance with the theory and the proper functioning of the model's contacts. Figure 14 shows the deformation map corresponding to the bushing most affected by the application of the load.

Figure 15 shows the stress map of the Ti6Al4V model with distal loading. As is presented in Figure 15, the maximum stress obtained was 68.505 MPa. After generating the von Mises equivalent stress map, we focus on examining the most critical area relevant to the experiment's practical application: the region surrounding the inserts. More specifically, Figure 16 presents the most loaded bushing of the Ti6Al4V model with distal loading.

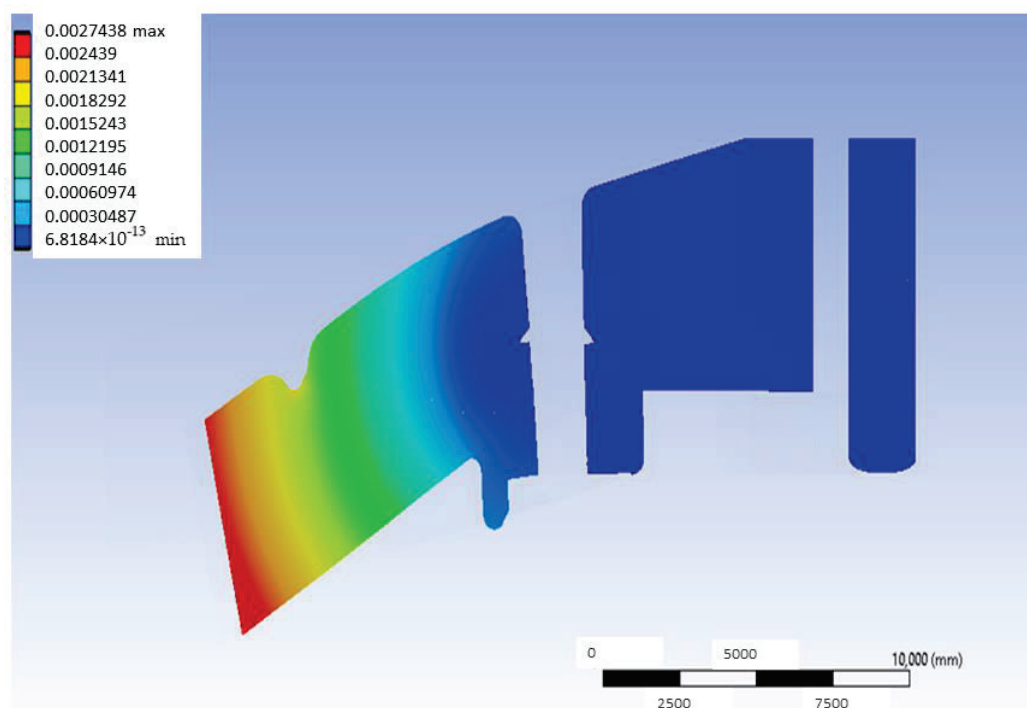


Figure 13. Mid-plane deformation map of the distally loaded Ti6Al4V model (units: mm).

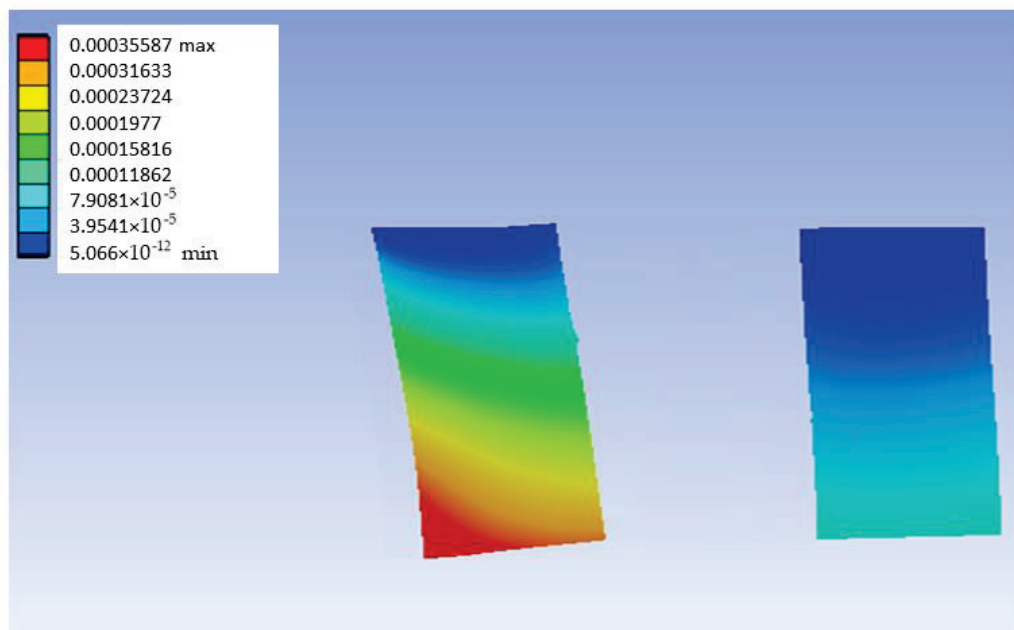


Figure 14. Deformation map of the Ti6Al4V bushing most affected by the application of the distal load in the Ti6Al4V model (units: mm).

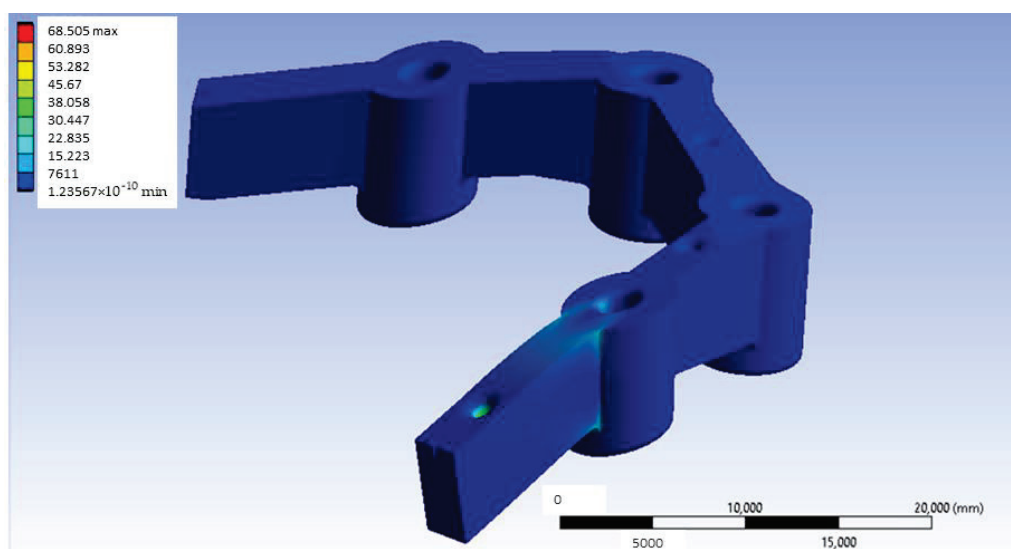


Figure 15. Stress map of the Ti6Al4V model with distal loading (units: MPa).

As with the other materials, there is a concentration of stress in the upper left zone. For the case of the Ti6Al4V models, the equivalent tables for the same load conditions as in the two previous materials will be presented. Table 14 contains a summary of the results obtained for the distal load Ti6Al4V model, while Table 15 presents the results obtained for the most loaded bushing of such model. Table 16 summarizes of results obtained for medial load Ti6Al4V model. Table 17 shows the results obtained for the most loaded bushing in the medial load TiAlV4 model. Tables 18 and 19 show the results obtained for central load and the results obtained for the most loaded bushing in the central load Ti6Al4V model, respectively.

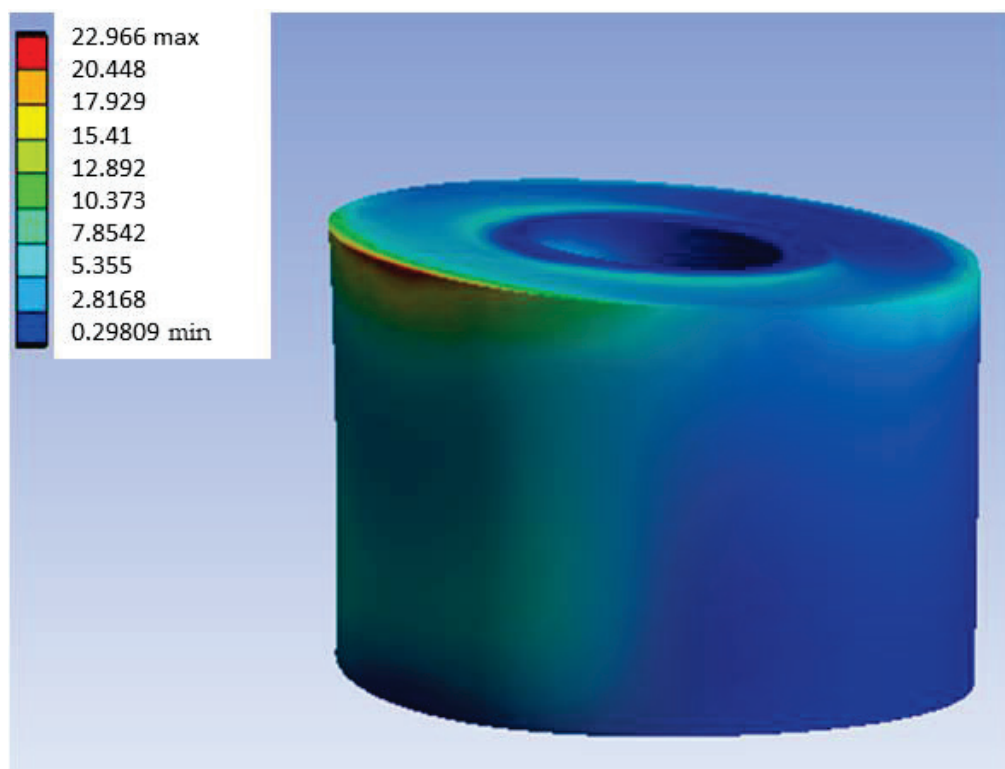


Figure 16. Stress map of the most loaded bushing of Ti6Al4V in the Ti6Al4V model with distal load.

Table 14. Summary of results obtained for distal load Ti6Al4V model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	$1.36 \cdot 10^{-13}$	$5.49 \cdot 10^{-4}$	$2.51 \cdot 10^{-11}$	13.701	$1.21 \cdot 10^{-4}$
10 N	$2.73 \cdot 10^{-13}$	$1.10 \cdot 10^{-3}$	$5.03 \cdot 10^{-11}$	27.402	$2.42 \cdot 10^{-4}$
15 N	$4.09 \cdot 10^{-13}$	$1.65 \cdot 10^{-3}$	$7.54 \cdot 10^{-11}$	41.103	$3.63 \cdot 10^{-4}$
20 N	$5.45 \cdot 10^{-13}$	$2.20 \cdot 10^{-3}$	$1.01 \cdot 10^{-10}$	54.804	$4.84 \cdot 10^{-4}$
25 N	$6.82 \cdot 10^{-13}$	$2.74 \cdot 10^{-3}$	$1.26 \cdot 10^{-10}$	68.505	$6.05 \cdot 10^{-4}$

Table 15. Results obtained for the most loaded bushing in the distal load Ti6Al4V model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	$1.01 \cdot 10^{-12}$	$7.12 \cdot 10^{-5}$	$5.09 \cdot 10^{-7}$	4.5933
10 N	$2.03 \cdot 10^{-12}$	$1.42 \cdot 10^{-4}$	$1.02 \cdot 10^{-6}$	9.1866
15 N	$3.04 \cdot 10^{-12}$	$2.14 \cdot 10^{-4}$	$1.53 \cdot 10^{-6}$	13.78
20 N	$4.05 \cdot 10^{-12}$	$2.85 \cdot 10^{-4}$	$2.03 \cdot 10^{-6}$	18.373
25 N	$5.07 \cdot 10^{-12}$	$3.56 \cdot 10^{-4}$	$2.54 \cdot 10^{-6}$	22.966

Table 16. Summary of results obtained for medial load Ti6Al4V model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	$1.04 \cdot 10^{-25}$	$9.20 \cdot 10^{-5}$	$2.00 \cdot 10^{-10}$	13.365	$1.18 \cdot 10^{-4}$
10 N	$2.08 \cdot 10^{-25}$	$1.84 \cdot 10^{-4}$	$4.01 \cdot 10^{-10}$	26.730	$2.36 \cdot 10^{-4}$
15 N	$3.12 \cdot 10^{-25}$	$2.76 \cdot 10^{-4}$	$6.01 \cdot 10^{-10}$	40.095	$3.53 \cdot 10^{-4}$
20 N	$4.16 \cdot 10^{-25}$	$3.68 \cdot 10^{-4}$	$8.01 \cdot 10^{-10}$	53.460	$4.71 \cdot 10^{-4}$
25 N	$5.20 \cdot 10^{-25}$	$4.60 \cdot 10^{-4}$	$1.00 \cdot 10^{-9}$	66.825	$5.89 \cdot 10^{-4}$

Table 17. Results obtained for the most loaded bushing in the medial load TiAlV4 model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	$1.04 \cdot 10^{-25}$	$5.82 \cdot 10^{-6}$	$5.17 \cdot 10^{-6}$	0.90975
10 N	$2.08 \cdot 10^{-25}$	$1.16 \cdot 10^{-5}$	$1.03 \cdot 10^{-5}$	1.8195
15 N	$3.12 \cdot 10^{-25}$	$1.75 \cdot 10^{-5}$	$1.55 \cdot 10^{-5}$	2.7292
20 N	$4.16 \cdot 10^{-25}$	$2.33 \cdot 10^{-5}$	$2.07 \cdot 10^{-5}$	3.639
25 N	$5.20 \cdot 10^{-25}$	$2.91 \cdot 10^{-5}$	$2.59 \cdot 10^{-5}$	4.5487

Table 18. Summary of results obtained for central load Ti6Al4V model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa, Strain max: maximum strain value.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.	Strain Max.
5 N	0	$1.11 \cdot 10^{-4}$	$3.67 \cdot 10^{-9}$	13,188	$1.16 \cdot 10^{-4}$
10 N	0	$2.21 \cdot 10^{-4}$	$7.33 \cdot 10^{-9}$	26,376	$2.33 \cdot 10^{-4}$
15 N	0	$3.32 \cdot 10^{-4}$	$1.10 \cdot 10^{-8}$	39,564	$3.49 \cdot 10^{-4}$
20 N	0	$4.42 \cdot 10^{-4}$	$1.47 \cdot 10^{-8}$	52,752	$4.65 \cdot 10^{-4}$
25 N	0	$5.53 \cdot 10^{-4}$	$1.83 \cdot 10^{-8}$	65,940	$5.82 \cdot 10^{-4}$

Table 19. Results obtained for the most loaded bushing in the central load Ti6Al4V model, being Def. min: Minimum deformation value in mm, Def. max: maximum deformation value in mm, VM min: minimum von Mises stress value in MPa, VM max: maximum von Mises stress value in MPa.

Force	Def. Min.	Def. Max.	VM Min.	VM Max.
5 N	0	$6.22 \cdot 10^{-6}$	$7.80 \cdot 10^{-5}$	0.91538
10 N	0	$1.24 \cdot 10^{-5}$	$1.56 \cdot 10^{-4}$	1.8308
15 N	0	$1.87 \cdot 10^{-5}$	$2.34 \cdot 10^{-4}$	2.7461
20 N	0	$2.49 \cdot 10^{-5}$	$3.12 \cdot 10^{-4}$	3.6615
25 N	0	$3.11 \cdot 10^{-5}$	$3.90 \cdot 10^{-4}$	4.5769

3.4. Comparison of the Results Obtained for the Models and Load Conditions Under Study

In this section and as a summary, a series of graphs are presented with the maximum values of stresses and deformations both in the set of models under analysis and in the bushings that support the higher stresses values in each of the models.

The maximum deformation values in the three models for the loading conditions under study are presented in Figure 17. As can be observed, in all the materials analyzed, the highest deformation values correspond to the load applied in the distal position, followed

by the applied in a central position and, finally, by the load applied in a medial position. Furthermore, the largest deformations on average for the three load conditions correspond to the models made of G-PMMA, followed by the PEEK models and, finally, by those made entirely of Ti6Al4V. With regard to the order of the materials, it can be said that it was the expected one, given that it corresponds to the same order as their Young's modulus values, from lowest to highest. Also, with regard to the magnitude of the maximum deformation of the assembly versus the place of load application, it is also consistent that the position that produces the greatest deformations is the distal one, given that a lever arm is created in the structure that leads to higher displacement values induced by the load.

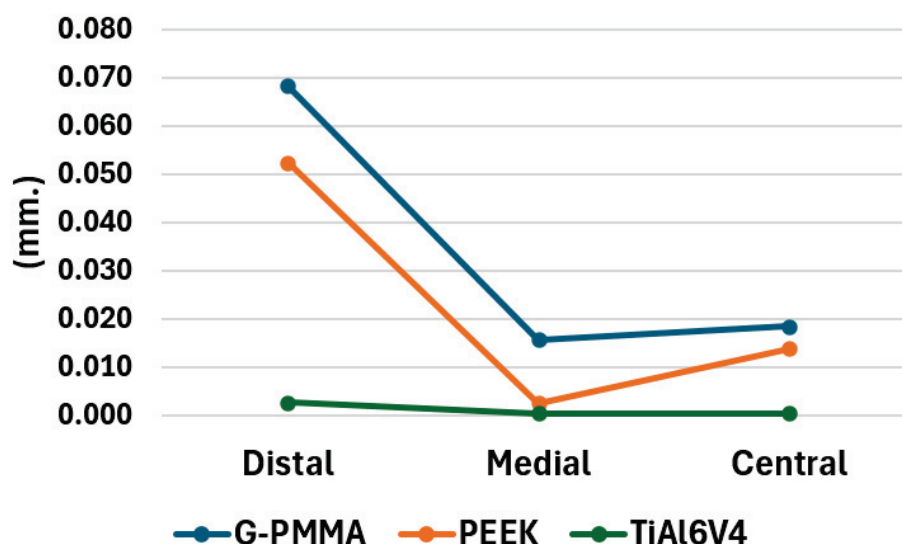


Figure 17. Maximum deformation of all the models (in mm.) by materials and force position.

The comparison of the maximum stress values achieved according to the von Mises criteria in the models of the three materials and for the three loading conditions under study are presented in Figure 18. Taking into account what is shown in the referred Figure, it is possible to affirm that, in all cases, the maximum stress value occurs for the models with loads applied in the distal position, followed by the models with load in the medial position and, finally, the models with load in the central position. Furthermore, with the same position of the load, in all cases, the highest stress values are reached in those models that are made entirely of Ti6Al4V, followed by the models made of G-PMMA, except for its bushings and, finally, the PEEK models. That is, the inverse order to that of the deformations and which means that those models in which greater displacements occur generally reach lower values of maximum stresses.

Figure 19 presents the maximum deformation values reached in the most loaded bushings of all the models under study. As can be seen in the referred Figure, the deformations of the most loaded Ti6Al4V bushings in each of the models under study follow a similar trend, but the greatest deformations correspond to those loaded in the distal position, followed by the central and medial positions. Also, the bushings that present the greatest deformation are those corresponding to the models made of G-PMMA and PEEK, with very similar results. Although the deformation values in the medial and central position are similar, those in the medial position are somewhat higher. In this case, the highest strain values are also reached for the G-PMMA model, followed by the PEEK and, finally, the Ti6Al4V. Regarding the models with loading in the central position, almost indistinguishably, there are the bushings corresponding to the G-PMMA and PEEK models, while the maximum strain in the Ti6Al4V model is somewhat lower.

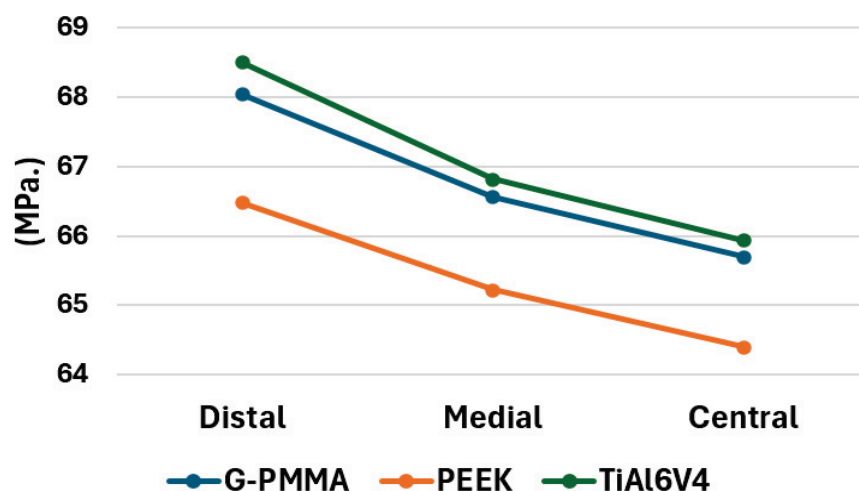


Figure 18. Maximum stress values according to the von Mises criteria of all the models (MPa.) by materials and force position.

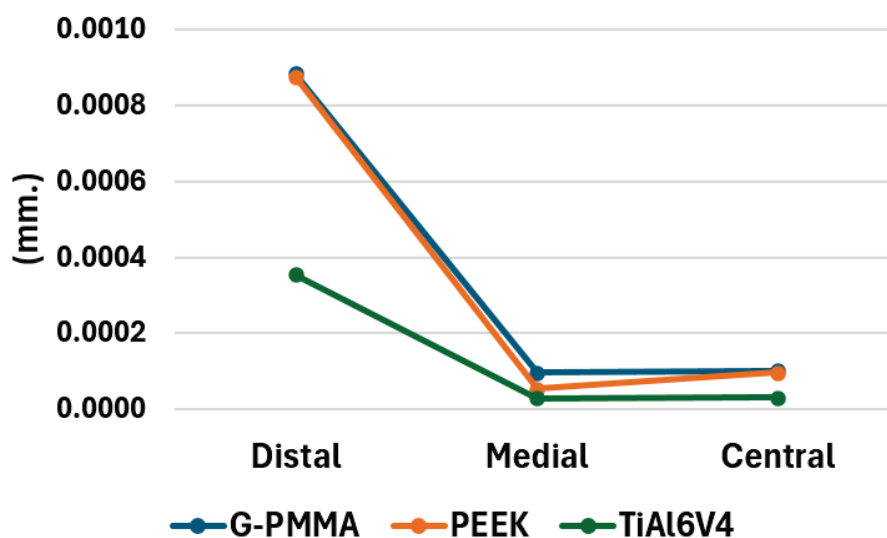


Figure 19. Maximum deformation values (in mm.) achieved in the most loaded bushes of each model by materials and force position.

Figure 20 presents the maximum von Mises stresses reached in the most loaded bushings of each model, which, in all cases, are made of Ti6Al4V. As can be observed in this figure, the highest values correspond to the models to which load has been applied in the distal position, followed by those with the load in medial position and, finally, those with the load in a central position. Regarding the materials of the different models, the highest values in all cases correspond to models made of G-PMMA, followed by slightly lower values for those models of PEEK, while the maximum deformation values of the bushing inserted in the Ti6Al4V models are noticeably lower. Figure 21 shows the values of the mechanical tests performed and the determination of the displacement in millimeters of the sample. Displacement values were obtained when the force applied in each of the positions was 150 N. As can be seen from the results, the distal position is the one that offers the greatest deformation, as in the case of the simulation that can be seen in Figures 17 and 19. The displacements decrease sharply for the medial and central positions, with the Ti6Al4V material offering the lowest displacement values in all the positions studied. The structures made of G-PMMA and PEEK show similar values although PEEK offers higher displacement in the distal position equaling the values in the medial position and slightly increases the displacement in the central position for the G-PMMA. From the

results obtained, it can be said that the mechanical tests reproduce the behavior simulated by the finite element method both in the positions of application of the mechanical load on the structure and in the type of material studied in the fabrication of the prosthetic structures.

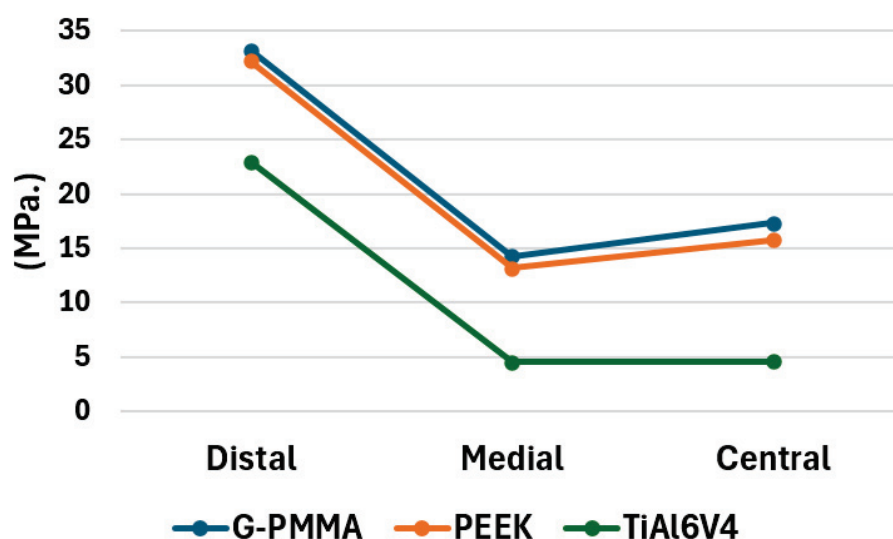


Figure 20. Maximum stress by von Mises criteria (in MPa) achieved in the most loaded bushes of each model by materials and force position.

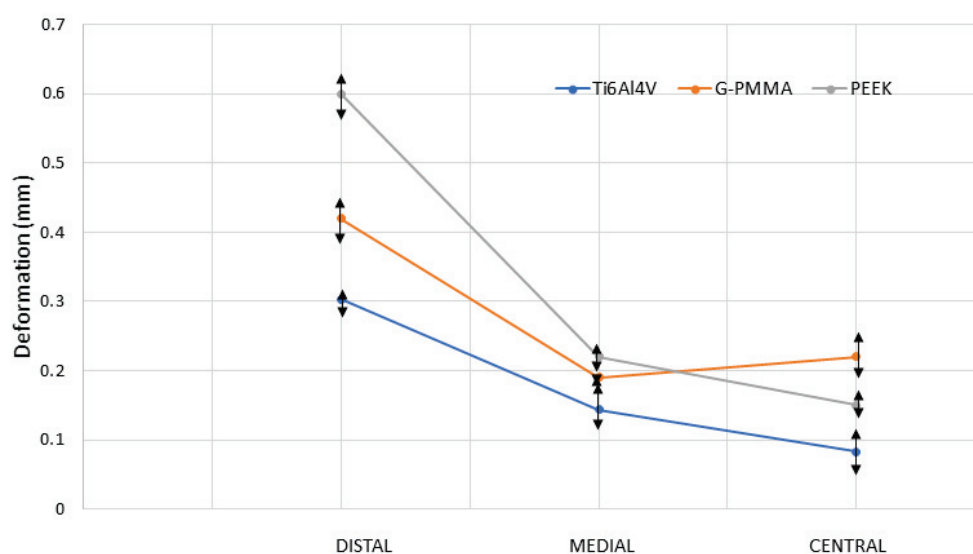


Figure 21. Deformation in millimeters obtained under a load of 150 N in flexural tests.

4. Discussion

One of the strengths of the present research is that all the models analyzed share the same mesh and geometry which makes the comparison among materials and load conditions easy. That is because the discrepancies observed among models will arise solely from variations in the materials and the different applied loading conditions. When speaking about different materials models, the one with the most different behavior when compared with other materials is Ti6Al4V [43].

The general behavior of G-PMMA and PEEK in the test performed is similar. About maximum deformations in the model, it has been found that deformations are higher in the G-PMMA models when compared to those made of PEEK. This result was expected, primarily due to differences in their mechanical properties. PEEK is a high-performance thermoplastic with significantly superior stiffness, strength, and thermal stability. These

characteristics enable PEEK to resist deformation under mechanical stress, particularly at elevated temperatures, making it more rigid and resistant to strain. In contrast, G-PMMA, although reinforced with graphene fibers to enhance its strength, retains the inherent limitations of a polymeric material, with a lower modulus of elasticity and tensile strength compared to PEEK. As a result, G-PMMA is more prone to deformation when subjected to mechanical loads. Therefore, the higher deformation observed in G-PMMA is a predictable outcome based on the inherent mechanical properties of these two materials [44–46].

The highest values of maximum stress according to the von Mises criteria are achieved in models made of Ti6Al4V, followed by G-PMMA and PEEK. These stress values are primarily governed by the distinct mechanical properties inherent to each material. Ti6Al4V, being a titanium alloy, possesses a significantly higher strength-to-weight ratio compared to the other two materials. This alloy is designed to withstand high mechanical stresses, exhibiting excellent tensile strength and resistance to fatigue and deformation. As a result, when subjected to loads, Ti6Al4V can absorb a higher amount of stress before undergoing plastic deformation, which explains why it shows the highest stress values among the three materials. Its strength is further enhanced by its metallic structure, which allows for efficient load distribution and retention of structural integrity under extreme conditions [47,48].

In contrast, G-PMMA is a polymer composite, reinforced with graphene fibers to improve its mechanical properties over pure PMMA. While the addition of graphene fibers enhances its strength, it remains a thermoplastic material with lower intrinsic stiffness and tensile strength compared to titanium alloys. As such, G-PMMA is more prone to plastic deformation under stress compared to Ti6Al4V. However, due to its relatively higher stiffness compared to other common polymers, G-PMMA is able to withstand moderate stress levels before significant deformation occurs, placing it in the intermediate position between Ti6Al4V and PEEK in terms of stress capacity. Lastly, PEEK, although a high-performance thermoplastic with excellent mechanical properties, including high strength and thermal stability, is still a polymer material and thus inherently more susceptible to deformation under high loads compared to metals. While PEEK is known for its excellent chemical resistance, toughness, and dimensional stability, its intrinsic modulus of elasticity and tensile strength are lower than those of both Ti6Al4V and G-PMMA. Consequently, PEEK experiences higher deformation under applied loads and reaches a lower stress value before yielding. Its behavior under load is governed by the molecular structure of the polymer, which, although optimized for performance, cannot match the superior stress resistance provided by metallic materials such as Ti6Al4V.

In the case of the maximum von Mises stress and deformation values achieved in the most loaded bushing of each model, values of G-PMMA and PEEK models are really close and in some cases they would even be considered within the 0.2% of the uncertainty confidence interval of error due to the model convergence. Please note that in all models, bushes are made of Ti6Al4V alloy.

Also, it should be noted that there is also a difference in the results obtained depending on the applied load, whether distal, medial, or central, showing that in all simulations, it is the distal test that offers the worst results in terms of presenting a higher value for both displacement and tension. This situation would be mitigated if the prosthesis cantilever is reduced. Likewise, it is also advisable to check whether the base model in PEEK or G-PMMA with Ti6Al4V bushings can afford a displacement map 10 times greater (an order of magnitude) than that obtained with the model made entirely of Ti6Al4V. Therefore, the further the loading point is from a strong bond, the more global displacement there is, and this is proportional to the unitary displacement, which is proportional to Young's modulus, which depends on the material, which is why what we see happens. The displacements in the bushings are similar, because Ti6Al4V is much more rigid than the other two materials.

Overall, the graph suggests that the distal loading position should be studied in order to find if it can be restricted in some way so that the stress does not generate as much deformation compared to the rest.

The present study is subject to certain limitations inherent to finite element analyses in biomechanics. A key simplifying assumption is the use of linear-elastic material models for all the pieces involved in these models. In fact, materials have been considered homogeneous and isotropic. While these assumptions facilitate computational efficiency and are widely used in similar studies, they do not account for the non-linear and time-dependent behavior of biological tissues, such as viscoelasticity or plastic deformation under high loads.

The use of linear models, simplifying the contacts between elements, can also partially distort the results, an effect that we have tried to minimize using loads that ensure that the material works below the elastic limit at all conditions. Also, the geometries used, faithful reproductions of the real models, generate stress concentrations, which in tests and simulations by the finite element method often do not occur because the geometry used has been simplified (the fact being that maintaining visual verisimilitude has hardly been practiced in these models). This generates load values higher than the real ones, concentrated in the “corners” and in other singularities, which, in a laboratory test, would disappear practically immediately since the physical object would suffer from erosion or deformation.

Boundary conditions were also idealized to represent a controlled *in silico* environment with fully fixed supports, although these idealizations may not fully replicate the complex mechanical interactions occurring *in vivo*. Additionally, this analysis does not incorporate critical biological factors such as bone adaptation, remodeling, or degradation over time, which can significantly influence the long-term performance of dental prostheses. The effects of wear and fatigue, which are known to play a pivotal role in the lifespan of dental materials under repetitive loading, were also excluded.

From a clinical point of view, bruxing patients with high mechanical loads should be advised to use Ti6Al4V frameworks as polymeric frameworks may have limitations in mechanical properties. In patients with low to moderate load levels where dental esthetics is also an important part of the treatment, the polymeric materials studied can be used. In the latter case, the clinician should evaluate the type of structure and assess whether there are stresses in the distal positions with stresses with a large cantilever. In these cases, polymeric materials would not be advisable, as has been demonstrated in the mechanical tests carried out and confirmed by the simulation of the different structures by means of finite elements [36,48–50].

Despite these limitations, the study provides valuable insights into the biomechanical behavior of the prosthetic system under different loading conditions. Future research could integrate more advanced material models, non-linear contact interactions, and simulations of biological processes to enhance the clinical relevance of the findings.

5. Conclusions

Regarding the mechanical response of the materials, it was observed that the PEEK material presented a superior capacity to support high magnitude loads without compromising the structural integrity of the prosthesis. On the other hand, the G-PMMA material showed adequate resistance to deformation under moderate loads, but its performance decreased in high load scenarios, suggesting that its use is more indicated for prostheses in areas with less exposure to extreme masticatory forces. Finally, the PEEK material showed intermediate behavior in the three loading conditions, balancing strength, and flexibility, suggesting a potential for versatile use in prosthetic applications. However, it was noted

that this material may require additional adjustments to maintain long-term stability in patients with variable masticatory loads. The validation of the simulation by the finite element method was carried out by means of mechanical tests and the validity of the simulation was verified. It is possible to observe that in the application of 150N of the load, the distal position is the one with the highest deformation, with a displacement of 0.6 mm for PEEK, 0.43 mm for G-PMMA, and the lowest for Ti6Al4V, which corresponds to 0.30 mm. The deformations decrease sharply for the medial and central positions decreasing to values close to 0.1 mm. G-PMMA presents the highest deformation values (0.21 mm) in the central position. The choice of the appropriate material should consider not only the strength of the material under various loading conditions, but also the adaptability of the material to the specific needs of the patient. The findings of this study provide valuable information for decision making in the selection of prosthetic materials and establish a basis for future research that can further optimize the design of dental prostheses based on realistic conditions of use.

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

Funding: The authors are grateful to Filipe Fernandes y Susana Leite from laboratorio Symmetrya in Porto (Portugal) and the Spanish Government for its support through the research project MINECO (PID2022-137496OB-I00) and to the Klockner dental system.

Institutional Review Board Statement: Not applicable.

Informed Consent Statement: Not applicable.

Data Availability Statement: The data that support the findings of this study are available from the corresponding author upon reasonable request. Data are contained within the article.

Acknowledgments: The authors are grateful to Oscar Valencia and Rui Pedro Marques.

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

References

1. Lekholm, U.; van Steenberghe, D.; Herrmann, I.; Bolender, C.; Folmer, T.; Gunne, J.; Henry, P.; Higuchi, K.; Laney, W.; Lindén, U. Osseointegrated implants in the treatment of partially edentulous jaws: A prospective 5-year multicenter study. *Int. J. Oral Maxillofac. Implants* **1994**, *9*, 627–635.
2. Aparicio, C.; Manero, J.M.; Conde, F.; Pegueroles, M.; Planell, J.A.; Vallet-Regí, M.; Gil, F.J. Acceleration of apatite nucleation on microrough bioactive titanium for bone-replacing implants. *J. Biomed. Mater. Res. Part A* **2007**, *82*, 521–529. [CrossRef] [PubMed]
3. Buser, D.; Janner, S.F.; Wittneben, J.G.; Bragger, U.; Ramseier, C.A.; Salvi, G.E. 10-year survival and success rates of 511 titanium implants with a sandblasted and acid-etched surface: A retrospective study in 303 partially edentulous patients. *Clin. Implants Dent. Relat. Res.* **2012**, *14*, 839–851. [CrossRef] [PubMed]
4. Grandi, T.; Guazzi, P.; Samarani, R.; Grandi, G. Immediate loading of four (all-on-4) post-extractive implants supporting mandibular cross-arch fixed prostheses: 18-month follow-up from a multicenter prospective cohort study. *Eur. J. Oral Implantol.* **2012**, *5*, 277–285. [PubMed]
5. Padrós, R.; Punset, M.; Molmeneu, M.; Velasco, A.B.; Herrero-Climent, M.; Rupérez, E.; Gil, F.J. Mechanical properties of cochr dental-prosthesis restorations made by three manufacturing processes. influence of the microstructure and topography. *Metals* **2020**, *10*, 788. [CrossRef]
6. Skalak, R. Biomechanical considerations in osseointegrated prostheses. *J. Prosthet. Dent.* **1983**, *49*, 843–848. [CrossRef]
7. Sailer, I.; Karasan, D.; Todorovic, A.; Ligoutsikou, M.; Pjetursson, B.E. Prosthetic failures in dental implant therapy. *Periodontology* **2000** **2022**, *88*, 130–144. [CrossRef] [PubMed]

8. Manero, J.M.; Gil, F.J.; Padrós, E.; Planell, J.A. Applications of Environmental Scanning Electron Microscopy (ESEM) in Biomaterials Field. *Microsc. Res. Tech.* **2003**, *61*, 469–480. [CrossRef] [PubMed]
9. Brizuela-Velasco, A.; Álvarez-Arenal, Á.; Gil-Mur, F.J.; Herrero-Climent, M.; Chávarri-Prado, D.; Chento-Valiente, Y.; Dieguez-Pereira, M. Relationship Between Insertion Torque and Resonance Frequency Measurements, Performed by Resonance Frequency Analysis, in Micromobility of Dental Implants: An In Vitro Study. *Implants Dent.* **2015**, *24*, 607–611. [CrossRef]
10. Dieguez-Pereira, M.; Brizuela-Velasco, A.; Chavarri-Prado, D.; Perez-Pevida, E.; deLlanos-Lanchares, H.; Alvarez-Arenal, A. The Utility of Implant-Supported Fixed Dental Prosthesis Material for Implant Micromovement and Peri-implant Bone Microstrain: A Study in Rabbit Tibia. *Int. J. Oral Maxillofac. Implants* **2020**, *35*, 1132–1140. [CrossRef] [PubMed]
11. Sirandoni, D.; Leal, E.; Weber, B.; Noritomi, P.Y.; Fuentes, R.; Borie, E. Effect of Different Framework Materials in Implant-Supported Fixed Mandibular Prostheses: A Finite Element Analysis. *Int. J. Oral Maxillofac. Implants* **2019**, *34*, e107–e114. [CrossRef] [PubMed]
12. Ortorp, A.; Jönsson, D.; Mouhsen, A.; Vult von Steyern, P. The fit of cobalt-chromium three-unit fixed dental prostheses fabricated with four different techniques: A comparative in vitro study. *Dent. Mater.* **2011**, *27*, 356–363. [CrossRef] [PubMed]
13. Rödiger, M.; Schneider, L.; Rinke, S. Influence of Material Selection on the Marginal Accuracy of CAD-CAM-Fabricated Metal- and All-Ceramic Single Crown Copings. *BioMed Res. Int.* **2018**, *2018*, 2143906. [CrossRef] [PubMed]
14. Lövgren, N.; Roxner, R.; Klemendz, S.; Larsson, C. Effect of production method on surface roughness, marginal and internal fit, and retention of cobalt-chromium single crowns. *J. Prosth. Dent.* **2017**, *118*, 95–101. [CrossRef] [PubMed]
15. Rekow, E.D.; Erdman, A.G.; Riley, D.R.; Klamecki, B. CAD-CAM for dental restorations—some of the curious challenges. *IEEE Trans. Biomed. Eng.* **1991**, *38*, 314–318. [CrossRef] [PubMed]
16. Alqurashi, H.; Khurshid, Z.; Syed, A.U.Y.; Rashid Habib, S.; Rokaya, D.; Zafar, M.S. Polyetherketoneketone (PEKK): An emerging biomaterial for oral implants and dental prostheses. *J. Adv. Res.* **2020**, *28*, 87–95. [CrossRef] [PubMed]
17. Punset, M.; Brizuela, A.; Pérez-Pevida, E.; Herrero-Climent, M.; Manero, J.M.; Gil, J. Mechanical Characterization of Dental Prostheses Manufactured with PMMA-Graphene Composites. *Materials* **2022**, *15*, 5391. [CrossRef] [PubMed]
18. Vila, M.M.; Ginebra, M.P.; Gil, F.J.; Planell, J.A. Effect of porosity and environment on the mechanical behavior of acrylic bone cement modified with acrylonitrile-butadiene-styrene particles: I. Fracture toughness. *J. Biomed. Mater. Res.* **1999**, *48*, 121–127. [CrossRef]
19. Vazquez, B.; Ginebra, M.P.; Gil, F.J.; Planell, J.A.; Bravo, A.L.; San Roman, J. Radiopaque acrylic cements prepared with a new acrylic derivative of iodo-quinoline. *Biomaterials* **1999**, *20*, 2047–2053. [CrossRef] [PubMed]
20. Kotina, E.; Hamilton, A.; Lee, J.D.; Lee, S.J.; Grieco, P.C.; Pedrinaci, I.; Griseto, N.T.; Gallucci, G.O. Milled PMMA: A Material for Long-Term Implant-Supported Fixed Complete Dental Prostheses. *Int. J. Prosthodont.* **2024**, *37*, 225–231. [CrossRef] [PubMed]
21. Zafar, M.S. Prosthodontic Applications of Polymethyl Methacrylate (PMMA): An Update. *Polymers* **2020**, *12*, 2299. [CrossRef]
22. Colling, E.W. *The Physical Metallurgy of Titanium Alloys*; American Society for Metals: Geauga County, OH, USA, 1984; pp. 394–401.
23. Breme, J.; Biehl, V. *Metallic Biomaterials, Handbook of Biomaterials Properties*; Black, J., Hastings, G., Eds.; Chapman & Hall: London, UK, 1998; p. 137.
24. Wennerberg, A.; Albrektsson, T. On implant surfaces: A review of current knowledge and opinions. *Int. J. Oral Maxillofac. Implants* **2010**, *25*, 63–74. [PubMed]
25. Brunette, D.M.; Tengvall, P.; Textor, M.; Thomsen, P. (Eds.) *Titanium in Medicine: Material Science, Surface Science, Engineering, Biological Responses and Medical Applications*; Springer: Berlin, Germany, 2001; pp. 354–365.
26. Gil, F.J.; Planell, J.A. Aplicaciones biomédicas del titanio y sus aleaciones. *Biomecánica* **1993**, *2*, 34–43. [CrossRef]
27. Barbosa, G.A.; Bernardes, S.R.; de França, D.G.; das Neves, F.D.; de Mattos Mda, G.; Ribeiro, R.F. Stress Over Implants of One-Piece Cast Frameworks Made With Different Materials. *J. Craniofac Surg.* **2016**, *27*, 238–241. [CrossRef] [PubMed]
28. Gil, F.J.; Rodríguez, D.; Planell, J.A.; Cortada, M.; Giner, L.; Costa, S. Galvanic corrosion behaviour of Titanium implants coupled to dental alloys. *J. Mater. Sci. Mater. Med.* **2000**, *11*, 287–293.
29. Gil, F.J.; Sánchez, L.A.; Espias, A.; Planell, J.A. In vitro corrosion behaviour and metallic ion release of different prosthodontic alloys. *Int. Dent. J.* **1999**, *49*, 347–351. [CrossRef] [PubMed]
30. Al-Hity, R.R.; Kappert, H.F.; Viennot, S.; Dalard, F.; Grosogeat, B. Corrosion resistance measurements of dental alloys, are they correlated? *Dent. Mater.* **2007**, *23*, 679–687. [CrossRef]
31. Aparicio, C.; Gil, F.J.; Fonseca, C.; Barbosa, M.; Planell, J.A. Corrosion behaviour of commercially pure titanium shot blasted with different materials and sizes of shot particles for dental implant applications. *Biomaterials* **2003**, *24*, 263–273. [CrossRef]
32. Gautam, C.; Joyner, J.; Gautam, A.; Rao, J.; Vajtai, R. Zirconia based dental ceramics: Structure, mechanical properties, biocompatibility and applications. *Dalton Trans.* **2016**, *45*, 19194–19215. [CrossRef]
33. Chopra, D.; Guo, T.; Gulati, K.; Ivanovski, S. Load, unload and repeat: Understanding the mechanical characteristics of zirconia in dentistry. *Dent. Mater.* **2024**, *40*, e1–e17. [CrossRef] [PubMed]

34. De Angelis, F.; Vadini, M.; Buonvivere, M.; Valerio, A.; Di Cosola, M.; Piattelli, A.; Biferi, V.; D'Arcangelo, C. In Vitro Mechanical Properties of a Novel Graphene-Reinforced PMMA-Based Dental Restorative Material. *Polymers* **2023**, *15*, 622. [CrossRef] [PubMed]
35. Díez-Pascual, A.M. PMMA-Based Nanocomposites for Odontology Applications: A State-of-the-Art. *Int. J. Mol. Sci.* **2022**, *23*, 10288. [CrossRef] [PubMed]
36. Zidan, S.; Silikas, N.; Haider, J.; Alhotan, A.; Jahantigh, J.; Yates, J. Evaluation of equivalent flexural strength for complete removable dentures made of zirconia-impregnated PMMA nanocomposites. *Materials* **2020**, *13*, 2580. [CrossRef] [PubMed]
37. Wang, R.; Tao, J.; Yu, B.; Dai, L. Characterization of multiwalled carbon nanotube-polymethyl methacrylate composite resins as denture base materials. *J. Prosthet. Dent.* **2014**, *111*, 318–326. [CrossRef]
38. Horsch, L.; Kronsteiner, D.; Rammelsberg, P. Survival and complications of implant-supported cantilever fixed dental prostheses with zirconia and metal frameworks: A retrospective cohort study. *Clin. Implants Dent. Relat. Res.* **2022**, *24*, 621–629. [CrossRef]
39. Paratelli, A.; Perrone, G.; Ortega, R.; Gómez-Polo, M. Polyetheretherketone in Implant Prosthodontics: A Scoping Review. *Int. J. Prosthodont.* **2020**, *33*, 671–679. [CrossRef] [PubMed]
40. Tribst, J.P.M.; Campanelli de Moraes, D.; Melo de Matos, J.D.; Lopes, G.D.R.S.; Dal Piva, A.M.O.; Souto Borges, A.L.; Bottino, M.A.; Lanzotti, A.; Martorelli, M.; Ausiello, P. Influence of Framework Material and Posterior Implant Angulation in Full-Arch All-on-4 Implant-Supported Prosthesis Stress Concentration. *Dent. J.* **2022**, *10*, 12. [CrossRef] [PubMed]
41. Yu, W.; Li, X.; Ma, X.; Xu, X. Biomechanical analysis of inclined and cantilever design with different implant framework materials in mandibular complete-arch implant restorations. *J. Prosthet. Dent.* **2022**, *127*, 783.e1–783.e10. [CrossRef] [PubMed]
42. Ahmadi, A.; Dörsam, I.; Stark, H.; Hersey, S.; Bourauel, C.; Keilig, L. The all-on-4 concept in the maxilla-A biomechanical analysis involving high performance polymers. *J. Biomed. Mater. Res. B Appl. Biomater.* **2021**, *109*, 1698–1705. [CrossRef] [PubMed]
43. Choi, A.H.; Matinlinna, J.P.; Ben-Nissan, B. Finite element stress analysis of Ti-6Al-4V and partially stabilized zirconia dental implant during clenching. *Acta Odontol. Scand.* **2012**, *70*, 353–361. [CrossRef] [PubMed]
44. Altıparmak, N.; Polat, S.; Onat, S. Finite element analysis of the biomechanical effects of titanium and Cfr-peek additively manufactured subperiosteal jaw implant (AMSJI) on maxilla. *J. Stomatol. Oral Maxillofac. Surg.* **2023**, *124*, 101290. [CrossRef] [PubMed]
45. Hussein, M.O. Biomechanical Performance of PEEK and Graphene-Modified PMMA as Telescopic Removable Partial Denture Materials: A Nonlinear 3D Finite Element Analysis. *Int. J. Prosthodont.* **2022**, *35*, 793–800. [CrossRef] [PubMed]
46. Moncayo-Matute, F.P.; Vázquez-Silva, E.; Peña-Tapia, P.G.; Torres-Jara, P.B.; Moya-Loaiza, D.P.; Vilorio-Ávila, T.J. Finite Element Analysis of Patient-Specific 3D-Printed Cranial Implant Manufactured with PMMA and PEEK: A Mechanical Comparative Study. *Polymers* **2023**, *15*, 3620. [CrossRef]
47. Selva-Otaolaurruchi, E.J.; Fernández-Estevan, L.; Solá-Ruiz, M.F.; García-Sala-Bonmati, F.; Selva-Ribera, I.; Agustín-Panadero, R. Graphene-Doped Polymethyl Methacrylate (PMMA) as a New Restorative Material in Implant-Prosthetics: In Vitro Analysis of Resistance to Mechanical Fatigue. *J. Clin. Med.* **2023**, *12*, 1269. [CrossRef] [PubMed]
48. Unsal, G.S.; Yusufoglu, S.I. Finite Element Analysis of Endocrown and Post-and-Core Abutments for Removable Partial Dentures with Different Framework Materials. *Int. J. Prosthodont.* **2023**, *36*, 203–215. [CrossRef]
49. Wang, B.; Huang, M.; Dang, P.; Xie, J.; Zhang, X.; Yan, X. PEEK in Fixed Dental Prostheses: Application and Adhesion Improvement. *Polymers* **2022**, *14*, 2323. [CrossRef] [PubMed]
50. Peng, W.M.; Liu, Y.F.; Jiang, X.F.; Dong, X.T.; Jun, J.; Baur, D.A.; Xu, J.J.; Pan, H.; Xu, X. Bionic mechanical design and 3D printing of novel porous Ti6Al4V implants for biomedical applications. *J. Zhejiang Univ. Sci. B* **2019**, *20*, 647–659. [CrossRef] [PubMed]

Disclaimer/Publisher's Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

Article

Biomechanical Finite Element Analysis of Two Types of Short-Angled Implants Across Various Bone Classifications

Mario Ceddia ^{1,†}, Tea Romasco ^{2,3,†}, Nilton De Bortoli, Jr. ⁴, Bruno Freitas Mello ⁵, Adriano Piattelli ^{6,7}, Eitan Mijiritsky ^{8,9}, Natalia Di Pietro ^{2,3,*,‡} and Bartolomeo Trentadue ^{1,‡}

¹ Department of Mechanics, Mathematics and Management, Politecnico di Bari University, 70125 Bari, Italy; marioceddia1998@gmail.com (M.C.); bartolomeo.trentadue@poliba.it (B.T.)

² Department of Medical, Oral and Biotechnological Sciences, “G. d’Annunzio” University of Chieti-Pescara, 66100 Chieti, Italy; tea.romasco@unich.it

³ Center for Advanced Studies and Technologies (CAST), “G. d’Annunzio” University of Chieti-Pescara, 66100 Chieti, Italy

⁴ Department of Oral Implantology, Associação Paulista dos Cirurgiões Dentistas-APCD, São Bernardo do Campo 02011-000, Brazil; bortoliusp@gmail.com

⁵ Department of Periodontics and Implant Dentistry, University of Vale do Itajaí, Itajaí 88302-901, Brazil; brunofmellofloripa@gmail.com

⁶ School of Dentistry, Saint Camillus International University of Health and Medical Sciences, 00131 Rome, Italy; apiattelli51@gmail.com

⁷ Facultad de Medicina, UCAM Universidad Católica San Antonio de Murcia, 30107 Murcia, Spain

⁸ Department of Head and Neck Surgery and Maxillofacial Surgery, Tel-Aviv Sourasky Medical Center, School of Medicine, Tel-Aviv University, Tel Aviv 64239, Israel; mijiritsky@bezeqint.net

⁹ Goldschleger School of Dental Medicine, Faculty of Medicine, Tel-Aviv University, Tel Aviv 39040, Israel

* Correspondence: natalia.dipietro@unich.it

† These authors contributed equally to this work.

‡ These authors also contributed equally to this work.

Abstract: The aim of this finite element analysis (FEA) was to investigate the distribution of von Mises stress within dental implant components, as well as trabecular and cortical bone. The study considered various bone qualities that influence cortical thickness in contact with the implant, specifically examining cortical thicknesses of 0.5, 1.5, and 3 mm, corresponding to Bergkvist’s classifications IV, III, and II, respectively. A simplified 3D model of the bone was developed for the analysis. Two short implants were inserted into the model: one with a 30° inclined abutment (IA) and another positioned at a 30° angle featuring a straight abutment (II). A vertical force (120 N) was applied to the upper surface of the abutments. FEA software was employed to assess the stresses on the peri-implant tissues and the implants. The findings indicated that a reduction in cortical bone thickness results in an increase in stress within the cortical bone. For IA, the stresses recorded 32.56, 56.12, and 96.14 MPa for cortical thicknesses of 3, 1.5, and 0.5 mm, respectively. Conversely, II exhibited increased stresses across all bone qualities (52.32, 76.15, and 126.32 MPa for the same cortical thicknesses). It is advisable to avoid II in cases of poor bone quality and thin cortical due to the heightened risk of overload-induced bone resorption; however, it may be preferable to use IA in scenarios involving good bone quality and thicker cortical.

Keywords: finite element analysis (FEA); dental implants; dental stress analysis; short implants; angled implants; inclined abutments

1. Introduction

Teeth are firmly anchored within the alveolar bone, which possesses mechanical properties that enable it to withstand the forces generated during chewing [1–6]. The alveolar bone exhibits the capacity to adapt to mechanical stresses through a process known as remodeling. According to Wolff’s law, the bone responds adaptively to external forces [7].

Subjecting the bone to mechanical stimuli is essential for maintaining its structure and density. The loss of a tooth, however, results in the absence of stimulation to the residual bone, thereby causing a decrease in density and structural integrity, which can lead to a reduction in width, height, and thickness. This reduction in bone volume may result in atrophic ridges, complicating the use of traditional prosthetics [7,8].

Bone augmentation represents a widely utilized technique in dentistry, which may employ either osteoinductive or osteoconductive materials to promote bone regeneration and facilitate the placement of standard dental implants [9]. Short implants have emerged as a viable treatment option for patients with atrophic or resorbed bone, whether partially or fully edentulous. The definition of short implants varies among authors; some define them as implants measuring less than 10 mm in length, while others consider implants shorter than 7 mm to be classified as short [10–13]. In 2016, a European agreement established that short implants are classified as those measuring 8 mm or less, while ultra-short implants are categorized as those measuring less than 6 mm [14]. Numerous studies have documented the success and survival rates associated with short implants, which historically exhibited lower success rates [15–18]. However, advancements in surface modifications, the optimization of macro-geometry, and improvements in surgical techniques have significantly enhanced these success rates, resulting in their more frequent application by dental professionals.

The nature of the implant–abutment connection also plays a critical role in determining success rates; implants that utilize internal Morse cone connections demonstrate a more uniform stress distribution in comparison to those featuring external hexagonal connections [19–21]. The Morse cone connection facilitates cold welding through friction between the abutment and implant surfaces, which minimizes micro gaps and consequently reduces bacterial infiltration [22,23]. The integration of the conical connection with short implants has resulted in a balanced distribution of loads across the implant, abutment, and surrounding alveolar bone [24,25].

To achieve a high success rate, it is crucial to establish a favorable biomechanical environment that enables the effective distribution of masticatory loads to the surrounding tissues. In this regard, finite element analysis (FEA) serves as a valuable tool for simulating the behavior of the implanted device within the bone, allowing researchers to model dental implants and bone interactions with increased accuracy and efficiency, thus mitigating the challenges associated with *in vivo* testing [16,26–32]. Specifically, the assessment of von Mises stress distribution serves as a crucial indicator of the stability of dental implants and the integrity of surrounding bone, both of which directly influence the longevity and long-term success of implant procedures. Indeed, the biomechanics of the chewing process generate axial and bending forces that produce a series of moments transmitted to the dental implant and subsequently to the surrounding bone. In this regard, several factors are instrumental in influencing the distribution of bone stress, which is pivotal to ensuring stability. For instance, excessive stress can lead to bone resorption and potential implant failure [33]. According to Frost's theory [7], bone can initiate growth and remodeling processes in response to biomechanical demands based on the intensity and direction of applied stress. Moderate mechanical stress may stimulate bone remodeling, thereby enhancing bone density and strength, while also promoting Sema3A secretion and osteocyte proliferation. Conversely, excessive stress may induce pathological damage, as illustrated in Table 1 and discussed in a study by Uto et al. [34].

Table 1. Bone behavior according to Frost's theory [7].

Bone Stress	State of Bone
<1–2 MPa	Bone stops growing and resorbs
1–2 MPa	Bone remodeling may occur, but the bone mass remains unchanged
20–40 MPa	Bone remodeling occurs, and bone mass increases
40–60 MPa	Bone remodeling and bone mass significantly increase
>60 MPa	There is an overload with potential risks of resorption
120 MPa	The bone may break due to a fracture

The stress exerted on bone tissue is also influenced by the geometric characteristics of dental implants [35–37]. A FEA study indicated that an increase in the implant diameter significantly reduces stress, in contrast to modifications in the length of the implant [37]. Furthermore, research conducted by Baggi et al. [35] demonstrated that the length of the implant predominantly affects the distribution of stress within the trabecular bone, thereby enhancing its overall distribution. For instance, a study by Zupancic Cepic et al. [38] specifically explored the biomechanical interactions of short implants (measuring 4, 6, and 8 mm) in supporting fixed partial dentures within the posterior mandible. The findings revealed that splinted configurations exhibited superior stress distribution compared to non-splinted configurations. Additionally, the results indicated that oblique loading at 30° resulted in greater stress distribution than axial loading. This observation suggests that oblique forces may generate higher and broader stress patterns around implants, potentially increasing the risk of damage to both the surrounding bone tissue and the implants themselves.

It is crucial to acknowledge that the bone type significantly impacts implant stability and stress distribution. Azcarate-Velázquez et al. [36] conducted an FEA to assess the effect of bone quality on two different types of dental implants. Their analysis demonstrated that a decline in bone quality led to poorer stress distribution and the overloading of the cortical bone at the implant collar. Another FEA study by Liu et al. [39] illustrated that the application of short implants in atrophied edentulous mandibles considerably affects the stress distribution within the surrounding bone tissue, where peak stress values tended to increase as the bone quality diminished. The classification system established by Lekholm and Zarb in 1985 [40] served as the foundational framework for examining the impact of bone quality and cortical thickness in this context. This classification categorizes bone into four primary types, ranging from Type I, characterized by dense cortical bone with a substantial amount of trabecular bone, to Type IV, which consists chiefly of trabecular bone with minimal cortical bone. While Lekholm and Zarb's classification is centered on the four most clinically relevant bone categories, it may limit the study of more complex cases. To address this limitation, Misch [41] proposed an alternative classification in 1997, encompassing six categories that consider bone density and morphology, as well as the proportions of cortical and trabecular bone. This revised classification facilitates a more nuanced evaluation of bone quality, particularly in cases of diminished bone quality. Nevertheless, since short implants are primarily situated in medium- to high-quality bone, the classification established by Lekholm and Zarb remains applicable. It is essential to clarify that Lekholm and Zarb's classification does not directly account for trabecular bone density, as the radiographic techniques employed for classification, such as panoramic and periapical radiographs, do not provide detailed information regarding trabecular bone density [40]. While these radiographs can reveal the amount of bone present, they do not yield precise assessments of bone density, which necessitates advanced imaging techniques, such as computed tomography (CT) or micro-CT. Furthermore, Lekholm and Zarb's classification is more practical than that proposed by Misch, as it also incorporates the surgeon's tactile perception during the site preparation for implantation.

The implantation process itself is a significant factor influencing the distribution of stress within both the bone and the implants. Research conducted by Morita et al. [42] has demonstrated that the inclination of dental implants can lead to a reduction in stress levels, with decreases of 8.4% under a 90° load and 9.7% under a 60° load when compared to a vertical implant. On the other hand, the use of an inclined abutment resulted in a stress reduction of 15.7% under a 90° load and 30% under a 60° load relative to a vertical implant with a straight abutment.

Notably, the factors discussed thus far, including the quality of the bone, do not currently provide a definitive indication regarding the most suitable positioning system: either inclined implants with straight abutments (II) or straight implants with inclined abutments (IA). Conducting an FEA study comparing II and IA configurations holds significant clinical relevance within the field of implant dentistry, as it offers insights into

how variations in implant angulation and abutment alignment affect biomechanics, stability, and clinical outcomes. The application of II and IA is particularly relevant in clinical scenarios characterized by limited bone volume or unfavorable anatomical structures. Patients frequently encounter challenges such as inadequate posterior bone support due to resorption, which complicates the placement of straight implants without extensive bone grafting. The employment of II, IA, or short implants can enhance prosthetic support by leveraging existing bone while reducing the necessity for additional procedures that may elevate patient morbidity [43]. Additionally, tilted implants or abutments may enable longer prosthetic extensions without the corresponding cantilever risks associated with straight implants, thus improving both prosthetic support and functionality. This advantage is clinically significant, as it may lead to the more effective restoration of masticatory function for patients missing considerable posterior teeth. Clinical outcomes suggest that treatments utilizing angled implants, whether II or IA, contribute to enhanced patient satisfaction by mitigating perceived prosthetic instability—a prevalent concern among individuals relying on traditional denture solutions [44]. A systematic review revealed no significant difference in implant survival and success rates between strategies employing angled implants and conventional techniques over medium- to long-term follow-ups [44]. This finding implies that both strategies may yield clinically comparable results for patients facing anatomical challenges.

Therefore, the objective of the present finite element study was to investigate the bone stress distributions induced by two short implants, considering various cortical thicknesses (0.5 mm, 1.5 mm, and 3 mm) and utilizing the ANSYS Workbench 2023 FEA software (R23[®], Canonsburg, PA, USA). This study examined two implant configurations: II positioned at a 30° angle and 30° IA, both of which featured Morse cone connections. These specific configurations were selected for investigation due to the scarcity of studies in the literature that have simultaneously assessed the effects of implant inclination or abutment in relation to bone type for short implants. A limited number of studies have evaluated the impact of the inclination of short implants corresponding to the bone class for different implant configurations (IA and II). For instance, a study conducted by Anitua et al. [45] indicated that the inclination of the implant is a more significant determinant of bone stress than the implant length itself. Specifically, a 17° inclination doubled the von Mises equivalent stress within the bone tissue, while a 45° inclination resulted in a 1300% increase in stress. Therefore, an inclination angle of 30° was chosen for this study, as it serves as an intermediate position that allows for the observation of inclination effects without reaching extreme angles such as 45°, which could induce excessive stress or fail to represent common clinical scenarios [45].

The null hypothesis proposed in this study posits that neither the configuration type (IA or II) nor the bone type influences the stress distribution.

2. Materials and Methods

2.1. Modeling

The three-dimensional (3D) geometric models of the implants and associated components were procured from the implant company (Implacil De Bortoli, Sao Paulo, Brazil). These models were subsequently assembled utilizing computer-aided design (CAD) software (Autodesk Inventor 2023.1, San Francisco, CA, USA). Figure 1 represents an illustration detailing the short implants (Maestro) employed in this study.

The implants in question possessed a diameter of 3.5 mm and a length of 7 mm. These dimensions were selected as they aligned with the designation of “short” implants as classified by the manufacturer (Implacil De Bortoli, Sao Paulo, Brazil). Additionally, the implants were engineered with healing chambers incorporated into their macro-geometric design to facilitate and expedite osseointegration, reduce bone compression during the insertion process, and improve the quality of the newly established bone tissue. Both implants are equipped with an internal Morse cone connection, which is intended to enhance implant stability and strength while mitigating the risk of micromovement and potential implant

failure. Notably, the IA variant was designed with a Morse cone connection that includes a retention screw, whereas the II variant featured the standard implant–abutment Morse cone connection.

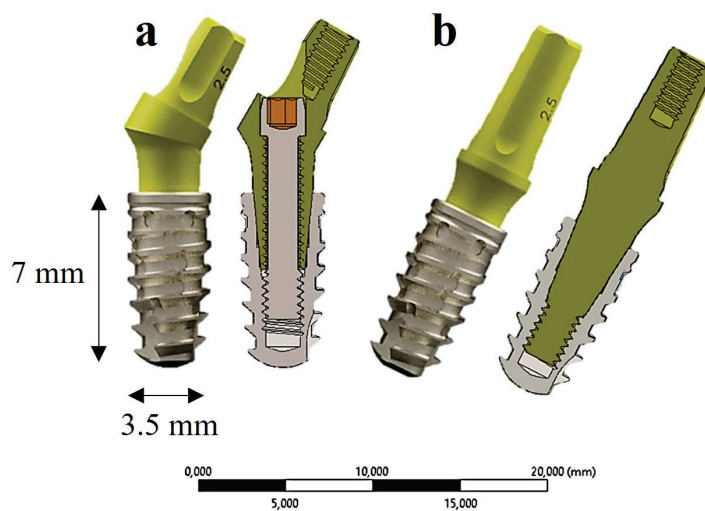


Figure 1. A three-dimensional (3D) model of the implants with abutments. (a) The implant with a 30° inclined abutment (IA); (b) the 30° inclined implant with a straight abutment (II).

Subsequently, the mandibular block model was constructed and designed according to the dimensions depicted in Figure 2.

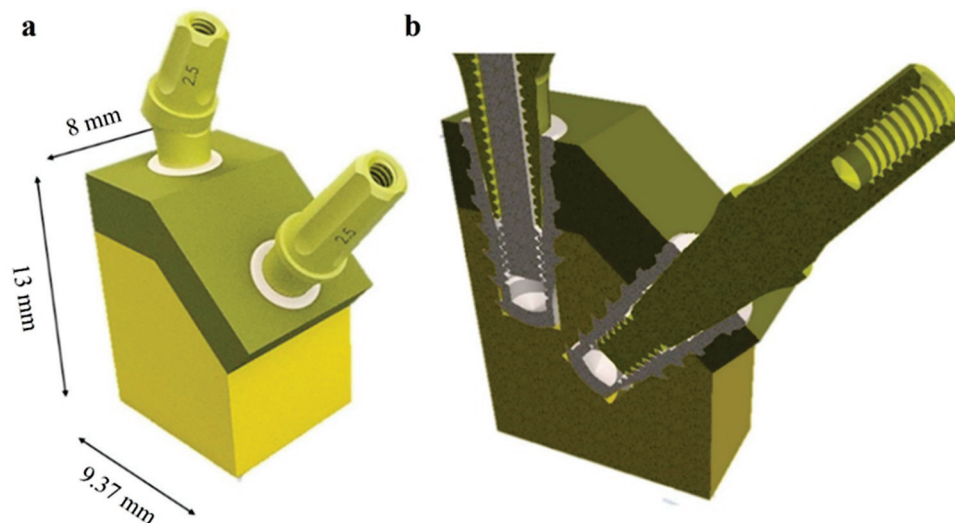


Figure 2. A 3D cortical and trabecular bone model. (a) The model with the inserted implants; (b) its sectional view.

The 3D bone model was generated utilizing the same CAD software (Autodesk Inventor 2023.1, San Francisco, CA, USA) based on a CT scan [25]. Then, this model was employed to subtract the geometric thread profile following the insertion of the implant in a crestal position, ensuring a precise fit between the implant and the bone. It was assumed that, upon the insertion of the implant into the bone block, all surrounding bone surfaces are in contact with the outer surface of the implant. This assumption supports the concept of ideal osseointegration, thereby simplifying the modeling process. Full osseointegration indicates that the implant is stabilized through bony attachment, facilitating optimal load transfer during functional loading. In clinical practice, achieving successful osseointegration represents a fundamental objective in implant therapy, thereby proving a logical framework for FEA simulations that model the best-case scenario, wherein the implant

functions as a unified entity with the bone. Utilizing simplified models is especially advantageous in exploratory studies, as it allows for an initial understanding of stress behavior without imposing excessive demands on computational resources.

2.2. Materials

A modification of the classification proposed by Lekholm and Zarb has been documented in previous research studies [46–48], confirming its validity through histomorphometric analysis, the measurement of bone mineral density (BMD), and micro-CT variables. For example, Pereira et al. [49] identified a correlation between the classification by Lekholm and Zarb and various histomorphometric parameters of bone.

Bergkvist et al. [50] calculated the BMD using Hounsfield Units (HU) obtained from micro-CT scans and observed a significant correlation between BMD and Lekholm and Zarb's classification [48,50]. This classification assesses bone density and cortical bone thickness measured in HU, categorizing bone into three distinct types based on the density and thickness of the cortical bone:

- Case 1 corresponds to a Type IV bone, characterized by bone density ranging from 400 to less than 500 HU, with a cortical bone thickness of 0.5 mm encasing the spongy bone (Figure 3a);
- Case 2 represents a Type III bone, exhibiting a bone density ranging from 550 to less than 850 HU and a cortical bone thickness of 1.5 mm surrounding the spongy bone (Figure 3b);
- Case 3 pertains to a Type II bone with a bone density ranging from 850 to less than 1000 HU, featuring a cortical bone thickness of 3 mm encasing the spongy bone (Figure 3c) [48,50].

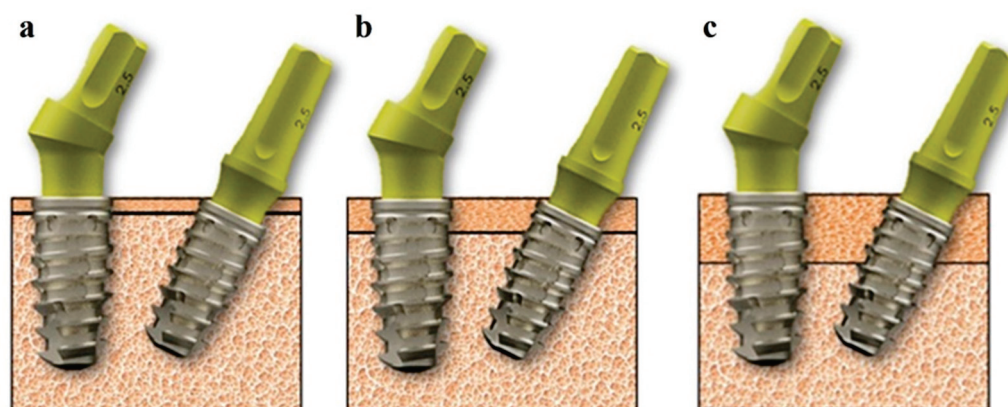


Figure 3. The cortical thickness for the three bone types studied with IA and II. (a) Type IV bone—cortical thickness: 0.5 mm; (b) Type III bone—cortical thickness: 1.5 mm; (c) Type II bone—cortical thickness: 3 mm.

Bone serves as a prime example of an anisotropic material characterized by its heterogeneous composition and intricate microarchitecture. The mechanical properties of bone, including stiffness and strength, exhibit significant variability along different axes, attributable to its complex arrangement of cortical and trabecular structures. This inherent anisotropic nature results in distinct responses during compression and tension testing [51]. Furthermore, deformation and stress outcomes are influenced by the direction of loading. Non-invasive techniques, such as the integration of the vibrational response with CT or magnetic resonance imaging (MRI) data, hold promise for the development of more realistic bone models [52,53]. Bones are classified as cortical or trabecular, with the primary distinction resting in their mechanical characteristics; cortical bone is denser than trabecular bone. This difference in density is critical for assessing the bone's capacity to support dental implants and to endure applied loads. Generally, denser bone correlates with greater stiffness and strength, which positively influences the stability of dental implants. Conversely,

trabecular bone, characterized by a higher volume of internal trabecular structures, exhibits reduced density and, subsequently, lower mechanical properties [51,54].

To achieve accurate numerical results in FEA studies, it is imperative to model the properties of both biological and non-biological structures with precision. For instance, O'Mahony et al. [55] conducted a comparative assessment of the mandible utilizing fully isotropic and transversely isotropic (anisotropic) models. Their findings indicated that the anisotropic model yielded higher stress levels (up to 20%) compared to the isotropic model, particularly at the crestal region. However, there is a notable lack of studies in the literature that have thoroughly evaluated the anisotropic characteristics of mandibular and maxillary bone [56]. Consequently, in most studies present in the literature, the materials are represented with uniform, isotropic, and linear properties to simplify computational models [25,57–59].

The isotropic bone model utilized in this study serves as a foundational reference for comparisons with more complex models. Young's modulus, as presented in Table 2, is a measure of the stiffness of a material and indicates the extent to which a material deforms under an applied load. A high value of Young's modulus signifies that the material is stiff and experiences minimal deformation under stress [60]. The relationship between stress and Young's modulus is articulated by Hooke's law, which asserts that, within elastic limits, stress is directly proportional to deformation. Therefore, a bone with a higher Young's modulus is capable of withstanding greater loads without significant deformation. According to Table 2, cortical bone exhibits a higher Young's modulus (13.7 GPa) in comparison to trabecular bone (1.37 GPa), reflecting its greater density and stiffness. As a result, cortical bone is better equipped to support higher loads and withstand increased stresses compared to trabecular bone [25,57].

Table 2. Mechanical properties of the materials used [25,57].

Component	Young's Modulus (GPa)	Poisson's Ratio
Abutment/implant (Ti6Al4V)	110	0.35
Cortical bone	13.7	0.3
Trabecular bone	1.37	0.3

2.3. Finite Element Analysis (FEA) Simulation

The 3D model CAD file, saved in .stp format, was imported into ANSYS Workbench 2023 FEA software (R23[®], Canonsburg, PA, USA). The model was primarily discretized using 10-node tetrahedral elements (SOLID 187), with a maximum element size set at 0.5 mm (Figure 4) [25].

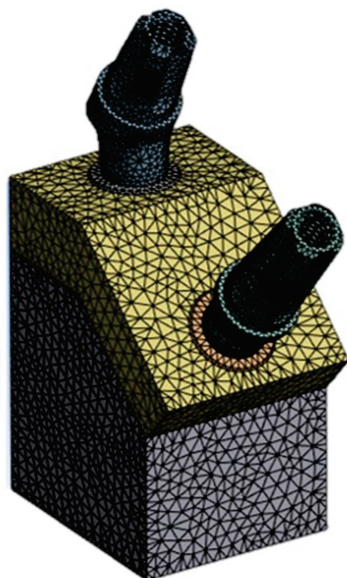


Figure 4. Schematic model of the mesh with IA and II configurations.

A sensitivity analysis was conducted to assess the influence of the mesh dimension on the results. Previous FEA studies have indicated that a mesh size of 0.5 mm yields reliable outcomes [25,57]. It has been determined that an element size of less than 0.3 mm is appropriate for accurately modeling the bone-implant interface [61]. Consequently, a mesh size of 0.3 mm was employed for the contact region between the implant and bone to enhance the estimation of the stress distribution.

2.4. Loads and Constraints

The constraints established required the immobilization of the bone block in all spatial directions (Figure 5).

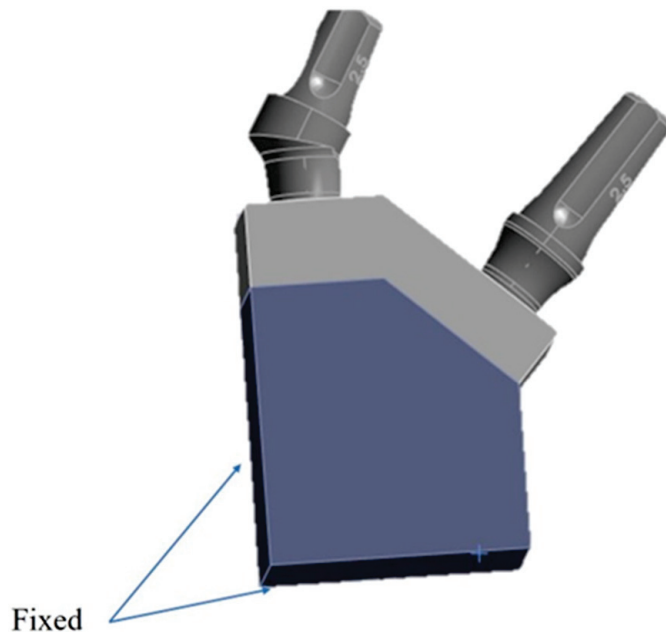


Figure 5. Constraint conditions.

The implant was securely affixed to the bone, eliminating any potential for relative movement. Furthermore, a frictional contact was implemented on the outer surface of the abutment, utilizing a coefficient of friction of 0.3. However, it is important to note that the connection between the abutment and implant did not account for micro-movements or material fatigue at the interface.

Molars are designed to withstand substantial vertical loads resulting from food compression and the activity of masticatory muscles during the process of chewing. Their morphology, characterized by prominent cusps and extensive masticatory surfaces, is optimized to accommodate these forces. Nonetheless, it is important to recognize that inclined forces, in addition to vertical ones, may also act on the molars, arising from lateral jaw movements or eccentric forces during the mastication of harder foods. Despite this, for the purposes of this study, the relatively low intensity of oblique loads, in comparison to vertical loads, was regarded as negligible [62]. A compressive force of 120 N was applied to the upper surface of the abutment along its long z-axis direction, simulating the forces generated during mastication on molars [42] (Figure 6).

The application of a vertical force induces a moment that generates mechanical stress in both cortical and cancellous bone, which can lead to deformations and an increased risk of fractures, particularly in regions with compromised bone quality. The primary distinctions between the IA and II configurations pertain to the fulcrum and the corresponding lever arm. In the case of the II, insertion is achieved by drilling a hole that is inclined at an angle of 30° relative to the surface of the alveolar ridge. The bending resulting from masticatory forces utilizes the apex of the implant as its fulcrum, thereby creating a lever arm that is greater than that of the IA, which operates with the fulcrum located at the Morse cone

connection. This enhanced lever arm may contribute to elevated bending moments and, consequently, increased stress on the implant.

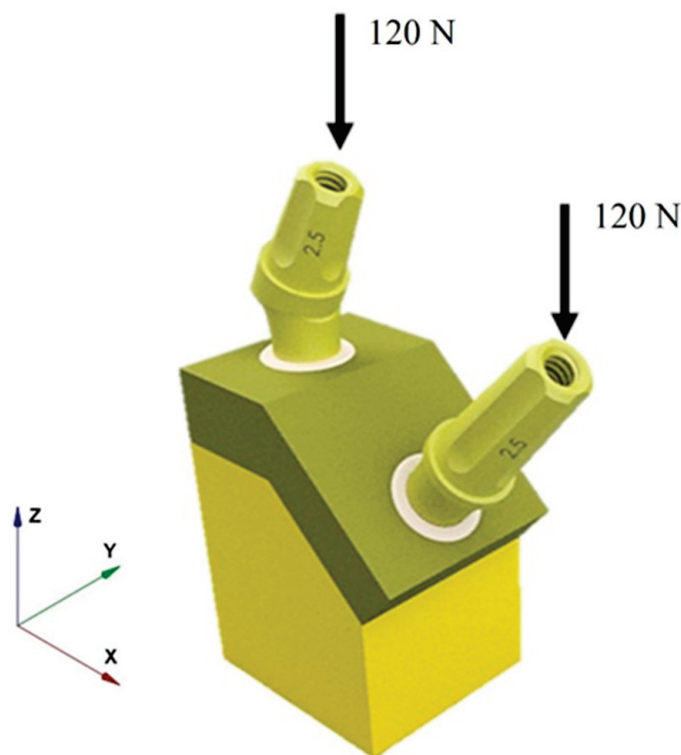


Figure 6. Application of a vertical load to the upper surface of IA and II configurations.

2.5. FEA Simulation

All models were analyzed on a Windows 10 64-bit system equipped with an Intel i7 processor and 16 GB of RAM. Following the analysis, the numerical results were converted into visual representations utilizing color maps. These color maps range from blue, indicating areas with the least stress, to red, which signifies areas of maximum stress. The assessment of stress levels was conducted using the von Mises criterion, which identifies the failure of a material when the equivalent stress exceeds its tensile strength. This criterion is instrumental in analyzing the transfer of loads from the implant to the surrounding bone. Previous FEA studies [63,64] have demonstrated that the stress concentration in particular regions can lead to bone resorption, thereby jeopardizing implant stability. Moreover, the distribution of von Mises stresses is influenced by factors such as the quality and thickness of the bone, whether cortical or trabecular [65]. Understanding these variations is essential for selecting the most suitable type of implant and surgical technique, which in turn promotes optimal bone integration and enhances overall health.

3. Results

3.1. Analysis of Stress on Implants

The research examined the mechanical performance of IA and II configurations in three distinct types of bone, each exhibiting varying cortical thicknesses, under a vertical load of 120 N (Figure 7).

As illustrated in Figure 7, the II configuration exhibited higher stress values compared to IA across all three bone types. Conversely, the IA configuration demonstrated elevated stress values solely at cortical thicknesses of 1.5 mm and 0.5 mm. For all implant types, the maximum stress occurred at the abutment neck, where the bending resulting from vertical chewing loads is more pronounced. Table 3 presents the numerical values of the stresses recorded at the neck of the implant for the three configurations of cortical bone thickness.

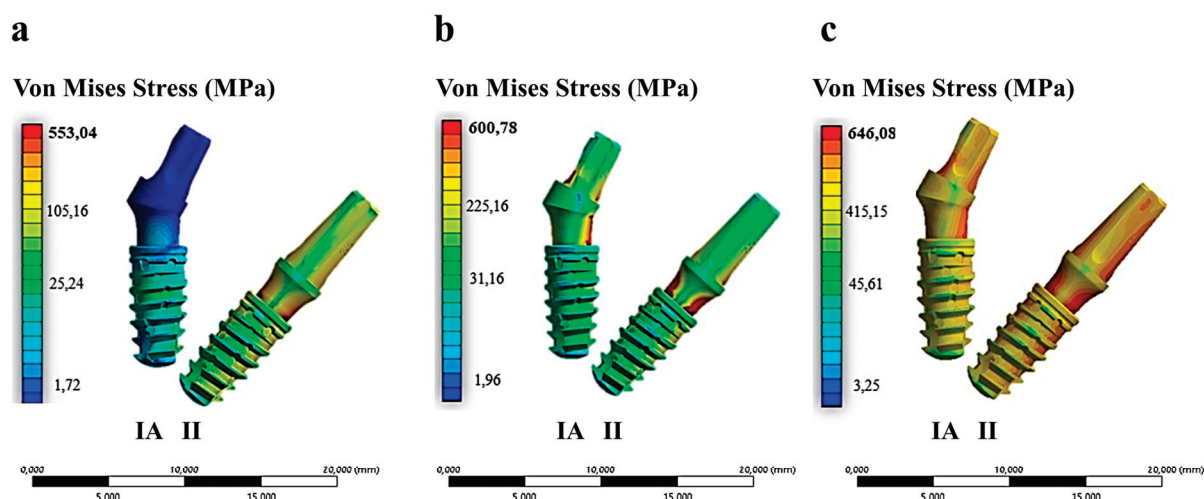


Figure 7. Von Mises stress distribution in implants placed in three different bone types and cortical bone thickness configurations. (a) Type II bone—3 mm; (b) Type III bone—1.5 mm; (c) Type IV bone—0.5 mm.

Table 3. Von Mises stress values in implants placed in three cortical bone thickness configurations evaluated at the implant neck.

Cortical Thickness	Von Mises Stress (MPa)	
	Inclined Implant with a Straight Abutment (II)	Implant with an Inclined Abutment (IA)
3 mm	533.04	65.12
1.5 mm	600.78	576.48
0.5 mm	646.70	602.11

3.2. Analysis of Stress on Bone with the Implant Featuring an Inclined Abutment (IA)

In Figure 8, the distribution of stress is presented for the implant exhibiting a 30° IA.

The decline in bone quality and the reduction in cortical thickness resulted in an increase in overall stress. It is evident that the stress on the implant is substantially higher than that on the surrounding cortical and trabecular bone, as illustrated in Figure 8. This phenomenon, referred to as “stress shielding”, is commonly observed in cases where a more rigid material is integrated into a less rigid substance, such as bone [66]. The rigidity of the implant, approximately 110 GPa, greatly exceeded that of cortical bone at 13.7 GPa and trabecular bone at 1.37 GPa. This disparity in rigidity results in the implant shielding the bone from stress distribution, thus creating a non-uniform force distribution between the implant and the bone. Furthermore, bone exhibits viscoelastic properties that enable it to absorb and dissipate certain stress levels. In contrast, the rigid implant tends to accumulate more stress, particularly in the neck region. While the higher stiffness of cortical bone (13.7 GPa versus 1.37 GPa for trabecular bone) protects the trabecular bone from excessive stresses, it leads to increased stress concentrations on the cortical bone. At a cortical thickness of 3 mm, the stress on the cortical bone surrounding the implant was measured at 32.56 MPa; at a 1.5 mm thickness, this stress increased to 56.12 MPa, and with a reduction to a 0.5 mm thickness, the stress tripled to approximately 96.14 MPa. This trend of heightened stress correlated with decreased cortical thickness was also prevalent in trabecular bone. For instance, at a thickness of 3 mm, the stress recorded was 16.21 MPa, which rose to 25.46 MPa at 1.5 mm, and at 0.5 mm, the maximum stress value reached 61.47 MPa.

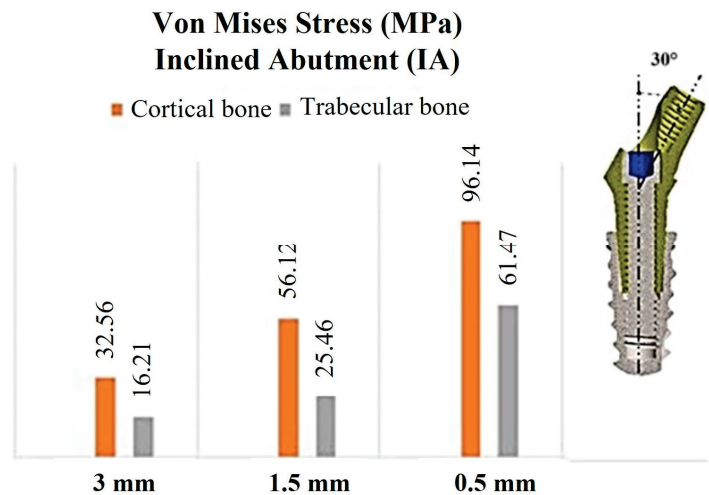


Figure 8. Von Mises stress values associated with the cortical and trabecular bones, in relation to cortical thicknesses of 3, 1.5, and 0.5 mm, with a 30° IA.

3.3. Analysis of Stress on Bone with the Inclined Implant (II)

The placement of the implant at an angle of 30° in the bone resulted in an increase in von Mises stress compared to the IA configuration (Figure 9).

For a cortical thickness of 3 mm, the stress exerted on the cortical bone was approximately 52.32 MPa. In contrast, with a thickness of 1.5 mm, the stress increased to 76.15 MPa, and at 0.5 mm of cortical bone, the stress reached critical levels of approximately 126.32 MPa. Given that the strength of cortical bone is estimated to be around 110 MPa [67], the latter scenario may precipitate significant concerns such as bone fracture.

Furthermore, a substantial difference in the stress distribution on the trabecular bone was observed when comparing the positioning of an IA to that of an II. The positioning of the II resulted in a more uniform stress distribution across the trabecular bone. Additionally, the von Mises stress values recorded were 23.15 MPa within the trabecular bone for a cortical thickness of 3 mm, 36.14 MPa for a cortical thickness of 1.5 mm, and 76.72 MPa for a cortical of 0.5 mm. Figure 10 demonstrates that, with the II placement, a greater volume of trabecular bone was subjected to stress, with peak stress observed around the apex of the implant.

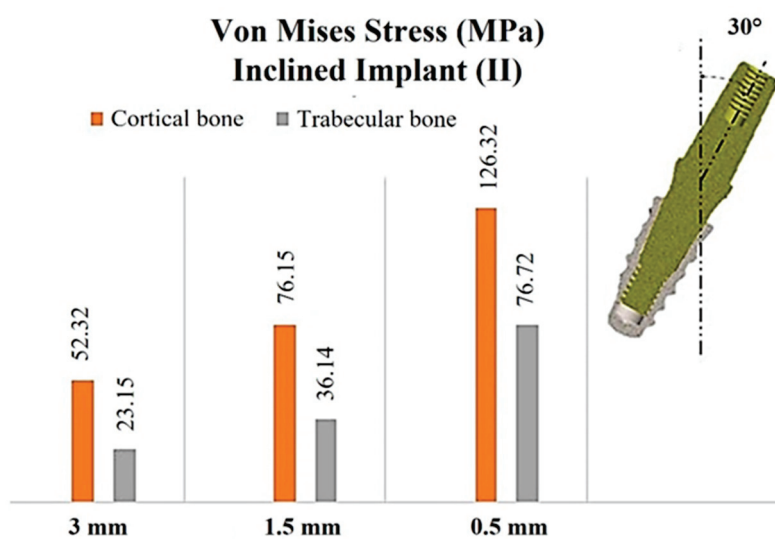


Figure 9. Von Mises stress values associated with the cortical and trabecular bones, in relation to cortical thicknesses of 3, 1.5, and 0.5 mm, with a 30° II.

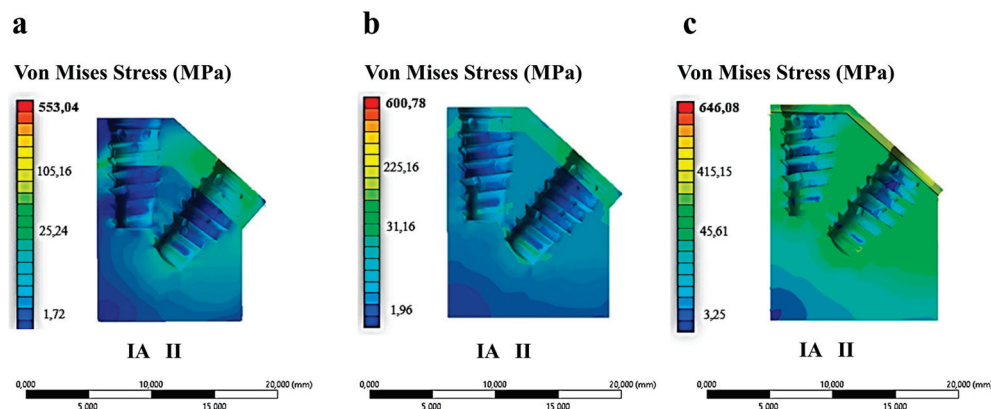


Figure 10. Von Mises stress distribution on IA and II as a function of cortical thickness: (a) 3 mm; (b) 1.5 mm; (c) 0.5 mm.

Figure 11 presents a comprehensive summary of the primary findings related to the distribution of stress in both cortical and trabecular bones.

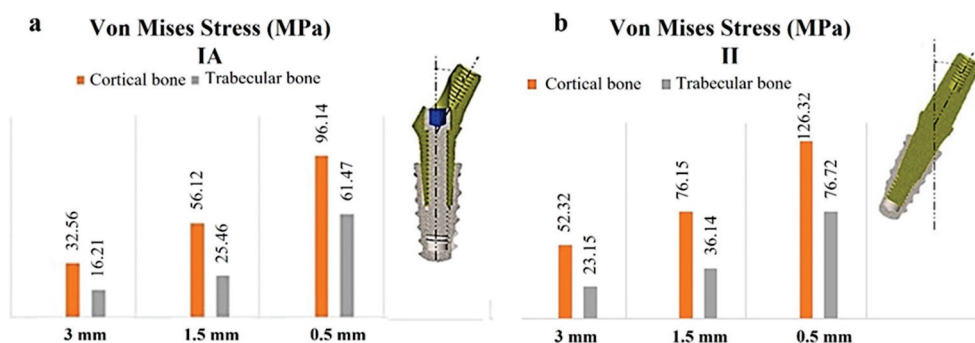


Figure 11. Von Mises stress values associated with cortical and trabecular bones in relation to cortical thickness. (a) IA; (b) II.

The left side of Figure 11 depicts an IA configuration, while the right side presents a configuration with an II.

The results derived from the FEA revealed several significant insights:

- A reduction in cortical thickness resulted in increased stress on both trabecular and cortical bone, particularly at the apex and neck regions, respectively;
- The II configuration exhibited higher stress levels compared to the IA; however, it also facilitated a more uniform stress distribution across the trabecular bone;
- The highest levels of stress were recorded within the cortical bone rather than the trabecular bone, especially for the II configuration at a cortical thickness of 0.5 mm, where stress reached 126.32 MPa. This value surpassed the maximum strength threshold of cortical bone (approximately 110 MPa), potentially leading to significant bone damage. In contrast, the stress for the IA configuration was measured at 96.14 MPa;
- Both implant insertion types did not present any complications for the implants, as the stresses encountered at the neck level remained below 900 MPa, which is acknowledged as the maximum strength capacity of titanium.

To mitigate the risk of bone resorption or fracture, the utilization of IA may be advisable, particularly in cases characterized by poor bone quality and thin cortical bone (approximately 0.5 mm). Conversely, when the quality of bone is deemed satisfactory and the cortical thickness exceeds 0.5 mm, an II can enhance the stress distribution across both cortical and trabecular bone, thereby decreasing the likelihood of bone resorption due to inappropriate bone stimulation. Additionally, the selection of the implant diameter is of paramount importance. An increase in the diameter by 0.5 mm corresponds to a 10–15%

enhancement in the functional surface area. This factor is especially relevant, as the crestal region experienced the highest levels of stress, indicating that the implant diameter plays a more significant role than the length in mitigating these stresses.

4. Discussion

The success of implant treatment is contingent upon the effective distribution of stresses to the supporting bone [68]. This distribution is influenced by several factors, including the implant geometry, bone type, type of loading, prosthetic design, as well as the healing and osseointegration processes [69]. Research has demonstrated that increasing the diameter of an implant diminishes the stress experienced by both the implant and the surrounding bone [24,35]. This reduction in stress is attributed to a more favorable distribution of forces across a larger functional area, which ultimately enhances stability [70–72]. Additionally, a larger implant diameter can mitigate bone loss by optimizing stress distribution at the implant collar [73]. Nonetheless, it is essential to acknowledge that excessively large diameters necessitate a substantial volume of bone, which may be in short supply due to bone atrophy, a prevalent condition resulting from tooth loss. The absence of a load on the teeth contributes to the resorption of bone in the mandible and maxilla [74]. This phenomenon was first described by Atwood in 1963 and was subsequently developed by Cawood and Howell in 1971, who categorized atrophy into six stages based on the quantity of residual bone available [75]. To address the challenges posed by bone atrophy and to facilitate the use of prostheses in affected patients, various techniques have been employed, including bone grafts, sinus lift, and guided bone regeneration, among others [76–78]. In cases of severe maxillary atrophy, short implants are frequently utilized when the quantity of available bone is limited. The application of short implants provides several advantages, such as the potential to circumvent more invasive and costly bone grafting procedures, a reduction in healing time, as well as a decrease in the risk of complications. Furthermore, these implants can be positioned in anatomically critical areas with constrained space, thereby stabilizing fixed or removable prostheses [18,79].

The type of bone surrounding dental implants is a crucial determinant of osseointegration, implant stability, and load resistance. Numerous studies have investigated the characteristics of bone and the various factors that may affect its quality and density [68,79]. For instance, Yang et al. [30] conducted a study that analyzed stress distribution on anatomically customized dental implants across different bone densities. The researchers modeled four types of maxillary bone utilizing Lekholm and Zarb's bone classification system [40]. Their findings underscored the significant influence of bone density on stress distribution at the surface of dental implants, revealing diminished density corresponding with elevated bone stresses.

Additionally, an FEA study by Shabanpour Kasari et al. [65] explored how varying levels of osseointegration influence the biomechanical response of the bone surrounding dental implants and stress distribution. This investigation highlighted the effects of osseointegration on stress values in both cortical and trabecular bone around dental implants. The results suggested that improved osseointegration enhances implant stability and diminishes bone deformation, thereby reducing the risk of implant failure due to excessive loading. Furthermore, the study emphasized the importance of bone quality and cortical thickness in the response of the short implant and the associated risk of bone resorption, particularly in low-quality bone regions, such as the posterior mandible.

Moreover, Liu et al. [39] examined the effects of the implant inclination on the stress distribution and support structures, highlighting the significance of proper implant and abutment configurations in minimizing stress concentration and ensuring uniform stress distribution. Indeed, the utilization of IA is recommended in scenarios involving poor bone quality and diminished cortical thickness to mitigate bone stress. Conversely, II and IA are advisable for facilitating better stress distribution in high-quality bone with a thick cortical layer. Notably, when subjected to oblique loads, II may lead to stress concentrations at critical locations, resulting in higher maximum stress values compared

to straight implants [39]. This phenomenon may compromise the durability and strength of support structures and elevate the risk of long-term implant failure. Certain studies indicated that stress in the platform of a tilted implant may increase by as much as five times compared to that observed in a straight implant [65,79–81]. Additional investigations have also addressed the influence of implant inclination on cantilever use. Guven et al. [82] discovered that employing tilted implants in conjunction with short cantilevers can decrease stress on cortical bone by minimizing the moment of force generated, thereby relieving stress on adjacent bone tissues.

However, in a numerical study [42], the application of IA was found to be more effective than II in reducing stress concentration. This increased efficacy arises because the head of the IA is situated closer to the occlusal surface, which serves as the primary loading point. This proximity facilitates a more equitable distribution of stress during loading and diminishes the concentration in specific areas.

Other variables may also influence the stress distribution within the implant and surrounding bone. For instance, a study conducted by Yao et al. [83] assessed the maximum values of the micro-gap between the implant and the abutment, which can serve as an indicator of the stability of the connection during loading. Specifically, a larger micro-gap may result in undesirable micromovements, negatively impacting the health of surrounding tissues and the longevity of the implant by creating localized stress concentration points where material failure is more likely to occur. Furthermore, the bending and torsional forces characteristic of mastication were shown to contribute to a deterioration in abutment stiffness, leading to progressive damage and an increased long-term risk of failure. In the present study, the application of the finite element method facilitated an examination of the mechanical response associated with two short implants positioned in two specific configurations (IA and II, both at an angle of 30°). This investigation encompassed varying cortical bone thicknesses and different bone types.

It is essential to recognize that bone is a dynamic tissue, adapting its structure in response to external stresses. Numerous studies have established two critical stress thresholds pertinent to bone: one associated with resorption and another linked to fracture [84,85]. The findings from this study indicated that the II configuration with a cortical thickness of 0.5 mm exhibited significant stress levels on the cortical bone, measuring 126.32 MPa. Consequently, the risk of bone fracture under these conditions is considerably greater when compared to a cortical thickness of 1.5 mm (76.15 MPa) or to another thickness of 0.5 mm (52.32 MPa). Thus, the null hypothesis posited in the previous analysis may be appropriately rejected.

Therefore, based on the outcomes of this research, the utilization of a 30° IA configuration is recommended in scenarios characterized by very thin cortical bone. Clinically, when bone quality is assessed to be high and the cortical thickness exceeds 0.5 mm, the placement of a 30° II configuration is advisable. Conversely, the IA configuration serves as a suitable solution across all categories of the bone class, as no stress values exceeding 120 MPa were recorded on the bone, while the stress on the implant itself measured 602.12 MPa, remaining below the strength threshold of the titanium alloy.

The results obtained, consistent with the existing literature, suggested that the implementation of straight implants and IA configurations is more advantageous in situations where bone quality is deficient. This is attributed to the reduced bone stress associated with this configuration compared to the II placement. This phenomenon can be elucidated by the hollow structure of the IA designed to accommodate the retention screw, which enhances stress dispersion. This characteristic effectively mitigates the stress concentration, promoting a more uniform distribution of stress throughout the connection and preventing areas of overload at the implant neck that could lead to bone resorption.

Concerning the limitations of this study, it is important to acknowledge that finite element models serve as a simplification of the actual structures they represent. Consistent with previous literature, this study assumed that bone is homogeneous and isotropic to facilitate the modeling process. Such assumptions, however, may lead to considerable

underestimations of stress distributions, thereby potentially compromising the accuracy and reliability of clinical outcomes [58,59]. Furthermore, a fixed contact between the bone and implant was assumed to simplify the analysis; yet, from a clinical perspective, achieving perfect osseointegration is implausible due to the inherent micromovements present at the implant interface [86]. Nonetheless, as delineated in a study by Kulkarni et al. [87], enhanced osseointegration is associated with increased implant stability and minimized bone deformation. The rationale for this assumption is derived from the clinical focus on attaining optimal mechanical stability and effective load transfer from the implant to the adjacent bone. Regarding load applications, it should be noted that these forces are time-varying and incorporate oblique components. For the purposes of this study, the oblique force components were neglected, as they are deemed to be less significant than the masticatory loads placed on molars. Additionally, the application of the load was simplified by directing force to the abutment, without accounting for the intricacies of the prosthetic crown. The failure to consider micro-movements and material fatigue at the abutment-implant interface may also represent a limitation, given that these factors can influence long-term stress distribution and overall implant success. A clinical investigation [83] has examined various phenomena pertinent to dental implant connections, including micro-movements and material fatigue. Such micro-movements at the implant interface could induce increased material wear and fatigue, thereby heightening the risk of mechanical failure and the subsequent loss of osseointegration, which could adversely affect the functionality of the implant. Nonetheless, tapered designs have demonstrated greater resistance to fatigue due to their capability to distribute forces evenly and mitigate micro-movements [88]. However, it is critical to note that should fatigue exceed a certain threshold, it may still lead to the fracture or failure of the connection, ultimately rendering the implant ineffective.

Future research will aim to evaluate the impact of these limitations on numerical results to enhance the modeling accuracy, alongside the integration of necessary *in vitro* and clinical trials. Future investigations should specifically assess the effects of modeled bone anisotropy by incorporating mechanical properties derived from patient histomorphometric data. Additionally, both oblique and time-varying loads should be applied based on data collected directly from patients, thereby augmenting the proposed numerical analysis and facilitating validation through forthcoming *in vitro* and *in vivo* mechanical testing. By utilizing this information, clinicians may be better equipped to select optimal implant placement strategies based on anatomical variables and clinical considerations.

5. Conclusions

In conclusion, the results of this study indicated that enhancing bone quality and cortical thickness is associated with a reduction in stress within the bone structure, provided that the system is not overloaded. Furthermore, the utilization of an IA, as opposed to employing an II, demonstrated a more significant reduction in stress, particularly in cortical bones with a thickness of 0.5 mm. As previously discussed, the fulcrum of bending generated by masticatory forces in the IA configuration was located at the Morse cone connection. This geometric arrangement facilitated an optimal load distribution between the implant and the surrounding bone. In contrast, with the II configuration, the fulcrum was positioned at the apex of the implant, which may lead to increased deflection in response to external forces.

Therefore, in cases where bone quality is compromised and cortical thickness is diminished during implant placement, it is advisable to favor an IA configuration. This methodology alleviates stress on the cortical bone, thereby reducing the likelihood of bone resorption due to overload. A comprehensive understanding of stress distribution within bone structures can assist clinicians in selecting the most appropriate implant placements, taking into account anatomical variables and clinical factors. Further validation through *in vitro* and *in vivo* testing is essential to enhance the precision of the numerical analysis conducted in this study.

Author Contributions: Conceptualization, B.T., M.C., N.D.B.J., B.F.M. and A.P.; methodology, M.C., T.R. and E.M.; software, M.C. and B.T.; validation, T.R., E.M., N.D.P. and B.T.; formal analysis, M.C. and T.R.; investigation, M.C., T.R., N.D.B.J. and B.F.M.; resources, B.T.; data curation, M.C., T.R. and N.D.P.; writing—original draft preparation, M.C., T.R. and A.P.; writing—review and editing, T.R. and N.D.P.; visualization, T.R. and N.D.P.; supervision, N.D.P. and B.T.; project administration, A.P., B.T. and N.D.P.; funding acquisition, N.D.P. and B.T. 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 original contributions presented in the study are included in the article and further inquiries can be directed to the corresponding author.

Acknowledgments: The authors express their gratitude to Implacil De Bortoli company (Sao Paulo, Brazil) for generously providing the mathematical data regarding the implants used in this study.

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

References

1. Carlsson, G.E.; Omar, R. The future of complete dentures in oral rehabilitation. A critical review. *J. Oral Rehabil.* **2010**, *37*, 143–156. [CrossRef] [PubMed]
2. Emami, E.; de Souza, R.F.; Kabawat, M.; Feine, J.S. The impact of edentulism on oral and general health. *Int. J. Dent.* **2013**, *2013*, 498305. [CrossRef] [PubMed]
3. Polzer, I.; Schimmel, M.; Müller, F.; Biffar, R. Edentulism as part of the general health problems of elderly adults. *Int. Dent. J.* **2010**, *60*, 143–155. [PubMed]
4. Holm-Pedersen, P.; Schultz-Larsen, K.; Christiansen, N.; Avlund, K. Tooth loss and subsequent disability and mortality in old age. *J. Am. Geriatr. Soc.* **2008**, *56*, 429–435. [CrossRef]
5. Kassebaum, N.J.; Bernabé, E.; Dahiya, M.; Bhandari, B.; Murray, C.J.; Marcenes, W. Global Burden of Severe Tooth Loss: A Systematic Review and Meta-analysis. *J. Dent. Res.* **2014**, *93* (Suppl. S7), 20S–28S. [CrossRef]
6. Kumakami-Sakano, M.; Otsu, K.; Fujiwara, N.; Harada, H. Regulatory mechanisms of Hertwig’s epithelial root sheath formation and anomaly correlated with root length. *Exp. Cell Res.* **2014**, *325*, 78–82. [CrossRef]
7. Frost, H.M. A 2003 update of bone physiology and Wolff’s Law for clinicians. *Angle Orthod.* **2004**, *74*, 3–15. [CrossRef]
8. Esposito, M.; Grusovin, M.G.; Maghaireh, H.; Worthington, H.V. Interventions for replacing missing teeth: Different times for loading dental implants. *Cochrane Database Syst. Rev.* **2013**, *2013*, CD003878. [CrossRef]
9. Tsuchida, S.; Nakayama, T. Recent Clinical Treatment and Basic Research on the Alveolar Bone. *Biomedicines* **2023**, *11*, 843. [CrossRef]
10. Weng, D.; Jacobson, Z.; Tarnow, D.; Hürzeler, M.B.; Faehn, O.; Sanavi, F.; Barkvoll, P.; Stach, R.M. A prospective multicenter clinical trial of 3i machined-surface implants: Results after 6 years of follow-up. *Int. J. Oral Maxillofac. Implant.* **2003**, *18*, 417–423. [PubMed]
11. Tawil, G.; Younan, R. Clinical evaluation of short, machined-surface implants followed for 12 to 92 months. *Int. J. Oral Maxillofac. Implant.* **2003**, *18*, 894–901. [CrossRef] [PubMed]
12. Testori, T.; Wiseman, L.; Woolfe, S.; Porter, S.S. A prospective multicenter clinical study of the Osseotite implant: Four-year interim report. *Int. J. Oral Maxillofac. Implant.* **2001**, *16*, 193–200. [PubMed]
13. Friberg, B.; Gröndahl, K.; Lekholm, U.; Brånemark, P.I. Long-term follow-up of severely atrophic edentulous mandibles reconstructed with short Brånemark implants. *Clin. Implant. Dent. Relat. Res.* **2000**, *2*, 184–189. [CrossRef]
14. Lombardo, G.; Marincola, M.; Signoriello, A.; Corrocher, G.; Nocini, P.F. Single-Crown, Short and Ultra-Short Implants, in Association with Simultaneous Internal Sinus Lift in the Atrophic Posterior Maxilla: A Three-Year Retrospective Study. *Materials* **2020**, *13*, 2208. [CrossRef] [PubMed]
15. Nisand, D.; Picard, N.; Rocchietta, I. Short implants compared to implants in vertically augmented bone: A systematic review. *Clin. Oral Implant. Res.* **2015**, *26* (Suppl. S11), 170–179. [CrossRef]
16. Vidya Bhat, S.; Premkumar, P.; Kamalakanth Shenoy, K. Stress Distribution Around Single Short Dental Implants: A Finite Element Study. *J. Indian Prosthodont. Soc.* **2014**, *14* (Suppl. S1), 161–167. [CrossRef]
17. Badaró, M.M.; Mendoza Marin, D.O.; Pauletto, P.; Simek Vega Gonçalves, T.M.; Porporatti, A.L.; De Luca Canto, G. Failures in Single Extra-Short Implants (≤ 6 mm): A Systematic Review and Meta-analysis. *Int. J. Oral Maxillofac. Implant.* **2021**, *36*, 669–689. [CrossRef]
18. Comuzzi, L.; Tumedei, M.; Petrini, M.; Romasco, T.; Lorusso, F.; De Angelis, F.; Piattelli, A.; Tatullo, M.; Di Pietro, N. Clinical and Radiological Evaluation of a Self-Condensing Bone Implant in One-Stage Sinus Augmentation: A 3-Year Follow-Up Retrospective Study. *Int. J. Environ. Res. Public Health* **2023**, *20*, 2583. [CrossRef] [PubMed] [PubMed Central]

19. de Faria Almeida, D.A.; Pellizzer, E.P.; Verri, F.R.; Santiago, J.F., Jr.; de Carvalho, P.S. Influence of tapered and external hexagon connections on bone stresses around tilted dental implants: Three-dimensional finite element method with statistical analysis. *J. Periodontol.* **2014**, *85*, 261–269. [CrossRef]
20. Santiago Junior, J.F.; Verri, F.R.; Almeida, D.A.; de Souza Batista, V.E.; Lemos, C.A.; Pellizzer, E.P. Finite element analysis on influence of implant surface treatments, connection and bone types. *Mater. Sci. Eng. C Mater. Biol. Appl.* **2016**, *63*, 292–300. [CrossRef]
21. Sheridan, R.A.; Decker, A.M.; Plonka, A.B.; Wang, H.L. The Role of Occlusion in Implant Therapy: A Comprehensive Updated Review. *Implant. Dent.* **2016**, *25*, 829–838. [CrossRef] [PubMed]
22. Castro, C.G.; Zancopé, K.; Veríssimo, C.; Soares, C.J.; Neves, F.D. Strain analysis of different diameter Morse taper implants under overloading compressive conditions. *Braz. Oral Res.* **2015**, *29*, S1806–S83242015000100225. [CrossRef]
23. Lemos, C.A.A.; Verri, F.R.; Santiago, J.F.; Júnior Almeida, D.A.F.; Batista, V.E.S.; Noritomi, P.Y.; Pellizzer, D.P. Retention System and Splinting on Morse Taper Implants in the Posterior Maxilla by 3D Finite Element Analysis. *Braz. Dent. J.* **2018**, *29*, 30–35. [CrossRef] [PubMed]
24. Anitua, E.; Tapia, R.; Luzuriaga, F.; Orive, G. Influence of implant length, diameter, and geometry on stress distribution: A finite element analysis. *Int. J. Periodontics Restor. Dent.* **2010**, *30*, 89–95. [PubMed]
25. Di Pietro, N.; Ceddia, M.; Romasco, T.; De Bortoli Junior, N.; Mello, B.F.; Tumedei, M.; Specchiulli, A.; Piattelli, A.; Trentadue, B. Finite Element Analysis (FEA) of the Stress and Strain Distribution in Cone-Morse Implant–Abutment Connection Implants Placed Equicrestally and Subcrestally. *Appl. Sci.* **2023**, *13*, 8147. [CrossRef]
26. Ceddia, M.; Solarino, G.; Tucci, M.; Lamberti, L.; Trentadue, B. Stress Analysis of Tibial Bone Using Three Different Materials for Bone Fixation Plates. *J. Compos. Sci.* **2024**, *8*, 334. [CrossRef]
27. Ceddia, M.; Solarino, G.; Cassano, G.D.; Trentadue, B. Finite Element Study on Stability in the Femoral Neck and Head Connection to Varying Geometric Parameters with the Relates Implications on the Effect of Wear. *J. Compos. Sci.* **2023**, *7*, 387. [CrossRef]
28. Ceddia, M.; Trentadue, B.; De Giosa, G.; Solarino, G. Topology Optimization of a Femoral Stem in Titanium and Carbon to Reduce Stress Shielding with the FEM Method. *J. Compos. Sci.* **2023**, *7*, 298. [CrossRef]
29. Comuzzi, L.; Ceddia, M.; Di Pietro, N.; Inchingolo, F.; Inchingolo, A.M.; Romasco, T.; Tumedei, M.; Specchiulli, A.; Piattelli, A.; Trentadue, B. Crestal and Subcrestal Placement of Morse Cone Implant–Abutment Connection Implants: An In Vitro Finite Element Analysis (FEA) Study. *Biomedicines* **2023**, *11*, 3077. [CrossRef]
30. Yang, Y.; Liu, Y.; Yuan, X.; Ren, M.; Chen, X.; Luo, L.; Zheng, L.; Liu, Y. Three-dimensional finite element analysis of stress distribution on short implants with different bone conditions and osseointegration rates. *BMC Oral Health* **2023**, *23*, 220. [CrossRef]
31. Chang, Y.; Tambe, A.A.; Maeda, Y.; Wada, M.; Gonda, T. Finite element analysis of dental implants with validation: To what extent can we expect the model to predict biological phenomena? A literature review and proposal for classification of a validation process. *Int. J. Implant. Dent.* **2018**, *4*, 7. [CrossRef] [PubMed]
32. Brown, A.; Zaky, S.; Ray, H., Jr.; Sfeir, C. Porous magnesium/PLGA composite scaffolds for enhanced bone regeneration following tooth extraction. *Acta Biomater.* **2015**, *11*, 543–553. [CrossRef]
33. Zhang, C.; Zeng, C.; Wang, Z.; Zeng, T.; Wang, Y. Optimization of stress distribution of bone-implant interface (BII). *Biomater. Adv.* **2023**, *147*, 213342. [CrossRef] [PubMed]
34. Uto, Y.; Kuroshima, S.; Nakano, T.; Ishimoto, T.; Inaba, N.; Uchida, Y.; Sawase, T. Effects of mechanical repetitive load on bone quality around implants in rat maxillae. *PLoS ONE* **2017**, *12*, e0189893. [CrossRef] [PubMed]
35. Baggi, L.; Cappelloni, I.; Di Girolamo, M.; Maceri, F.; Vairo, G. The influence of implant diameter and length on stress distribution of osseointegrated implants related to crestal bone geometry: A three-dimensional finite element analysis. *J. Prosthet. Dent.* **2008**, *100*, 422–431. [CrossRef]
36. Azcarate-Velázquez, F.; Castillo-Oyagüe, R.; Oliveros-López, L.G.; Torres-Lagares, D.; Martínez-González, Á.J.; Pérez-Velasco, A.; Lynch, C.D.; Gutiérrez-Pérez, J.L.; Serrera-Figallo, M.Á. Influence of bone quality on the mechanical interaction between implant and bone: A finite element analysis. *J. Dent.* **2019**, *88*, 103161. [CrossRef]
37. García-Braz, S.H.; Prados-Privado, M.; Zanatta, L.C.S.; Calvo-Guirado, J.L.; Prados-Frutos, J.C.; Gehrke, S.A. A Finite Element Analysis to Compare Stress Distribution on Extra-Short Implants with Two Different Internal Connections. *J. Clin. Med.* **2019**, *8*, 1103. [CrossRef]
38. Zupancic Cepic, L.; Frank, M.; Reisinger, A.; Pahr, D.; Zechner, W.; Schedle, A. Biomechanical finite element analysis of short-implant-supported, 3-unit, fixed CAD/CAM prostheses in the posterior mandible. *Int. J. Implant. Dent.* **2022**, *8*, 8. [CrossRef]
39. Liu, C.; Xing, Y.; Li, Y.; Lin, Y.; Xu, J.; Wu, D. Bone quality effect on short implants in the edentulous mandible: A finite element study. *BMC Oral Health* **2022**, *22*, 139. [CrossRef]
40. Oliveira, M.R.; Gonçalves, A.; Gabrielli, M.A.C.; de Andrade, C.R.; Vieira, E.H.; Pereira-Filho, V.A. Evaluation of Alveolar Bone Quality: Correlation Between Histomorphometric Analysis and Lekholm and Zarb Classification. *J. Craniofacial Surg.* **2021**, *32*, 2114–2118. [CrossRef]
41. Misch, C.E.; Judy, K.W. Classification of partially edentulous arches for implant dentistry. *Int. J. Oral Implantol.* **1987**, *4*, 7–13. [PubMed]

42. Morita, J.; Wada, M.; Mameno, T.; Maeda, Y.; Ikebe, K. Ideal placement of an implant considering the positional relationship to an opposing tooth in the first molar region: A three-dimensional finite element analysis. *Int. J. Implant. Dent.* **2020**, *6*, 31. [CrossRef] [PubMed]
43. DeVico, G.; Bonino, M.; Spinelli, D.; Schiavetti, R.; Sannino, G.; Pozzi, A.; Ottria, L. Rationale for tilted implants: FEA considerations and clinical reports. *Oral Implant.* **2011**, *4*, 23–33.
44. Cortés-Bretón Brinkmann, J.; García-Gil, I.; Pedregal, P.; Peláez, J.; Prados-Frutos, J.C.; Suárez, M.J. Long-Term Clinical Behavior and Complications of Intentionally Tilted Dental Implants Compared with Straight Implants Supporting Fixed Restorations: A Systematic Review and Meta-Analysis. *Biology* **2021**, *10*, 509. [CrossRef]
45. Anitua, E.; Larrazabal Saez de Ibarra, N.; Morales Martín, I.; Saracho Rotaache, L. Influence of Implant Tilting and Length on the Biomechanics of Single-Tooth Restoration: A Finite Element Analysis in Atrophic Mandible. *Dent. J.* **2022**, *10*, 77. [CrossRef]
46. Anitua, E. The biomechanics of the short implant. In *Short and Extra-Short Implants*; Anitua, E., Ed.; Team Work Media España: Vitoria, Spain, 2017; p. 57.
47. Anitua, E.; Alkhraisat, M.H.; Piñas, L.; Orive, G. Efficacy of biologically guided implant site preparation to obtain adequate primary implant stability. *Ann. Anat.* **2015**, *199*, 9–15. [CrossRef]
48. Lekholm, U.; Zarb, G.A.; Albrektsson, T. Patient selection and preparation. In *Tissue Integrated Prostheses*; Quintessence Publishing Co., Inc.: Chicago, IL, USA, 1985; pp. 199–209.
49. Pereira, A.C.; Souza, P.P.; Souza, J.A.; Silva, T.A.; Batista, A.C.; Ribeiro-Rotta, R.F. Histomorphometrical and molecular evaluation of endosseous dental implants sites in humans: Correlation with clinical and radiographic aspects. *Clin. Oral Implant. Res.* **2013**, *24*, 414–421. [CrossRef]
50. Bergkvist, G.; Koh, K.J.; Sahlholm, S.; Klintström, E.; Lindh, C. Bone density at implant sites and its relationship to assessment of bone quality and treatment outcome. *Int. J. Oral Maxillofac. Implant.* **2010**, *25*, 321–328. [PubMed]
51. van Eijden, T.M. Biomechanics of the mandible. *Crit. Rev. Oral Biol. Med.* **2000**, *11*, 123–136. [CrossRef]
52. Camacho, N.P.; Rinnerthaler, S.; Paschalis, E.P.; Mendelsohn, R.; Boskey, A.L.; Fratzl, P. Complementary information on bone ultrastructure from scanning small angle X-ray scattering and Fourier-transform infrared microspectroscopy. *Bone* **1999**, *25*, 287–293. [CrossRef]
53. Mastrogiacono, M.; Papadimitropoulos, A.; Cedola, A.; Peyrin, F.; Giannoni, P.; Pearce, S.G.; Alini, M.; Giannini, C.; Guagliardi, A.; Cancedda, R. Engineering of bone using bone marrow stromal cells and a silicon-stabilized tricalcium phosphate bioceramic: Evidence for a coupling between bone formation and scaffold resorption. *Biomaterials* **2007**, *28*, 1376–1384. [CrossRef] [PubMed]
54. Li, P.; Shen, L.; Li, J.; Liang, R.; Tian, W.; Tang, W. Optimal design of an individual endoprosthesis for the reconstruction of extensive mandibular defects with finite element analysis. *J. Maxillofac. Surg.* **2014**, *42*, 73–78. [CrossRef] [PubMed]
55. O'Mahony, A.M.; Williams, J.L.; Spencer, P. Anisotropic elasticity of cortical and cancellous bone in the posterior mandible increases peri-implant stress and strain under oblique loading. *Clin. Oral Implant. Res.* **2001**, *12*, 648–657. [CrossRef]
56. Bonnet, A.S.; Postaire, M.; Lipinski, P. Biomechanical study of mandible bone supporting a four-implant retained bridge: Finite element analysis of the influence of bone anisotropy and foodstuff position. *Med. Eng. Phys.* **2009**, *31*, 806–815. [CrossRef]
57. Ceddia, M.; Comuzzi, L.; Di Pietro, N.; Romasco, T.; Specchiulli, A.; Piattelli, A.; Trentadue, B. Finite Element Analysis (FEA) for the evaluation of retention in a Conometric Connection for Implant and Prosthesis. *Osteology* **2023**, *3*, 140–156. [CrossRef]
58. Macedo, J.P.; Pereira, J.; Faria, J.; Souza, J.C.M.; Alves, J.L.; López-López, J.; Henriques, B. Finite element analysis of peri-implant bone volume affected by stresses around Morse taper implants: Effects of implant positioning to the bone crest. *Comput. Methods Biomech. Biomed. Engin.* **2018**, *21*, 655–662. [CrossRef] [PubMed]
59. Chen, X.; Mao, B.; Zhu, Z.; Yu, J.; Lu, Y.; Zhang, Q.; Yue, L.; Yu, H. A three-dimensional finite element analysis of mechanical function for 4 removable partial denture designs with 3 framework materials: CoCr, Ti-6Al-4V alloy and PEEK. *Sci. Rep.* **2019**, *9*, 13975. [CrossRef]
60. Ibrahim, A.; Jiang, Z.; Shirvani, K.; Dalili, A.; Abdel Hamid, Z. A Novel Viscoelastic Deformation Mechanism Uncovered during Vickers Hardness Study of Bone. *J. Funct. Biomater.* **2024**, *15*, 87. [CrossRef]
61. Rangert, B.; Gunne, J.; Glantz, P.O.; Svensson, A. Vertical load distribution on a three-unit prosthesis supported by a natural tooth and a single Brånemark implant. An in vivo study. *Clin. Oral Implant. Res.* **1995**, *6*, 40–46. [CrossRef]
62. Jörn, D.; Kohorst, P.; Besdo, S.; Rücker, M.; Stiesch, M.; Borchers, L. Influence of lubricant on screw preload and stresses in a finite element model for a dental implant. *J. Prosthet. Dent.* **2014**, *112*, 340–348. [CrossRef]
63. Bing, L.; Mito, T.; Yoda, N.; Sato, E.; Shigemitsu, R.; Han, J.M.; Sasaki, K. Effect of peri-implant bone resorption on mechanical stress in the implant body: In vivo measured load-based finite element analysis. *J. Oral Rehabil.* **2020**, *47*, 1566–1573. [CrossRef] [PubMed]
64. Kitamura, E.; Stegaroiu, R.; Nomura, S.; Miyakawa, O. Influence of marginal bone resorption on stress around an implant—A three-dimensional finite element analysis. *J. Oral Rehabil.* **2005**, *32*, 279–286. [CrossRef] [PubMed]
65. Shabanpour Kasari, M.; Yousefi, H.; Emamian Shirazi, A.; Geramy, A. Effect of bone quality and osseointegration on stress distribution in the bone tissue surrounding dental implant. A finite element analysis. *J. Osseointegration* **2023**, *15*, 106–112. [CrossRef]
66. Naghavi, S.A.; Lin, C.; Sun, C.; Tamaddon, M.; Basiouny, M.; Garcia-Souto, P.; Taylor, S.; Hua, J.; Li, D.; Wang, L.; et al. Stress Shielding and Bone Resorption of Press-Fit Polyether-Ether-Ketone (PEEK) Hip Prosthesis: A Sawbone Model Study. *Polymers* **2022**, *14*, 4600. [CrossRef] [PubMed]

67. Morgan, E.F.; Unnikrisnan, G.U.; Hussein, A.I. Bone Mechanical Properties in Healthy and Diseased States. *Annu. Rev. Biomed. Eng.* **2018**, *20*, 119–143. [CrossRef]
68. Anitua, E.; Larrazabal Saez de Ibarra, N.; Morales Martín, I.; Saracho Rotaache, L. Influence of Dental Implant Diameter and Bone Quality on the Biomechanics of Single-Crown Restoration. A Finite Element Analysis. *Dent. J.* **2021**, *9*, 103. [CrossRef]
69. Esposito, M.; Hirsch, J.M.; Lekholm, U.; Thomsen, P. Biological factors contributing to failures of osseointegrated oral implants. (I). Success criteria and epidemiology. *Eur. J. Oral Sci.* **1998**, *106*, 527–551. [CrossRef]
70. Okumura, N.; Stegaroiu, R.; Kitamura, E.; Kurokawa, K.; Nomura, S. Influence of maxillary cortical bone thickness, implant design and implant diameter on stress around implants: A three-dimensional finite element analysis. *J. Prosthodont. Res.* **2010**, *54*, 133–142. [CrossRef]
71. Qian, L.; Todo, M.; Matsushita, Y.; Koyano, K. Effects of implant diameter, insertion depth, and loading angle on stress/strain fields in implant/jawbone systems: Finite element analysis. *Int. J. Oral Maxillofac. Implant.* **2009**, *24*, 877–886. [PubMed]
72. Romanos, G.E.; Delgado-Ruiz, R.A.; Sacks, D.; Calvo-Guirado, J.L. Influence of the implant diameter and bone quality on the primary stability of porous tantalum trabecular metal dental implants: An in vitro biomechanical study. *Clin. Oral Implant. Res.* **2018**, *29*, 649–655. [CrossRef]
73. Urdaneta, R.A.; Leary, J.; Lubelski, W.; Emanuel, K.M.; Chuang, S.K. The effect of implant size 5 × 8 mm on crestal bone levels around single-tooth implants. *J. Periodontol.* **2012**, *83*, 1235–1244. [CrossRef] [PubMed]
74. Faot, F.; Thomé, G.; Bielemann, A.M.; Hermann, C.; Melo, A.C.; Padovan, L.E.; de Mattias Sartori, I.A. Simplifying the Treatment of Bone Atrophy in the Posterior Regions: Combination of Zygomatic and Wide-Short Implants-A Case Report with 2 Years of Follow-Up. *Case Rep. Dent.* **2016**, *2016*, 5328598. [CrossRef] [PubMed]
75. Misch, C.E. *Contemporary Implant Dentistry*, 2nd ed.; Mosby: St. Louis, MO, USA, 1993.
76. Huang, H.Y.; Ogata, Y.; Hanley, J.; Finkelman, M.; Hur, Y. Crestal bone resorption in augmented bone using mineralized freeze-dried bone allograft or pristine bone during submerged implant healing: A prospective study in humans. *Clin. Oral Implant. Res.* **2016**, *27*, e25–e30. [CrossRef] [PubMed]
77. Urban, I.A.; Ravidà, A.; Saleh, M.H.A.; Galli, M.; Lozada, J.; Farkasdi, S.; Wang, H.L. Long-term crestal bone changes in implants placed in augmented sinuses with minimal or moderate remaining alveolar bone: A 10-year retrospective case-series study. *Clin. Oral Implant. Res.* **2021**, *32*, 60–74. [CrossRef]
78. Edher, F.; Nguyen, C.T. Short dental implants: A scoping review of the literature for patients with head and neck cancer. *J. Prosthet. Dent.* **2018**, *119*, 736–742. [CrossRef]
79. Watanabe, F.; Hata, Y.; Komatsu, S.; Ramos, T.C.; Fukuda, H. Finite element analysis of the influence of implant inclination, loading position, and load direction on stress distribution. *Odontology* **2003**, *91*, 31–36. [CrossRef]
80. Çağlar, A.; Aydin, C.; Ozen, J.; Yilmaz, C.; Korkmaz, T. Effects of mesiodistal inclination of implants on stress distribution in implant-supported fixed prostheses. *Int. J. Oral Maxillofac. Implant.* **2006**, *21*, 36–44. [PubMed]
81. Sütçüdel, M.; Eckert, S.E.; Zobitz, M.; An, K.N. Finite element analysis of effect of prosthesis height, angle of force application, and implant offset on supporting bone. *Int. J. Oral Maxillofac. Implant.* **2004**, *19*, 819–825. [PubMed]
82. Guven, S.; Atalay, Y.; Asutay, F.; Ucan, M.C.; Dundar, S.; Karaman, T.; Gunes, N. Comparison of the effects of different loading locations on stresses transferred to straight and angled implant-supported zirconia frameworks: A finite element method study. *Biotech. Biotechnol. Equip.* **2015**, *29*, 766–772. [CrossRef]
83. Yao, K.T.; Kao, H.C.; Cheng, C.K.; Fang, H.W.; Huang, C.H.; Hsu, M.L. Mechanical performance of conical implant-abutment connections under different cyclic loading conditions. *J. Mech. Behav. Biomed. Mater.* **2019**, *90*, 426–432. [CrossRef]
84. Sugiura, T.; Horiuchi, K.; Sugimura, M.; Tsutsumi, S. Evaluation of threshold stress for bone resorption around screws based on in vivo strain measurement of miniplat. *J. Musculoskelet. Neuronal Interact.* **2000**, *1*, 165–170. [PubMed]
85. Hsieh, Y.F.; Robling, A.G.; Ambrosius, W.T.; Burr, D.B.; Turner, C.H. Mechanical loading of diaphyseal bone in vivo: The strain threshold for an osteogenic response varies with location. *J. Bone Min. Res.* **2001**, *16*, 2291–2297. [CrossRef] [PubMed]
86. Száva, D.T.; Száva, A.; Száva, J.; Gálfi, B.; Vlase, S. Dental Implant and Natural Tooth Micro-Movements during Mastication—In Vivo Study with 3D VIC Method. *J. Pers. Med.* **2022**, *12*, 1690. [CrossRef]
87. Kulkarni, P.G.; Paudel, N.; Magar, S.; Santilli, M.F.; Kashyap, S.; Baranwal, A.K.; Zamboni, P.; Vasavada, P.; Katiyar, A.; Vikram Singh, A. Overcoming Challenges and Innovations in Orthopedic Prosthesis Design: An Interdisciplinary Perspective. *Biomed. Mater. Devices* **2024**, *2*, 58–69. [CrossRef]
88. Diaz-Lopez, R.A.; Wen, P.H.; Shelton, J.C. Influence of taper design and loading on taper micromotion. *J. Mech. Behav. Biomed. Mater.* **2022**, *128*, 105106. [CrossRef]

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

Article

Mechanical, Adhesive and Surface Properties of a Zirconia-Reinforced Lithium Silicate CAD/CAM Ceramic Exposed to Different Etching Protocols

Fabián Murillo-Gómez ^{1,2,*}, José Roberto Hernández-Viquez ¹, José Roberto Sauma-Montes de Oca ³, Cristina Vargas-Vargas ¹, Natalia González-Vargas ¹, José Roberto Vega-Baudrit ⁴ and Daniel Chavarría-Bolaños ^{3,4}

¹ Restorative Dentistry Department, School of Dentistry-FOd, University of Costa Rica-UCR, San José 11502, Costa Rica

² Dental Materials Research Laboratory (LIMD), School of Dentistry-FOd, University of Costa Rica-UCR, San José 11502, Costa Rica

³ Dentistry Graduate Program, University of Costa Rica-UCR, San José 11502, Costa Rica; jose.saumamontesdeoca@ucr.ac.cr (J.R.S.-M.d.O.); daniel.chavarria@ucr.ac.cr (D.C.-B.)

⁴ National Laboratory of Nanotechnology-LANOTEC, National Centre of High Technology-CENAT, San José 10109, Costa Rica; jvegab@gmail.com

* Correspondence: fabian.murillogomez@ucr.ac.cr; Tel.: +506-2511-4359; Fax: +506-2511-5420

Abstract: The aim of this investigation was to evaluate the effect of etching protocols on bond strength, surface roughness, and mechanical properties of a zirconia-reinforced lithium silicate (ZLS) CAD/CAM-ceramic. In total, 100 bars (ISO 6872), 75 plaques, and 25 cubes were cut from ZLS blocks (Vita Suprinity®). The surfaces were standardized, crystallized and divided into five groups: 1. control (no/treatment-C), 2. 5%-Hydrofluoric-acid (HF)/20 s (HF5%20s), 3. HF5%60s, 4. HF10%20s, and 5. HF10%60s. Flexural strength (FS) (three-point bending test, 1 mm/min), roughness (Pa), and micro-shear bond-strength (μ SBS) tests were performed. The data were statistically analyzed with one-way ANOVA, Tukey's test ($p < 0.05$) and Weibull (FS data). C showed higher Pa ($1.176 \pm 0.370 \mu\text{m}$) than HF10%60s ($0.627 \pm 0.236 \mu\text{m}$) and all other groups. Groups C and 20 s showed the most irregular surface patterns. The FS results were not influenced by etching protocols, while the Weibull modulus was, with the 5%HF groups being the most reliable (m: 5.63/6.70), while C and HF10%60s (m: 2.78/2.73) were the least reliable. All fractures originated from surface defects on the treated side of specimens. The 5%HF groups showed higher μ SBS (20 s: $21.35 \pm 4.70 \text{ MPa}$; 60 s: $23.50 \pm 4.27 \text{ MPa}$) than the 10%HF groups (20 s: $14.51 \pm 2.47 \text{ MPa}$; 60 s: $16.54 \pm 3.12 \text{ MPa}$) and C ($6.46 \pm 2.71 \text{ MPa}$). The most prevalent failure pattern was “mixed” for etched groups, and “adhesive” for C. Etching protocols affect the evaluated properties by roughening materials' surface and, in some cases, regularizing surface defects. The best overall outcomes were achieved when applying 5%HF.

Keywords: zirconia-reinforced lithium silicate; CAD/CAM materials; glass-ceramics; flexural strength; surface treatment; hydrofluoric acid

1. Introduction

Ceramic indirect dental restorations have been popular for many years now; however, their production process has faced significant improvements recently owing to fast growth in automated CAD/CAM technologies. It has revolutionized the traditional laboratory processes, turning them into faster, more reliable, profitable, and controlled procedures, allowing the clinician and patient to obtain high-quality products while reducing chair time and improving treatment convenience [1,2].

This technological evolution has also influenced the dental materials' development industry, boosting the release of new materials directed specifically to CAD/CAM technologies. Such materials show, in some cases, better properties than their analog versions [3–5].

CAD/CAM materials comprise glass-containing ceramics, polycrystalline ceramics, polymers, and hybrid materials [3,6]. These options give the clinician several possibilities of matching materials' properties with the most appropriate clinical use.

Historically, the industry has tried to provide materials to fulfill mainly two aspects: suitable mechanical properties and proper aesthetical capabilities. One of the best candidates to achieve an ideal balance between these two aspects is lithium disilicate-reinforced ceramic. This material combines a translucent glass matrix with strong crystals, providing outstanding aesthetics and high mechanical properties [4,7]. As this combination of glassy and crystalline-reinforcing phases has proven successful, manufacturers have recently developed similar proposals following this approach [8].

The zirconia-reinforced lithium silicate glass-ceramic (ZLS) was developed as an alternative to lithium disilicate ceramic. Given the excellent results reported in the literature regarding its balanced combination of aesthetics and mechanical properties, and the fact that this material was patented for several years, other manufacturers sought to use lithium silicate technology to create a new material. In this case, Dentsply and VITA developed ZLS using lithium silicate crystalline technology (along with other crystalline structures) and increased zirconia content, aiming to combine the advantages of lithium silicate ceramics with the well-known superior mechanical properties of zirconia-based ceramics. It consists of lithium metasilicate and lithium orthophosphate crystals inserted in a zirconia-powder-added glassy phase [3,9,10]. This material is available in two commercial presentations: press-ingots and CAD/CAM blocks [6]. Its overall properties and behavior are like those of lithium disilicate [3–5,8,11–16]. However, some adverse outcomes related to its microstructure, Weibull modulus, fatigue lifetime predictions, and fracture behavior have been reported [4,5,15,17–21].

As ceramic bonding with resin cement and tooth tissues is one of the most important factors to ensure long-lasting restorations, the adhesion capabilities of ZLS have also been tested [22–25]. The suggested surface treatment is to apply a hydrofluoric acid (HF) followed by a silane-containing primer, same as for other glass-ceramics [24–34]. Both strategies work synergistically. Hydrofluoric acid (HF) dissolves part of the glassy matrix in glass-containing ceramics, increasing the surface energy of the material, while the silane acts as a bridge between silica-based ceramics and resin-based materials. Although HF has raised concerns due to its potential toxicity to soft tissues, despite some alternative approaches, it remains the gold standard for improving glass-ceramic bonding when used alongside a silane primer [22–25]. However, there are still questions about the most suitable etching parameters (concentration/application time) due to significant variability in the studies' methods and observed outcomes [23,25]. Although ceramic surface treatment's primary goal is to ensure a reliable bonding of the indirect restoration, concern about its effect on materials' mechanical properties has been recurrent. This concern has led researchers to investigate the influence of different HF concentrations and application times on the physical–mechanical behavior of glass-ceramics as previously studied [7,30,35–51].

Only a few studies have investigated the influence of HF etching on the mechanical properties of a ZLS ceramic [38,45,49]. Moreover, none have yet evaluated this issue, to the authors' knowledge, related to other essential properties of this novel material such as roughness and bond strength. Thus, this study aimed to evaluate the effect of different etching protocols on the bond strength, surface roughness and morphology, flexural strength, and Weibull parameters of a zirconia-reinforced lithium silicate CAD/CAM ceramic material. The null hypothesis set was that the etching protocol does not influence the bond strength, surface roughness, flexural strength, and Weibull parameters of a zirconia-reinforced lithium silicate CAD/CAM ceramic material.

2. Materials and Methods

2.1. Specimen Preparation

CAD/CAM blocks of a zirconia-reinforced lithium silicate ceramic material (Suprinity, Vita Zahnfabrik H. Rauter GmbH & Co., Bad Säckingen, Germany; Lot No. 48521) were

used in this study. The typical composition of this material, described by the manufacturer, is the following: SiO₂: 56–64%, Li₂O: 15–21%, K₂O: 1–4%, P₂O₅: 3–8%, Al₂O₃: 1–4%, ZrO₂: 8–12%, CeO₂: 0–4%, La₂O₃: 0.1%, Pigments: 0–6%.

Samples were fabricated by cutting, from the blocks, 100 bars ($4 \pm 0.2 \text{ mm} \times 1.2 \pm 0.2 \text{ mm} \times 14 \pm 0.2 \text{ mm}$, according to ISO 6872 [52]), 75 plaques ($5 \pm 0.2 \text{ mm} \times 4 \pm 0.2 \text{ mm} \times 2 \pm 0.2 \text{ mm}$) and 25 cubes ($4 \pm 0.2 \text{ mm} \times 4 \pm 0.2 \text{ mm} \times 4 \pm 0.2 \text{ mm}$), with the aid of a diamond saw and a precision cutting machine under constant water cooling (Isomet 1000, Buehler, Lake Bluff, IL, USA). The samples were calibrated and dimensionally controlled using a digital caliper (CALDI-6 MP, Truper, Shanghai, China). Sample surfaces were then standardized using SiC 600# grit sandpaper to simulate CAD/CAM machining process and homogenize all specimens' roughness baseline. Crystallization of samples was carried out following the manufacturer's firing parameters (starting temperature/drying time: 400 °C/4 min, heating time: 8 min, temperature rise rate: 55 °C/min until 840 °C for a holding time of 8 min, vacuum holding time: 8 min and long-term cooling until 680 °C) using a multipurpose ceramic furnace (VIP Executive 500, Jetrus/Whip Mix, Louisville, KY, USA). After crystallization, the samples were dimensionally controlled, registered and stored for group division and further treatment.

Sample size determination was performed employing a power/sample size calculation test (Minitab v17.2.1, Minitab Inc.; State College, PA, USA) with a confidence interval of 95%. The standard deviation and expected confidence interval error, both necessary to perform this test, were collected from a pilot study and previous research (for numerical variables: flexural strength, bond strength, and roughness measurement). For the flexural strength test, the minimum significant sample size was 20 specimens, for bond strength, it was 15, and for roughness evaluation, it was 5 specimens. Following this analysis, sample sizes were defined as follows: flexural strength (bars), $n = 20$; bond strength (plaques), $n = 15$; and for roughness assessment (cubes), $n = 5$. A schematic representation of specimens' shapes and distributions is displayed in Figure 1.

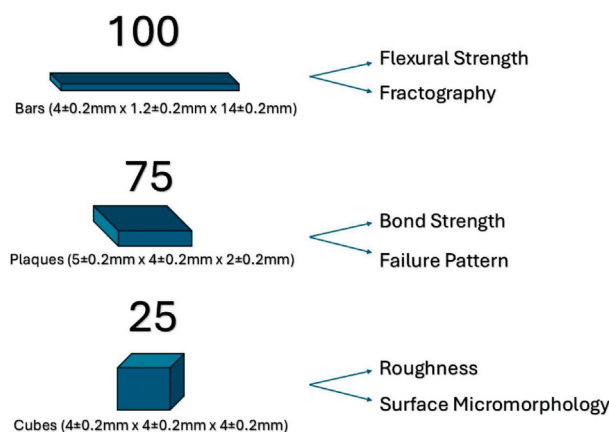


Figure 1. Schematic representation of all specimens prepared for this study.

Bars, plaques, and cubes were equally distributed into five groups to be treated according to different etching protocols: 1. control: no other treatment than SiC paper regularization was performed; 2. HF5%20s: 5% hydrofluoric acid (Condac porcelana 5%, FGM, Joinville, Brazil; Lot No. 080217) was applied for 20 s; 3. HF5%60s: 5% HF was applied for 60 s; 4. HF10%20s: 10% HF (Condac porcelana 10%, FGM, Brazil; Lot No. 310717) was applied for 20 s; and 5. HF10%60s: 10% HF was applied for 60 s. According to the manufacturer's instructions, all treatments were carried out, washing all specimens with air-powered water from the three-way syringe after the respective etching time elapsed. The specimens were also cleaned in an ultrasonic bath with distilled water for 5 min, dried, and preserved in a hermetically closed environment until the respective tests' performance for each specimen. The plaques were additionally treated with a silane-MDP-containing

ceramic primer (Monobond N, Ivoclar Vivadent, Schaan, Lichtenstein) for 60 s, air-dried for 30 s, and reserved for 3 min before being used. A single-blinded and previously calibrated operator performed sample preparation, surface treatment, and evaluation methods.

2.2. Roughness Assessment

Treated specimens (cubes) were submitted to surface roughness analysis using a contact profilometer (Dektak XT, Bruker, Billerica, MA, USA) employing a 0.5 μm tip. Five scanning lectures were performed, equally distributed through each central specimen area ($3000 \times 3000 \mu\text{m}$) at a cross speed of 8.7 $\mu\text{m}/\text{sec}$ and a resolution of 0.14 $\mu\text{m}/\text{pt}$.

The profile roughness mean (Pa) was selected to analyze the samples' surface roughness, and an average was calculated from the five lectures performed to extract each specimen's final measurement. Data were recorded and tabulated to be statistically analyzed.

2.3. Surface Micromorphology Evaluation

The same cubes used for roughness evaluation were also analyzed in a scanning electron microscope (SEM). They were sputter-coated with a gold-palladium powder alloy, placed on aluminum stubs with the aid of carbon adhesive tape, and taken into the microscope (JSM 5000, JEOL, Tokyo, Japan) operating at 15 kV and a working distance of 20 mm. Each group's representative images were obtained at various magnifications ($100\times$, $500\times$, $2000\times$, and $5000\times$) to analyze the surface micromorphology. For greater morphological detail, only $5000\times$ magnification images are presented in this report.

2.4. Flexural Strength Analysis

Treated bars were submitted to the three-point bending test following all the indications described in the ISO 6872 [52]. It was performed using a universal testing machine (H10 KS, Tinius Olsen, PA, USA). A specific jig consisting of two support rollers was used, where the specimen lied (1.5 mm diameter, positioned 12 mm apart from their centers) and a third equal roller which applied the load (at the midpoint between the support rollers) at a crosshead speed of 1 mm/min. The flexural strength (σ) was calculated in MPa according to ISO 6872 equation: $3Pl/(2wb^2)$, where P is the fracture load (N), l is the distance between the support rollers (mm), w is the specimens' width, and b is the thickness of the specimen (mm). Data were recorded and tabulated to be statistically analyzed.

2.5. Fractographic Analysis

Fragments from the flexural strength experiment were collected and evaluated using a digital stereomicroscope (Jiusion Ltd., Shenzhen, China) using magnification ranging from $40\times$ to $500\times$. Representatives from each group were chosen and prepared to be evaluated in SEM (JSM 5000, JEOL, Tokyo, Japan) using magnifications from $60\times$ to $350\times$ to determine the fracture origin and associated features. Specimens were prepared to employ the same procedure described in Section 2.3.

2.6. Microshear Bond Strength Evaluation (μSBS)

Seventy-five resin cement cylinders (Variolink Esthetic LC, Ivoclar Vivadent, Schaan, Lichtenstein) (1 ± 0.1 mm in diameter and 1 ± 0.1 mm in height) were built up (one cylinder was bonded to each ceramic plaque) on treated ceramic surfaces using a circled silicon mold ($n = 15$). Resin cement was injected onto the mold with the aid of an auto-mixing tip until the compartment was filled. A glass slide (0.5 mm thick) was placed over the top of the mold, and each cylinder was exposed for 20 s using an LED light-curing unit (Elipar Deep Cure, 3M, Saint Paul, MN, USA; light output: $1100 \text{ mW}/\text{cm}^2$). The molds were removed after 5 min, and the specimens were water-rinsed, dried, and stored in relative humidity at 37°C for 24 h. Afterward, each ceramic plaque was attached to a holding device previously adapted to a universal testing machine (Instron Electropuls E3000, Instron Corp, Canton, MA, USA) to perform microshear bond testing (μSBS), following the method developed by Shimada et al. [53]. A thin wire (0.2 mm) attached to the testing machine was placed

surrounding the extreme base of each resin cement cylinder parallel for and in contact with the bonding area to apply an uplifting shear load at a crosshead speed of 0.5 mm/min. Bond strength was calculated considering the cross-sectional bonded area of each specimen and the load needed to separate both materials. All values were expressed in MPa.

2.7. Microshear Bond Strength Failure Pattern Evaluation

De-bonded areas of resin cement/ceramic plaques assemblies were analyzed after performing the bond strength experiment, using a digital stereomicroscope (Jiusion Ltd., China) using magnification ranging from 40× to 500×. Representatives from the most prevalent failure patterns were chosen and prepared to be evaluated in SEM (JSM 5000, JEOL, Tokyo, Japan) using magnifications from 50× to 60×. Specimens were prepared to employ the same procedure described in Section 2.3. Fractures were classified as adhesive failure between both materials (A), cohesive failure within the ceramic material (CC), cohesive failure within the resin cement layer (CR) or mixed (M) when a combination of more than one failure type occurred.

2.8. Statistical Analysis

Roughness means, bond strength and flexural strength data were submitted to the analysis of the parametric assumptions (normality by employing the Anderson–Darling test; and variance homogeneity by applying the Bartlett test) to evaluate the viability of performing such a statistical test. Afterwards, one-way ANOVA and Tukey’s post hoc tests were applied to all data sets (Minitab v17.2.1, Minitab Inc.; State College, PA, USA). Additionally (as described in the ISO 6872), the Weibull analysis was also executed with flexural strength data, calculating the Weibull modulus (m , numerically and graphically) and characteristic strength (σ_0), both with their respective confidence intervals, using the maximum likelihood ratio (Minitab v17.2.1, Minitab Inc.; State College, PA, USA) [54]. All tests were performed using a confidence interval of 95%. Within the Weibull parameters, differences between groups were determined by rank order statistics, which means that if two groups presented any overlap between their confidence intervals, they were considered not statistically different from each other.

3. Results

3.1. Roughness Assessment

Roughness profile mean values (Pa) were not normally distributed (Anderson–Darling, $p < 0.05$). Data were transformed by applying the Box–Cox procedure ($\lambda = -0.331012$) to fit them into a normal distribution (Anderson–Darling, $p = 0.522$). Homoscedasticity was also confirmed after this process (Bartlett, $p = 0.425$), allowing for a parametric statistical test, using one-way ANOVA. It revealed that the factor “etching protocol” was statistically significant for this variable ($p = 0.003$). According to Tukey’s post hoc test, the control group (C) showed the highest Pa value, not statistically different from groups using HF5% or HF10%20s, while group HF10%60s presented the lowest Pa means. However, this group did not differ from groups using HF5% and HF10%20s (Table 1, Figure 2).

Table 1. Roughness profile mean values and standard deviations (Pa) recorded for each etching protocol group.

Etching Protocol	Pa (μm)
Control	1.176 (0.370) A
HF5%20s	0.873 (0.245) AB
HF5%60s	0.726 (0.195) AB
HF10%20s	0.884 (0.583) AB
HF10%60s	0.627 (0.236) B

Different letters mean statistical differences among groups (Tukey, $p < 0.05$).

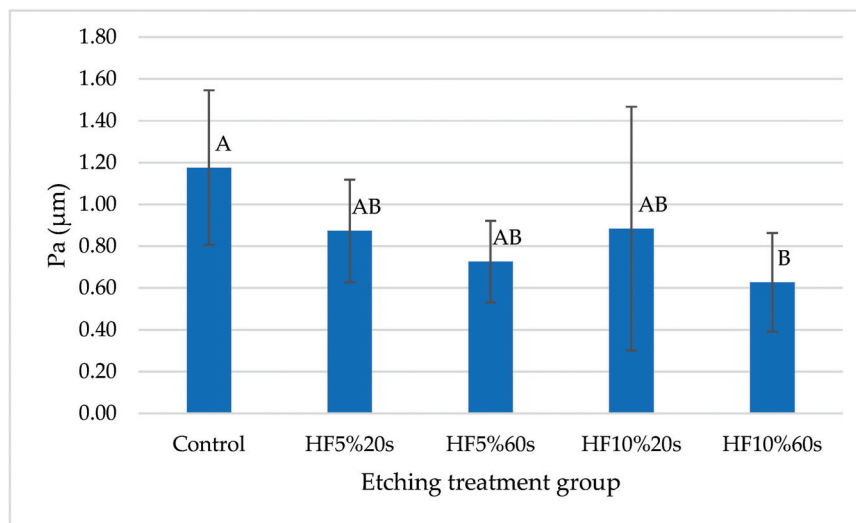


Figure 2. Graphical representation of Table 1 containing roughness mean values and standard deviations (expressed in MPa) from each group. Different letters represent significant statistical differences.

3.2. Surface Micromorphology Evaluation

Morphological analysis revealed that the surface pattern is etching-protocol-dependent, as each treatment produced different characteristics within the material's surface. The control group (no etching, only SiC sandpaper regularization prior firing) showed a very irregular surface where the glass-matrix exhibited some continuity disruptions such as scratches and voids (Figure 3a). The other groups presented a different morphology from the control group because the glass-matrix is not visible; instead, the crystalline phase shows many irregular-shaped tiny crystals (most of them below micron-scale) distributed along the whole surface. The HF5%60s etching protocol produced the most regular surface morphology (Figure 3c), while the most irregular seemed to be the surface treated under the HF10%20s protocol (Figure 3d), exhibiting more “dark areas” compared with the other groups, which may be interpreted as deeper depressions in SEM images. Both groups to which the etchant was applied for 60 s also presented those “dark areas”, these being darker (more profound defects) but more focused on the group using 10% HF than on the group employing 5% HF (Figure 3b,e).

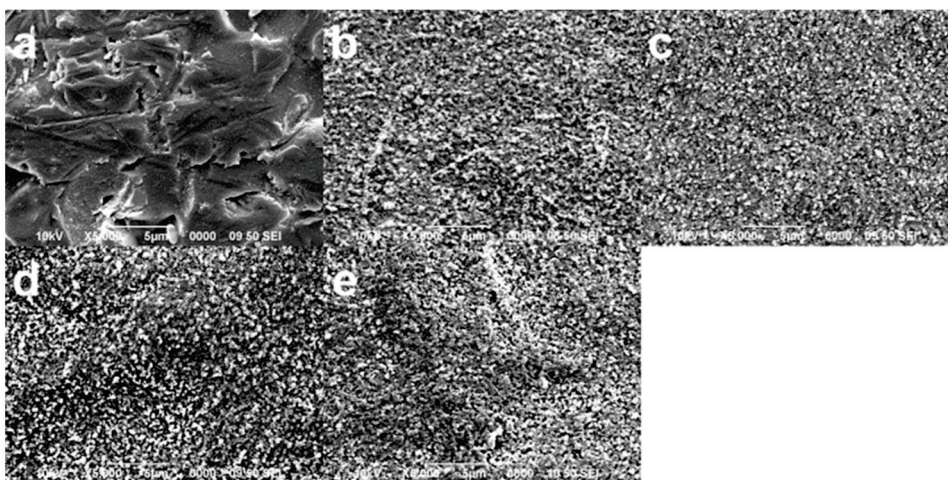


Figure 3. Surface micromorphology of zirconia-reinforced lithium silicate ceramic (ZLS) after being treated with the different etching protocols. SEM (Scanning Electron Microscope) images showing the typical surface morphology produced by the experimental groups: (a) control group (no etching); (b) HF5%20s group; (c) HF5%60s group; (d) HF10%20s group; and (e) HF10%60s group.

3.3. Flexural Strength Analysis

After confirming the normality (Anderson–Darling, $p = 0.165$) and homoscedasticity (Bartlett, $p = 0.640$) of the flexural strength data, ANOVA revealed that the “etching protocol” factor was not statistically significant ($p = 0.473$). Regarding the Weibull parameters, no significant differences were detected within the characteristic strength (σ_0) among groups either; however, significant differences were noted within the Weibull modulus (m) for some groups (Table 2). Groups employing 5% HF obtained a significantly higher Weibull modulus than the HF10%60s and C groups, these last two not being statistically different from group HF10%20s (Table 2). Weibull modulus plots for all groups are displayed in Figure 4.

Table 2. Flexural strength (σ) in MPa with the respective standard deviation, Weibull modulus (m) and characteristic strength in MPa (σ_0) with their respective confidence intervals for all groups.

Group	σ	(m)	(σ_0)
Control	312.24 (29.12)	2.78 (2.04–3.79) B	330.43 (310.35–354.16)
HF5%20s	316.38 (22.25)	6.70 (4.65–9.65) A	329.14 (319.79–339.22)
HF5%60s	321.89 (22.81)	5.63 (3.97–7.99) A	332.27 (321.27–344.28)
HF10%20s	314.08 (24.74)	4.69 (3.39–6.48) AB	322.53 (310.68–335.65)
HF10%60s	308.58 (20.99)	2.73 (1.96–3.79) B	329.53 (308.82–354.18)

Different letters mean statistical differences among groups (Tukey, $p < 0.05$).

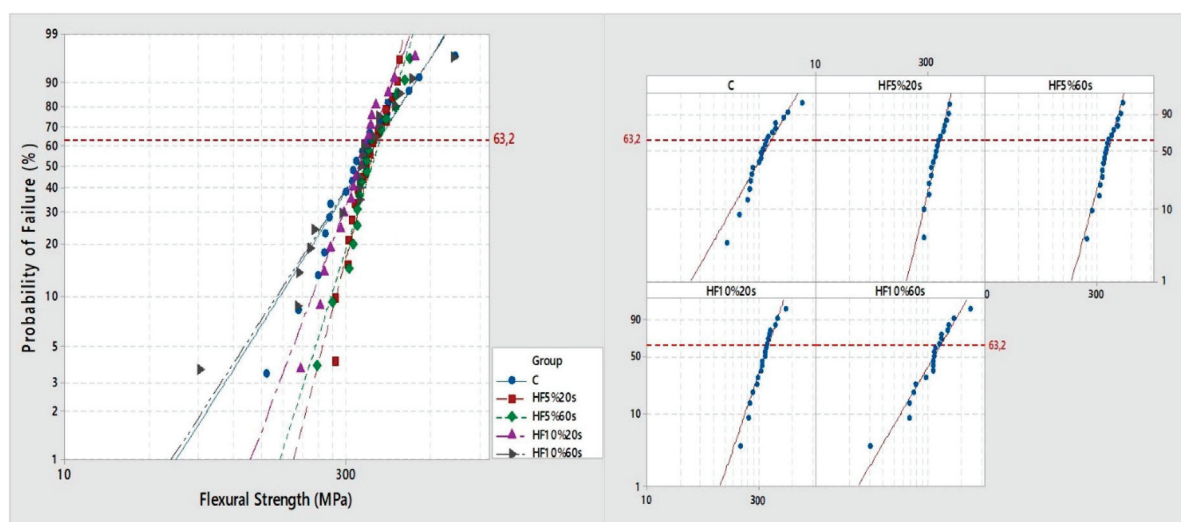


Figure 4. Weibull modulus plots for each group. The left plot shows the graphical Weibull modulus corresponding to all experimental groups (control: blue; HF5%20s: red; HF5%60s: green; HF10%20s: purple; HF10%60s: gray). Right side plots show a separate graphical representation for each groups' Weibull modulus. All Weibull plots represent a curve relating the probability of failure % (Y-axis) and flexural strength in MPa (X-axis). The Weibull characteristic strength is pointed with the discontinuous red horizontal line, across a 63.2% probability of failure.

3.4. Fractographic Analysis

All analyzed fragments showed that fractures originated from a defect on the treated surface (tensile side, opposite to the side where the load was applied). All cracks propagated from defects within such a surface to the compression side, showing different features (Figure 5).

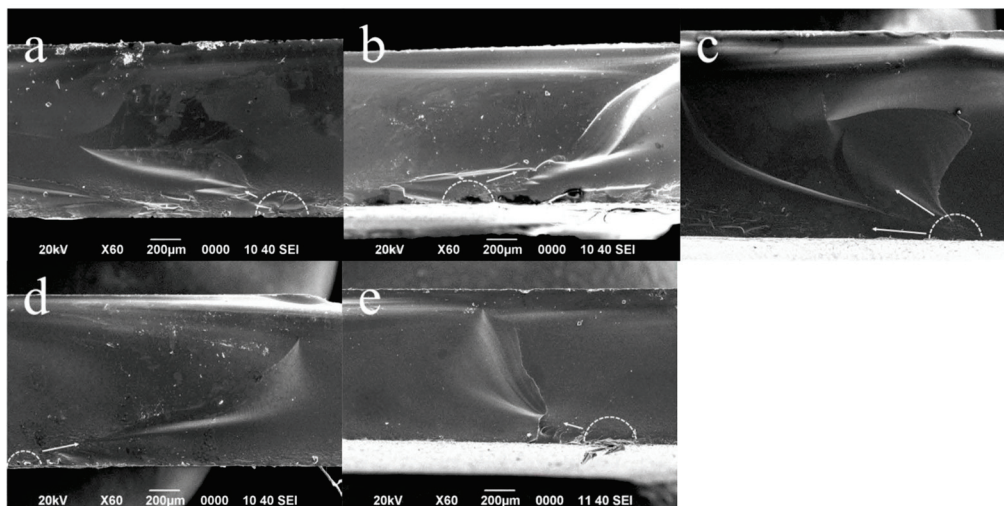


Figure 5. Representative SEM images (60 $\times$) from each group regarding fractographic analysis: (a) control group (no etching); (b) HF5%20s group; (c) HF5%60s group; (d) HF10%20s group; and (e) HF10%60s group. The discontinuous semicircle points to the origin of fracture for each group, which was located at surface defects on the treated side (tensile side), showing that the main direction of fracture propagation (arrows) to the compression side compression curl area is also notorious.

3.5. Microshear Bond Strength Evaluation (μ SBS)

Normality (Anderson–Darling, $p = 0.473$) and homoscedasticity (Bartlett, $p = 0.073$) were confirmed. ANOVA revealed that the “etching protocol” factor was statistically significant ($p < 0.0001$). Tukey’s test revealed that the groups treated with 5%HF resulted in higher μ SBS compared to the groups treated with 10%HF, while all HF-treated groups showed higher μ SBS mean values than the control group (Table 3 and Figure 6).

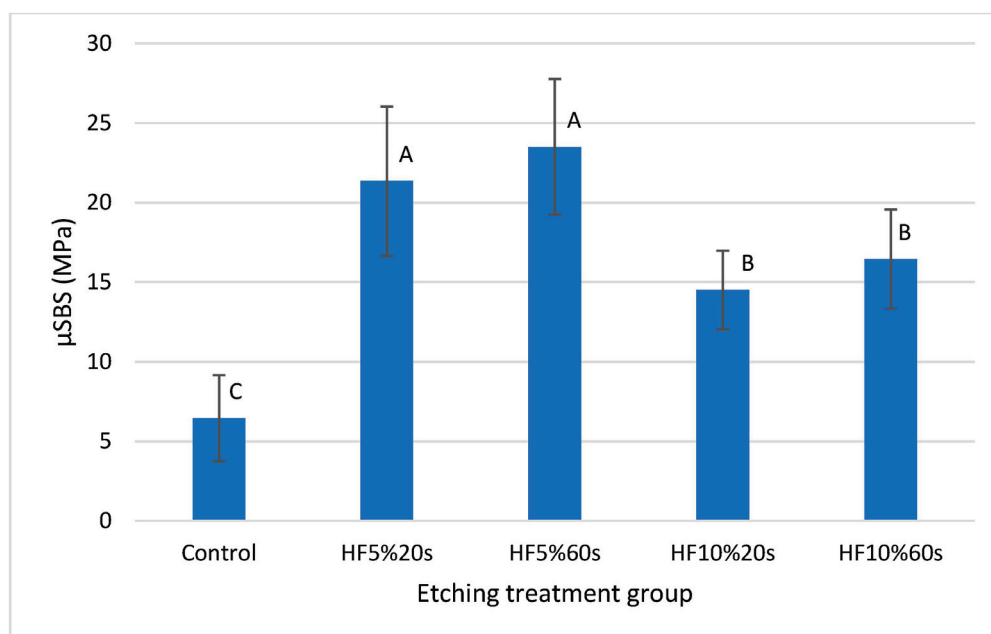


Figure 6. Graphical representation of Table 3 containing microshear bond strength mean values and standard deviations (expressed in MPa) from each group. Different letters represent significant statistical differences.

Table 3. Microshear bond strength mean values and standard deviations (expressed in MPa) correspond to each group.

Etching Protocol	μ SBS (MPa)
Control	6.46 (2.71) C
HF5%20s	21.35 (4.70) A
HF5%60s	23.50 (4.27) A
HF10%20s	14.51 (2.47) B
HF10%60s	16.45 (3.12) B

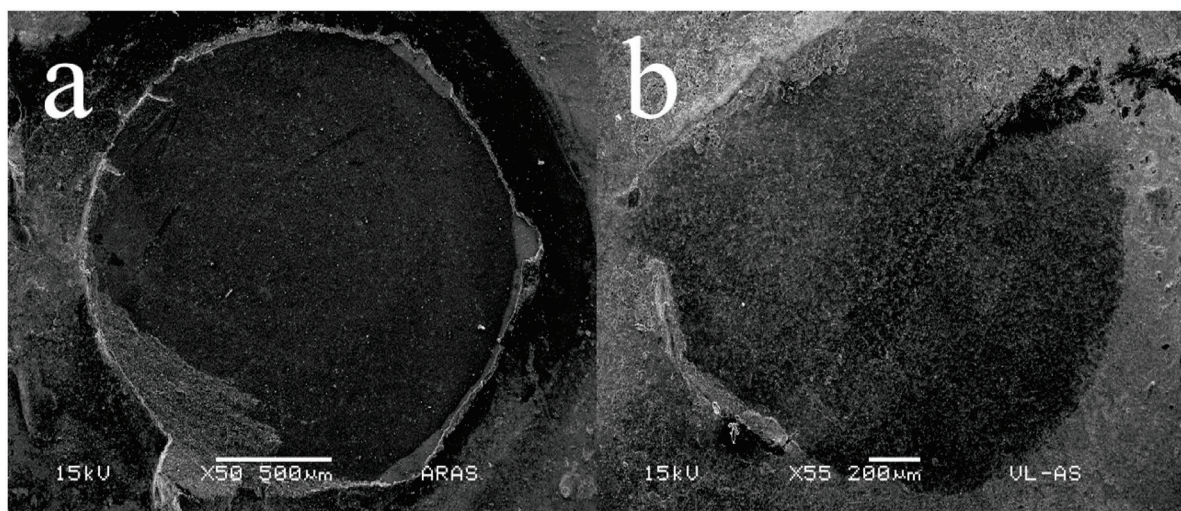
Different letters mean statistical differences among groups (Tukey, $p < 0.05$).

3.6. Microshear Bond Strength Failure Pattern Evaluation

The most prevalent failure pattern for the control group was adhesive failure (A), while groups treated with HF showed a mixed failure pattern (M) mostly, this pattern being more prevalent on the 5%HF groups (Table 4 and Figure 7).

Table 4. Failure type prevalence (%) of de-bonded specimens for each group.

Etching Protocol	Failure Type Prevalence (%)			
	M	CC	CR	A
Control	20	0	0	80
HF5%20s	87	0	0	13
HF5%60s	100	0	0	0
HF10%20s	73	0	0	27
HF10%60s	60	5	0	35

**Figure 7.** Representative SEM photo-micrographs of the most prevalent failure patterns obtained from microshear bond strength test: (a) adhesive failure, (b) mixed failure.

4. Discussion

Statistical analysis revealed that employing different HF etching protocols affected the surface roughness, bond strength, and Weibull modulus of zirconia-reinforced lithium silicate CAD/CAM ceramic; however, the etching protocol did not influence the material's flexural strength (nor the Weibull characteristic strength). Therefore, the null hypothesis has to be partly rejected.

CAD/CAM materials in dentistry are constantly evolving, and this is the reason why research assessing the effectiveness of clinical procedures in recently developed materials is still relevant. Thus, ZLS ceramic was chosen as the main subject of this work. Additionally, its similarity to one of the most successful materials available nowadays

(lithium disilicate) makes this material of great interest as an alternative to the first [3–5]. The surface treatments studied here were chosen because 4–5% and 9–10% are the most available concentrations of HF (commercially speaking), and 20 and 60 s are the most used application times within glass-ceramics [28–31]. To investigate bonding performance, μ SBS was chosen due to its convenience in employing highly resistant materials such as CAD/CAM ceramics while still being an accurate and very used method [28,32,33]. Some issues of concern have been highlighted in tests employing shear. However, they can be minimized by taking some precautions such as those applied here. Roughness evaluation was performed to measure the effect of the chosen treatments on the material's surface, relating it to SEM topographical evaluation and its mechanical properties to measure the extension of such effects on the structural behavior of this material. A three-point bending test was used due to its specimens' design suitability with CAD/CAM blocks' geometrical shape, also a widely validated method [4].

Regarding roughness, the current outcomes revealed that the roughest surfaces were recorded among the control group (no HF, only SiC sandpaper regularization) and the lowest within the HF10%60s group (Table 1 and Figure 2). SEM micromorphology analysis showed some sharper scratches in the control group, product of the regularization made to all specimens with SiC sandpaper. The procedure may explain the rise in the control group's roughness, as reported previously for other glass-ceramics [47,49]. It is also likely and potentially dangerous because ZLS ceramic has been noted to be susceptible to developing residual crack growth when roughened in the pre-crystallized stage, as happens when milling a restoration [4,18].

HF acts by dissolving part of the SiO_2 content on glass-ceramics as follows: $\text{SiO}_2 + 4\text{HF} \rightarrow \text{SiF}_4 + 2\text{H}_2\text{O}$ / $4\text{SiF}_4 + 3\text{H}_2\text{O} + 2\text{HF} \rightarrow 3\text{H}_2\text{SiF}_6 + \text{H}_2\text{SiO}_3$. This will erode the surface of the ceramic, exposing the crystalline structures beneath the original glassy matrix [33,39]. It appears that besides dissolving part of the glassy matrix of this material, HF etching can also regularize its macrosurface morphology, disguising sharper defects, caused, in this case, by SiC sandpaper [4]. Surface roughness remained statistically comparable within groups employing 5%HF and HF10%20s group. These methods show a macro-homogenizer effect on materials' surface morphology [45]. However, it is striking that, at least numerically speaking, the roughness was slightly higher when HF was applied for 20 s than for a longer time. It probably happened because 20 s may not be enough time for 10%HF to regularize this material's surface macro-topography, leaving some sharper defects (Figure 3). In addition, 10% concentration applied during shorter periods may produce sharper, more profound, and localized defects within materials' surface instead of producing a regular etching pattern within the horizontal extension of the surface [30]. This effect could be more significant if the material already shows previous defects or micro-scratches, as produced in the current work by the SiC sandpaper or in an actual situation by the milling bur [30].

On the other hand, when 10%HF was applied, it produced smoother surfaces. In this regard, it is essential to point out that ZLS is constituted by a conjunction of three phases comprising mostly spherical homogenous tiny crystals ($\sim 0.5 \mu\text{m}$) [8]. Such composition may favor easier and greater crystal dislodgement when 10%HF is applied for a longer time to this material, producing a smoother and more stable macrosurface [3,4,15,19].

The microstructure and surface roughness of glass-ceramics are indeed not isolated factors. Both are related and could affect materials' mechanical properties, and this is the reason why it is well documented that achieving smoother surfaces improves their mechanical behavior [20,35]. Additionally, a proper bonding between the ceramic material and the luting agent also influences the whole system's mechanical behavior [29]. Following this rationale, flexural strength and bonding performance were analyzed when applying variations on HF etching parameters.

Several studies have investigated this relationship between etching treatments and mechanical properties, mainly evaluating flexural strength or fatigue failure load resistance within other glass-ceramics. Outcomes in this regard are controversial. Some au-

thors have reported a significant negative effect of HF etching on materials' mechanical behavior [7,36,38,39,50,51], while some others reported no significant influence of HF etching [37,40–42,44,47]. In contrast, fewer authors have determined a positive effect of HF etching on glass-ceramics mechanical properties [45,46,49]. Specifically, about ZLS, only a few of those previous works addressed this issue but with significant variability between study parameters and results. For example, Barchetta et al. and Monteiro et al. reported a positive effect of increased application time [45,49], while Hallman et al. found that the recommended etching protocol for this material (HF5%20s) reduced its flexural strength by around 20.9% compared to the control [38].

The current outcomes revealed that different HF etching protocols did not significantly affect the flexural strength of a ZLS CAD/CAM ceramic. However, significant differences were detected in the Weibull modulus (m) of some groups (Table 2, Figure 4). Weibull analysis is one of the most used tests to characterize ceramics' mechanical behavior by associating a prediction of failure associated with two parameters. The characteristic strength or scale (σ_0) is the strong point, where 63.2% of the specimens would have failed. And the Weibull modulus describes the variation in strength data (m , shape), related to a calculation of the distribution of defects or flaws within a materials' microstructure to infer its mechanical consistency [54].

The flexural strength values and Weibull parameters obtained here are similar to those previously reported for this type of material (ZLS) [17,20,21,45,55]. The control and HF10%60s groups showed the lowest Weibull modulus. In the case of the control group, visible surface defects, a product of SiC sandpaper regularization (as previously discussed), are evident (Figure 3) and also consistent with the roughness results (Table 1). The greater prevalence of those flaws probably increased the variability in this group's mechanical reaction to bending forces, making it less reliable than the 5%HF groups (Table 2, Figure 4). The lowest Weibull modulus produced by 10%HF may be explained by a possible, more substantial effect of this concentration, producing deeper and sharper defects within the surface [28,30]. Such zones (Figure 3) may accumulate tension and develop a faster crack growth rate than smoother or more uniform regions within a surface [4,15,18,43]. It can also be supported by the fractographic analysis, which revealed that most of the fractures originated in surface defects within the treated side of the specimens (Figure 5).

Regarding the application time, both 5%HF groups showed a statistically comparable Weibull modulus, and roughness means. However, judging by the morphological analysis (Figure 3), it can be said that 60 s exerted a more homogenous macro-effect on the materials' surface than 20 s did. HF in contact with the material's surface for a short time probably produces more localized etching in vulnerable zones, such as previous defects (Figure 3). At the same time, a longer application time may etch materials' surfaces more horizontally than vertically. A macro-smoother surface has been recommended in terms of mechanics for this specific material [3–5,15,18,49].

Bond strength was also significantly affected by HF etching protocols (Table 3). The control group showed poor bonding with resin cement compared to all treatments employing HF and silane as previously reported [23,26,32]. This result was consistent with the greater prevalence of the adhesive failure type recorded for this group (Table 4, Figure 7). It reaffirms the beneficial effect of promoting micromechanical and chemical interactions between ceramic materials and resin types of cement. The most prevalent failure type (mixed) obtained by HF-treated groups also reflects this interaction (Table 4, Figure 7). Groups employing 5%HF produced better bonding values than the 10%HF groups. This probably occurred because 10%HF may have produced deeper and sharper cone-shaped surface defects [30], hindering the resin cement from infiltrating them fully and thus affecting ceramic/cement bonding [34]. Analyzing failure patterns in bond strength testing is essential, as the debonded area provides critical information about the interaction between the bonded components and their behavior during debonding. The literature suggests that adhesive failures indicate poor interaction between the substrate and the adherent material, as no remnants of either part remain attached after separation [23,26,32]. Clinically, this

could result in complete debonding of an indirect restoration. In contrast, mixed failures show areas where the adherent material remains attached to the substrate after separation, indicating an intimate interaction between the two materials in certain regions. These areas likely experienced cohesive failures within the adherent material (in this case, resin cement), meaning that the adhesive bond in those spots was stronger than the material's cohesive strength. Clinically, this scenario is preferred to ensure long-term reliable bonding of the indirect restoration. While fully cohesive failures within the adherent material also suggest strong bonding, in microshear bond strength tests, they may indicate a potential bias in test execution. Specifically, such failures could suggest that the shear force applied during testing was not fully parallel to the bonded area. Therefore, a higher prevalence of mixed failure types, as observed in this study, is preferred over fully cohesive failures.

From previous outcomes, it can be said that the most suitable etching protocol for glass-ceramics is material-dependent. Moreover, no automatic extrapolation should be made from one type of glass-ceramic to another. That scenario would apply to materials belonging to the same sub-class, for example, lithium disilicate and lithium silicate-based ceramics. According to their glassy content and bond strength to resin-based materials, the previous tendency in the literature has been to recommend an etching time to treat glass-ceramics [28]. However, the rising concern of the effect of such treatments on the structural integrity and mechanical behavior of CAD/CAM materials has led to an integral approach [30,38]. Therefore, an etching treatment employing 5%HF (followed by a silane coupling agent) can be recommended to treat ZLS CAD/CAM ceramics based on the current results. This recommendation agrees with a recent study that determined that 4.9%HF produced better ceramic/cement bonding on this material, compared to 9.5%HF [23]. So, within the limitations of this *in vitro* study, this information may be considered clinically significant and should be taken into account when looking for proper surface treatment for this material.

Some limitations of this *in vitro* study are the use of a static mechanical test instead of a dynamic method, a test set-up directed to simulate only the pre-cementation performance of the tested material instead of reproducing a methodological approach directed to emulate the clinical–mechanical performance of this material in long-term service, and a lack of comparison between ZLS and other similar materials, such as lithium disilicate ceramics. However, the aim of this specific work was successfully achieved, and these outcomes can be considered a starting point for further investigations in this regard. Future research could address these aspects by evaluating the influence of these etching protocols on the fatigue resistance and lifetime predictions of cemented restorations made from this material, as well as comparing them to lithium disilicate restorations.

5. Conclusions

Within the limitations of this *in vitro* study and based on the current results, it can be concluded that variations in hydrofluoric acid (HF) etching protocols do not influence the ZLS CAD/CAM ceramics' flexural strength; however, those variations do affect its surface roughness, bond strength, and Weibull modulus. HF etching is necessary to create a rough surface topography on this ceramic to improve interlocking with the luting material and regularize some previous sharp surface defects. The best overall outcomes were achieved when using 5%HF to treat this material.

Author Contributions: Conceptualization, F.M.-G., J.R.H.-V., J.R.S.-M.d.O., C.V.-V., N.G.-V., J.R.V.-B. and D.C.-B.; Data Curation, F.M.-G.; Formal Analysis, F.M.-G.; Investigation, F.M.-G., J.R.H.-V., J.R.S.-M.d.O., C.V.-V., N.G.-V., J.R.V.-B. and D.C.-B.; Methodology, F.M.-G.; Project Administration, F.M.-G.; Supervision, F.M.-G.; Validation, F.M.-G.; Writing—Original Draft, F.M.-G.; Writing—Review and Editing, F.M.-G., J.R.V.-B. and D.C.-B. All authors have read and agreed to the published version of the manuscript.

Funding: This research was partly funded by the Vice-rectory of Research (VIn) of the University of Costa Rica, project number: B9260.

Data Availability Statement: The original contributions presented in the study are included in the article, further inquiries can be directed to the corresponding author.

Acknowledgments: This work is part of the research project number B9260, registered by Murillo-Gómez at the Vice-rectory of Research (VIn) of the University of Costa Rica. Some data included in this work and belonging to the original root project were used by JRH-V, JRS-M, CV-V and NG-V as partial fulfillment of the requirements to obtain the DDS degree. Beside the institutions of affiliations, the authors thank the Materials Science and Engineering Research Center (CICIMA) of the UCR for logistic and equipment support for this research.

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

References

1. Dolev, E.; Bitterman, Y.; Meirowitz, A. Comparison of marginal fit between CAD-CAM and hot-press lithium disilicate crowns. *J. Prosthet. Dent.* **2019**, *121*, 124–128. [CrossRef] [PubMed]
2. Kollmuss, M.; Kist, S.; Goeke, J.E.; Hickel, R.; Huth, K.C. Comparison of chairside and laboratory CAD/CAM to conventional produced all-ceramic crowns regarding morphology, occlusion, and aesthetics. *Clin. Oral. Investig.* **2016**, *20*, 791–797. [CrossRef] [PubMed]
3. Belli, R.; Wendler, M.; de Ligny, D.; Cicconi, M.R.; Petschelt, A.; Peterlik, H.; Lohbauer, U. Chairside CAD/CAM materials. Part 1: Measurement of elastic constants and microstructural characterization. *Dent. Mater.* **2017**, *33*, 84–98. [CrossRef] [PubMed]
4. Wendler, M.; Belli, R.; Petschelt, A.; Mevec, D.; Harrer, W.; Lube, T.; Danzer, R.; Lohbauer, U. Chairside CAD/CAM materials. Part 2: Flexural strength testing. *Dent. Mater.* **2017**, *33*, 99–109. [CrossRef] [PubMed]
5. Wendler, M.; Belli, R.; Valladares, D.; Petschelt, A.; Lohbauer, U. Chairside CAD/CAM materials. Part 3: Cyclic fatigue parameters and lifetime predictions. *Dent. Mater.* **2018**, *34*, 910–921. [CrossRef]
6. Gracis, S.; Thompson, V.; Ferencz, J.; Silva, N.; Bonfante, E. A New Classification System for All-Ceramic and Ceramic-like Restorative Materials. *Int. J. Prosthodont.* **2016**, *28*, 227–235. [CrossRef]
7. Kurtulmus-Yilmaz, S.; Cengiz, E.; Ongun, S.; Karakaya, I. The Effect of Surface Treatments on the Mechanical and Optical Behaviors of CAD/CAM Restorative Materials. *J. Prosthodont.* **2019**, *28*, e496–e503. [CrossRef]
8. Elsaka, S.E.; Elnaghy, A.M. Mechanical properties of zirconia reinforced lithium silicate glass-ceramic. *Dent. Mater.* **2016**, *32*, 908–914. [CrossRef]
9. de Carvalho Ramos, N.; Campos, T.M.B.; de La Paz, I.S.; Machado, J.P.B.; Bottino, M.A.; Cesar, P.F.; De Melo, R.M. Microstructure characterization and SCG of newly engineered dental ceramics. *Dent. Mater.* **2016**, *32*, 870–878. [CrossRef]
10. Furtado de Mendonca, A.; Shahmoradi, M.; de Gouvêa, C.V.D.; De Souza, G.M.; Ellakwa, A. Microstructural and Mechanical Characterization of CAD/CAM Materials for Monolithic Dental Restorations. *J. Prosthodont.* **2019**, *28*, e587–e594. [CrossRef]
11. Sen, N.; Us, Y.O. Mechanical and optical properties of monolithic CAD-CAM restorative materials. *J. Prosthet. Dent.* **2018**, *119*, 593–599. [CrossRef] [PubMed]
12. Kilinc, H.; Turgut, S. Optical behaviors of esthetic CAD-CAM restorations after different surface finishing and polishing procedures and UV aging: An in vitro study. *J. Prosthet. Dent.* **2018**, *120*, 107–113. [CrossRef] [PubMed]
13. Ludovichetti, F.S.; Trindade, F.Z.; Adabo, G.L.; Pezzato, L.; Fonseca, R.G. Effect of grinding and polishing on the roughness and fracture resistance of cemented CAD-CAM monolithic materials submitted to mechanical aging. *J. Prosthet. Dent.* **2019**, *121*, 866.e1–866.e8. [CrossRef] [PubMed]
14. Stawarczyk, B.; Liebermann, A.; Eichberger, M.; Güth, J.F. Evaluation of mechanical and optical behavior of current esthetic dental restorative CAD/CAM composites. *J. Mech. Behav. Biomed. Mater.* **2015**, *55*, 1–11. [CrossRef]
15. Wendler, M.; Kaizer, M.R.; Belli, R.; Lohbauer, U.; Zhang, Y. Sliding contact wear and subsurface damage of CAD/CAM materials against zirconia. *Dent. Mater.* **2020**, *36*, 387–401. [CrossRef]
16. Scherer, M.M.; Prochnow, C.; Venturini, A.B.; Pereira, G.K.R.; Burgo, T.A.d.L.; Rippe, M.P.; Valandro, L.F. Fatigue failure load of an adhesively-cemented lithium disilicate glass-ceramic: Conventional ceramic etching vs etch & prime one-step primer. *Dent. Mater.* **2018**, *34*, 1134–1143.
17. Liu, C.; Eser, A.; Albrecht, T.; Stournari, V.; Felder, M.; Heintze, S.; Broeckmann, C. Strength characterization and lifetime prediction of dental ceramic materials. *Dent. Mater.* **2021**, *37*, 94–105. [CrossRef]
18. Wendler, M.; Belli, R.; Lohbauer, U. Factors influencing development of residual stresses during crystallization firing in a novel lithium silicate glass-ceramic. *Dent. Mater.* **2019**, *35*, 871–882. [CrossRef]
19. Vichi, A.; Fabian Fonzar, R.; Goracci, C.; Carrabba, M.; Ferrari, M. Effect of finishing and polishing on roughness and gloss of lithium disilicate and lithium silicate zirconia reinforced glass ceramic for CAD/CAM systems. *Oper. Dent.* **2018**, *43*, 90–100. [CrossRef]
20. Venturini, A.B.; Prochnow, C.; Pereira, G.K.R.; Segala, R.D.; Kleverlaan, C.J.; Valandro, L.F. Fatigue performance of adhesively cemented glass-, hybrid- and resin-ceramic materials for CAD/CAM monolithic restorations. *Dent. Mater.* **2019**, *35*, 534–542. [CrossRef]

21. Riquieri, H.; Monteiro, J.B.; Viegas, D.C.; Campos, T.M.B.; de Melo, R.M.; de Siqueira Ferreira Anzaloni Saavedra, G. Impact of crystallization firing process on the microstructure and flexural strength of zirconia-reinforced lithium silicate glass-ceramics. *Dent. Mater.* **2018**, *34*, 1483–1491. [CrossRef] [PubMed]
22. Al-Thagafi, R.; Al-Zordk, W.; Saker, S. Influence of surface conditioning protocols on reparability of CAD/CAM zirconia-reinforced lithium silicate ceramic. *J. Adhes. Dent.* **2016**, *18*, 135–141. [PubMed]
23. Fonzar, R.F.; Goracci, C.; Carrabba, M.; Louca, C.; Ferrari, M.; Vichi, A. Influence of acid concentration and etching time on composite cement adhesion to lithium-silicate glass ceramics. *J. Adhes. Dent.* **2020**, *22*, 175–182.
24. Ustun, S.; Ayaz, E.A. Effect of different cement systems and aging on the bond strength of chairside CAD-CAM ceramics. *J. Prosthet. Dent.* **2020**, *125*, 334–339. [CrossRef]
25. Sato, T.P.; Anami, L.C.; Melo, R.M.; Valandro, L.F.; Bottino, M.A. Effects of surface treatments on the bond strength between resin cement and a new zirconia-reinforced lithium silicate ceramic. *Oper. Dent.* **2016**, *41*, 284–292. [CrossRef]
26. Murillo-Gómez, F.; De Goes, M.F. Effect of Different Silane-Containing Solutions on Glass-Ceramic/Cement Bonding Interacting with Dual-Cure Resin Cements. *Odontol. Int. J. Dent. Sci.* **2015**, *16*, 87–105. [CrossRef]
27. Itthipongsatorn, N.; Srisawasdi, S. Dentin microshear bond strength of various resin luting agents to zirconia-reinforced lithium silicate ceramics. *J. Prosthet. Dent.* **2020**, *124*, 237.e1–237.e7. [CrossRef]
28. Tian, T.; Tsoi, J.K.H.; Matinlinna, J.P.; Burrow, M.F. Aspects of bonding between resin luting cements and glass ceramic materials. *Dent. Mater.* **2014**, *30*, e147–e162. [CrossRef]
29. Murillo-Gómez, F.; Wanderley, R.B.; De Goes, M.F. Impact of Silane-containing Universal Adhesive on the Biaxial Flexural Strength of a Resin Cement/Glass-ceramic System. *Oper. Dent.* **2019**, *44*, 200–209. [CrossRef]
30. Murillo-Gómez, F.; Palma-Dibb, R.G.; De Goes, M.F. Effect of acid etching on tridimensional microstructure of etchable CAD/CAM materials. *Dent. Mater.* **2018**, *34*, 944–955. [CrossRef]
31. Amaral Colombo, L.D.; Murillo-Gómez, F.; De Goes, M.F. Bond Strength of CAD/CAM Restorative Materials Treated with Different Surface Etching Protocols. *J. Adhes. Dent.* **2019**, *21*, 307–317.
32. Murillo-Gómez, F.; Rueggeberg, F.A.; De Goes, M.F. Short- and long-term bond strength between resin cement and glass-ceramic using a silane-containing universal adhesive. *Oper. Dent.* **2017**, *42*, 514–525. [CrossRef] [PubMed]
33. El-Damanhoury, H.M.; Gaintantzopoulou, M.D. Self-etching ceramic primer versus hydrofluoric acid etching: Etching efficacy and bonding performance. *J. Prosthodont. Res.* **2018**, *62*, 75–83. [CrossRef] [PubMed]
34. Sundfeld Neto, D.; Naves, L.Z.; Costa, A.R.; Correr, A.B.; Consani, S.; Borges, G.A.; Correr-Sobrinho, L. The effect of hydrofluoric acid concentration on the bond strength and morphology of the surface and interface of glass ceramics to a resin cement. *Oper. Dent.* **2015**, *40*, 470–479. [CrossRef]
35. Dal Piva, A.M.d.O.; Tribst, J.P.M.; Venturini, A.B.; Anami, L.C.; Bonfante, E.A.; Bottino, M.A.; Kleverlaan, C.J. Survival probability of zirconia-reinforced lithium silicate ceramic: Effect of surface condition and fatigue test load profile. *Dent. Mater.* **2020**, *36*, 808–815. [CrossRef]
36. Venturini, A.B.; Prochnow, C.; May, L.G.; Bottino, M.C.; Felipe Valandro, L. Influence of hydrofluoric acid concentration on the flexural strength of a feldspathic ceramic. *J. Mech. Behav. Biomed. Mater.* **2015**, *48*, 241–248. [CrossRef]
37. Prochnow, C.; Venturini, A.B.; Guilardi, L.F.; Pereira, G.K.R.; Burgo, T.A.L.; Bottino, M.C.; Kleverlaan, C.J.; Valandro, L.F. Hydrofluoric acid concentrations: Effect on the cyclic load-to-failure of machined lithium disilicate restorations. *Dent. Mater.* **2018**, *34*, e255–e263. [CrossRef]
38. Hallmann, L.; Ulmer, P.; Gerngross, M.D.; Jetter, J.; Mintrone, M.; Lehmann, F.; Kern, M. Properties of hot-pressed lithium silicate glass-ceramics. *Dent. Mater.* **2019**, *35*, 713–729. [CrossRef]
39. Zogheib, L.V.; Della Bona, A.; Kimpura, E.T.; McCabe, J.F. Effect of hydrofluoric acid etching duration on the roughness and flexural strength of a lithium disilicate-based glass ceramic. *Braz. Dent. J.* **2011**, *22*, 45–50. [CrossRef]
40. Porto, T.S.; Park, S.J.; Faddoul, A.J.; Faddoul, F.F.; Cesar, P.F. Evaluation of the surface roughness and accelerated aging of CAD/CAM materials. *Int. J. Prosthodont.* **2020**, *33*, 418–428. [CrossRef]
41. Salazar-Marcho, S.M.; de Melo, R.M.; Macedo, L.G.S.; Valandro, L.F.; Bottino, M.A. Strength of a feldspar ceramic according to the thickness and polymerization mode of the resin cement coating. *Dent. Mater. J.* **2011**, *30*, 323–329. [CrossRef] [PubMed]
42. Menees, T.S.; Lawson, N.C.; Beck, P.R.; Burgess, J.O. Influence of particle abrasion or hydrofluoric acid etching on lithium disilicate flexural strength. *J. Prosthet. Dent.* **2014**, *112*, 1164–1170. [CrossRef] [PubMed]
43. Belli, R.; Wendler, M.; Zorzin, J.I.; Lohbauer, U. Practical and theoretical considerations on the fracture toughness testing of dental restorative materials. *Dent. Mater.* **2018**, *34*, 97–119. [CrossRef]
44. Fraga, S.; Valandro, L.F.; Bottino, M.A.; May, L.G. Hard machining, glaze firing and hydrofluoric acid etching: Do these procedures affect the flexural strength of a leucite glass-ceramic? *Dent. Mater.* **2015**, *31*, e131–e140. [CrossRef]
45. Barchetta, N.; Amaral, M.; Prochnow, C.; Rodrigues, F.; Bottino, M.; Valandro, L.; De Melo, R.M. Strength of a Zirconia-Reinforced Lithium Silicate Ceramic: Acid-Etching Time and Resin Cement Application Effects. *Int. J. Periodontics Restor. Dent.* **2019**, *39*, 431–437. [CrossRef]
46. Venturini, A.B.; Prochnow, C.; Pereira, G.K.R.; Werner, A.; Kleverlaan, C.J.; Valandro, L.F. The effect of hydrofluoric acid concentration on the fatigue failure load of adhesively cemented feldspathic ceramic discs. *Dent. Mater.* **2018**, *34*, 667–675. [CrossRef]

47. Prochnow, C.; Venturini, A.B.; Grasel, R.; Bottino, M.C.; Valandro, L.F. Effect of etching with distinct hydrofluoric acid concentrations on the flexural strength of a lithium disilicate-based glass ceramic. *J. Biomed. Mater. Res. Part B Appl. Biomater.* **2017**, *105*, 885–891. [CrossRef]
48. Addison, O.; Marquis, P.M.; Fleming, G.J.P. Resin strengthening of dental ceramics-The impact of surface texture and silane. *J. Dent.* **2007**, *35*, 416–424. [CrossRef]
49. Monteiro, J.B.; Oliani, M.G.; Guilardi, L.F.; Prochnow, C.; Rocha Pereira, G.K.; Bottino, M.A.; de Melo, R.M.; Valandro, L.F. Fatigue failure load of zirconia-reinforced lithium silicate glass ceramic cemented to a dentin analogue: Effect of etching time and hydrofluoric acid concentration. *J. Mech. Behav. Biomed. Mater.* **2018**, *77*, 375–382. [CrossRef]
50. Soares, L.D.; Basso, G.R.; Spazzin, A.O.; Griggs, J.; Moraes, R.R. Mechanical reliability of air-abraded and acid-etched bonded feldspar ceramic. *Dent. Mater.* **2016**, *32*, 433–441. [CrossRef]
51. Xiaoping, L.; Dongfeng, R.; Silikas, N. Effect of etching time and resin bond on the flexural strength of IPS e.max Press glass ceramic. *Dent. Mater.* **2014**, *30*, e330–e336. [CrossRef] [PubMed]
52. ISO 6872:2015; Dentistry—Ceramic Materials. International-Organization-for-Standardization: Geneva, Switzerland, 2015.
53. Shimada, Y.; Yamaguchi, S.; Tagami, J. Micro-shear bond strength of dual-cured resin cement to glass ceramics. *Dent. Mater.* **2002**, *18*, 380–388. [CrossRef] [PubMed]
54. Quinn, J.B.; Quinn, G.D. A practical and systematic review of Weibull statistics for reporting strengths of dental materials. *Dent. Mater.* **2010**, *26*, 135–147. [CrossRef] [PubMed]
55. Vichi, A.; Zhao, Z.; Paolone, G.; Scotti, N.; Mutahar, M.; Goracci, C.; Louca, C. Factory Crystallized Silicates for Monolithic Metal-Free Restorations: A Flexural Strength and Translucency Comparison Test. *Materials* **2022**, *15*, 7834. [CrossRef]

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.

MDPI AG
Grosspeteranlage 5
4052 Basel
Switzerland
Tel.: +41 61 683 77 34

Materials Editorial Office
E-mail: materials@mdpi.com
www.mdpi.com/journal/materials



Disclaimer/Publisher's Note: The title and front matter of this reprint are at the discretion of the Guest Editors. The publisher is not responsible for their content or any associated concerns. The statements, opinions and data contained in all individual articles are solely those of the individual Editors and contributors and not of MDPI. MDPI disclaims responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.



Academic Open
Access Publishing

mdpi.com

ISBN 978-3-7258-7582-5