Effect of Proximal Wall Height on All-Ceramic Crown Core Stress Distribution: A Finite Element Analysis Study

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Purpose: Mechanical analyses of idealized crown-cement-tooth systems through finite element analysis (FEA) has provided valuable insight concerning design parameters and materials that favor lower stress patterns. However, little information regarding variation of basic preparation guidelines in stress distribution has been available. The primary objective of this study was to evaluate maximum principal stresses on a molar crown veneer plus core system natural tooth configuration preparation with variations in the ratio of proximal axial length (PAL) to buccal axial length (BAL) as well as loading condition and position. ***Materials and Methods:*** Three-dimensional models comprising a crown veneer (porcelain), crown core (zirconia), cement layer, and tooth preparation (4.2 mm BAL with PAL reductions of 0.8 mm, 1.0 mm, and 1.2 mm) yielding BAL:PAL ratios of 1.23, 1.31, and 1.4 were designed by computer software (Pro/Engineering). The models were imported into an FEA software (Pro/Mechanica), with all degrees of freedom constrained at the root surface of the tooth preparation. Each tooth preparation crown configuration was evaluated under a vertical (axial) 200 N load, and under a combined vertical 200 N and horizontal (buccally) 100 N load applied at different positions from the central fossa to the cusp tip. Maximum principal stress (MPS) was determined for the crown core for each crown BAL:PAL ratio, loading condition, and position. ***Results:*** Under both vertical and combined loading conditions, the highest MPSs were located at the occlusal region and in the occlusogingival region of the ceramic core. MPS values increased in the proximal region as the BAL:PAL ratio increased. Combined loading resulted in a general increase in MPS compared to vertical loading. ***Conclusion:*** Increasing the BAL:PAL ratio (reducing the proximal axial length of the preparation) acted as a stress concentrator at regions near the crown margins, suggesting this area may be vulnerable to damage from fit adjustment as well as during function. Such increases in stress concentration should be considered in clinical scenarios, especially when inherent flaws are present in the material, since extensive high-magnitude tensile stress fields have been noted under all loading conditions. *Int J Prosthodont 2009;22:78–86.*

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Replacement of tooth structure following pathology or trauma has been of interest to dental practitioners and researchers since dentistry's early days. The selection of a restorative technique and material usually involves the assessment of the remaining dental structure with respect to type and location of damaged tissue (tooth-specific) and the overall systemic condition of the patient.[1] Although small, mineralized tissue loss in anterior and posterior teeth may be restored through minimally invasive adhesive procedures with acceptable longevity,[2] more extensive tissue loss requires full-crown restorative procedures for the long-term replacement of form and function.[3]

According to a survey conducted by the American Dental Association, more than 45 million dental crowns, of which 37 million were porcelain-based, were placed in private dental offices in the United States in 1999.[4] Although clinically acceptable service life for metal-ceramic crown systems has been reported for decades, their esthetic and biocompatible properties have led dental biomaterials and clinical investigators to concentrate research efforts in all-ceramic crown systems.[3]

Most current systems are based upon esthetic porcelain veneers applied to core structural ceramics. The failure rate of posterior all-ceramic crowns is reported as 3% to 4% each year,[5–9] despite recent significant improvements in dental ceramics strength (ie, high strength alumina and zirconia cores). This failure rate, irrespective of all-ceramic crown system strength, indicates that complex mechanical scenarios other than overload catastrophic failure play significant roles in system damage initiation, accumulation, and failure.[3]

Mechanical testing of all-ceramic crowns has concentrated on the determination of failure modes under a single load to failure.[10–12] Unlike long-term clinical observations,[5–9] single load mechanical studies have shown that crown survival rates increase with a material's mechanical strength.[10,11] In addition, it has been speculated that crack origin in clinically failed monolithic crowns occurrs at internal surfaces of the restoration,[13] leading to the rationale of reinforcing all-ceramic crown structures with high-strength ceramic cores.[13] Thus, from a material selection perspective, the use of glass-infiltrated alumina or zirconia core materials has resulted in moderately stronger and tougher structures.[12,13] However, despite the improvement in the ceramic materials' strength, relatively high failure rates have been reported.

Given the ability to vary material and specimen configurations in computer-aided design (CAD) and mechanical modeling software, finite element analysis (FEA) has been used to predict stress distributions and the mechanical behavior of dental crowns,[3,14–20] fixed partial dentures,[21,22] and restorations.[14,23–26] The majority of mechanical modeling investigations concerning dental crowns have addressed the influence of load type and position on stress distribution with different crown materials, system material configurations,[3,16–18,20,27] and variations in tooth preparation configurations.[14,16]

FEA investigations have addressed inadequate tooth structure removal during preparation and resulting crown thickness effects on stress distributions in anterior[14] and posterior[16] crowns. Characteristic solid model designs are often employed in place of intricate clinical shapes due to time-consuming solid CAD design and potential increases in meshing and computing time. Realistic models representing clinical details of full-crown tooth preparations are rare in the dental biomechanics field.

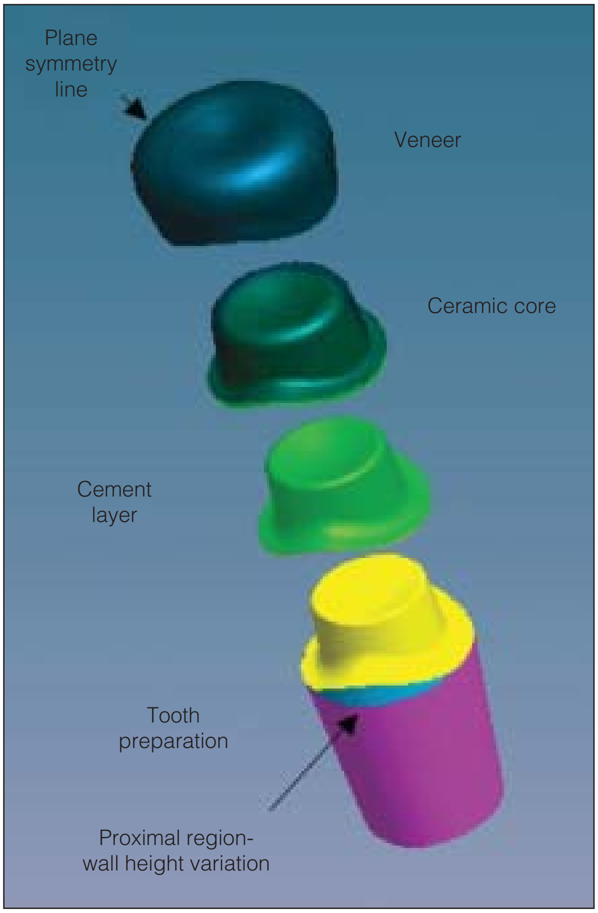

Fig 1 Tooth/all-ceramic crown system components created in CAD software.

Posterior dental crowns have been evaluated as a function of load position,[3,16–18,20,27] crown system configuration,[17,19] and material properties,[3,16–18,20,27] but with little regard to the ratio of the preparation proximal to buccal axial length, which varies as a natural consequence of alveolar ridge and tooth anatomy. The purpose of this study was to evaluate maximum principal stresses in all-ceramic crown veneer-core ystems as a function of the ratio of buccal axial length (BAL) to proximal axial length (PAL) with variations in loading condition and position.

Materials and Methods

A three-dimensional (3-D) finite element model of a plane-symmetric posterior tooth preparation and bilayer all-ceramic crown with different PALs was created for this investigation. The components of the tooth/all-ceramic crown system comprised a veneer layer (porcelain), a crown core layer (zirconia) of a uniform 0.5 mm thickness, a 100-µm-thick cement layer (resin cement), and a tooth preparation (dentin) component created in CAD software (Pro/Engineer Wildfire, PTC) (Fig 1).

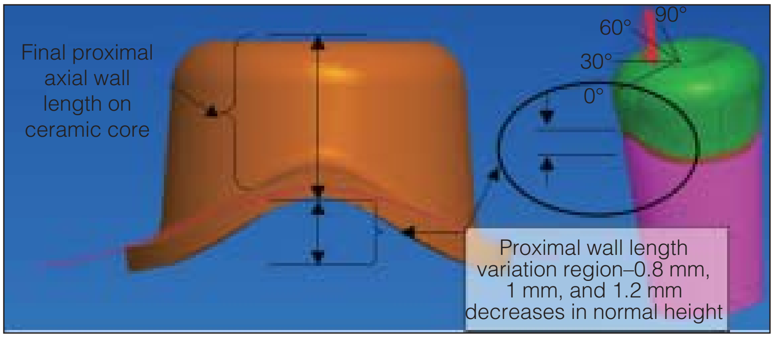

Fig 2 Schematic representation of the proximal axial length (PAL) variation region in the ceramic core *(left)* and tooth/all-ceramic system *(right)*. Total proximal wall length reductions of 0.8 mm, 1 mm, and 1.2 mm were used, leading to buccal axial wall length to proximal wall length ratios (BAL:PAL) of 1.23, 1.31, and 1.4, respectively.

Table 1 Material Properties Input for FEA[3]

Component	Material	Young's modulus (E) (GPa)	Poisson's ratio	Density (g/mL)
Veneer	Porcelain	70	0.19	2.40
Ceramic core	Zirconia	200	0.19	2.40
Cement layer	Resin cement	8	0.33	2.19
Tooth preparation	Dentin	16	0.31	2.14

The original preparation height in the buccal and ligual aspects was 4.2 mm. According to Goodacre et al,[28] a minimum of 4 mm of axial wall length for molars and 3 mm for other teeth is recommended. The PAL variations used in this study were incremental decreases of 0.8 mm (PAL = 3.4 mm), 1 mm (PAL = 3.2 mm), and 1.2 mm (PAL = 3 mm) in total proximal axial length (Fig 2). This gave BAL:PAL ratios of 1.23, 1.31, and 1.4, respectively.

The rationale for the incremental decrease in the proximal wall height was to allow investigation of crown biomechanical behavior with anatomically dictated changes in margin contour (patient-specific anatomy would require different amounts of proximal wall reduction to avoid restoration interference with surrounding soft and hard tissues).

The components were assembled, imported into FEA software (Pro/Mechanica, PTC), meshed (~8,048 tetrahedral elements), and tested for convergence prior to mechanical simulation.

The following assumptions were included in the FEA model: *(1)* all solids were homogeneous, isotropic, and linear elastic, *(2)* no slip conditions (perfect bonding) between components, *(3)* uniform 100-μm-thick cement layer, *(4)* uniform 0.5-mm core thickness, *(5)* absence of flaws in all components, *(6)* 6 degrees of freedom (full) constraint on the root component surface. The material properties used are presented in Table 1. Each tooth preparation configuration was evaluated under an axial 200-N load and under a combined 200-N axial (vertical) and 100-N horizontal (along the x-axis in the radial direction, towards the buccal flange) load applied over a 1-mm-diameter surface area on the veneer layer. Load position on the veneer layer surface was varied as a function of distance from the crown center (from 0 to 4.5 mm in 0.25-mm increments) and as a function of angle (from 0 degrees [proximal flanges] to 90 degrees [buccal or lingual flange], in 30-degree increments, Fig 2). Due to the buccolingual and mesiodistal symmetry of the crown design, loads were applied to a single quadrant of the crown and were then expanded to create a 3-D maximum principal stress (MPS) graphic output as a function of load position on the veneer (Fig 3).

Results

Overall, the highest MPS levels within the ceramic core were located in 2 different regions as a function of load position on the veneer. These locations were the occlusal wall region, as the loads were applied up to 2 mm outward from the veneer center (Fig 4a), and the marginal region, as loads were applied starting at 2 mm from the veneer center and working toward the periphery (Fig 4b). Also, it was observed that the zirconia core occlusal wall MPS levels were high as loads were applied from 0 to 2 mm away from the veneer central region and steadily dropped as loads were applied beyond 2 mm toward 4.5 mm from the veneer center (Fig 3).

The MPS levels obtained from the core as a function of load position in the veneer (Fig 3a) were expanded to 3-D plots through quarter symmetry (Fig 3b).

The expanded MPS for the ceramic core with a 200-N vertical load on the system with varied BAL:PAL

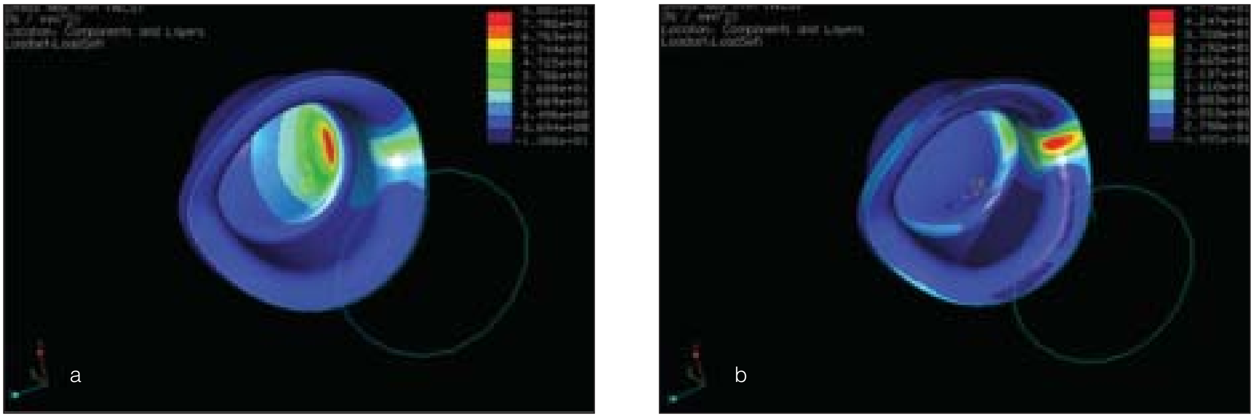

Fig 3 **(a)** The MPS levels as a function of load position in the veneer (load offset distance and angle) resulting from a 200-N axial orientation in a BAL:PAL = 1.31 model. The MPS data as a function of load position in the veneer layer obtained was used to produce **(b)** 3-D MPS plots (in this case, not MPS on the core geometry, but MPS obtained in the core as a function of load position in the veneer layer). The 3-D plot was generated through the expansion of the results obtained for one quadrant due to the model quarter symmetry. Irrespective of angle, a steady drop in MPS as loads are applied from 2 mm toward more off-axis positions with respect to the veneer layer component center, and the higher MPS observed as loads are positioned at (0 degrees) or in proximity (30 degrees) of the reduced proximal wall at load offset positions ranging from 3.5 to 4.5 mm.

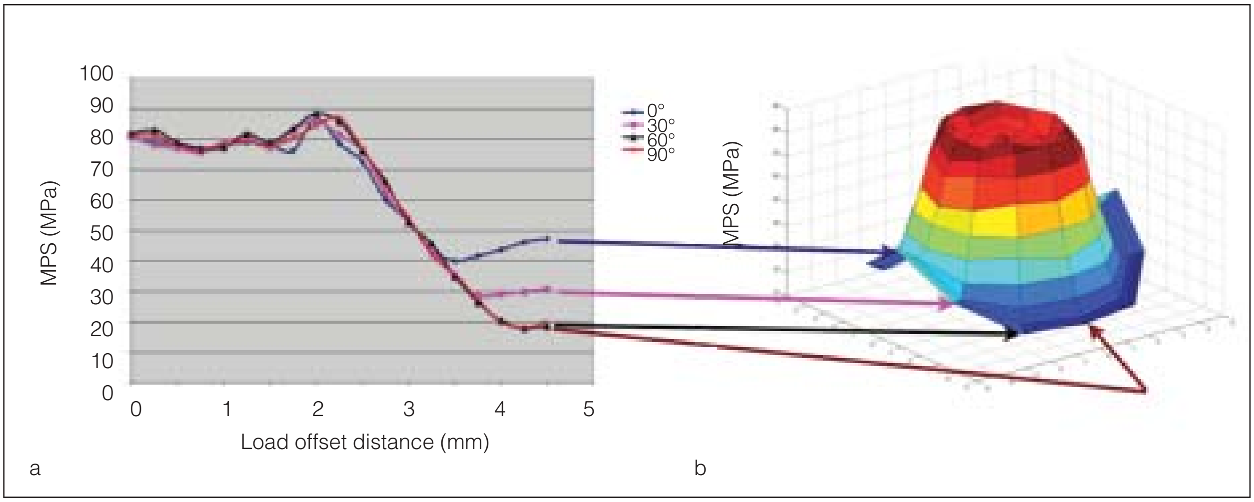

Fig 4 Representation of MPS location shift in the zirconia core as the "0" degree load position (Fig 2) varied **(a)** from the center to 2 mm (MPS location at occlusal wall), and **(b)** from 2 mm toward more peripheral regions of the veneer layer component (MPS location at marginal regions) (BAL:PAL = 1.4).

ratios as a function of load position in the veneer is presented in Fig 5a. The MPS at peripheral regions (3.5 to 4.5 mm from the veneer center) as a function of load position at the veneer is presented separately (Fig 5b). This depicts the relative increase in MPS levels as a function of BAL:PAL ratio increase at the ceramic core proximal region (0 degrees and 30 degrees). No changes in MPS levels were observed as the 200-N vertical load position changed from 2.5 to 4.5 mm at 60 degrees and 90 degrees for the different BAL:PAL ratios, ie, as loads were positioned toward the buccal or lingual flanges (Fig 5b).

The ceramic core MPSs resulting from a 200-N vertical and 100-N horizontal load along the x-axis as a function of load position in the veneer component for different BAL:PAL ratios are presented in Fig 6a. Ceramic core MPS levels resulting from loads applied 3.5 to 4.5 mm relative to the veneer center as a function of angle are presented separately (Fig 6b). Peripheral region MPSs showed small increases in MPS levels as a function of increasing BAL:PAL at the ceramic core proximal region (0 degrees and 30 degrees) (Fig 6b).

Comparisons between MPS levels obtained for a 200-N vertical load versus a combined 200-N vertical

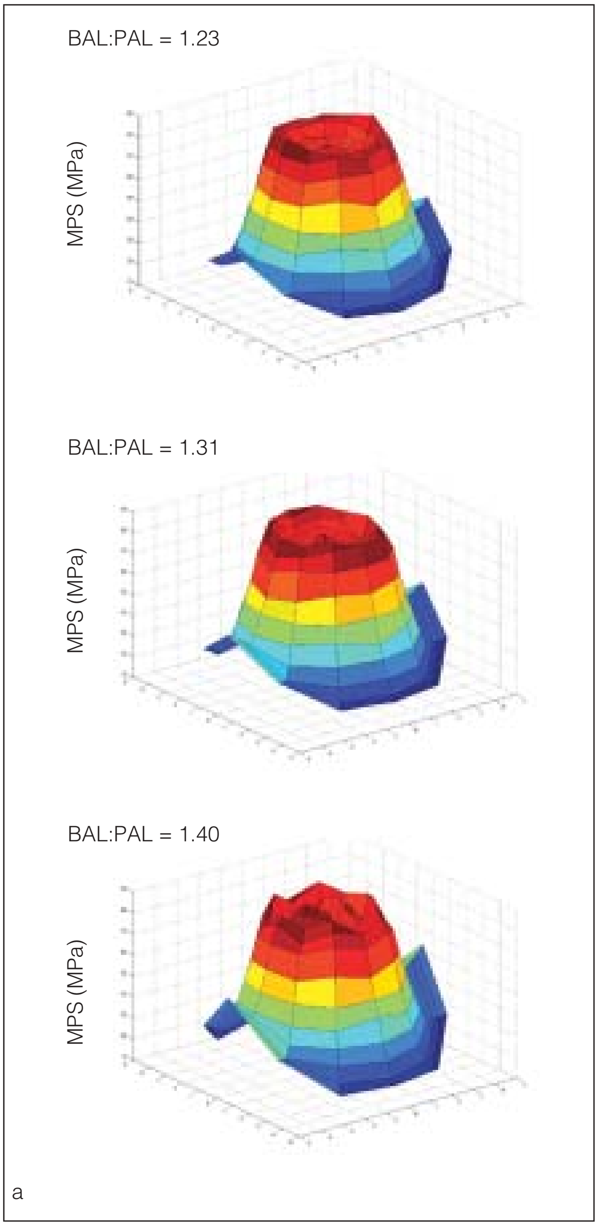

Fig 5a 3-D MPS obtained in the core as a function load position on the veneer with a 200-N axial load. (MPS scale:10 to 90 MPa, increments of 10 MPa.)

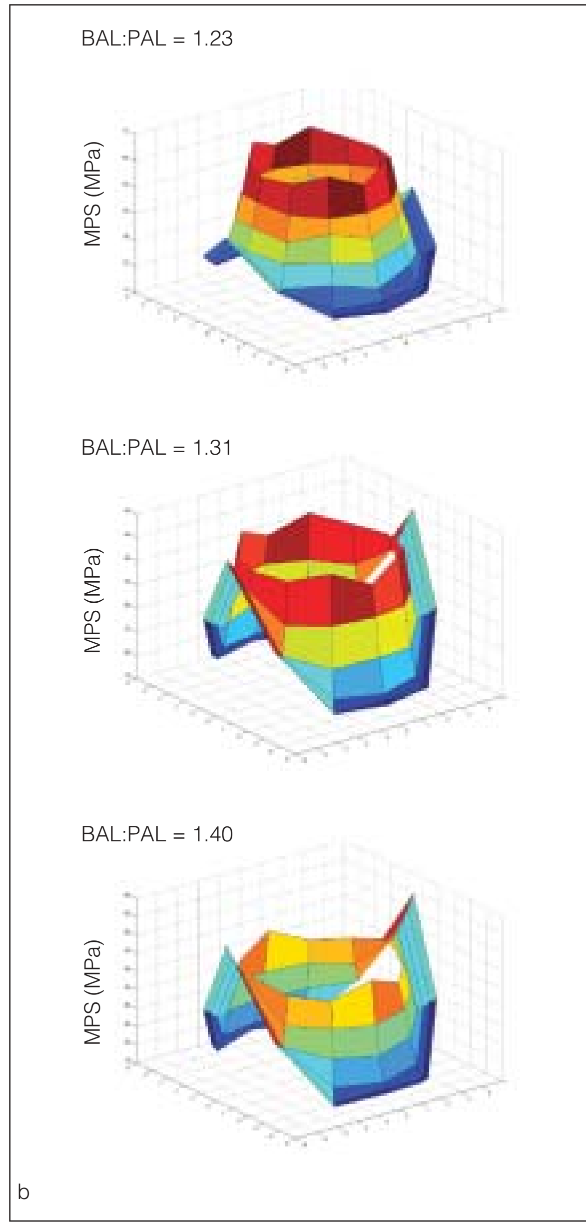

Fig 5b A detailed view of the MPS resulting from positioning the 200 N vertical load at regions 2.5 to 4.5 mm from the veneer center. (MPS scale: BAL:PAL = 1.23, 10 to 70 MPa, increments of 10 MPa; BAL:PAL = 1.31, 15 to 50 MPa, increments of 5 MPa; BAL:PAL = 1.4, 15 to 60 MPa, increments of 5 MPa.) Note that the MPS values increase as the BAL:PAL ratio is increased–when the load is positioned towards the proximal regions of the veneer.

and 100-N horizontal load showed a general increase in MPS levels at load positions from the veneer center to 2 mm outward irrespective of BAL:PAL ratio and angle (Figs 7 and 8). A representation of the MPS variations due to the horizontal load component addition is presented for a BAL:PAL ratio of 1.23 in the buccal/lingual (90 degrees) (Fig 7) and proximal (0 degrees) regions (Fig 8).

Other observations included an increase in MPS observed as the combined load was placed 3.5 to 4.5 mm relative to the veneer center at the 60 degree to 90 degree position, compared to the 200-N vertical load (Fig 7). No increase in MPS was observed due to the 100-N horizontal load addition along the model x-axis at proximal regions (0 degrees to 30 degrees) (Fig 8).

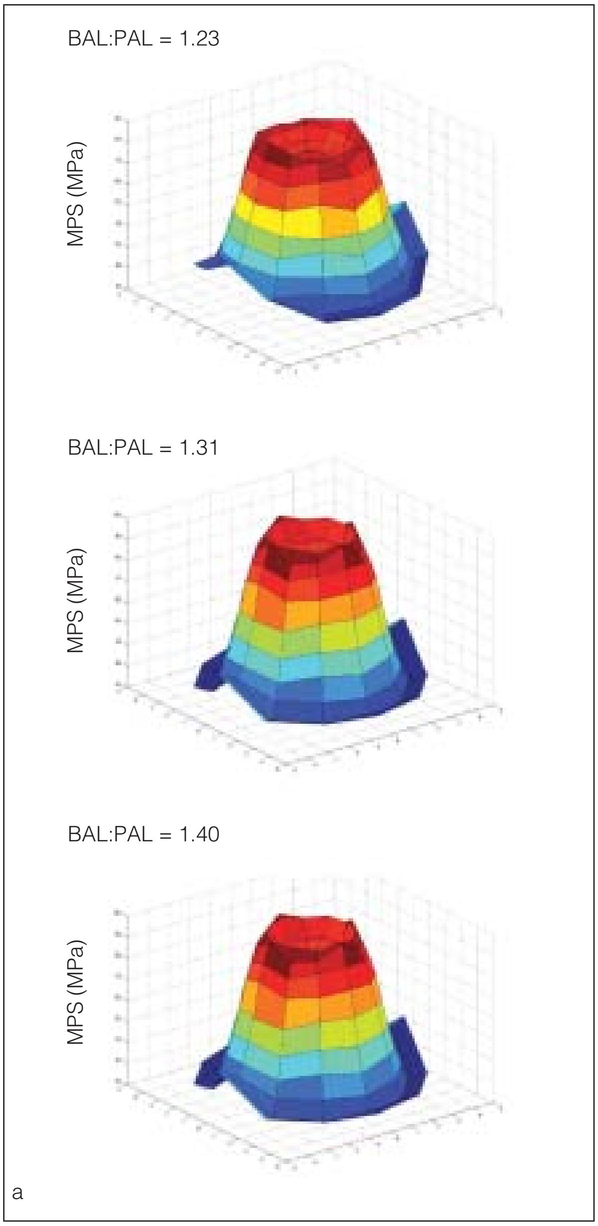

Fig 6a 3-D MPS obtained in the core as a function of a combined 200-N axial and 100-N horizontal (occlusal plain) load position on the veneer. (MPS scale: BAL:PAL = 1.23, 10 to 90 MPa, increments of 10 MPa; BAL:PAL = 1.31 and 1.4, 10 to 100 MPa, increments of 10 MPa.)

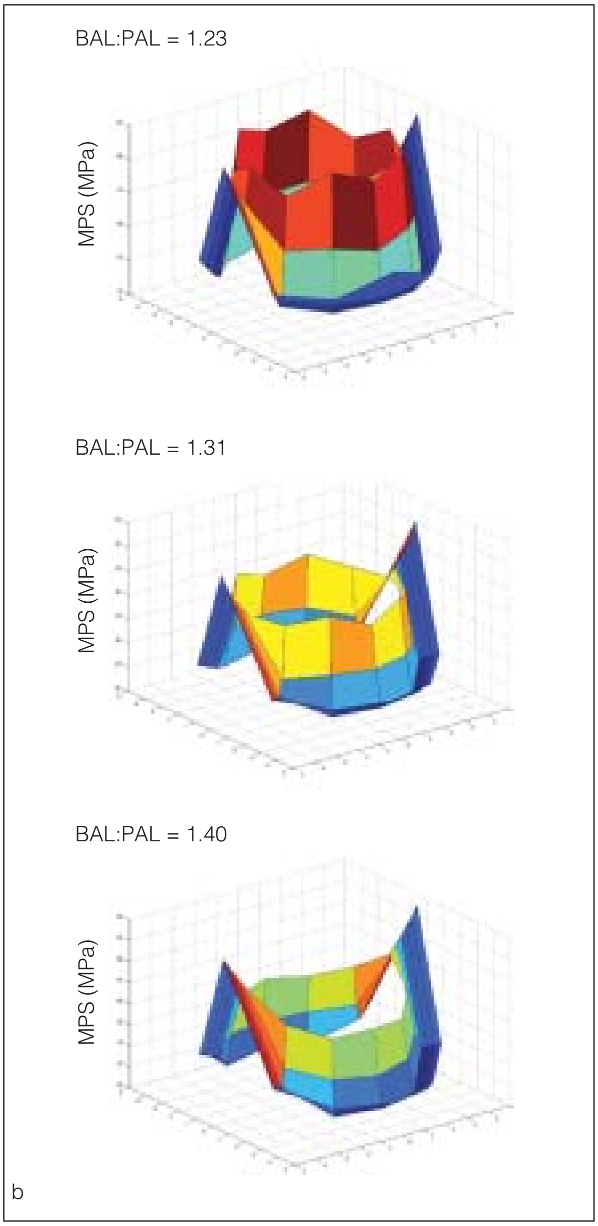

Fig 6b A detailed view of the MPS resulting from positioning the 200-N vertical load and 100-N horizontal load at regions 2.5 to 4.5 mm from the veneer center. (MPS scale: BAL:PAL = 1.23, 20 to 45 MPa, increments of 5 MPa; BAL:PAL = 1.31, 20 to 55 MPa, increments of 5 MPa; BAL:PAL = 1.4, 20 to 60 MPa, increments of 5 MPa.)

Discussion

Laboratory and mechanical simulations of tooth-crown systems have been previously performed in an attempt to determine what materials and basic preparation configurations result in the best mechanical performance.[3,16–18,20,27] However, while information concerning a particular set of tooth preparation variables and materials is collected during mechanical testing, investigations of larger numbers of combinations between preparation configurations and materials are not practical.

Mechanical simulation of restorative systems can be a useful tool for researchers attempting to develop robust restorative system designs, since it allows the simulation of the interplay between variables and their

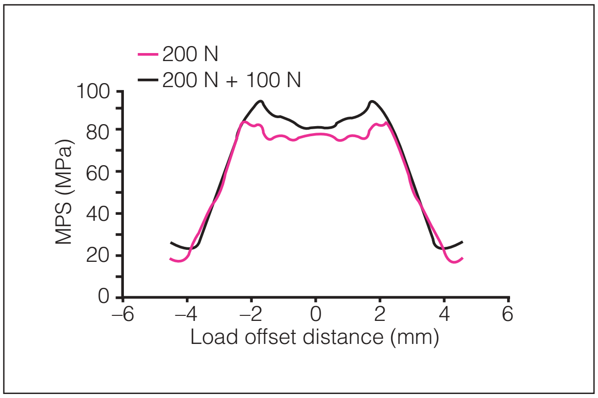

Fig 7 200-N vertical load versus a combined 200-N vertical and 100-N horizontal load as a function of load offset distance for a BAL:PAL = 1.23 at the buccal/lingual region loading angle (90 degrees). Note the increase in MPS due to the addition of a 100-N horizontal load. Also note the increase from the center to ~2 mm due to the addition of a 100-N horizontal load.

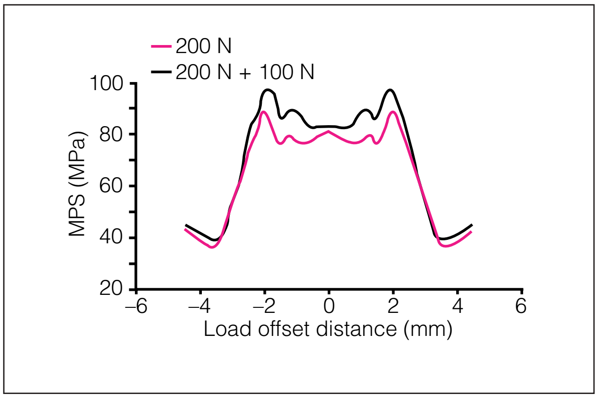

Fig 8 200-N vertical load versus a combined 200-N vertical and 100-N horizontal loads as a function of load offset distance for a BAL:PAL = 1.23 at the proximal region loading angle (0 degrees). Note that the 100-N horizontal force addition along the model x-axis did not increase MPS at the proximal region margin.

influence in the system biomechanics. FEA has been used to study crown systems,[3,14–20] fixed partial dentures,[21,22] and tooth restorations.[14,23–26] Specific to tooth-crown systems, only one study has addressed a large number of variable combinations.[3]

Rekow et al[3] performed a factorial analysis evaluating the relative contribution of crown system configuration, materials, anatomic variables, load position, and a combination of variables on all-ceramic crown MPS levels. Their study showed that MPS was significantly affected by crown material, crown thickness, cement modulus, load position, and crown-supporting structure. They also showed that a combination of variables, such as crown thickness–cusp inclination and crown thickness–load position, had significant contributions to MPS. They also pointed out that variables considered alone are inadequate to predict stress distribution in crowns for biomechanical configuration design purposes. This study indicates that an additional variable that must be considered is the proximal margin contours (BAL:PAL ratio).

Although multiple configurations and interaction between variables may be evaluated through factorial analysis,[3] this type of investigation lacks detailed information regarding relative MPS changes in distribution/location due to factor variation. For instance, this type of analysis would depict significant changes in MPS values as a function of tooth preparation height. However, it would not provide information concerning stress location as a function of tooth preparation height. Extent and location of stress concentration, however, could be critical, particularly if stress concentrates in areas where damage may have been introduced during fabrication and/or in laboratory and clinical procedures.

Axial wall preparation length has been classically considered a crucial factor for crown retention. Goodacre et al[28] recommend a minimum of 4 mm of axial wall length for molars and 3 mm for the other teeth. However, the literature concerning the influence of axial wall length on system biomechanics is sparse. Scherrer and de Rijk[29] reported a significant increase in fracture resistance as a result of increasing the axial length of crowns with highly stylized, flat occlusal surface crowns of feldspathic porcelain, glass-ceramic, and alumina-reinforced glass. However, this study did not provide information on stress distribution and fracture patterns with load position, load direction, or variation in axial wall length around the gingival margin.

Occlusal reduction during preparation varies with different types of crowns and teeth. In addition, the resultant axial wall length obtained after crown preparation also depends on the position and alignment of teeth in the arch, occlusal relationships, esthetics, periodontal architecture, and tooth morphlogy.[28] Bindl et al[30] evaluated the survival of 208 CAD/CAM–fabricated crowns on molars and premolars grouped by an axial wall length of 3 mm or more (classic design, n = 70), less than 3 mm (reduced, n = 52), and no axial wall (endodontically treated teeth, n = 86). The same authors reported that only 7 specimens failed by fracture. Among the failed specimens, 3 were of the classic design and 4 were of the reduced design.[30] The ratio of interproximal to buccal or lingual wall length was not specified.[30]

Based upon our FEA results from both axial and combined axial and horizontal loading on the veneer, high MPS levels were observed in 2 regions of the zirconia core. As expected, one region was in the occlusal surface beneath the loading area (Figs 3 and 4a). A second region of high stress developed in the core in close proximity to the margin (Fig 4b). It is important to note that previous stylized models with no variation in axial length in the proximal region showed lower MPS intensity and less stress concentration irrespective of load position variation.[3] The MPS patterns observed in the present study for loads applied at 60 degrees and 90 degrees from the proximal wall (buccally or lingually) for both axial and combined loads followed the same pattern for the different BAL:PAL ratios (Figs 5 and 6).

Load position plays a significant role in stress concentrations in complex anatomy structures such as full crowns. The magnitude and location of the highest MPS varied with the offset distance of the load with respect to the veneer central axis (Fig 4). Specific to the 0-degree and 30-degree angulation series of simulations, as load position moved toward the periphery of the veneer, the MPS concentrated at the center and gingival regions of the core proximal axial wall. When the horizontal load component was added, stress locations remained relatively unchanged but the magnitude increased (Figs 7 and 8). For either loading condition, the stresses in the marginal area increased as the BAL:PAL ratio increased (proximal wall height reduction). It should be noted that since the horizontal force component was placed along the x-axis of the model (buccally or lingually), it resulted in very subtle alterations in MPS in the regions of PAL variation (Figs 6 and 8) compared to axial load conditions (Figs 5 and 7). Our mechanical simulation results are further supported by laboratory testing (load to failure) of dome glass shapes with different indenter and load distribution,[31] where distributed loads resulted in a stress concentration shift and subsequent failure at the crown margin region.[31] This stress concentration shift with distributed loading (while in function) may be exacerbated by PAL reduction and could be the basis for crack initiation from crown margins, particularly those vulnerable to damage from fabrication, laboratory, and clinical procedures. The large tensile fields observed due to variations in axial wall height, along with the high tensile fields in the occlusal wall region observed in all simulations, may be further increased if stress-raising features such as flaws and inclusions are present in the material.[32–34]

Conclusion

Tooth function involves both tooth-to-tooth contacts and occlusally distributed loads during mastication. Increasing the BAL:PAL ratio (reducing the proximal axial length of the preparation) acted as a stress concentrator at regions near the crown margins, suggesting this area may be vulnerable if damaged during fit adjustment or function. Since extensive tensile stress fields of high magnitude were noted under all loading conditions, these stress concentrations should be considered in clinical scenarios, especially when inherent flaws may be present in the material.

Acknowledgment

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Literature Abstract

Bone level changes at axial- and nonaxial-positioned implants supporting fixed partial dentures. A 5-year retrospective longitudinal study

This retrospective study analyzed the influence of implant inclination on marginal bone loss at freestanding, implant-supported FPDs over a 5-year period of functional loading. Thirty-eight partially dentate, periodontally compromised patients with 42 freestanding FPDs supported by 111 Astra implants were included. Twenty-four implants (57%) were placed in the maxilla. Fifteen FPDs were supported by two implants and 27 FPDs were placed on three implants. Twenty-two (52%) FPDs were designed with a cantilever extension. All implants used had a diameter of 3.5 mm with the length varying between 8 and 19 mm. Standardized photographs were taken for implant inclination measurements. The first was taken at the implant sites of the occluded original master casts, the second was taken with the guide pins abutment pick-up in place, and the third was obtained when the second image was superimposed with precision on the first image. The third photograph showed the image of the two casts in occlusion with the guide pins revealing the inclination of the implants in relation to the occlusal plane. The inclination in the mesiodistal direction of each individual implant, in relation to a vertical axis perpendicular to the occlusal plane, was measured. For cases with an FPD supported by two implants, an additional photograph of the cast with the guide pins in place was taken in a transversal direction. Assessments of the interimplant inclination in both mesiodistal and buccolingual directions were performed. The methodological error of the whole recording procedure as well as the interexaminer reproducibility for inclination assessments was determined. The marginal bone level, in relation to the marginal edge of the fixture, was assessed using standardized radiographs. It was shown that the axial-positioned implants had a mean angulation of 2.4 degrees while the mean value for nonaxial-positioned implants was 17.1 degrees. The mean bone loss during 5 years in function was 0.4 mm (SD: 0.97) and 0.5 mm (SD: 0.95) for the axial- and nonaxial-positioned implants, respectively. Thirty-nine percent of the axial-positioned implants demonstrated no bone loss after 5 years in function, compared with 37% of the nonaxial-positioned implants. Thirty percent of the axial-positioned and 33% of the nonaxial-positioned implants showed more than 1 mm peri-implant bone loss. No statistically significant differences in marginal bone change were found between axial- and nonaxial-positioned implants. The interimplant inclination for the FPDs supported by two implants varied between 1 degree and 36 degrees (mean: 7.4 degrees, SD: 8.8) in the mesiodistal direction and between 0 degrees and 24 degrees (mean: 6.9 degrees, SD: 7.3) in the buccolingual direction. No significant correlations were found between inter-implant inclination and 5-year bone level changes. The findings of the study (with moderately tilted implants (≤ 30 degrees) indicated that a tilted position of the implant did not render an increase risk for bone loss during functional loading. However, the results may not be extrapolated to single implant replacements because of different loading conditions.

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