Exploring the limitations of fibre-reinforced composite fixed dental prostheses: Fibres (un)limited

Keulemans, F.

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CHAPTER 4

Three-dimensional finite element analysis of anterior two-unit cantilever resin-bonded fixed dental prostheses
4.1 Abstract

Objectives: The aim of this study was to evaluate, by finite element analysis (FEA), the influence of different framework materials on the biomechanical behaviour of anterior two-unit cantilever resin-bonded fixed dental prostheses (RB-FDPs).

Materials and Methods: The 3D FEA model consisted of a two-unit cantilever RB-FDP replacing a maxillary lateral incisor with a wing-shaped retainer on the central incisor and an adjacent canine. Five different framework materials were compared: direct fibre-reinforced composite (FRC-Z250), laboratory fibre-reinforced composite (FRC-ES), metal (M), glass-ceramic (GC) and zirconia (ZI). The isotropic materials were veneered with isotropic feldspathic porcelain, while the anisotropic material was veneered with isotropic particulate filler composite. A stress of 90 MPa at a 45° angle was applied to the incisal edge of the pontic.

Results: A similar stress pattern, with tensile stresses in the connector area, was observed in RB-FDPs for all materials. Maximal principal stress showed a decreasing order: ZI (239.6 MPa) > M (197.1 MPa) > GC (178.4 MPa) > FRC-ES (177.1 MPa) > FRC-Z250 (156.9 MPa). The maximum displacement of RB-FDPs was higher for FRC-Z250 (0.048 mm) and FRC-ES (0.035 mm) than for M (0.019 mm), GC (0.019 mm) and ZI (0.017 mm). Stress analysis depicted differences in location of the maximum stress at the luting cement interface between materials. For FRC-Z250 and FRC-ES the maximum stress was located in the upper part of the proximal area of the retainer, whereas for M, GC and ZI the maximum stress was located at the cervical outline of the retainer.

Conclusions: Within the limitations of this study, FEA revealed differences in biomechanical behaviour between RB-FDPs made of different framework materials. The general observation was that a RB-FDP made of FRC provided a more favourable stress distribution.
4.2 Introduction

Resin-bonded fixed dental protheses (RB-FDPs) have proven to be a reliable treatment alternative for the replacement of missing teeth [1] especially in cases were conservation of tooth tissue is needed and limited financial resources are available. According to a recent systematic review, RB-FDPs exhibit an estimated survival rate of 87.7% (95% confidence interval: 81.6%-91.9%) after 5 years [2]. Notwithstanding their good clinical performance, the most frequent complication was debonding, which occurred in 19.2% (95% CI: 13.8-26.3%) of RB-FDPs over an observation period of 5 years [2].

The use of more extensive preparation of the abutment teeth, including palatal or lingual coverage with 180-degree wrap-around, chamfer, cingulum rests, and proximal guide planes and grooves, is a way to improve the retention of RB-FDPs [3]. Another way to minimize debonding is to design RB-FDPs as a two-unit cantilever. This approach came into focus after the observation that many partially debonded three-unit fixed-fixed RB-FDPs could be successfully converted into a two-unit cantilever design after removal of the debonded retainer [4]. Elimination of interfacial stresses, induced by a combination of dynamic tooth contacts and differential movements of the abutment teeth, is the most widely accepted explanation for their successful clinical performance [3,5]. Several clinical studies of the last decade have demonstrated that two-unit cantilever RB-FDPs performed as well as or even better as their three-unit fixed-fixed counterparts [4,6-10].

The framework of RB-FDPs is traditionally made of metal alloys, but their poor aesthetics and the growing awareness towards possible adverse health effects of dental alloys [11] stimulated the interest in metal-free restorations. Nowadays, all-ceramics [10] and fibre-reinforced composites (FRC) [12] are viable alternatives for framework fabrication of RB-FDPs. Some clinical cases reported promising results for all-ceramic RB-FDPs [13,14]. In addition Kern et al. reported 5-year survival rates of 73.9 % for three-unit fixed-fixed designs and 92.3% for two-unit cantilever designs [10]. A recently published systematic review reported for FRC-FDPs a survival rate of 73.4% (95% CI: 69.4-77.4%) after 4.5 year [15]. During a 5 year multicenter clinical study FRC RB-FDPs exhibited a survival rate of 64% [16]. The differences in material properties, especially elastic modulus, adhesive properties and thermal expansion coefficient are believed to affect the mechanical and clinical performance of RB-FDPs [17]. In order to better understand the failure mechanism of two-unit cantilever RB-
FDPs, increased knowledge on the biomechanical behaviour of these restorations is needed.

The aim of the present study was to compare, by means of three-dimensional finite element analysis (3D FEA), the biomechanical behaviour of anterior two-unit cantilever RB-FDPs made of various framework materials.

4.3 Material and Methods

Definition of structures, geometric conditions, and materials

In order to create a FE model, a physical model of a single tooth gap in the anterior right maxilla, consisting of a central incisor, a missing lateral incisor and a canine (Figure 4.1A), was created. The central incisor served as the abutment tooth, but was not provided with a retainer preparation. The missing lateral incisor was replaced by a two-unit cantilever RB-FDP (Figure 4.1C) with a retainer on the central incisor. A wing-shaped retainer design, which enwrapped the palatal and distal surface of the abutment tooth, was selected and the pontic was shaped according a modified ridge lap design.

A dental CAD/CAM system (Dental Cadim 107D, Advance Co. Ltd., Tokyo, Japan) was used for measuring the model of the single tooth gap and the replica of the FDP at 0.25mm intervals, where after the captured data points were plotted in a 3D CAD software (VX 7.5, VX Co. Ltd., Florida, USA) in order to construct the 3D model. The 3D model of the single tooth gap and the RB-FDP were joined together and subsequently the cement layer was created manually. The model was converted to 3D solid models (ANSYS 11 Sp, ANSYS Inc., Houston, TX, USA).

The geometry of the healthy standard tooth as abutment has been previously described [18]. Not only the natural tooth geometry, but also the composition was mimicked, by including enamel, dentin and pulp tissues into the models. On the basis of the contours of the solid model, root under the bone, periodontal ligaments and alveolar bone volumes were not created. Three-dimensional FE model of the cement layer is shown in Figure 4.1B. The thickness of the cement layer was maintained at 100 μm.
Materials properties are deviated from clinically used materials (reference brand between parentheses): hybrid particulate filler composite (PFC) for laboratory use (Estenia C&B; Kuraray medical Inc., Tokyo, Japan), hybrid PFC for chairside use (Filtek Z250; 3M ESPE, MN, USA), unidirectional FRC for laboratory use (Estenia C&B EG fiber; Kuraray medical Inc., Tokyo, Japan), unidirectional fibre-reinforced composite for direct and chairside use (everStick C&B; StickTech Ltd., Turku, Finland), Au-Pd alloy (Olympia; J.F. Jelenko, Armork, NY, USA), lithium disilicate glass-ceramic (IPS Empress 2; Ivoclar-Vivadent, Schaan, Liechtenstein), zirconia (InCeram Zirconia; Vita, Bad Säckingen, Germany), feldspathic porcelain (Creation; Klema, Meiningen, Austria), resin-based luting cement (Variolink 2; Ivoclar-Vivadent, Schaan, Liechtenstein), enamel, dentin and pulp. The material properties, mostly obtained from existing literature, are summarised in Table 4.1. The materials were assumed to be isotropic, homogeneous, and linear-elastic, except for the FRC. The
mechanical behaviour of a unidirectional continuous FRC, influenced by their anisotropic (orthotropic) properties, can be described by 3 young’s moduli, 3 Poisson’s ratios and 3 shear moduli [19]. Twenty-node brick element as solid 95 in ANSYS has the anisotropic material option. Anisotropic material directions corresponded to the element coordinate directions. The orientation of the element coordinate system was altered in such a way it matched the fibre direction.

Table 4.1 Elastic properties of the materials used in the FE model.

<table>
<thead>
<tr>
<th></th>
<th>E modulus (GPa)</th>
<th>Poisson’s ratio</th>
<th>Shear modulus (MPa)</th>
<th>References</th>
</tr>
</thead>
<tbody>
<tr>
<td>Enamel</td>
<td>80.0</td>
<td>0.30</td>
<td>-</td>
<td>[20]</td>
</tr>
<tr>
<td>Dentin</td>
<td>17.6</td>
<td>0.25</td>
<td>-</td>
<td>[21]</td>
</tr>
<tr>
<td>Pulp</td>
<td>0.002</td>
<td>0.45</td>
<td>-</td>
<td>[22,23]</td>
</tr>
<tr>
<td>Resin luting cement</td>
<td>8.3</td>
<td>0.24</td>
<td>-</td>
<td>[24]</td>
</tr>
<tr>
<td>Chairside PFC</td>
<td>11.5</td>
<td>0.31</td>
<td>-</td>
<td>[25,26]</td>
</tr>
<tr>
<td>Laboratory PFC</td>
<td>22.0</td>
<td>0.27</td>
<td>-</td>
<td>[19]</td>
</tr>
<tr>
<td>Chairside FRC</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>longitudinal (X)</td>
<td>46.0</td>
<td>0.39</td>
<td>16.5</td>
<td>a</td>
</tr>
<tr>
<td>transverse (Y,Z)</td>
<td>7.0</td>
<td>0.29</td>
<td>2.7</td>
<td></td>
</tr>
<tr>
<td>Laboratory FRC</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>longitudinal (X)</td>
<td>39.0</td>
<td>0.35</td>
<td>14.0</td>
<td>[19]</td>
</tr>
<tr>
<td>transverse (X,Y)</td>
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<td>0.11</td>
<td>5.4</td>
<td></td>
</tr>
<tr>
<td>Lithium disilicate glass-ceramic</td>
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<td>0.25</td>
<td>-</td>
<td>[24]</td>
</tr>
<tr>
<td>Zirconia</td>
<td>205</td>
<td>0.22</td>
<td>-</td>
<td>[24]</td>
</tr>
<tr>
<td>Au-Pd alloy</td>
<td>103</td>
<td>0.33</td>
<td>-</td>
<td>[27,28]</td>
</tr>
</tbody>
</table>

a Data obtained by StickTech Ltd. (Turku, Finland)

Five different two-unit cantilever RB-FDP models of various framework materials were generated:
1) FRC-Z250: a FRC-FDP made of a continuous unidirectional E-glass FRC framework (Figure 4.2) veneered with hybrid PFC for direct and chairside use;
2) FRC-ES: a FRC-FDP made of a continuous unidirectional E-glass FRC framework veneered with hybrid PFC for laboratory use;
3) M: a metal-ceramic FDP made of type 3 Au-Pd alloy framework veneered with feldspathic porcelain;
4) GC: an all-ceramic FDP made of a lithium disilicate glass-ceramic framework veneered with feldspathic porcelain;
5) ZI: an all-ceramic FDP made of a zirconia framework and veneered with feldspathic porcelain.
A FRC framework was designed with thickness of 0.6 mm and a height of 3.0 mm [29]. The three-dimensional FE model of the FRC framework and its position in relation to the RB-FDP is shown in Figure 4.2.

![Image](image_url)

**Figure 4.2** 3D FE model of a two-unit cantilever FRC RB-FDP: position of the FRC framework in relation to the FDP and the abutment teeth is shown.

*Mesh generation, boundary conditions, and data processing*

In order to avoid quantitative differences in the stress value in the models, all solid models were derived from a single mapping mesh pattern that generated 103,861 twenty-node brick element (Solid 95 in ANSYS) and 154,784 nodes. The loading and boundary conditions are depicted in Figure 4.3. A stress of 90 MPa was applied at a 45° angle to the incisal edge of the pontic. The final element in all directions of FE model abutment tooth was fixed and distal direction of contact area to canine was fixed. FE analysis was presumed to be linear static. FE model construction and FE analysis were performed on PC workstation (Precision Work Station M90, Dell Inc., Texas, USA) using FE analysis software ANSYS 11. The locations and magnitudes of the principal stress (MPa) and displacement (mm) were identified and used for evaluating the biomechanical behaviour. Maximum principal stress describes the highest in-plane stress and can be regarded to be a tensile stress.
4.4 Results

Stresses in the FDP

Differences in maximum principal stress were observed (Figure 4.4 and Table 4.2) between the different framework materials and showed a decreasing order: ZI (239.6 MPa) > M (197.1 MPa) > GC (178.4 MPa) > FRC-ES (177.1 MPa) > FRC-Z250 (156.9 MPa). Maximum principal stress concentrations were located in the connector area, more precisely at the occlusal embrasure, for all framework materials. However, additional stress concentrations were observed at the contact area with the adjacent tooth for all framework materials and at the mesio-cervical edge of the retainer for GC (20-30 MPa), M (30-40 MPa) and ZI (50-70 MPa). The principal stresses at the contact area with the adjacent tooth were lower for FRC-ES and FRC-Z250 (30-40 MPa) in comparison to GC (50-70 MPa), M and ZI (>70 MPa).

Stresses at the cement-retainer interface

Differences in maximal principal stress were also observed (Figure 4.5 and Table 4.2) at the cement-retainer interface between all framework materials and showed a decreasing order: ZI (60.8 MPa) > M (36.1 MPa) > GC (32.7 MPa) > FRC-ES (23.9 MPa) > FRC-Z250 (17.5 MPa). Their location differed between all the
framework materials. Stress concentrations were observed in the upper part of the proximal area for FRC-Z250 and FRC-ES, while they were located in a semi-circular way around the connector and at the cervical edge of the retainer for M, GC and ZI.

**Figure 4.4** Principal stress distribution within two-unit cantilever RB-FDPs of various framework materials.

**Figure 4.5** Principal stress distribution at the cement-retainer interface for two-unit cantilever RB-FDPs of various framework materials.
**Stresses in the cement layer.**

FEA revealed (Figure 4.6 and Table 4.2) only slight differences in maximal principal stress between all framework materials and showed a decreasing order: FRC-Z250 (31.3 MPa) > ZI (27.5 MPa) > FRC-ES (27.3 MPa) > M (24.5 MPa) > GC (23.7 MPa). However, they were located in a different area of the cement layer. Highest stress concentrations were located in the upper part of the proximal area for FRC-Z250 and FRC-ES, while they were located at the cervical margin for M, GC and ZI.

**Stresses in the abutment tooth**

At the abutment tooth only slight differences in maximal principal stress were observed (Figure 4.7 and Table 4.2) between the different framework materials. Highest value was 34.9 MPa for FRC-Z250 and the lowest value was 30.9 MPa for FRC-ES. Once again, their location showed some differences. Highest maximal principal stress concentrations for FRC-Z250 and FRC-ES were observed at the upper middle part of the proximal area and were surrounded by a large area of stress concentration (17-31 MPa) which extended into the palato-cervical area. Highest maximal principal stress concentrations, on the other hand, for M, GC and ZI were located in a small region of the palato-cervical area of the abutment tooth.

**Displacement**

Differences in maximum displacement were observed in the pontic part of the RB-FDP between the different materials (Table 4.2). Higher displacement of the RB-FDP was encountered with FRC-Z250 (0.048 mm) and FRC-ES (0.035 mm) then with M (0.019 mm), GC (0.019 mm), and ZI (0.017 mm). Although, the maximum displacement of the retainer, cement layer, and abutment tooth revealed the same trend as those for RB-FDPs, a difference of 0.001 mm between highest (0.010 mm) and lowest (0.009 mm) value could not be regarded as clinically relevant. For that reason, maximum displacements of the retainer, cement layer and abutment tooth were regarded similar for all different materials.
Figure 4.6  Principal stress distribution within the cement layer for two-unit cantilever RB-FDPs of various framework materials.

Figure 4.7  Principal stress distribution at the abutment tooth for two-unit cantilever RB-FDPs of various framework materials.
4.5 Discussion

A static fracture strength test, during which a FDP is vertically loaded till failure, is the most common way to evaluate the mechanical behaviour of FDPs in laboratory conditions. The drawbacks of this approach are reckoned by researchers familiar with it. One of these drawbacks is the difficulty to fabricate uniform FDPs in terms of shape and dimensions. Although, FEA can be regarded as a relative easy and cost-effective way to evaluate the mechanical behaviour of complex structures, some limitations of our approach should be acknowledged. Some of these limitations can be drawn back to the simplifications made to the finite element models, eg, tooth model without roots, periodontal ligament [30] and, bone, and the assumptions made related to the material properties [31]. The latter illustrated by the fact that all materials, except FRC, were assumed to be isotropic, homogenous and linear elastic, despite the anisotropic nature of tooth tissue like dentine [32]. Therefore, one should be aware of the fact that the reported values regarding principal stress and displacement can not be regarded as absolute values, which was not the aim of this study. The main purpose of this study was to compare the biomechanical behaviour of anterior two-unit cantilever RB-FDP made of different framework materials. Nevertheless, the ideal approach is to use the results from both FEA and mechanical testing simultaneously, which may be able to provide more reliable and validated data than either method alone [33]. So mechanical testing on two-unit cantilever RB-FDPs in the same condition as this study could be a valuable asset.

In the present study, the FE model was loaded by applying a stress of 90 MPa in a 45° angle to the incisal edge of the pontic tooth. An applied stress of 90 MPa to a 5.5 mm² incisal area corresponds to a load of 495 N. The applied load is significantly higher than previously reported maximum anterior mastication loads of 108-382 N [34,35] and therefore can be regarded as the worst case scenario. In clinical circumstances, an anterior occlusal contact more closely resembles an area than a point, for that reason it was chosen to apply the load to a loading area.

Roots, periodontal ligament and bone, which are responsible for physiologic tooth mobility, were not included in the FE model. Under clinical conditions, a part of the loading is transferred via the roots and the periodontal ligament into the bone. The lack of physiologic tooth mobility in the present FE model negatively influences the outcome of the FEA, in such a way the principal stress values are overestimated. The effect of tooth mobility was illustrated by Rosentritt et al. [36], who found higher fracture strengths for anterior cantilever RB-FDPs when luted to abutment teeth with
high mobility [36]. Clinically, the rationale to use a cantilever design instead of fixed-fixed design is related to the teeth mobility. The risk for debonding of three-unit fixed-fixed RB-FDP from one end is relatively high, when teeth with increased mobility are involved abutment. A debonded retainer may result in secondary caries which is not diagnosed in time.

The present FEA revealed differences in biomechanical behaviour, more precisely stress distribution and displacement, between RB-FDPs made of different framework materials (Table 4.2).

Although the location of the maximum principal stresses and displacement, observed at the FDP level, was identical for all framework materials, the values differed. The differences in displacement and principal stress can be explained by the differences in elastic modulus (stiffness) between the framework materials. RB-FDPs made of materials with a higher stiffness suffered less displacement, but higher principal stress than those made of less stiff materials, which can be illustrated by comparison of zirconia and chairside FRC. Zirconia has a elastic modulus of 205 GPa and showed 0.017 mm displacement with 239.6 MPa maximum principal stress in comparison to the 0.048 mm and the 156.9 MPa by the chairside FRC with an elastic modulus between 11 GPa (chairside hybrid composite) and 46 GPa (FRC). The highest maximum principal stress was located at the occlusal embrasure of the connector. It has to be noticed that the connector in our FE model was designed with a sharp embrasure and that stresses at the occlusal embrasure of the connector can be significantly decreased by changing the connector design [37] and the radius of curvature of the connector strongly affects the fracture resistance of a FDP [37,38]. Recently, Plengsombut et al. confirmed this finding by revealing a significant lower fracture strength for specimens with a round connector in comparison to those with a sharp connector [39].

A comparable situation with regard to stress values was found at the level of cement-retainer interface. Far more interesting were the differences in location between FRC on one hand and M, GC and ZI on the other hand (Figure 4.4). A possible explanation is the difference in design between both groups of FDPs. In a FRC-FDP the stiffer fibres transfer the stress from the pontic to the central part of the retainer corresponding to the connector location, in contrast to FDPs (M, GC and ZI) with a uniform framework design were the stress is transferred to an area around the connector and towards the cervical margin of the retainer. Debonding of the FDPs due to premature failure of the adhesive interface between retainer and cement layer, is likely to be caused by such unfavourable stress location in combination with direct
exposure to the oral environment. Especially zirconia, known for it’s questionable adhesion to resin luting cements [40,41], will be prone to adhesive failure. At the level of the cement layer there was a only a slight difference between maximum principal stress values of all framework materials, but as expected the differences in location, as seen at the cement-retainer interface, between FRC on one hand and M, GC and ZI on the other hand (Figure 4.6) became more pronounced at the cement layer. It is interesting to notice that the cement layer, in the case of M, GC and ZI, is able to absorb the stresses in the area surrounding the connector and to dissipate those stresses towards the cervical outline. Such unfavourable stress transfer can result in premature failure of the cement layer.

The difference in maximum principal stress value between different framework materials was even lower at the level of the abutment tooth. However, the location of the stress concentration, as depicted in Figure 4.7, was different. Adhesive failure at the enamel-cement interface is not very likely to occur, as enamel bonding is a reliable procedure with reported values for resin luting cements, like Variolink 2, of 49.3 MPa [42].

<table>
<thead>
<tr>
<th>Table 4.2</th>
<th>Maximum and minimum principal stress (MPa) and displacement (μm) for two-unit cantilever RB-FDPs of various framework materials.</th>
</tr>
</thead>
<tbody>
<tr>
<td>FDP</td>
<td>Cement-retainer interface</td>
</tr>
<tr>
<td>max</td>
<td>min</td>
</tr>
<tr>
<td>FRC-Z250</td>
<td>156.9</td>
</tr>
<tr>
<td>FRC-ES</td>
<td>177.1</td>
</tr>
<tr>
<td>GC</td>
<td>178.4</td>
</tr>
<tr>
<td>ZI</td>
<td>239.6</td>
</tr>
<tr>
<td>M</td>
<td>197.1</td>
</tr>
</tbody>
</table>

Based on the results of this study the predominant failure mode of two-unit cantilever RB-FDPs for each framework material might be predicted. Zirconia and metal RB-FDPs are suspected to fail most likely because of debonding. A multitude of clinical research on cantilever metal RB-FDPs corroborates this prediction [7-9], since debonding was reported as the major reason of failure. Metal alloys exhibits plasticity,
which can explain this mode of failure. On the other hand, only a limited amount of \textit{in vitro} studies on zirconia RB-FDPs are available. It was shown that minimal invasive cantilever RB-FDPs subjected to fatigue loading, predominantly failed due to debonding \cite{36,43}. However, the same studies showed a decrease in percentage of debonding in favour of retainer fractures, when a more retentive retainer design was used. Although, one should be aware that the high stress concentrations at the mesio-cervical edge of the retainer indicates (Figure 4.5) that retainer fracture is most probably the result of partial debonding. Due to partial debonding more complex torque and bending forces acts on the retainer, which results in retainer fracture.

Glass ceramic and FRC RB-FDPs might be more susceptible for connector fractures. Since no studies on cantilever glass ceramic RB-FDPs are conducted, the only studies available are those on cantilever alumina RB-FDPs \cite{10,44,45}. These cantilever alumina RB-FDPs exhibited a 5-year survival rate of 92.3\% \cite{10}. During their study only one cantilever RB-FDP was lost due to fracture of the connector. Koutayas \textit{et al.} reported connector fracture as the predominant fatigue failure of cantilever alumina RB-FDPs \cite{44,45}. Since glass ceramic exhibits flexure strength of 252 MPa \cite{46}, which is inferior to the flexure strength of alumina (429 MPa) reported by Tinschert \textit{et al.} \cite{47} and their bond strength to resin luting cements is superior to that of alumina \cite{48}, the previous described studies can be regarded as representative for the affirmation of their expected failure mode. Clinical \cite{49} and \textit{in vitro} \cite{50,51} findings on FRC RB-FDPs also confirms this prediction. In comparison to glass ceramic and zirconia, were connector fracture results in an immediate aesthetic problem, this is not the case for FRC. From an aesthetic point of view the fibre reinforcement fulfils a fail-safe situation, because even after connector fracture the fibre reinforcement protects the FDP from complete debonding.

The results of this study on anterior two-unit cantilever RB-FDPs can be compared to those of Shinya \textit{et al.} \cite{17} on anterior three-unit fixed-fixed RB-FDPs. It should be noticed that the FE model and material properties were exactly the same for both studies, but that only FRC-, and metal-based three-unit fixed-fixed RB-FDPs were evaluated by Shinya \textit{et al.} \cite{17}. It is interesting to observe that the difference in principal stresses between various framework materials is higher for three-unit fixed-fixed designs than for two-unit cantilever designs. This suggests that the influence of framework material is less important for two-unit cantilever designs.

Metal-based anterior two-unit cantilever RB-FDPs, proven to be a clinically viable treatment option \cite{4,6-9}, can be regarded to be the gold standard for comparison with the other materials. Although acceptable bond strength to resin luting cements
can be achieved by glass ceramics, their low strength in combination with the less even stress distribution from loading area towards abutment tooth makes it not to be a suitable material for the fabrication of anterior two-unit cantilever RB-FDPs. FRC-based RB-FDPs seems to be more promising as they exhibits a good bond strength to resin luting cement and more even stress distribution. Nevertheless, they are at the moment only suitable as low cost temporary alternative due to the low strength of the veneering composite. Further improvements can be expected from modified framework designs [52] and improved resin composites [53]. Zirconia, regardless of its high strength, does not seems to be the ideal material for cantilever RB-FDPs, due to the unfavourable stress distribution and low bond strength to resin luting cement leading to premature debonding. Recent improvement of the adhesive performance of zirconia by selective infiltration etching increased the bond strength to Panavia F2.0 up to 49.8 MPa [41]. The achievement of a strong and durable bond with zirconia-based materials, makes it a most promising alternative to metal-based anterior two-unit cantilever RB-FDPs.

4.6 Conclusions

Within the limitations of this study, FEA revealed differences in biomechanical behaviour between RB-FDPs made of different framework materials:

1. The general observation was that a RB-FDP made of FRC provided a more evenly distributed stress pattern from loading area towards abutment tooth.
2. Maximum principal stress was located at the occlusal embrasure of the connector for all framework materials: highest value was found for ZI, while the lowest for FRC-Z250.
3. Advanced stress analyses suggest a possible difference in predominant failure mode: connector fracture for FRC-, and glass ceramic-based RB-FDPs and debonding for metal-, and zirconia-based RB-FDPs.
4. A stress concentration was found at the contact area with the adjacent tooth, indicating that the applied load is partially transferred to the adjacent tooth.
4.7 References


