

MECHANICAL BEHAVIOUR OF THREE-UNIT ZIRCONIA INFRASTRUCTURES WITH MODIFIED COPINGS

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ABSTRACT

Aim of the study The purpose of this in vitro study was to analyse the influence of various types zirconia copings on flexural strength of zirconia frameworks designated for posterior tooth-supported fixed dental prostheses.

Materials and methods Zirconia samples (Pre-sintered Zirconia, 3Y-TZP, NOVAZir® ST, Novadent, Hamburg, Germany) were designed as three-unit bridge infrastructures, replacing the first right lower molar. The samples (n=4) were divided into two testing groups: the first group included two samples with copings having even thickness of 0.6mm (coded P1 and P2), and second group (coded P3 and P4) comprised modified copings with thickness varying from 0.80-0.84mm in specific occlusal-axial areas. The cross sectional area of all connectors was established at 9mm² with an elliptical shape. Mechanical tests (three-point flexural tests) were conducted on all samples using a universal testing machine. **Results** The evaluation of zirconia samples' mechanical behaviour revealed group-dependent differences: zirconia samples with modified copings demonstrated higher flexural strength in the applied mechanical tests (0.894kN for P3 and 1.109kN for P4) compared to zirconia samples with even coping thickness (0.561kN for P1 and 0.772kN for P2). **Conclusions** Modified zirconia copings have shown potential as an alternative to enhance the flexural strength of zirconia infrastructures, and could provide a more suitable support and an even thickness for veneering ceramic.

Key words: zirconia, modified copings, flexural strength, dental prosthesis, CAD-CAM

INTRODUCTION

Computerized technology has an increasingly significant applicability in modern dental practice, and zirconia represents one of the widely used bio-materials for the fabrication of dental prosthetic restorations. Zirconia (Zirconium dioxide / ZrO₂) has been developed as an alternative for

metal-free infrastructures for fixed dental prostheses (FDPs) due to its favourable mechanical properties related to the transformation toughening mechanism. Moreover, it exhibits superior physical characteristics, chemical stability, as well as good biocompatibility and aesthetic properties [1,2].

Zirconia is increasingly gaining popularity, with its usage expanding across a wide range of applications; this trend is further facilitated by the introduction of highly translucent zirconia variants, enabling monolithic processing [3,4]. However, the design and dimensions of FDPs must be customized based on their location within the dental arch (topographical considerations) and the particularities of the clinical case, as these factors are essential for the long-term survival of the fixed prostheses [3]. Poor mechanical behaviour of zirconia-based FDPs often occurs not only from issues with veneered porcelain but also from the zirconia infrastructures [5]. The failure of veneered porcelain (chipping problems or fracture) could be attributed to various factors, including porosities and surface conditions of the porcelain, excessive occlusal or lateral load, the presence of parafunctions, low bond strength between zirconia coping and porcelain, improper zirconia infrastructure design, lack of coping support for veneered porcelain, and thermal stress during porcelain firing. [5]. On the other hand, the mechanical behaviour of zirconia infrastructures is influenced by various elements, such as: the design/shape and thickness of the zirconia copings [6,7], the shape and cross-sectional surface of rigid connectors [8-13], the type of dental abutment preparation (anatomical reduction, rounded or sharp edges, with or without cervical shoulder) [14,15], the thickness of the spacer [16,17], the applied thermo-mechanical treatments [18,19], the grinding processing of zirconia [8], or the applied luting agent [5].

In the initial phase of introducing the high-strength structural ceramic Y-TZP, copings were typically fabricated with a uniform thickness that conformed to the prepared tooth. This practice could lead to chipping or fracturing of the veneered porcelain, as its thickness would vary depending on the configuration of the prepared

tooth [5]. The use of a modified zirconia coping design enables the achievement of uniform porcelain thickness, providing porcelain with resistance to occlusal and lateral loads and thereby reducing the risk of fracture [5]. Scientific literature findings highlight that an anatomically oriented framework design, combined with an appropriately uniform porcelain thickness, has the potential to enhance the clinical performance of zirconia-based FDPs, effectively reducing the incidence of porcelain chipping or fracture [20-22]. In the light of the aforementioned aspects, the purpose of this *in vitro* experimental study was to investigate the influence of zirconia modified coping designs on the fracture load of tooth-supported three-unit zirconia infrastructures from the posterior area of the mandibular dental arch. The study hypothesis was that the modified design of zirconia copings would positively influence the fracture load of the analysed three-unit zirconia infrastructures.

MATERIALS AND METHODS

Sample preparation

A didactic mandibular plaster model (coded PM) (**Fig. 1**), from the authors' archive, was scanned using an extraoral scanner (E4 Crown & Bridge 3Shape Extraoral Scanner, 3Shape, Copenhagen, Denmark), resulting in a virtual dental model. From this, a segmental digital dental arch model (representing abutment teeth 4.5 and 4.7 – both prepared for ceramic-veneered zirconia crowns, and 4.6 – edentulous space) was obtained. Based on this segmental digital model, a Cr-Co metallic alloy segmental model (coded SM) (NOVACHROME®, Novadent, Hamburg, Germany) (**Fig. 1**) was obtained by computerized milling. Successively, the Cr-Co model was scanned with the same scanner used previously, resulting in the virtual segmental model (coded VSM) (**Fig. 1**). Subsequently, based on the virtual segmental

model (VSM), four zirconia samples (Pre-sintered Zirconia, 3Y-TZP, NOVAZir® ST, Novadent, Hamburg, Germany) were produced using CAD-CAM technology.

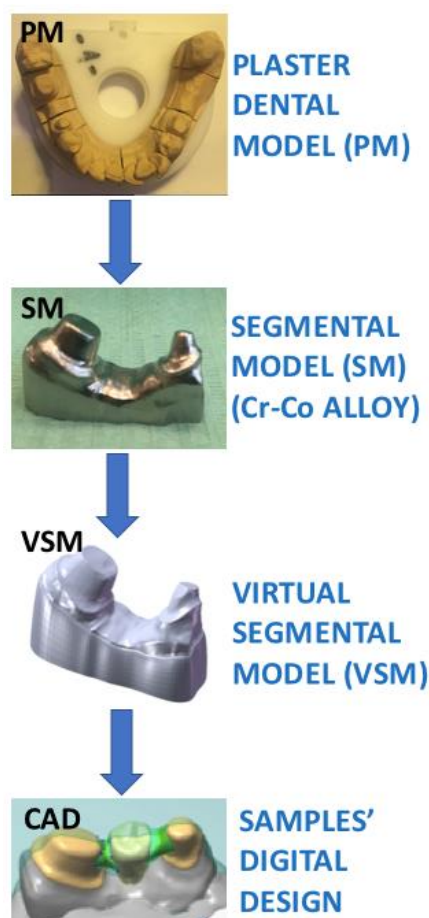


Figure 1. Initial sequences in obtaining zirconia samples

The zirconia samples represented three-unit bridge infrastructures replacing the first right lower molar (4.6); the pontics were designed as premolars due to the narrow edentulous space, and the rigid connectors had a cross-sectional area of 9mm² with an elliptical shape. Two of the zirconia infrastructures (coded P1 and P2) were designed with even thickness copings of 0.6mm (**Fig. 2**), while the other two zirconia infrastructures (coded P3 and P4) had modified copings (**Fig. 3**). These copings had a greater thickness (varying from 0.80-0.84mm) in specific occlusal-axial areas: occlusal-distal (OD), occlusal-lingual (OL), and occlusal-mesial (OM) areas (**Fig. 3**). These areas were deliberately selected to be

thickened at the zirconia coping level since, in previous studies on the mechanical behaviour of zirconia infrastructures [6-13,21], these specific zones exhibited the lowest mechanical resistance and were the first to fail in mechanical tests.

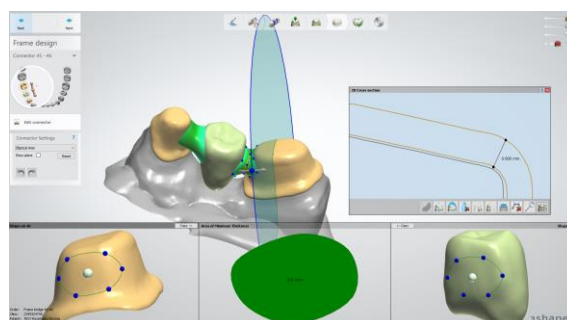


Figure 2. Virtual design for P1 and P2 zirconia samples (coping thickness of 0.6mm)

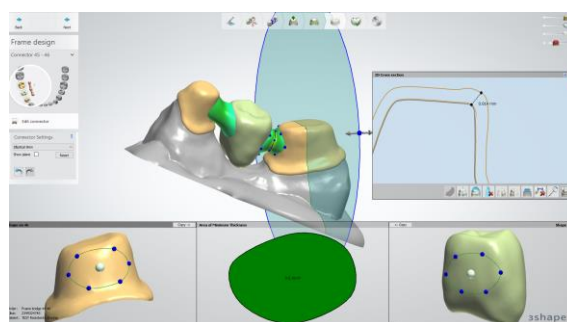


Figure 3. Virtual design for P3 and P4 zirconia samples (modified copings of 0.80-0.84mm in occlusal-axial areas)

The resulting zirconia infrastructures were subsequently sintered in an electric furnace (iSINT HT, Mihm-Vogt GmbH & Co KG, Stutensee-Blankenloch, Germany), following the manufacturer's specifications. The obtained zirconia infrastructures can be observed in **Fig. 4**.

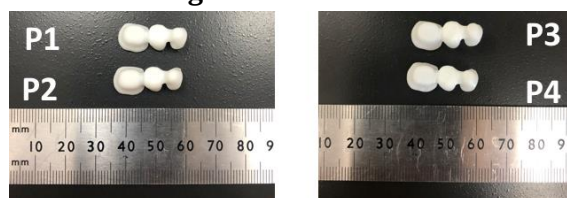


Figure 4. The obtained zirconia samples

Flexural tests

The 3-point flexural tests were performed on all zirconia samples corresponding to the three-unit fixed dental

prosthesis, using a universal testing machine (LFV 300kN, Walter+Bai AG, Löhningen Switzerland). Each of the four samples was placed onto the Cr-Co metallic alloy segmental model (SM) and the load was applied on the pontic, at its highest feature (the tip of the buccal cusp), using a steel cylindrical roller (of 20mm), in order to reproduce the occlusal force. The test was performed by applying a 1mm/min crosshead speed until complete fracture of samples occurred (**Fig. 5**).

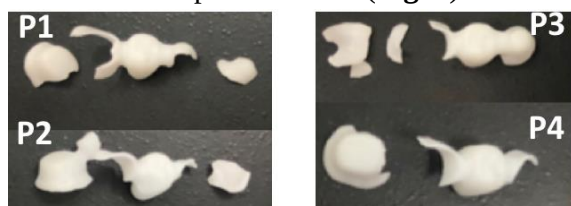


Figure 5. The fractured zirconia samples

RESULTS

Table I contains values of forces at both initial and at catastrophic failure of samples obtained during the mechanical tests (three-point flexural tests). Additionally, the curves obtained during the performed mechanical tests for all zirconia samples are presented in **Fig. 6-9**.

Table 1. The values of forces obtained during the mechanical testing of zirconia infrastructures

Sample	Slope kN/mm	Initial failure/kN	Catastrophic failure/kN
P1	4.657	0.561	1.426
P2	4.86	0.772	0.812
P3	5.861	0.894	1.247
P4	6.471	1.109	1.241

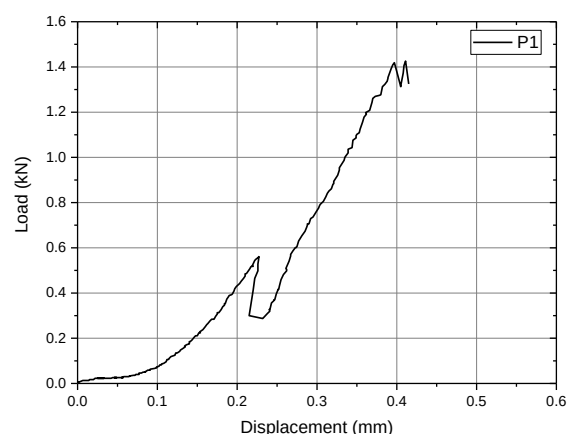


Figure 6. Load-displacement curve for P1

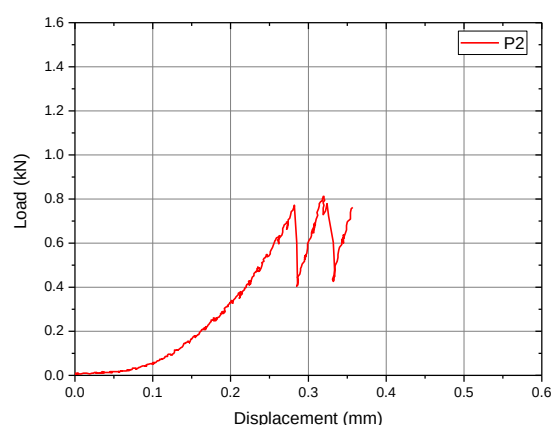


Figure 7. Load-displacement curve for P2

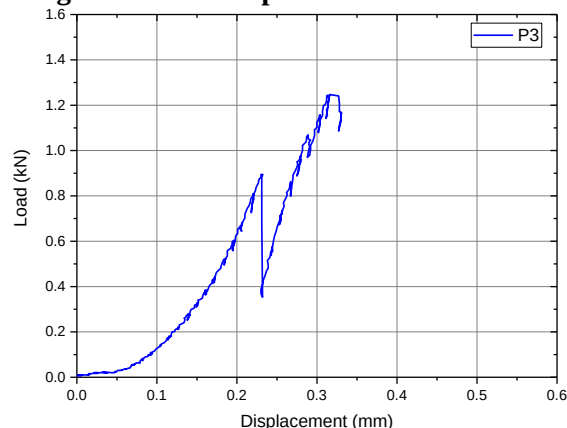


Figure 8. Load-displacement curve for P3

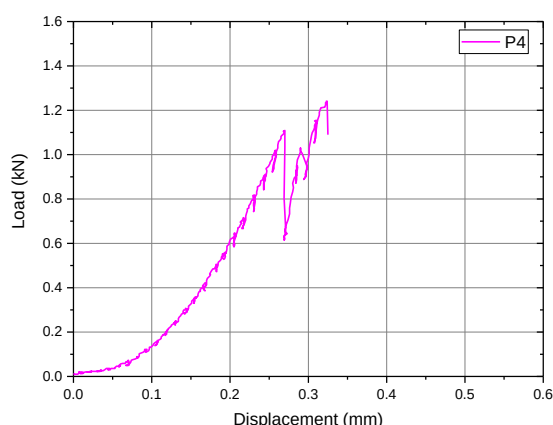


Figure 9. Load-displacement curve for P4

The obtained results were analysed with respect to the study hypothesis. The analysis of each curve corresponding to each zirconia sample (**Fig. 6-9**), showed a brittle behaviour of the material, without significant plastic deformations. The behaviour of the samples in the flexural tests indicates that failure occurs in two stages.

A significant initial failure occurs at a value of approximately 0.561kN for P1, 0.772kN for P2, 0.894kN for P3, and 1.109kN for P4. This initial failure is the most significant one, as it appears when the samples fail immediately after exceeding the linear-elastic region. In the second stage, stress redistribution occurs, which can be observed by the change in the slope of the curve (the zirconia infrastructure is supported by the Cr-Co model and continues to "resist"). Thus, catastrophic failure occurs at the maximum force on the curve: 1.426kN for P1, 0.812kN for P2, 1.247kN for P3 and 1.241kN for P4.

Comparative analysis of the results obtained under different bending loads on zirconia infrastructures with different designs indicates a group-depended rigidity of samples P1 and P2, respectively P3 and P4. Sample P2 fails in the first stage at a force value 0.161kN (approx. 16.1kgF) higher than that of sample P1, and catastrophic failure of P2 occurs at values close to those of the force for the initial failure, at considerably lower values than those of sample P1. On the other hand, sample P4

fails in the first stage at a force 0.215kN (approximately 21.5kgF) higher than that of sample P3, and catastrophic failure occurs at nearly equal force values. In a general comparison (**Fig. 10**), it is evident that samples P3 and P4 have similar rigidity, higher than that of samples P1 and P2. The second type of design (corresponding to samples P3, P4) creates the conditions for a more favourable stress distribution than the first configuration (corresponding to samples P1 and P2), with samples P3 and P4 resisting higher forces in the flexural test.

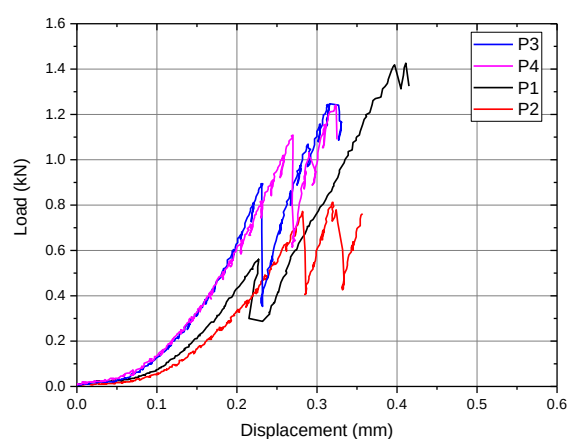


Figure 10. Load-displacement curve for all zirconia samples

DISCUSSIONS

In light of the obtained and compared results a superior performance can be observed for the samples with the second configuration (P3, P4 – with modified copings) indicating considerably higher force values at the initial failure compared to those in the first configuration (P1, P2 – with 0.6mm thickness copings). Samples P3 and P4 are also more rigid, implying a more uniform distribution of stresses throughout the entire assembly. The most favourable situation was identified for sample P4, exhibiting the highest rigidity and bearing the greatest load at the initial failure. The study hypothesis, that the modified design of zirconia copings would positively influence the fracture load of the analysed three-unit zirconia infrastructures was accepted.

The mechanical strength and biocompatibility characteristics of zirconia recommend this material for use in the fabrication of fixed dental prostheses, particularly in the posterior region of dental arches [23-25]. Various factors play a role in achieving successful zirconia-based dental prostheses [26,27]. The outcomes of our in vitro study align with information found in scientific dental literature, where certain studies emphasize the significance of a proper design of zirconia infrastructure to enhance flexural resistance. For example, in 2012, Kim et al. [28] demonstrated that zirconia infrastructures with a 0.7mm thickness coping exhibit higher fracture strength than infrastructures with a 0.5mm thickness coping. Similarly, Sawada et al. (2016) [6] indicated that modifying the zirconia infrastructure by thickening the zirconia copings or adopting a suitable design significantly increases the prosthetic restoration's fracture resistance. The authors demonstrated that, due to these modifications, complete fractures can be prevented or only incomplete fissures occur within the thickness of the restoration. Moreover, anatomical zirconia copings with a modified design reduce stress concentration at the interface with veneered ceramic (Anami et al., 2015) [29]. In a recent study, Koltz et al. (2024) [3] indicated that zirconia 3Y-TZP infrastructures, with thicknesses ranging from 0.7-1mm, are recommended for posterior tooth-supported cantilever fixed dental prostheses; the authors specified that the use of FDPs manufactured by gradient technology (4Y-PZS or 5Y-PZS) "may not be unreservedly recommended" for this specific clinical use.

From another perspective, a partially fenestration of zirconia copings reduces the resistance of a prosthetic restoration without adversely affecting the colour of the final prosthesis [7]; such a partially fenestrated design may be accepted as a variation of the

traditional coping design, with overall restoration strength being clinically acceptable. Additionally, Rosentritt et al. [16] assert that the correct design of the zirconia infrastructure enhances its fracture resistance, yet they did not observe a significant influence of different thicknesses of the space between the abutment and the restoration ("spacer") on the overall resistance of the structure.

Furthermore, Ramos et al. [30] concluded in their in vitro study that, following cyclic mechanical stress on zirconia crowns, out of the three crown model groups: traditional crown (feldspathic ceramic on zirconia infrastructure), modified crown (partial veneering of the zirconia infrastructure with feldspathic ceramic), and full-monolithic zirconia crown, the last two groups showed significantly better results. Additionally, another relatively recent study in 2020 [17] indicates that monolithic zirconia fixed prostheses with reduced occlusal thickness of 0.5 mm, made on abutments without a gingival margin preparation, exhibited higher fracture resistance compared to those made on chamfer-type gingival limit abutments. Despite the reduced resistance of prosthetic restorations with reduced occlusal thickness compared to those with optimal thickness, the study demonstrated that they can tolerate occlusal forces.

At the same time, the marginal collar of zirconia copings is considered an important element in the infrastructures' design, influencing the overall resistance of the prosthetic structure. In 2011, Cho et al. [31] concluded that a zirconia coping design with a 2.0mm marginal collar width is recommended for long-term longevity and aesthetics of ceramic-veneered zirconia crowns. Later, in 2016, the studies of Ha et al. [32,33] suggested that the cervical collar height should be greater than 3mm, especially on the lingual side, to prevent fracture of posterior prosthetic restorations and reduce stress in veneering

ceramics. In a relatively recent, Shetty et al. (2020) [34] indicated that a minimum collar height of 1mm is required to achieve better fracture strength of ceramic-veneered zirconia crowns, but for obtaining optimum fracture strength, a 1.5mm collar height is recommended.

Obtaining a successful zirconia-based dental prosthetic restoration is also significantly dependent on various processing parameters such as milling, sintering, and finishing [35]. Understanding the mechanical characteristics of zirconia infrastructures is essential for achieving optimal performance in every clinical scenario. The present work aligns with data found in dental literature and also has clinical-practical relevance: depending on the needs and specificities of the clinical case, incorporating modified zirconia copings into the prosthetic design could enhance the mechanical strength of the prostheses, facilitate a more uniform redistribution of forces, and consequently, provide greater overall resistance for the dental prosthesis. This study could contribute to guiding dental practitioners in selecting the proper design for zirconia infrastructures for each distinct clinical case. Nevertheless, the absence of a statistical analysis, the limited number of tested samples in the mechanical tests, along with the fact that the study does not

simulate all oral conditions, represent limitations of the present study.

CONCLUSIONS

Following the analysis of the obtained results and considering the limitations of this study, we reached the following conclusions:

1. Samples P3 and P4 (zirconia infrastructures with modified thickness copings) exhibited superior mechanical behaviour when compared to samples P1 and P2 (zirconia infrastructures with 0.6mm thickness copings);
2. Samples P3 and P4 showed considerably higher force values at initial failure (0.894kN and 1.109kN) compared to P1 and P2 samples (0.561kN and 0.772kN);
3. In comparison to samples P1 and P2, samples P3 and P4 also demonstrated increased rigidity, leading to a more uniform stress distribution throughout the entire infrastructure;
4. There is potential for modified zirconia copings to serve as an alternative, contributing to the improvement of fracture resistance of zirconia infrastructures and, consequently, offering better support for veneering ceramic

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