Polymerization and loading stress distribution in adhesive resin-based composite class II restorations
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Citation for published version (APA):
Ausiello, P. (2002). Polymerization and loading stress distribution in adhesive resin-based composite class II restorations

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CHAPTER 3

Effect of adhesive layer properties on stress distribution in composite restorations – a 3D finite element analysis

Introduction

Restoration means a change of the natural tooth biomechanical balance. This is particularly true of resin-based composite restorations since the stiffnesses of the materials involved can vary greatly and do not fully match those of natural teeth. Problems will arise when they are submitted to stressing conditions. Polymerization shrinkage on one hand (Davidson, 1984; Magne, 1999a) and cyclical (fatigue) loading (Darbyshire, 1988; Ausiello, 1999) on the other hand can disorganize the restoration’s coherence. The polymerization shrinkage, ranging from 1.5% to 3% of the total material volume (De Gee et al., 1993) is a major problem in adhesive filling techniques. The restrained contraction produces stresses, which can exceed the cohesive and adhesive strength values of the restorative materials themselves (Feilzer et al., 1987). Stress can also interfere with the adhesive interface, enamel or dentine substrate. The use of low modulus restorative materials or the application of flexible adhesive linings showed to render release of such stresses and thus can be adopted as a tool to reduce composite restoration deterioration (Kemp-Scholte and Davidson, 1990; Davidson and Abdalla, 1993). Utilization of low modulus filling materials is not always possible in stress bearing areas, the restoration

Published in Dental Materials, 2002; 4: 8-16.
has to be strong and wear resistant, which implies highly loaded and thus high modulus composites (Peters, 1999; Davidson, 1986). The second option, using flexible linings, can be realized more easily as the dentine bonding systems (DBS) are unfilled or only lightly filled resins and thus possess a low modulus. A relatively thick bonding layer of 50-150 micron proved to be effective in levelling the mismatch of modulus values at the restoration-tooth structure interface (Davidson and Davidson-Kaban, 1998). The aim of the present study was to apply engineering tools to identify the adhesive lining thickness and flexibility (sufficient stress absorbency). By doing this procedure we aim to prevent undesirable cusp displacement and critical interfacial stresses that might lead to premature failure of the adhesively restored tooth by reason of polymerization shrinkage and mechanical loading.

Materials and methods

**CAD/FEM model.** A 3D CAD model of a human upper premolar was realized by digitising a plaster model with a laser scanner and the resulting profiles were used in solid tooth model generation. Literature data on the tooth morphology for the definition of the dentine and enamel volumes (Braden, 1976) and a plaster model (Thanaka model, Japan, 1978) for the external shape definition were used. Crown and roots were constructed in two different phases and then assembled. The crown was build up by digitising an upper premolar tooth plaster model on the scale of one to five by a Cyberware laser scanner. Over two hundred profiles were generated at 0.33 mm increments by vertical and horizontal directions scanning. Amongst the profiles, only 34 were collected, 17 vertical and 17 horizontal at 2 mm increments and they were assembled in a 3D wire-frame structure by means of a 3D CAD (Autocad 12, Autodesk, Inc., Neuchatel, Swiss, 1992).
The 3D curves were exported in Pro-Engineer 16.0 (Parametric Technology Co., Waltham, MA, USA, 1994), where a solid model was generated by fitting the horizontal and vertical profiles. The model was cut in the cervical area in order to obtain the final crown. The roots were modelled by their mesial-distal and buccal-lingual representations taken from literature (Wheeler, 1974). The two representations were scanned and 8 vertical profiles were generated imitating the scanned images. The roots were constructed by fitting the vertical profiles (Ausiello et al., 2001). The pulp region was obtained in an analogous way and subtracted from the roots. The crown and the roots, with the pulp chamber, were assembled in the final model. A parametric cutting plane was chosen to generate different cavities and MOD preparations. In Fig. 1, the class II MOD cavity is shown (3.5 mm occlusal width). The cavity design was characterized by a flat floor and sharp internal line angles. No bevel was done at the approximal and occlusal margins. The preparation derived was flat from proximal to proximal surface.

The solid model was transferred into a FEM program (ANSYS Rel. 5.3, ANSYS Inc., Houston, USA, 1994) where a 3D mesh was

![Crown wire frame and cut in cervical area](image1)

![Tooth dentine](image2)

![3D solid model](image3)

![3D FEA model](image4)

**Fig. 1** - Solid and 3D models generation.
created (Ausilio et al., 2001). In the previous work we explained that the volumes were redefined and meshed with 8 node brick and 4 node tetrahedral elements, resulting in 7282 elements (3376 hexahedral and 3906 tetrahedral shape elements) and 5236 nodal structures. Different material properties were assigned to the elements according to the volume definition (Fig. 1). The adhesive layer was modelled in the FEM program using spring elements connecting the nodes from the cavity wall of the natural tooth with those of the composite restoration (Fig. 2).

Normal and shear stiffness of the adhesive layer were simulated by connecting each node pair with three different springs, one in the direction normal to the composite-tooth interface and the other two parallel to the interface (Fig. 1). The spring constant $K_i$ for the $i^{th}$ element normal to the interface layer was evaluated by:

$$K_i = \frac{A_i E}{l}$$  \hspace{1cm} (1)

where $A_i$ is the nodal average area evaluated as the average value of the concurring element areas to each node, $E$ is the Young’s modulus.

![Fig. 2 - Adhesive interface modelling by means of spring elements.](image)
of the adhesive material and \( l \) is the thickness of the interface. A nodal average value for the area was assumed in order to obtain a homogeneously distributed rigidity inside the adhesive, overcoming the effect of the mesh non homogeneity or edge presence.

For the springs placed in the interface plane, the rigidity was defined as:

\[
K_i = \frac{A_j G}{l} \tag{2}
\]

where \( G \) is the shear modulus of the adhesive material. For each node pair the two springs parallel to the interface had the same rigidity. By the end, 1836 springs were arranged in the tooth model, every one with its own constant.

The above formulas show that the layer rigidity increases with the elastic modulus of the adhesive material and decreases with its thickness. A thin interface layer of an adhesive, characterized by a low modulus, could then act as a thicker one with a higher modulus adhesive since they are mechanically equivalent.

**Experimental model validation**

In order to validate this class II MOD finite element model, a compression test was performed on a class II MOD restored human upper premolar until fracture of the samples. Ten caries-free human upper premolars were considered. Class II MOD cavities were prepared with a diamond bur at high speed under water coolant. Axial and gingival walls were cut non retentive, approximately at 90° angle. No bevel was prepared at the cavosurface enamel angles. The material combination used was: composite Prodigy (Kerr, USA), with a Young’s modulus of 12.5 GPa and polymerization shrinkage of 2.73 ± 0.31% by volume, in combination with Optibond FL adhesive (Kerr, USA). The samples were inserted up to the cementum-enamel junction in a
steel cylindrical ring with the apical root area in contact with the steel ring floor. Subsequently it was filled with rigid resin composite material, so that only the material deformation within the tooth was recorded. The cylinder was clamped to the test machine and the load was applied vertically by means of a 6 mm diameter steel cylinder with the axis parallel to the tooth axis, in order to simulate one main important force which develops during occlusion. A 1 mm/min constant rate was imposed to the loading cylinder and the vertical displacement and the axial load were acquired until the restored tooth fractured. The same test was simulated using the FE analysis and the results matched. Under loading, restored teeth mostly fail along the tooth-restoration interface. Fracture was simulated by means of a special ANSYS feature element named “birth and death”. Using this feature, the load was applied to the model by steps: if in a single load step, the force at the ends of a spring reaches a greater value than the critical one, that spring will be killed in the next step. Killing a spring means reducing rigidity by a factor of $10^{-6}$, so that it does not effectively contribute to the global structure stiffness in the following calculations.

In Tab. 1 the FEM model material properties are listed. These data were collected from literature (Braden, 1976; Wheeler, 1974; Versluis, 1996) and from Kerr Inc.

For every spring $i$, the critical force $F_{\text{crit},i}$ was evaluated by:

$$F_{\text{crit},i} = \sigma_i A_i$$

where $A_i$ is again the local average adhesive area and $\sigma_i$ is the adhesive strength (tensile strength for the springs normal to the composite-natural tooth interface and shear strength for those parallel to the interface). A higher layer thickness leads to a lower rigidity as well as to a higher layer compliance, while the adhesive strength remains unaltered. In fact, the adhesive thickness $l$ does not appear in the
formula (3) for the critical force, meanwhile it appears in the formulas (1) and (2) for the spring rigidities. In this way restored teeth with the same adhesive material but different layer thickness will fail under the same load while exhibiting different deformations.

**Numerical simulations**

The load conditions applied to the restored tooth structure were: vertical occlusal load only, polymerization shrinkage (for the evaluation of the stresses arising from the composite polymerization shrinkage, a volumetric contraction was applied to the composite) and both conditions. In particular, in Tab. 2 the 6 different models used in the numerical simulations are listed together with the various load conditions applied. Moreover, in all the models the external roots nodes were constrained in all directions. Adhesive mechanical properties were already listed in Tab. 1 and were identical for all the restored tooth models. In total 11 different numerical analyses were performed. The comparison test loading cylinder was modelled with 3D elastic
Fig. 3 - Experimental and theoretical stress-strain curves.

beams (Fig. 3). The beams have one end in common and the other one on a cusp. End rotations were not constrained. The common end was displaced in the central position of the loading cylinder section in the experimental test. The load was applied on the tooth in two points (Fig. 3) through the beams elements (blue lines) on the cusps (red

Fig. 4 - Crack growth in compression test.
areas). The resulting force $F$ crossing the central position of the loading cylinder section was 400 N (Fig. 3, red arrow). The beam elastic properties were thought to be infinitely rigid compared to the tooth. Resin support was not modelled and it was considered to be as rigid as the loading system.

Moreover, the following assumptions were made:

- The pulp chamber was modelled as a void because of its negligible stiffness and strength;

- A static linear analysis was performed: all materials were considered elastic throughout the entire deformation, which is a reasonable assumption for brittle materials in non failure conditions (Williams and Edmunson, 1984);

- Dentine is an elastic and isotropic material while enamel is an elastic and anisotropic material from a mechanical point view (Rees and Jacobsen, 1995);

![Crack simulation](image)

**FIG. 5** - Tooth failure in compression.
Fig. 6 - Stress due to occlusal loading and stress due to polymerization shrinkage for sound tooth and restored teeth with different elastic modulus.

- The anisotropic enamel behaviour is neglected in this paper and it is assumed homogeneous and isotropic according to Darendeiler et al. (1998) and Versluis (1996);

Fig. 7 - Stress due to polymerization shrinkage for restored teeth with a different adhesive layer thickness.
The numerical analyses performed were all linear static, except the axial load simulation that was a 10 sub-step non-linear analysis. At every sub-step a constant load increment was applied.

Model A (Fig. 6) is the sound tooth, while Model B (Fig. 7) is a restored tooth where the adhesive interface is assumed infinitely rigid. The three models C (C1, C2 and C3) and Model D (Fig. 6) represent a restored premolar with an elastic adhesive interface, respectively of different layer thickness and different composite modulus. Regarding the adhesive layer, it is more correct to distinguish the different models using the layer normal rigidity per unit area $\bar{K}_A$ rather than the layer thickness. In fact we already mentioned that different combinations of adhesive elastic modulus and layer thickness could lead to identical interface rigidities. The parameter $\bar{K}_A$ can be obtained by:

$$\bar{K}_A = \frac{E}{l}$$ (4)

Tab. 3 lists the layer rigidities for the restored premolar models. In Model B the nodes from the side of the composite are “merged” with

<table>
<thead>
<tr>
<th>Model</th>
<th>Material properties</th>
<th>Load conditions</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Adhesive Thickness (µm)</td>
<td>Composite elastic modulus (GPa)</td>
</tr>
<tr>
<td>A</td>
<td>Sound tooth</td>
<td></td>
</tr>
<tr>
<td>B</td>
<td>Merge</td>
<td>25</td>
</tr>
<tr>
<td>C1</td>
<td>50</td>
<td>25</td>
</tr>
<tr>
<td>C2</td>
<td>100</td>
<td>25</td>
</tr>
<tr>
<td>C3</td>
<td>150</td>
<td>12.5</td>
</tr>
<tr>
<td>D</td>
<td>50</td>
<td>12.5</td>
</tr>
</tbody>
</table>

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those of the side of the natural tooth; it means they are strictly joined together. A "merged" interface can be seen as an interface with zero thickness and infinite rigidity, as equation (4) demonstrates. If the thicknesses reported in Table 2 are considered, an elastic modulus of 1 GPa should be taken for the adhesive in order to obtain the normalized stiffness reported in Table 3. However, for adhesives of higher modulus, an appropriate thickness could be calculated from (4).

### Results

The experimental and theoretical curves are compared in Fig. 3. The two similar stress-strain behaviours confirm the validity of the model. Even the ultimate properties could be fitted by the theoretical curve. The experimental test showed a mild non-linear behaviour near to the starting point and a linear behaviour in the rest of the curve. This effect is not due to the real material or geometrical non-linearity but to the initial system assessment that include, above all, contact and sliding effects. These phenomena are not important in order to validate the FEM model.

In Fig. 4 Von Mises stress maps are shown for three of the 10 sub-steps chosen across the failure zone: the corresponding load condi-

<table>
<thead>
<tr>
<th>Model</th>
<th>$\bar{K}_i = (E/l)(10^3)$/mm - mm$^2$</th>
</tr>
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<tbody>
<tr>
<td>B</td>
<td>$\infty$ (infinite rigidity)</td>
</tr>
<tr>
<td>C1</td>
<td>20</td>
</tr>
<tr>
<td>C2</td>
<td>10</td>
</tr>
<tr>
<td>C3</td>
<td>6.67</td>
</tr>
<tr>
<td>D</td>
<td>20</td>
</tr>
</tbody>
</table>
tions are also illustrated. In order to more clearly represent the crack occurrence, the number of elements that reached the failure load value (*killed springs*) and the lingual cuspal deflection were plotted both related to occlusal load (Fig. 5). Von Mises stress maps were also extracted from the other numerical simulations. In Fig. 6 the sound tooth behaviour at 400 N occlusal loading (Mod. A) is compared to the behaviour of two restored teeth at the same load condition. The restored tooth models present the same adhesive layer thickness (50 μm) but differ in composite modulus, one being (Mod. C1) more rigid than the other (Mod. D). For the same restored teeth the stress maps, obtained by simulating the shrinkage load, are also shown. In Fig. 7 tooth structure responses to polymerization shrinkage are also depicted. Stress maps were evaluated for 4 high modulus composite restored teeth, differing in adhesive layer thickness (0 μm, 50 μm, 100 μm and 150 μm corresponding, respectively, to Mod. B, Mod. C1, Mod. C2 and Mod. C3). Evaluating stress difference maps is another valid way to rapidly depict tooth model behaviour differences. In Fig. 8 two of these Von Mises stress difference maps are shown. The left coloured map of Fig. 8 concerns the difference between a restored tooth with a highly rigid adhesive layer and another one with a low rigid one, both under polymerization shrinkage. The first model

![Image of stress maps](image_url)

**Fig. 8** - Shrinkage load: stress difference map between a more rigid adhesive layer restored tooth and a less rigid adhesive layer restored tooth.
presents a 50 μm adhesive layer thickness (Mod. C1) and the second a 150 μm one (Mod. C3). In Fig. 8 the coloured map on the right shows the difference between a high modulus composite restored tooth (Mod. C1) and a low modulus composite restored tooth (Mod. D), both with a 50 μm adhesive interface thickness and under the combination of shrinkage and occlusal load.

**Discussion**

Recently, the FE analysis has been extensively used in dentistry because it shows the tooth mechanical behaviour in detail. In fact stress, strain and some other quantity value in every nodes of the structure can be known. Its great velocity represents the other interesting aspect of the numerical FE analysis when the behaviour of the structure must be evaluated under different boundary conditions (applied loads and constrains).

Numerical results for the axial test simulation predict that tooth fracture occurs between a 700 N and 800 N compression load (Fig. 4). This condition is in agreement with experimental data which provide an 804 N mean fracture load at a 264 N standard deviation (Ausiello et al., 1997). It demonstrates the validity of our assumptions and the FEA capability of complex structure simulation in critical conditions. Additionally, a more accurate failure load evaluation can be made with numerical simulation by simply increasing the sub-step number of the non-linear analysis. But it would provide a useless analysis because the experimental data have such a high intrinsic variance. On the other hand, it is the high experimental dispersion which suggests a numerical approach for a biological system mechanical analysis. The mechanical behaviour of restored teeth by a stress representation in two-dimensional plane-strain finite elements has been already analysed by Spears (1998) for adhesive class II restorations. The results indicated

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that, for an ideally bonded restoration, a modulus of 30 MPa absorbs the stresses during composite shrinkage. Any explanation or considerations on the effect of a lining or on the thickness of the adhesive layer was still absent. To some extent, numerical results can be obtained by 2D modelling, but it has some significant shortcomings. Firstly, human teeth are irregular so that they cannot be represented in 2D volumes (Sakaugchi et al., 1991; Khera, 1991; Magne, 1999b; Toparli et al., 1999b). Furthermore, the structure has no symmetry so a 2D modelling approach could neglect some important result details. The 3D FEM is preferred to obtain an optimal realistic analysis (Toparli et al., 1999a) and undoubtedly represents a more detailed way to obtain useful mechanical information on the stress re-distribution of Von Mises scales at the dentine-composite adhesive interface. Other methods can hardly cover the elastic behaviour of the various components in a very complicated anatomy. Our previous study (Ausiello et al., 2001) considered 11165 elements and 7340 node structure. Through this kind of analysis, it is possible to obtain reliable results which can provide realistic indications for clinical aspects. Under occlusal loading, our model predicts crack start at the top of the adhesive layer interface, near the lingual cusp tip where the force component, normal to the tooth-composite interface, reaches its maximum and then propagates towards the bottom. At a critical load value, the number of killed springs rapidly increases as the lingual cusp deflection (relative displacement between the cusps vertices) decreases (Fig. 5). Under the same occlusal loading condition, the sound tooth model exhibits a quite different mechanical behaviour from the restored teeth (Fig. 6). Sound teeth can distribute the applied (high) stresses more homogeneously because under these loading conditions, the rigid enamel does not deform significantly but transfers the deformation to the lower, more resilient dentine, and thus can rigidly “move” over it. In restored teeth, the restoration cavity interrupts the tooth bi-elastic structure and causes a stress concentration at the base of the
lingual cusp, where the applied force component is maximal. Instead, a considerable difference is noted in the case of polymerization shrinkage. The high modulus composite generates higher stresses compared to the low modulus one. The maximum stress areas accumulate in the lower adhesive part of the restoration, at the cervical zone, where micro cracking and leakage are often observed (Ausiello et al., 1999). This occurrence was also noted in a previous work but there the stress values were not significant (being too high) because the FEA models were realized without adhesive interfaces (Ausiello et al., 2001). This last result was also noted in previous studies by photoelastic analyses (Kinomoto and Torii, 1998) where the distribution of the internal stresses in a box-shaped cavity composite restoration was investigated. From this study it can also be concluded that it is possible to carry out shrinkage stress distribution studies on resin-based composites in restored teeth. Specifically, the importance of the polymerization shrinkage stress release by an elastic adhesive layer was demonstrated. Fig. 7 shows the release effect of an adhesive layer as a function of its thickness at a specific intrinsic rigidity. An infinitely rigid interface layer (0 μm thick) produces very high stress areas all around the tooth-restoration interface. In the models with an adhesive layer of a given, finite rigidity, high stress areas are still distributed along the tooth-restoration interface, but a maximum is observed at the restoration base. Consequently, the average stress in the cusps is lower. In fact, from the comparison of the stress maps in Fig. 7 for the models C1, C2 and C3, it is possible to note that by increasing the adhesive thickness, the maximum stress area at the bottom of the restoration (red area) decreases as well as the lower stress area on the buccal side (light-blue area). This means that the higher the adhesive thickness is, the higher the elastic release effect; the stress difference is transformed in adhesive layer deformation. The influence of composite and adhesive layer rigidities on the biomechanical response are shown in Fig. 8 where stress difference maps of a restored tooth are depicted for
two restored teeth respectively with a different interface layer thickness (50 μm and 150 μm), Mod C1 - Mod C3, and composite modulus (25 and 12.5 MPa). Mod C1 - Mod D, as previously shown in Fig. 6 and Fig. 7. The difference maps were obtained by subtracting the stress value of the corresponding nodes of the two FEA results. It can be noted that the stress difference between the restored tooth with higher interface rigidity and the other one with lower interface rigidity reaches higher values at the bottom of the composite restoration (gingival wall angle). The high stresses induced by rigid composite polymerization shrinkage could be released by a more elastic interface: in this way a 15 MPa difference could be significant. A stress difference map (Fig. 8) was also calculated for a restoration using composites with a different elastic modulus (25 and 12.5 GPa) but with the same adhesive thickness (50 μm). The applied load is a combination of an occlusal loading and a shrinkage stress condition; the higher the composite modulus is, the more the stress is at the bottom of the composite restoration (red area with 40 MPa difference), while the lower modulus composite relieves the stresses on the dentine structure (see Fig. 8, right picture, light blue area, –20 MPa). As already noted, discussing Fig. 6, the composite modulus is a variable which can be used to reduce polymerization stress. The limit is the mechanical behaviour of the restored tooth which forces the composite to possess a sufficient rigidity. But Fig. 6, Fig. 7 and Fig. 8 let us imagine that an optimised adhesive-composite choice can produce a meaningful reduction of stress induced by polymerization.

The adhesive layer thickness and rigidity are important variables in defining the restored tooth mechanical behaviour. 3D finite element analysis has been successfully used to visualize failure processes in adhesively restored teeth during polymerization and occlusal loading. It could be demonstrated that structurally modified teeth show a complex biomechanical behaviour during the early stages of the restoration. When shrinkage and occlusal loading stresses act simultaneously.
it is shown that the composite rigidity and adhesive interface resilience have opposing effects on the stress relief. The more rigid the composites used in the restoration are, the higher the polymerization shrinkage stress and the lower the cusp movement under occlusal loading. An appropriate way to limit the intensity of the stress transmitted to the remaining natural tooth tissues is to employ an adhesive layer of substantial thickness as a lining, able to partially absorb the composite deformations. A thin layer of a more flexible adhesive (lower elastic modulus) exhibits the same rigidity as a thick layer of a less flexible adhesive (higher elastic modulus). For adhesives and composites of different rigidities, the FE analysis allows the determination of the optimal adhesive layer thickness leading to maximum stress release while preserving the interface integrity.