Design parameters for all-ceramic dental crowns

de Jager, N.

Link to publication

Citation for published version (APA):

General rights
It is not permitted to download or to forward/distribute the text or part of it without the consent of the author(s) and/or copyright holder(s), other than for strictly personal, individual use, unless the work is under an open content license (like Creative Commons).

Disclaimer/Complaints regulations
If you believe that digital publication of certain material infringes any of your rights or (privacy) interests, please let the Library know, stating your reasons. In case of a legitimate complaint, the Library will make the material inaccessible and/or remove it from the website. Please Ask the Library: http://uba.uva.nl/en/contact, or a letter to: Library of the University of Amsterdam, Secretariat, Singel 425, 1012 WP Amsterdam, The Netherlands. You will be contacted as soon as possible.
CHAPTER 5

Design parameters for all-ceramic dental crowns

5.1 Abstract

Introduction: Knowledge of factors which influence stress and its distribution is of key importance to the successful production of durable all-ceramic restorations. The objective of this study was to evaluate, by finite element analysis (FEA), the influence of the shape of the preparation and the cement layer on the stress distribution in CAD-CAM produced all-ceramic crowns and in their cement layer.

Material and methods: The CAD models of multi-layer crowns for posterior tooth 46 of three patients produced with CAD-CAM-technology were translated into a three-dimensional FEA program. The stress distribution due to the combined influences of bite forces, residual stresses caused by the difference in expansion coefficient of the two ceramic layers, and the influence of shrinkage of the cement was investigated.

Results: The tensile stresses in the crown for the chamfer knife-edge preparation might put the integrity of the currently available ceramic materials at risk, while a non-uniform cement layer might result in stresses exceeding the bond strength. It was concluded that for long lasting restorations in the posterior region it is advisable to make a chamfer with collar preparation, the cement layer as uniform, and the difference in thermal expansion for the two ceramics as small as possible.

Significance: This study indicates that for full ceramic crowns in the posterior region, specific design rules should be followed, and that FEA utilizing CAD-CAM data can be a successful tool to develop design guidelines for all-ceramic restorations.
5.2 Introduction

Despite the increased effort to prevent dental decay, there is still a need for prosthetic reconstructions. Because of their esthetics and biocompatibility, many patients prefer all-ceramic crowns to metal-ceramic crowns.

However, all-ceramic restorations, particularly when placed in the posterior region, have a history of being prone to brittle fracture. To overcome brittle fracture, strong ceramic core materials have been developed to support the weaker veneering ceramic materials. All-ceramic restorations that are produced from these new materials are still more brittle and less ductile than metal-ceramic restorations. As a consequence, the preparation and cementation procedures are more critical for all-ceramic restorations, than for metal-ceramic restorations. Moreover, the most important factor for the longevity of restorations is a good seal, since analyses of failures of fixed restorations have shown caries to be the most frequent cause of failure [1]. The shape of the preparation and of the cement layer may influence both the resistance of the restoration to fracture and the adhesion and consequent seal of the reconstruction, which is often the ‘Achilles heel’ of fixed restorative work. Adequate preparation guidelines are therefore of importance and should be based on sound data, taking all possible parameters into account. However, the literature contains no description of crowns of patients where the combined influences of bite forces, residual stresses due to the difference in thermal expansion of the two ceramic layers forming the crown, and the influence of shrinkage of the cement was studied. For manually produced restorations, design weaknesses are difficult to predict, as the ceramic layer shapes and sizes are operator dependent properties. However, for computer designed and manufactured restorations, these parameters, as well as those of the preparation, are digitally available and thus more amenable to stress analysis and failure prediction.

Finite element stress analysis (FEA) seems to be a proper tool for such an evaluation. FEA was originally developed in the aircraft industry [2] and has become widespread in the engineering field. In dentistry, FEA has been used to determine stress distributions in teeth by authors like Farah et al. [3]. Many authors have been making finite element analysis of dental restorations since then.
DeHoff \textit{et al.} [4] studied the influence of the residual stresses due to thermal contraction mismatch between two layers forming the crown. Hojjatie \textit{et al.} [5] and Palamara \textit{et al.} [6] studied the influence of occlusal loads on the dentin and the restoration with a three dimensional finite element analysis, where Kamousiora \textit{et al.} [7] and Shinohara \textit{et al.} [8] studied specifically the effect of the cement layer on the restoration, taking into account the occlusal loads. Proos \textit{et al.} did extensive work applied to margin design, different cement materials and cement layer design [9, 10] and different core materials and thicknesses [11, 12].

Based on these experiences it was hypothesized that FEA is a proper tool for the multi causal evaluation of mechanical failure of all-ceramic restorations.

The objective of this study was to evaluate, by finite element analysis, the influence of the shape of the preparation and the cement layer, on the stress distribution in CAD-CAM produced all-ceramic crowns and their cement layer.

5.2 Materials and methods

The CAD models of the crowns of three patients with crowns for posterior tooth 46 produced with CAD-CAM by Cicero, Elephant Dental B.V. (Hoorn, the Netherlands) were selected to be translated into a three dimensional FEA program. The crowns consisted of a core made of Synthoceram, a pressed high-strength, alumina-based porcelain; veneered with Synthagon, a leucite-free glass ceramic. The first crown had a chamfer knife-edge preparation and a cement layer with an uniform cement thickness of 0.140 mm except on the outline where the thickness was 0.025 mm. The second crown had a chamfer with collar preparation and the same cement layer thickness. The third crown had a chamfer with collar preparation and a cement layer which varied from 0.025 to 0.140 mm. In this crown the layer was thin on the outline and on flat horizontal surfaces to give maximum support to the crown on these surfaces. This crown was also analysed with a uniform cement thickness design.

Conversion CAD to FEA models

In the Cicero CAD program, the multi-layer crowns are described by the surfaces of each layer (inner, ceramic interface and outer surface). The design structure looks like an umbrella with the crossing vertical and horizontal lines forming surface elements. The horizontal lines have equal distances, giving a nice looking model.
However, in some places the horizontal lines fall together on the outline, while the vertical lines come together in one or more points. As a consequence, some surface elements are strongly deformed, some even with no surface, or with one or two sides with no length. Therefore, two steps were performed in order to transform the CAD model to FEA format. First, the various separate surfaces in the Cicero CAD-program were connected in the FEA model. Second, some crossing points were moved on the vertical lines, and some points were left out, in order to get a model with well-shaped surface elements.

The final model consisted of two ceramic layers (veneer and core), a cement layer, and the prepared tooth [Fig. 5.1] and consisted of 50,000 to 65,000 parabolic wedge and parabolic tetrahedron solid elements all together.

![Fig. 5.1 The layers composing the FEA model](image)

The finite element modeling and post processing was carried out with FEMAP software (FEMAP 8.10, ESP, Maryland Height, MO, USA), while the analysis was done with CAEFEM software (CAEFEM 7.3, CAC, West Hills, CA, USA). The material data used in this model is shown in Table 5.1.
Table 5.1
The relevant material properties of the materials used in the FEA model

<table>
<thead>
<tr>
<th>Material</th>
<th>Preparation</th>
<th>Cement</th>
<th>Core</th>
<th>Veneering Porcelain</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Dentine</td>
<td>RelyX ARC</td>
<td>Synthoceram</td>
<td>Synthagon</td>
</tr>
<tr>
<td>Tensile Strength (MPa)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Young's modulus (GPa)</td>
<td>18.3</td>
<td>5.1</td>
<td>300</td>
<td>140</td>
</tr>
<tr>
<td>Hardenings modulus (GPa)</td>
<td></td>
<td></td>
<td>120.0</td>
<td>68.0</td>
</tr>
<tr>
<td>Initial Yield Stress (MPa)</td>
<td></td>
<td></td>
<td>3.1</td>
<td></td>
</tr>
<tr>
<td></td>
<td>3.2</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Poisson ratio</td>
<td>0.30</td>
<td>0.27</td>
<td>0.27</td>
<td>0.30</td>
</tr>
<tr>
<td>Linear Shrinkage (%)</td>
<td>1.273</td>
<td>0.381</td>
<td></td>
<td>0.313</td>
</tr>
</tbody>
</table>

The material properties for the ceramics are data supplied by the supplier. The data for the dentine [13] and the cement [14, 15] are from literature.

The following assumptions were made in order to simplify the calculations:

1. The material of the ceramic crown layers, the tooth dentine, and the luting cement after setting were assumed to be homogeneous, linearly elastic, and isotropic. The anisotropy of the crown and the dentine was not taken into account, although due to the structure the mechanical properties of dentine do vary with orientation and location as shown by Konishi et al. [16];

2. The time dependent setting process of the luting cement was mimicked by a time independent elastic-plastic material property [15], although this model has defined properties for uniform cement layer thickness only and the cement layer thickness for the different models varied from 0.025 to 0.140 mm;

3. The influence of the periodontal ligament on the stresses in the crown is negligible, although Reese [17] found that the ligament and alveolar bone is of importance for the stress distribution;

4. The influence of the pulp chamber in the preparation on the stresses in the crown is negligible as found by Hojjatie et al. [5];

5. The distribution of the temperature during processing of the crown is uniform;

6. The visco-elastic behavior above the glass transition temperature (Tg) creates a stress free state in the ceramics. DeHoff et al. [18] showed that the stress above this temperature decrease quite rapidly, and the influence of the visco-elastic behavior of the porcelain near the softening temperature is negligible;
Chapter 5

7. The modulus of elasticity and the Poisson ratio are constant during the processing of the crown, although as has been shown by Käse et al. [19] that these properties are temperature dependent, especially near the softening temperature.

All nodes in the x-y plane, which corresponds to the root portion of the prepared tooth (Fig. 5.1), were assumed to be fixed; no translation or rotation was allowed in any direction.

There were three stress outputs: stresses due to bite forces, residual stresses due to the difference in expansion coefficient of the two layers forming the crown and the influence of shrinkage of the cement. The stresses caused by the influences were calculated separately, then the three outputs were combined using the linear combination facility of FEMAP, which recalculate the combined stresses based on the output vectors of the linearly combined components.

In post processing, the “contour options average elemental” without use of the “corner data” were used for visualizing the results of the Max Prin Stress and the Max Shear Stress.

Stresses due to bite forces

A calculation was done with bite forces as load. This study assumed a bite force on these molars of 665 N, which is about the maximum normal bite force [13], although it was reported by Nishigawa [20] that the maximum bite force during, sleep associated, bruxism can exceed 800 N for individuals. The bite force was distributed uniformly on the points of the crown in contact in occlusion perpendicular to the surface. The resulting vertical (z) component was made 665 N.

Residual stresses

To determine the residual stresses after the production process of the crown, caused by the differences in expansion coefficient of the two materials forming the crown, the temperature – expansion diagrams of these materials according to the supplier (Elephant Dental, Hoorn, the Netherlands) were used to calculate the expansion. The difference in thermal expansion coefficients, not those of both layers separately, was used to avoid as much as possible allocating stresses due to this calculation on the non-ceramic layers.
The temperature at the glass-transition point of Sintagon (veneering porcelain) is lower than that of Synthoceram (core material); only the difference in linear expansion of the two materials from the temperature at the glass-transition point of Sintagon to room temperature was used in calculating the stresses.

**Stresses due to shrinkage of the cement**

The setting of resin composites is a complex time dependant process, throughout which material properties undergo a dramatic change in a relatively short period. To determine the stresses due to shrinkage of the cement during hardening of the resin composite a time independent non-linear elastic-plastic material model was used according to the findings of De Jager et al. [15] for RelyX ARC of 0.140 mm layer thickness.

**5.3 Results**

Crown 3 (chamfer with collar and non-uniform cement layer design) failed shortly after placement at the distal-lingual side, where a part of the crown broke away. The crown was successfully replaced with a new crown of the same design. The three crowns are now one year in use.

Although calculated stresses are available for each element and each load condition, only overall results are presented.
Stresses in the veneering porcelain at the occlusal surface

Fig. 5.2 The stresses (in MPa) in the veneering porcelain at the occlusal surface

Fig. 5.2 shows the maximum principal stress of the combined stresses due to bite forces, difference in expansion coefficient of the two ceramics, and shrinkage of the cement at the occlusal surfaces of the three crowns. The stresses in Crown 3 with uniform cement layer (Crown 3*) differ slightly from the stresses in Crown 3; this is mainly caused by a slight difference in the occlusion points (Table 5.2). The bite forces are the main component of the principal tensile and compressive stresses of the combined stresses (Table 5.2).
### Table 5.2

The maximum principal tensile and compressive stresses in the crown and the maximum shear stresses in the cement

<table>
<thead>
<tr>
<th>Maximum Stress of the combined stresses and their components in:</th>
<th>Crown 1</th>
<th>Crown 2</th>
<th>Crown 3</th>
<th>Crown 3*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Veneering porcelain</td>
<td>Max Prin Stress (MPa)</td>
<td>Max Prin Stress (MPa)</td>
<td>Max Prin Stress (MPa)</td>
<td>Max Prin Stress (MPa)</td>
</tr>
<tr>
<td>Occlusal surface</td>
<td>Tensile stress due to:</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Bite forces</td>
<td>18</td>
<td>6</td>
<td>55</td>
<td>55</td>
</tr>
<tr>
<td>Different ceramics</td>
<td>2</td>
<td>3</td>
<td>6</td>
<td>5</td>
</tr>
<tr>
<td>Shrinkage cement</td>
<td>1</td>
<td>8</td>
<td>0</td>
<td>-1</td>
</tr>
<tr>
<td>Combined stresses</td>
<td>21</td>
<td>15</td>
<td>54</td>
<td>55</td>
</tr>
<tr>
<td>Compressive stress due to:</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Bite forces</td>
<td>-68</td>
<td>-200</td>
<td>-160</td>
<td>-155</td>
</tr>
<tr>
<td>Different ceramics</td>
<td>0</td>
<td>0</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Shrinkage cement</td>
<td>0</td>
<td>5</td>
<td>7</td>
<td>4</td>
</tr>
<tr>
<td>Combined stresses</td>
<td>-72</td>
<td>-201</td>
<td>-160</td>
<td>-163</td>
</tr>
<tr>
<td>Core-veneer interface</td>
<td>Tensile stress due to:</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Bite forces</td>
<td>23</td>
<td>14</td>
<td>44</td>
<td>39</td>
</tr>
<tr>
<td>Different ceramics</td>
<td>14</td>
<td>21</td>
<td>2</td>
<td>1</td>
</tr>
<tr>
<td>Shrinkage cement</td>
<td>4</td>
<td>2</td>
<td>-2</td>
<td>0</td>
</tr>
<tr>
<td>Combined stresses</td>
<td>31</td>
<td>27</td>
<td>37</td>
<td>35</td>
</tr>
<tr>
<td>Core</td>
<td>Cement-core interface</td>
<td>Tensile stress due to:</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Bite forces</td>
<td>57</td>
<td>15</td>
<td>101</td>
<td>101</td>
</tr>
<tr>
<td>Different ceramics</td>
<td>54</td>
<td>58</td>
<td>-2</td>
<td>-1</td>
</tr>
<tr>
<td>Shrinkage cement</td>
<td>3</td>
<td>12</td>
<td>-4</td>
<td>-1</td>
</tr>
<tr>
<td>Combined stresses</td>
<td>109</td>
<td>80</td>
<td>73</td>
<td>68</td>
</tr>
<tr>
<td>Crown</td>
<td>Cervical surface</td>
<td>Tensile stress due to:</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Bite forces</td>
<td>6</td>
<td>16</td>
<td>52</td>
<td>47</td>
</tr>
<tr>
<td>Different ceramics</td>
<td>12</td>
<td>39</td>
<td>3</td>
<td>2</td>
</tr>
<tr>
<td>Shrinkage cement</td>
<td>24</td>
<td>-7</td>
<td>-10</td>
<td>-8</td>
</tr>
<tr>
<td>Combined stresses</td>
<td>39</td>
<td>32</td>
<td>33</td>
<td>35</td>
</tr>
<tr>
<td>Cement</td>
<td>Layer</td>
<td>Shear Stresses due to:</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Max Shear Stress (MPa)</td>
<td>Max Shear Stress (MPa)</td>
<td>Max Shear Stress (MPa)</td>
<td>Max Shear Stress (MPa)</td>
<td></td>
</tr>
<tr>
<td>Bite Forces</td>
<td>11</td>
<td>4</td>
<td>29</td>
<td>26</td>
</tr>
<tr>
<td>Different Ceramics</td>
<td>1</td>
<td>2</td>
<td>0</td>
<td>1</td>
</tr>
<tr>
<td>Shrinkage Cement</td>
<td>22</td>
<td>29</td>
<td>34</td>
<td>25</td>
</tr>
<tr>
<td>Combined stresses</td>
<td>33</td>
<td>33</td>
<td>55</td>
<td>35</td>
</tr>
</tbody>
</table>

Crown 1: with chamfer-knife edge and uniform cement layer
Crown 2: with chamfer with collar and uniform cement layer
Crown 3: with chamfer with collar and non-uniform cement layer
Crown 3*: Crown 3 with uniform cement layer

Note: positive values are tensile stresses, where negative values are compressive stresses
Stresses at the core–veneer interface

Fig. 5.3 The stresses (in MPa) at the interface of the veneering porcelain and the core

Fig. 5.3 shows the maximum principal stress of the combined stresses at the core-veneer interface. The bite forces are an important component of the maximum principal tensile stress at the core-veneer interface, which is increased in crown 1 and 2 by the thermal contraction mismatch of the two ceramics (Table 5.2). There is no indication that the non-uniform cement layer lowers these stresses.

Stresses in the core at the cement-core interface

At the core surface (cement-core interface) the bite forces are an important component of the maximum principle tensile stress of the combined stresses; this stress is increased in crown 1 and 2 by the thermal contraction mismatch of the two ceramics (Table 5.2).
Design parameters for all-ceramic dental crowns

Stresses in the two ceramic layers

Fig. 5.4 The stresses (in MPa) in the two ceramic layers in a section cut in distal-mesial view

Fig. 5.4 shows a section cut of the three crowns in distal-mesial view. The occlusal thicknesses at the cross section with the highest tensile stress are 2.3 mm, 2.2 mm and 1.2 mm for crown 1, 2 and 3 respectively. Apparently, there is no relation between the maximum tensile stress in the core at the cement-core interface (Table 5.2) and the occlusal thickness, or between these stresses and the shape of the cement layer.
Stresses at the cervical surface

Fig. 5.5 The stresses (in MPa) at the cervical surface in distal-mesial view

Fig. 5.5 shows the crowns in distal-mesial view, showing that the highest stress at the cervical surface is located at the distal-lingual side. The relatively high principal tensile stress at the cervical surface of the core is caused by different influences (Table 5.2).
Stresses in the cement layer

Fig. 5.6 The stresses (in MPa) in the cement layer

Fig. 5.6 shows the maximum shear stress in the cement layer of the combined stresses, which is caused by the combination of the bite forces and the shrinkage of the cement (Table 5.2). The influence of the thermal mismatch of the two ceramics is negligible. The stress in the non-uniform cement layer is considerably higher than in the uniform cement layers. In crown 3 with a uniform cement layer, the maximum shear stress of the combined stresses is 35 MPa in comparison with 55 MPa in the clinically applied crown.

5.5 Discussion

For this study it was assumed that the FEA models did exactly correspond with the clinical crowns, although the clinically placed crowns might differ significantly from the FEA models. One possible error is introduced by the translation from the CAD into the FEA model; another possible error is introduced during the production of the crowns. For instance, glazing of crowns will round off sharp edges; this is not included in the design but can possibly be simulated with smoothing features of FEA-modeling software. These topics will be the subject of future studies. Possible errors arise from the assumptions listed in the Materials & Methods section.
Although the analyses indicate that the stresses did not exceed the strength, crown 3 (chamfer with collar) fractured shortly after placement. It was not possible to determine the initial flaw of this early failure. Production flaws or faults introduced during the cementation process in combination with the rather sharp preparation at the distal-lingual side may have initiated the fracture. The highest cervical surface stress was calculated to be at the distal-lingual side, and the fracture was consistent with initiation at this location.

The maximum tensile and compressive stresses at the occlusal surfaces of the three crowns (Fig. 5.2) in this study are the result of an applied bite force of 665 N on one element. The stresses are lower than the material strength of the applied materials (Table 5.1). A bite force of 665 N is of clinical relevance [13], however, such a point load, applied on only one element will hardly occur clinically. We applied such a point load for reason of simplicity. On the other hand, the produced crown might differ from the FEA model, as contact damage flaws and the surface roughness of the porcelain will affect the materials strength [21].

Kelly [22] examined clinically failed glass-ceramic crowns and showed that failures did not initiate from flaws on the occlusal surface. During loading, restorations develop features that have come to be called “wear facets”. Clinicians recognize that wear facets are usually not point contacts, as supposed in this study, but have dimensions of up to approximately 3 mm in diameter. This effect will result in a distribution of the bite forces over a greater area which will lower the stresses. Also in this study, the number of occlusion points influences the stresses developing in and under the occlusal surface layer.

For crowns 1 and 2, the thermal contraction mismatch of the two ceramics increases the maximum tensile stress in the veneering porcelain at the core-veneer interface (Fig. 5.3). Although the thermal contraction mismatch is intended to compress the veneering porcelain it can be seen that in contrast to this, in some places at the interface the rather complicated shape of the crowns causes tensile stresses in the veneering porcelain. Thompson et al. [23] found that Cerestore crowns mainly failed from sites located at or near the core-veneer interface, and estimated failure stresses from the initial flaw size for two crowns at 15 and 68 MPa. The last stress is higher than the stresses found at this location in this study; this might have been caused by a greater thermal mismatch.

The principal tensile stresses in the core at the cement-core interface (Fig. 5.4) were considerably lower than the strength of the core material.

In 1989 Kelly et al. [24] analyzed Dicor crowns that clinically failed at the cementation surface.
The maximum strength of the materials studied in 1989 would have led to a high failure probability with the stresses found in this study, but the strength of the core materials has been improved considerably since then. Apparently there is no relation between the maximum tensile stress in the core at the cementation surface (Table 5.2) and the occlusal thickness, nor between these stresses and the shape of the cement layer, which is in accordance with the findings of Hojjatie et al. [5] and Proos et al. [12] for the occlusal thickness and Proos et al. [10] and Sjögren et al. [25] for the cement layer.

The relatively high principal tensile stress at the cervical surface (Fig. 5.5) of crown 1 is strongly influenced by the shrinkage of the cement (Table 5.2). The principle tensile stress at this location is mainly tangential. Fracture could initiate at these locations in spite of the lower FEA calculated stress, because stress intensities can be high due to shape and size factors, as well as increased susceptibility to flaw formation at the crown margins. In our FEA models the veneering porcelain does not end exactly at the outline, which is often the case with the produced crowns; otherwise the shrinkage of the cement might have caused still higher stresses due to the more unfavorable load condition in those circumstances on the veneering porcelain.

Although crown 3 has a chamfer with collar preparation due to imperfect preparation at the distal-lingual side, the shape of the preparation in this location is rather knife-edge. Clinicians will recognize the difficulty in fabricating a perfect preparation, especially in the lower jaw on the lingual side. This is recognized by Begazo et al. [26]: from the preparations they studied, nearly all showed to have one or more locations with imperfections, most in the lower jaw.

The maximum shear stress in the cement layer (Fig. 5.6) is caused by the combination bite forces and shrinkage of the cement (Table 5.2). The stresses in the non-uniform cement layer are considerably higher than in the uniform cement layers. The assumption that the non-uniform layer is the main failure cause is supported by the results of the same crown 3 with uniform cement layer thickness design. The stresses in the non-uniform cement layer are higher than the bond strength as published by Cobb et al. [27] and the bond strength and its development in time of RelyX ARC as published by Braga et al. [28] indicating that in the clinical situation there is a considerable risk of bonding failure [25].

83
Conclusions:

With the improved strength of contemporary core materials, these analyses indicate that stresses in the core should not greatly influence the longevity of the restoration. In many configurations, the most critical sites will be in the veneering porcelain near the interface with the core and distal-lingual sites at the cervical surface.

Thermal mismatch of the two ceramics may increase tensile stress development in the veneering porcelain; therefore, it is advisable to have the difference in thermal expansion for the two ceramics as small as possible.

The outline is critical for the veneering porcelain, especially when it ends with a knife-edge; therefore, a chamfer with collar preparation is advisable. The non-uniform cement layer does not decrease the tensile stresses in the veneering porcelain at the interface between veneering porcelain and core, but can increase the maximum shear stresses in the cement layer at the bonding surfaces to values exceeding the bond strength of the cement layer to restoration and preparation. Therefore, the cement layer thickness should be as uniform and as thin [14] as possible.

This study indicates that following these specific design rules for full ceramic crowns may increase the longevity of these restorations.

5.6 Acknowledgements

The authors are grateful to W. de Ruiter (Cicero, Hoorn, the Netherlands) for the supporting work in the translation of the CAD model to the Finite Element Analysis software.

5.7 References


[26] Begazo CC, Van Der Zel JM, Waas MAJ and Feilzer AJ. Effectiveness of preparation guidelines for an all-ceramic restorative system. Am J Dent. in press.
