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ORIGINAL RESEARCH

Stress distribution in stainless steel crowns and zirconia crowns depending on different cement types: a finite element analysis

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Abstract

Background: A gap exists in the literature concerning the stress distribution in stainless steel crowns (SSCs) and zirconia crowns (ZrCs) when luted with different cement types. This study aimed to compare the stress distributions of SSCs and ZrCs luted with different cements using finite element analysis (FEA). **Methods**: Eight FEA models were created using two restoration types (SSC and ZrC) and four cement types: glass ionomer cement (GIC), resin-modified glass ionomer cement (RMGIC), zinc polycarboxylate cement (ZPC), and resin cement (RC). Stresses on crowns, teeth, and cements were analyzed. **Results**: In the SSC models, peak stresses on the crowns were concentrated at the functional cusp tips, whereas peak stresses on the tooth surface and luting cements were concentrated near the crown margins. Peak crown stress values were 31.21 MPa for SSC-GIC, 40.63 MPa for SSC-RMGIC, 35.97 MPa for SSC-ZPC, and 28.72 MPa for SSC-RC. Corresponding peak stresses on the tooth surface were 22.38, 16.54, 18.58, and 24.79 MPa, while peak stresses in the cement layer were 8.22, 5.50, 6.21, and 9.97 MPa, respectively. In the ZrC models, peak stresses on the crowns, tooth surface, and luting cements were concentrated near the crown margins. Peak crown stress values were 64.24 MPa for ZrC-GIC, 55.35 MPa for ZrC-RMGIC, 58.60 MPa for ZrC-ZPC, and 68.64 MPa for ZrC-RC. Peak stresses on the tooth surface were 22.60, 23.27, 22.98, and 22.37 MPa, while cement layer stresses were 18.23, 11.52, 13.11, and 22.36 MPa, respectively. **Conclusions**: The type of luting cement influenced stress levels and distributions within the crown, cement layer, and tooth structure in both SSC and ZrC models.

Keywords

Finite element analysis; Luting cement; Mixed dentition; Prefabricated pediatric crown; Stainless steel crown

1. Introduction

Stainless steel crowns (SSCs) and prefabricated zirconia crowns (ZrCs) are commonly recommended for pediatric patients to ensure long-term durability, occlusal function, and protection of structurally compromised teeth [1, 2]. These crowns are particularly useful in managing first permanent molars affected by extensive caries or developmental enamel defects such as molar-incisor hypomineralization [3]. Age, oral hygiene, defect size, and occlusal factors are important considerations in selecting the restoration material [3, 4]. SSCs are widely used due to their excellent durability, cost-effectiveness, and single-visit applicability [1]. ZrCs have gained popularity as aesthetic and biocompatible alternatives [5].

Permanent mandibular first molars are the earliest erupting permanent teeth and are highly susceptible to caries and developmental enamel defects during the mixed dentition period. Due to their early exposure to the oral environment, high occlusal load, and frequent structural compromise, these teeth commonly require full-coverage restorations in pediatric patients [3]. Therefore, the use of restorative systems and luting agents with reliable biomechanical performance is critical. In this context, understanding the biomechanical behavior of crown-cement-tooth complexes is essential to optimize restorative success [3]. Cementation is critical to the success of pediatric crowns [1]. Luting cements affect crown retention, marginal seal, stress distribution, and risk of failure under masticatory forces [6–8]. Glass ionomer cement (GIC), resin-modified glass ionomer cement (RMGIC), zinc polycarboxylate cement (ZPC), and resin cement (RC) are commonly used luting cements [5–7]. Studies suggest that RMGIC improves internal adaptation in SSCs, while RC enhances retention in ZrCs [6, 7]. However, the ideal cement for use with pediatric crowns remains a matter of clinical debate.

These cement types differ markedly in physicochemical and mechanical properties, including modulus of elasticity, polymerization characteristics, adhesive mechanisms, and humidity tolerance. Such differences are likely to influence stress transmission and dissipation patterns within the restored tooth structure. Despite this, while existing literature provides data on bond strength and clinical retention of different cements, few studies have investigated their biomechanical behavior within pediatric crowns [8].

Finite element analysis (FEA) provides an effective method for assessing biomechanical behavior under functional loads, offering valuable insights into stress and strain distributions within complex structures [9]. FEA enables the simulation of standardized loading conditions and controlled material environments that cannot be ethically or practically replicated *in vivo*. Furthermore, it allows precise comparison of material-dependent stress distributions by isolating variables that are difficult to control in laboratory experiments. Given the limited evidence on how different luting cements affect the biomechanical response of SSCs and ZrCs in young permanent molars, FEA is an appropriate and robust analytical tool. This study aimed to investigate stress levels and distributions in permanent first molars restored with SSCs and ZrCs, luted with different cement types, using FEA. The null hypothesis of this study was that cement type would not affect stress levels or stress distribution patterns within the same crown type.

2. Materials and methods

2.1 Study design

In this study, 8 FEA models were created using two pediatric crown materials (SSC and ZrC) and 4 luting cements: GIC, RMGIC, ZPC, and RC. The resulting groups were as follows: SSC-GIC, SSC-RMGIC, SSC-ZPC, SSC-RC, ZrC-GIC, ZrC-RMGIC, ZrC-ZPC, and ZrC-RC. The chemical compositions of the materials used are summarized in Table 1.

2.2 Modeling

Two identical artificial mandibular permanent first molars (Frasaco GmbH, Tettnang, Germany) were prepared for SSC and ZrC restorations. Preparations were performed using an airotor (Extra Torque 506C, Kavo, Kavo Dental GmbH & Co KG, Biberach, BW, Germany) and a flame-tipped bur (Komet, Paris, France). Crown preparations were planned according to the manufacturer's instructions, with both designs intended to be placed 1 mm subgingivally using a knife-edge finish. For the SSC, preparation depths were 1 mm at the developmental grooves on the occlusal surface, 1.5 mm at the cusp tips and cusp ridges, 1 mm in the proximal areas, and 0.5 mm on the buccal and lingual surfaces [10, 11]. For the ZrC, preparation depths were 1.5 mm on the occlusal surface, 1 mm in the proximal areas, and 1 mm on the buccal and lingual surfaces, with all undercuts removed. SSC (3M ESPE Dental Products, St. Paul, MN) and ZrC (NuSmile, NuSmile Ltd., Houston, TX) crowns were selected in sizes suitable for the teeth. The prepared teeth and selected crowns were scanned with a desktop scanner (Vinyl High Resolution, Smartoptics, Oslo, Norway). Scan data were imported into Geomagic Design X (version 2020, 3D Systems, Andover, MA, USA) in Standard Tessellation Language (STL) format. Solid models were generated using various solidification methods for prefabricated crowns and tooth structures (Fig. 1). The modeled tooth represents a vital, fully matured permanent mandibular first molar.

In the SSC models, it was assumed that 0.5 mm of enamel remained on the buccal and lingual surfaces after tooth preparation, and the enamel was modeled accordingly. In the ZrC models, no enamel was modeled, as the preparation was assumed to have completely removed the enamel tissue. For bone and soft tissue modeling, a mandible was obtained by sectioning a virtual skull model with SolidWorks software (version 2022, SolidWorks Inc., Concord, MA, USA). Using

TABLE 1. Chemical composition of the materials used in the study.

Material	Main Chemical Composition
Artificial tooth	Cross-linked thermoplastic resin
Stainless steel crown	Stainless steel alloy primarily composed of Fe 66.45 wt%, Cr 17.84 wt%, Ni 8.06 wt%, C 4.81 wt%, Al 2.05 wt%.
Zirconia crown	ZrO_2 80–96 wt%, Y_2O_3 4–10 wt%, HfO_2 <5 wt%, organic binder <5 wt%, pigments 1–4 wt%.
Zinc polycarboxylate cement	Powder: zinc oxide (ZnO) 85–95 wt%, stannous fluoride 1–10 wt%. Liquid: water 50–65 wt%, polyacrylic acid 40–50 wt%.
Glass ionomer cement	Powder: fluoro-aluminosilicate glass. Liquid: water 80–95 wt%, tartaric acid <20 wt%.
Resin cement	Organic resin matrix composed of Bis-GMA, UDMA, and/or TEGDMA, reinforced with inorganic fillers (silica, glass, or zirconia particles) and polymerization initiators.
Resin-modified glass ionomer cement	Paste A is composed of a radiopaque fluoroaluminosilicate glass, opacifying agent, hydroxylethyl methacrylate (HEMA), water, dispersion aid, and a reducing agent that allows for the self-cure methacrylate setting. Paste B is composed of nonreactive zirconia silica filler, methacrylated polycarboxylic acid, HEMA, resin monomers, water, potassium persulfate, and a photoinitiator.

Bis-GMA: Bisphenol A-glycidyl methacrylate; UDMA: Urethane dimethacrylate; TEGDMA: Triethylene glycol dimethacrylate; ZrO_2: zirconium dioxide; Y_2O_3: yttrium oxide; HfO_2: hafnium dioxide.

FIGURE 1. Solid model of the prepared tooth, pulp chamber, cement layer, and prefabricated crown.

this mandible model as a reference, solid modeling of cortical bone, trabecular bone, and gingival soft tissue was performed using surface modeling techniques (Fig. 2).

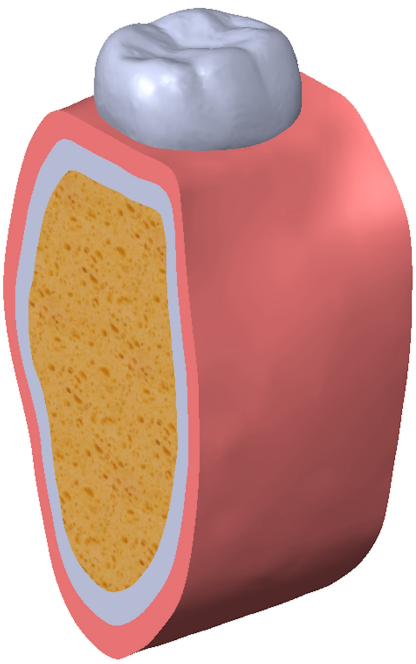

FIGURE 2. Solid model of cortical bone, trabecular bone, and gingival soft tissue.

A luting cement layer with a uniform thickness of 25 μm was modeled as a continuum solid layer between the crowns and the prepared tooth surfaces [12]. All structures were assembled into a single solid model that reflected their true morphology. The SolidWorks models were transferred to Ansys 18.1 (ANSYS Inc., Canonsburg, PA, USA) while preserving the 3D coordinates. Elastic modulus and Poisson's ratio values were defined for each of the mathematically modeled structures in Ansys software (Table 2, Ref. [13–16]). The mesh with 10-node quadratic tetrahedral elements was created. A mesh convergence analysis was performed by progressively refining the element size until further refinement resulted in negligible changes in the calculated stress values. The resulting numbers of nodes and elements were 774,814 and 509,830 for the SSC models, and 820,956 and 546,126 for the ZrC models, respectively. All models were assumed to be homogeneous, isotropic, and linear elastic materials.

TABLE 2. Elastic modulus and Poisson ratios.

Material	Elastic modulus (E) (GPa)	Poisson's ratio (μ)
Glass-ionomer cement (GIC) [13]	7.56	0.35
Resin-modified GIC [14]	3.70	0.30
Zinc polycarboxylate cement [13]	5.11	0.35
Resin cement [14]	9.60	0.33
Stainless steel crown [15]	210.00	0.33
Zirconia crown [13]	205.00	0.19
Enamel [15]	84.10	0.30
Dentin [15]	18.60	0.31
Cortical bone [16]	13.70	0.30
Trabecular bone [16]	1.37	0.30

2.3 Boundary and loading conditions

The models were fixed so that there was no movement from the lateral surface of the mandibular cross-section, and all modelled structures were firmly attached. The stress distribution within the models was evaluated under a loading condition that represents the bite forces of children aged 7–9 years, a clinically appropriate age group, as the treatment targeted pediatric patients with mixed dentition [17]. For this purpose, a combined occlusal force of 120 N was applied obliquely at a 30° angle in the buccolingual direction to mimic natural occlusal dynamics (Fig. 3).

2.4 Analysis

In FEA, structural integrity is evaluated by post-processing the solutions obtained for each substructure. Since the results obtained from FEA are based on mathematical calculations and do not exhibit variance, statistical analysis was not applicable [18]. Therefore, the evaluation was based on stress magnitudes at the nodes, cross-sectional views, and stress distribution patterns. Von Mises (vM) stress values were used to analyze stresses in crowns, teeth, and cements. For the analysis, the highest stress values in the structures were considered and recorded in megapascals (MPa).

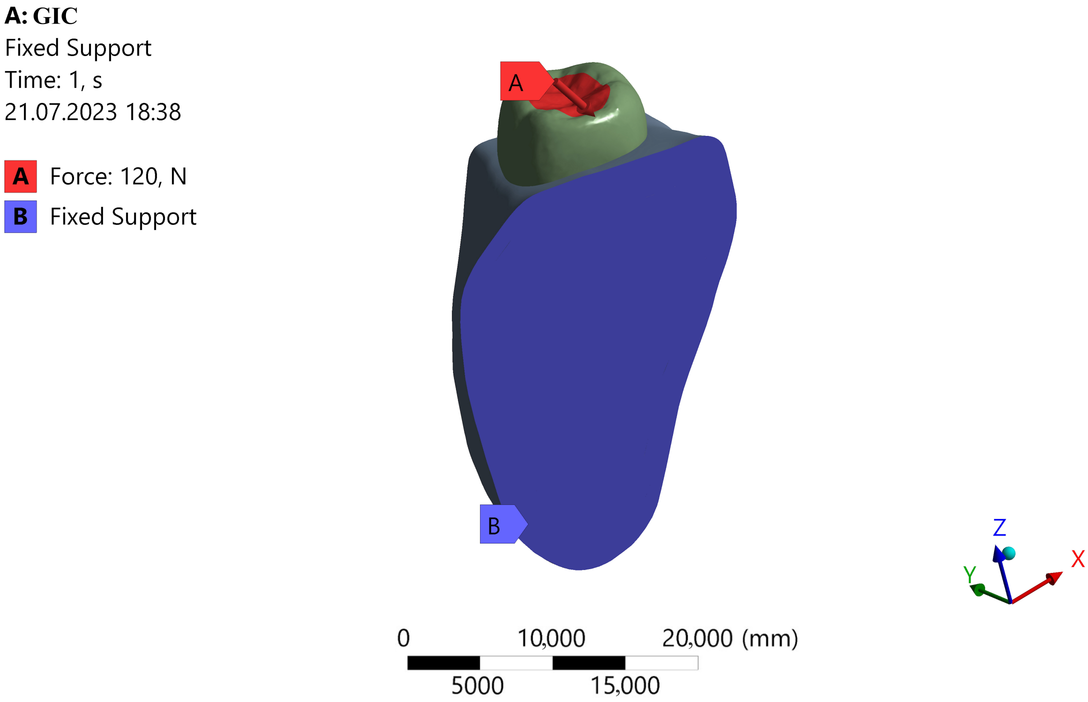

FIGURE 3. Application of bite force: the red arrow shows the direction of the force, and the red area on the occlusal area of the teeth indicates the region where the force was applied. GIC: glass ionomer cement.

3. Results

The peak vM stress values for the crowns, tooth surfaces, and cement layers are presented in Table 3. Generally, the highest stress values were observed in the crowns, followed by the tooth surfaces and the cement layers.

In the SSC models, peak stresses on the crowns were concentrated at the functional cusp tips, whereas peak stresses on the tooth surface and luting cements were concentrated near the crown margins (Fig. 4). Peak vM stress values on the crowns increased with decreasing elastic modulus of the luting cement. In contrast, the tooth surface and the cement layer exhibited opposite trends. Accordingly, peak vM stress values on the crowns were ranked as follows: SSC-RMGIC > SSC-ZPC > SSC-GIC > SSC-RC. For both the tooth surface and the cement layer, peak vM stress values showed a similar ranking: SSC-RC > SSC-GIC > SSC-ZPC > SSC-RMGIC (Table 3).

In the ZrC models, peak stresses on the crowns, tooth surface, and luting cements were concentrated near the crown margins (Fig. 5). Peak vM stress values on the crowns and cement layer decreased with decreasing elastic modulus of the luting cement, whereas the tooth surface exhibited the opposite trend. Accordingly, peak vM stress values on the crowns and cement layer were ranked as follows: ZrC-RC > ZrC-GIC > ZrC-ZPC > ZrC-RMGIC. In contrast, peak vM stress values on the tooth surface followed the order: ZrC-RMGIC > ZrC-ZPC > ZrC-GIC > ZrC-RC (Table 3).

4. Discussion

This study evaluated stress levels and stress distribution patterns in permanent first molars restored with SSCs and ZrCs, luted with different cement types, using FEA. The results demonstrated that, within the same crown type, stress levels and the distribution patterns varied depending on the cement used. Consequently, the null hypothesis that cement type would not affect stress levels or stress distribution patterns within the same crown type was rejected.

In FEA, both extracted natural teeth and artificial teeth can be used for geometric modeling [13, 15]. As the geometric models are generated based on scan data, the use of natural or artificial teeth does not affect the structural material properties incorporated into the simulations. However, artificial teeth offer an advantage by eliminating anatomical variability and ensuring geometric standardization across study groups. Therefore, in the present study, artificial teeth were preferred over extracted natural teeth to provide standardized and reproducible finite element modeling conditions.

The present model represents a vital, mature permanent first molar rather than an endodontically treated tooth. This choice reflects common clinical scenarios in pediatric dentistry, particularly in patients with molar-incisor hypomineralization (MIH), where full-coverage restorations are frequently indicated despite the absence of pulpal involvement [3]. Clinical guidelines and previous studies indicate that even noncarious permanent first molars with MIH affecting two or more cusps may require full coronal coverage to prevent posteruptive breakdown and hypersensitivity while preserving pulp vitality [19]. Therefore, the simulation of vital permanent molars in this study was intentionally designed to reflect routine clinical practice in pediatric dentistry, where full-coverage crown restorations are commonly applied to structurally compromised yet vital young permanent teeth. This approach ensures that the finite element model corresponds to a realistic and clinically relevant treatment scenario.

Enamel and dentin tissues do not exhibit uniform density

TABLE 3. Peak vM stress values on crowns, tooth surfaces, and cements (MPa).

Models	Crown	Tooth surface	Cement
SSC-GIC	31.21	22.38	8.22
SSC-RMGIC	40.63	16.54	5.50
SSC-ZPC	35.97	18.58	6.21
SSC-RC	28.72	24.79	9.97
ZrC-GIC	64.24	22.60	18.23
ZrC-RMGIC	55.35	23.27	11.52
ZrC-ZPC	58.60	22.98	13.11
ZrC-RC	68.64	22.37	22.36

SSC: stainless steel crown; GIC: glass ionomer cement; RMGIC: resin-modified glass ionomer cement; ZPC: zinc polycarboxylate cement; RC: resin cement; ZrC: zirconia crown.

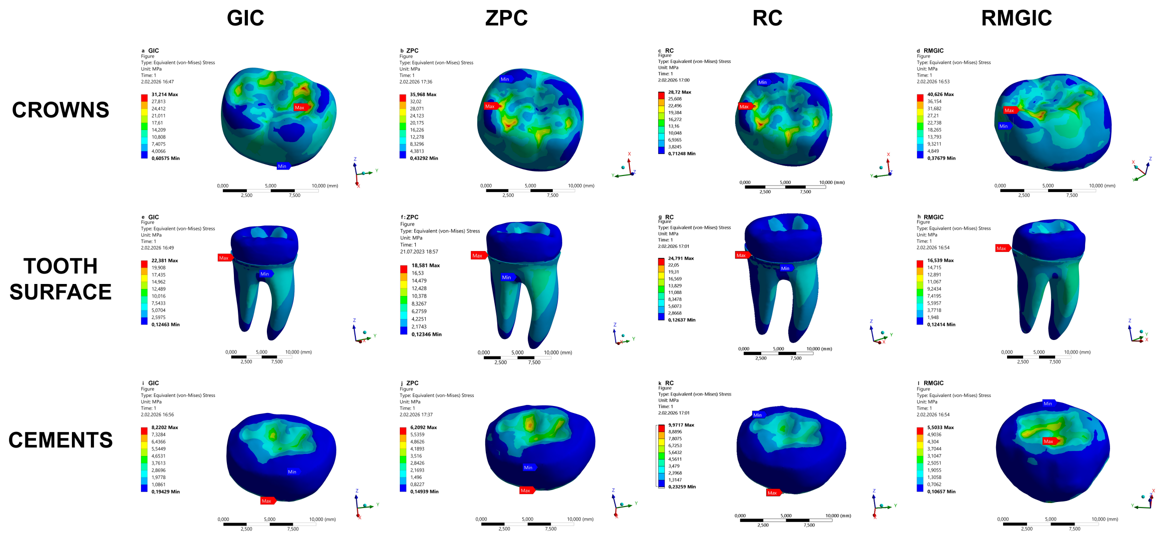

FIGURE 4. vM stress distributions in SSC models luted with different cements. The stress level was presented as a color-coded maps, expressed in MPa. Red represents high stress concentration, and blue represents lower stress concentration. GIC: glass ionomer cement; ZPC: zinc polycarboxylate cement; RC: resin cement; RMGIC: resin-modified glass ionomer cement; Min: Minimum; Max: Maximum.

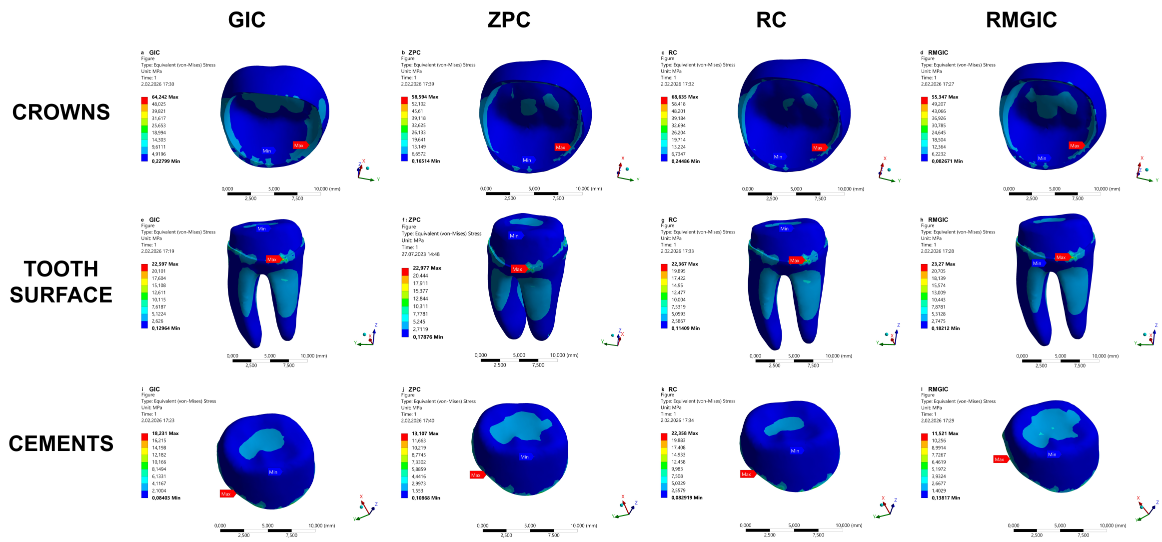

FIGURE 5. vM stress distributions in ZrC models luted with different cements. The stress level is presented as a color-coded map, expressed in MPa. Red represents high stress concentration, and blue represents lower stress concentration. GIC: glass ionomer cement; ZPC: zinc polycarboxylate cement; RC: resin cement; RMGIC: resin-modified glass ionomer cement; Min: Minimum; Max: Maximum.

or structure and are not isotropic [20]. Due to the complex structure of the tooth, it is challenging to produce geometric models with high accuracy. Therefore, as in most FEA studies, all materials were assumed to be homogeneous, isotropic, and linear elastic in the present study [21–23].

FEA can be used to investigate a single variable in a complex structure and identify stress-concentration areas. With the advent of FEA, it has become possible to demonstrate the distribution of stress throughout the tooth and restoration [24]. Knowing the relationship between natural tissues and artificial units placed in the mouth is important for successful clinical treatment. Compared to real model research, FEA has several advantages, such as eliminating the need for ethical considerations and allowing the model to be adjusted to the needs of the study. By analyzing the stress on bone and restoration components, tests can be repeated and used prior to restoration. Finite element analysis (FEA) allows researchers to perform multiple simulations without human experimentation, thereby improving load management, reducing patient discomfort, and supporting long-term clinical performance [25, 26]. The biomechanical study of the application of SSCs and ZrCs to permanent first molars presents several challenges, including their complex 3D structure and interacting variables. Several techniques have been used in clinics and laboratories to analyze the integrity of dental restorative materials under biting force [27]. However, it is difficult to establish a standard for applying these methods in pediatric clinical studies due to ethical concerns. The use of the vM criterion in FEA facilitates the determination of the effects of the combined force on restorations, converts compressive, tensile, and shear stresses into a single stress, and provides meaningful results [27]. Therefore, vM stress values were used in this study to analyze the stresses in crowns, tooth surfaces, and cements.

In the literature, the average masticatory force in the permanent molar region of children aged 7–9 years was reported to range from 94.92 N to 241.39 N [17]. In the present study, a loading force of 120 N, sufficient to produce a biomechanical effect, was preferred to ensure a clinically meaningful and consistent evaluation in line with the existing literature. FEA studies have shown that vertical loads yield similar stress distributions across models, whereas oblique loads produce more pronounced variations [28]. Furthermore, combined loads are recommended because they better mimic the bite directions and produce higher stress on the cortical bone [29]. In the present study, an oblique force was applied to the occlusal surface of the mandibular first molar at a 30° angle in the buccal-lingual direction.

Subgingival preparation for SSCs carries the risk of violating the biological width of the periodontal attachment. On the other hand, subgingival placement of SSCs is often necessary, especially in partially erupted molars with large carious lesions extending subgingivally. The patient and the parents may experience aesthetic concerns due to the metallic appearance, even in the posterior region [30]. Subgingival preparation for ZrCs provides aesthetics with a smooth transition between the crown and the tooth. It also ensures a tight crown fit to the tooth, increasing retention and reducing the possibility of crown dislodgment [31]. In this way, subgingival preparation makes ZrCs long-lasting and durable. However, along with these advantages, potential difficulties such as increased preparation time and the precise technique required to prevent gingival damage should be considered. In the present study, subgingival preparation was preferred for SSCs and ZrCs, as it is also commonly used in routine clinical practice and complies with the manufacturer's instructions for the prefabricated crowns used in the study.

Several studies have examined stress distribution in SSCs according to the type of luting cement used [15, 32]. Consistent with the present study, previous studies reported that peak stresses in SSCs were concentrated at the load application site. Although different luting cements generally yield similar stress distribution patterns, they may alter the magnitude of peak stresses. Furthermore, increasing the elastic modulus of the luting cement is associated with reduced peak stress within the crown and increased stress on the tooth surface, a trend that aligns with the results of the current study [15, 32]. Guler *et al.* [32] reported lower crown stress values in models using GIC, which has a higher elastic modulus than RC. Similarly, Waly *et al.* [15] reported crown stress values in ascending order for zinc phosphate, GIC, RC, and RMGIC, whereas stress values on the tooth surface followed the opposite trend. In agreement with these findings, the present study showed that RMGIC produced the highest stress concentrations in the crown and the lowest on the tooth surface. However, in contrast to previous reports, the current study found that RC resulted in lower peak crown stresses than GIC, accompanied by higher stress on the tooth surface. This discrepancy may be attributed to differences in the elastic modulus values assigned to the cements in the finite element models. Additionally, variations in model characteristics—including element and node density, loading conditions, cement layer thickness, and tooth preparation design—may further explain differences in the reported peak stress magnitudes.

Previous studies investigating the biomechanical behavior of ZrCs with different luting cement types confirmed that variations in cement elastic modulus influence stress distribution [8, 13]. Chung *et al.* [8] reported that RC, which has a higher elastic modulus than RMGIC, produced higher stress within the cement layer, whereas RMGIC produced higher stress within the tooth structure. Similarly, Ha *et al.* [13] suggested that luting cements with lower elastic moduli led to reduced stress concentrations in the cement layer. These findings are consistent with the results of the present study. Furthermore, the localization of peak stresses in ZrCs and within the luting cement layer was consistent with previous studies [8, 13].

Maintaining the integrity of the cement layer is essential, as microfractures can lead to microleakage, bacterial infiltration, or restoration dislodgement [33]. When stresses generated within the cement layer exceed its compressive strength, this structural integrity may be compromised. In the present study, the peak vM stress values observed in the luting cements were lower than the compressive strengths reported in the literature [34, 35], suggesting a low risk of cement failure under the applied loading conditions. However, each luting cement exhibits inherent advantages and limitations [34, 35]. GIC and ZPC chemically bond to tooth structure, whereas RC and RMGIC primarily rely on micromechanical adhesion, providing higher bond strength. In general, GICs exhibit superior mechanical properties compared with ZPC; however, their mechanical performance may be compromised by early moisture contamination. Additionally, overdrying of GICs can induce shrinkage, potentially leading to crack formation and postoperative hypersensitivity. RMGICs demonstrate higher fracture and wear resistance than conventional GICs; however, the release of hydroxyethyl methacrylate (HEMA) has been associated with potential biological effects, including pulpal inflammation and allergic reactions. RCs are insoluble and possess superior mechanical and physical properties compared with conventional luting agents. Nevertheless, some RCs have short working times, may induce polymerization shrinkage-related hypersensitivity, and pose challenges in the complete removal of excess cement, particularly at subgingival margins. Moreover, RCs are generally considered less biocompatible than GICs. Taken together, the selection of luting cement for pediatric crown restorations should be made on a case-by-case basis, considering biomechanical properties, adhesion mechanisms, polymerization behavior, and biological compatibility.

Although direct experimental validation was not performed in the present study, the material properties, loading conditions, and boundary constraints were defined based on previously published experimental and numerical studies. The stress distribution patterns observed are consistent with those reported in previously validated FEA studies: peak stresses in SSCs are concentrated at the load application site [15, 32], whereas peak stresses in ZrCs and within the luting cement layer are localized near the crown margins [8, 13]. These findings are further supported by *in vitro* modeling studies demonstrating that high stresses are imposed on luting cements, particularly in biologically critical marginal regions [33]. Nevertheless, future validation of the present model against experimental or clinical data would further enhance confidence in the results and strengthen the translational value of the findings.

This study had several limitations. Although the preparations were carried out according to the manufacturer's guidelines, minor variations inherent to manual techniques are unavoidable and may have contributed to slight differences in preparation geometry. Acknowledging the potential influence of these variations is therefore important when interpreting the findings. Only two crown materials were evaluated, and future studies could include alternatives such as lithium disilicate and composite resins. All materials in the FEA models were assumed to be linear elastic, homogeneous, and isotropic, which does not fully reflect the complex, anisotropic, and heterogeneous nature of dental tissues and restorative materials and therefore constitutes an inherent limitation of the modeling approach. In addition, polymerization shrinkage stresses associated with resin-based cements were not incorporated into the FEA models, representing another limitation of the present study. Finally, only one tooth type was analyzed, and patient-related factors such as anatomical variation, malocclusion, and dietary habits were not considered.

5. Conclusions

Within the limitations of this study, the findings demonstrate that the elastic modulus of luting cements influences stress levels and distribution patterns within pediatric crown restorations. As the elastic modulus of the luting cement decreased, stress transmitted to the tooth structure decreased in SSC models but increased in ZrC models. Peak stress values within the cement layer decreased with decreasing elastic modulus of the luting cements, regardless of crown material.

ABBREVIATIONS

FEA, finite element analysis; GIC, glass ionomer cement; MPa, megapascal; RC, resin cement; RMGIC, resin-modified glass ionomer cement; SSC, stainless steel crown; vM, Von Mises; ZPC, zinc polycarboxylate cement; ZrC, zirconia crown; ZnO, zinc oxide; HEMA, hydroxyethyl methacrylate; STL, Standard Tessellation Language; MIH, molar-incisor hypomineralization; Bis-GMA, Bisphenol A-glycidyl methacrylate; UDMA, Urethane dimethacrylate; TEGDMA, Triethylene glycol dimethacrylate.

AVAILABILITY OF DATA AND MATERIALS

All data generated or analyzed during this study are included in this published article.

AUTHOR CONTRIBUTIONS

ABS and HG—contributed to the study concept and design. ABS—performed the literature review and drafted the manuscript. EE—developed the finite element models, performed the analyses, and contributed to data interpretation. HG—supervised the study and critically revised the manuscript. All authors contributed to writing and revising the manuscript. All authors read and approved the final manuscript.

ETHICS APPROVAL AND CONSENT TO PARTICIPATE

Not applicable. This study did not involve human participants or animals.

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CONFLICT OF INTEREST

The authors declare no conflict of interest.

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