RETENTION OF A POSTERIOR RESIN-BONDED FIXED PARTIAL DENTURE WITH A MODIFIED DESIGN – AN IN VITRO STUDY.

by

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A thesis submitted in conformity with the requirements for the degree of Master of Science
Graduate Department of Dentistry
University of Toronto

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Abstract

*In vitro* retention of a RBFPD made with a modified design (MD) was compared to that of a current design (CD) using five experimental groups (G). G1 (CD) included proximal grooves (PG), occlusal rests (OR) and lingual wings (LW). G2 and G3 (MD) were similar to G1 however PG were replaced with retentive slots (SL). G4 had only SL. G5 had OR, LW and non-retentive SL. Air abraded castings (nonprecious alloy) were cemented with Cement-It (G1, G2, G4 and G5) and Panavia 21 (G3). Following storage (7d), prostheses were subjected to compressive load cycling and then separated in tension. Mean separation forces in N were:

G1: 360.9 - G2: 525.4 - G3: 562.2 - G4: 448.6 - G5: 417.1

Statistical analysis showed that G1 and G5 had separation values significantly lower than the value achieved for G2 and G3. It is concluded that the added means of retention significantly increased RBFPD resistance to dislodgement.
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Introduction

1. Overview

It has been the goal of dentistry over many years to develop conservative approaches for the restoration of decayed teeth and for replacement of missing ones. Buonocore accomplished this, in part, with the development of acid etching of enamel in 1955, which allowed micromechanical bonding between enamel and composite restorations. With the success of bonding to enamel, dentists were encouraged to attempt new innovative applications including the bonding of acrylic resin denture teeth to replace congenitally missing lateral incisors (Portnoy, 1973). The same technique was also used to construct fixed partial dentures (FPDs) of intermediate length that would replace two missing teeth (Stolpa, 1975). While such restorations were successful to a certain degree, cohesive fractures of the resin usually occurred after a short time. The bond to the enamel interface appeared strong enough, but the cohesive strength of the resin was not adequate to resist the stresses applied to the prosthesis. Bonding a natural or a denture tooth to the patient’s teeth with acrylic resin is now used only as a temporary measure.

In 1973, Rochette introduced the idea of bonding a cast metal framework to enamel tooth structure in order to immobilise teeth during periodontal therapy or to replace a missing tooth. The framework was cemented with an unfilled resin (Sevriton) and the attachment between the resin and metal was based on mechanical retention created in the metal through undercut perforations. Following this idea, Howe and Denehy (1977) described the use of a resin-bonded fixed partial denture (RBFPD), without tooth preparation, for tooth replacement in the anterior segment of the mouth. Later Livaditis (1980) presented the design for the posterior region, describing slight modifications to the tooth structure to accommodate the prosthesis. However, the holes created in the metal wings weakened the framework and were not
sufficient to achieve the desired long-term retention.

With the introduction of electrolytic etching for RBFPDs in 1981 by Thompson, Livaditis and Del-Castillo, the bonding between resin and metal became stronger, and the resin-bonded fixed partial dentures became more reliable.

**1.1. Advantages of resin-bonded fixed partial dentures.**

1. The major advantage of RBFPDs is that more tooth structure is preserved than with traditional partial or full-coverage restorations. In addition, RBFPD design utilises a supragingival margin. It is known that subgingival margins can have deleterious effects on the periodontal tissues.

2. Aesthetic advantages are also observed. Since only the lingual surfaces of anterior abutments are covered, the labial surface remains untouched and, therefore, there is no labial collar to hide in the gingival crevice.

3. Since tooth reduction is restricted to enamel, usually no anaesthesia is required. Lack of involvement of dentin eliminates postoperative sensitivity. And the fact that the margins are supragingival simplifies impression taking without need for retraction cord packing.

4. Cost effectiveness is also an advantage. The total treatment time is somewhat reduced compared to traditional fixed partial dentures. Besides, laboratory costs are less than for FPDs.

**1.2. Disadvantages of resin-bonded fixed partial dentures.**

1. The major disadvantage of RBFPD is that the longevity of this kind of prosthesis is still uncertain. Studies, which are discussed below, show a large variation in survival rates for RBFPDs, especially in the posterior segment of the mouth. It is important to note, however, that many of those studies have used the early designs of the prosthesis.

2. Debonding sometimes occur in only one retainer and the patient may remain unaware of the problem. Partial debonding may lead to plaque accumulation under the retainer and subsequently, to development of dental caries.
3. Since tooth reduction is minimal, the thickness of the retainer has to be well controlled to avoid overcontour. Also the use of nonprecious alloy, indicated for RBFPDs, may be a contraindication of the technique for patients with nickel sensitivity.

4. The use of RBFPD is limited to short span cases.

5. Another disadvantages include the difficulty for space correction (when the edentulous space is larger than that of the missing tooth), alignment correction as well as almost impossible temporization.

1.3. Evolution of resin-bonded fixed partial dentures.

There has been an ongoing attempt to improve the longevity of RBFPDs. Efforts are being made to enhance mechanical retention with modifications in tooth preparation and improved framework design, as well as to improve the resin-metal bonding. This has been accomplished both through the use of special adhesive resins, including some that chemically bond to the alloy, and through surface treatment of the bonding areas of the cast restoration.

All modifications of the original design by Howe and Denehy (1977) and Livaditis (1980) were aimed at conservation of tooth structure and avoidance of involving the gingival crevice in the RBFPD outline.

The first design for posterior abutments (Livaditis, 1980) included the preparation of guiding planes by reduction of the lingual and proximal surfaces, and one occlusal rest to ensure correct seating of the prosthesis and to prevent occluso-gingival movement. Although satisfactory clinical results were reported by Livaditis (1981) in a short-term study (1 year), the inclusion of grooves, as suggested by McLaughlin (1981), and 2 occlusal rests (Barrack, 1993), seems to significantly improve the longevity of RBFPDs in long-term studies (Barrack, 1993; Samama, 1996).

Based on long-term studies with good success rates, some authors consider resin-bonded FPDs a conservative and viable alternative to conventional FPDs (Crispin, 1991; Hussey, Pagni & Linden, 1991; Besimo, 1993; Barrack, 1993). On the other hand, some dentists have lost their confidence in such technique because there
are a number of studies that indicated poor prognosis for this kind of prosthesis (Berekally & Smales, 1993; Thayer, Williams, Diaz-Arnold & Boyer, 1993; Priest, 1995; Kerschbaum, Haaster & Marinello, 1996).

Currently, with the advent of bonding to dentin, more conservative cavity preparations and conservative techniques for replacement of missing teeth are being developed. Recently, with the goal of obtaining better retention, El-Mowafy (1996) presented a new design for posterior RBFPDs. It consists of inclusion of proximal slots, reduction of proximal and lingual surfaces, and occlusal rests. The metal framework is made with metal projections that seat on the gingival floor of the slot cavities prepared in both abutment teeth adjacent to the edentulous space. Following cementation, the cavities are restored with a resin composite material. The design has the advantage of creating a larger area for bonding and allowing mechanical interlocking between the resin composite restoration, the metal framework, and the cavity walls.

This new modified design, however, is less conservative than the designs that have been previously used. Therefore, prior to examining the effect of this design on long-term longevity of posterior RBFPD, it is prudent to evaluate its retentive effect in an in vitro study.

2. Developments in Surface Treatment of the Metal

The first resin-bonded splints, which were described by Rochette in 1973, relied on macromechanical-based attachment of the framework to the resin cement for success. To allow the attachment between the unfilled resin and the precious metal framework, Rochette used undercut perforations in the metal, which provided mechanical retention, and a silane coupling agent to ensure a chemical bond. The main advantage of this technique was the absence of tooth reduction.

Based on the concept introduced by Rochette (1973), Howe and Denehy (1977) described a technique for fabrication and attachment of cast fixed prosthesis for the replacement of a missing anterior tooth. The procedure consisted of using a resin composite (Adaptic) and enamel acid-etch procedure for adhesion, without tooth reduction of the abutments. The metal framework was also perforated. However, the
use of nonprecious alloy was preferred because of its greater strength resulting in a framework with minimal thickness (Howe & Denehy, 1977; Livaditis, 1980). Howe and Denehy were extremely conservative in their expectations regarding the longevity of the prosthesis, referring to it as a provisional restoration.

The perforated framework design cemented with resin composite to a tooth with minimal or no reduction was known as “Rochette Bridge”. Besides the advantages, this technique had some disadvantages. The resin was left exposed, which could lead to abrasion or leakage of fluids between metal and resin or a fracture of the resin tag, leaving the underlying tooth enamel exposed to plaque accumulation and vulnerable to decay (Berekally & Smale, 1993). More significantly, the retention was limited only to the regions around the perforations and not throughout the wings.

2.1. Electrolytic etching technique.

In an attempt to overcome the problems associated with the Rochette Bridge, Thompson et al. (1981) and McLaughlin (1981) suggested a method to eliminate the perforations and render the entire metal surface retentive. The fitted metal surface was electrolytically etched to create a retentive surface to which resin composite would adhere. The technique provides irregularities in the surface of nonprecious metal alloys by selective dissolution of the most corrosion-sensitive phases in the metal.

In preparation for electrolytic etching, all portions of the framework, except those that come into contact with the enamel, are coated with sticky wax. The remaining surface is cleaned with a 50μm Al₂O₃ air-blasted abrasive. The framework is attached to an electrode and immersed in an acid bath. The acid or electrolyte, the current density, and the etching time have to be chosen according to the particular alloy used. After etching, the wax is removed, and the framework is immersed in an 18% hydrochloric acid solution in an ultrasonic cleaner for 15 minutes.

Livaditis and Thompson (1982) evaluated the tensile bond strength of resin composite (Concise) to two different alloys electrolytically-etched and compared it to a control that was only sandblasted and cleaned in HCl for 10 minutes. The results
showed superior values for electrolytically etched alloys compared to their control. They also compared the bond strength obtained between etched metal/resin to the results found in the literature. They concluded that the electrolytic etching technique resulted in a stronger bonding in the interface resin/metal (27.3 ± 3.7 Mpa) when compared to the interface resin/enamel (8.5 to 9.9 Mpa).

Brady, Doukoudakis and Rasmussen (1985) compare the bond strength of electrolytically etched metal bonded to enamel to that achieved with perforated metal in an in vitro study. Their results showed that the etched metal was capable of withstanding more than four times the breaking load of the perforated metal.

The “Maryland bridge”, as this technique came to be called colloquially indicating its place of development, provided the same aesthetic result and preservation of tooth structure as the Rochette bridge, but offered the hope of improved retention. Superior survival rate for Maryland bridges was reported in some clinical studies: Creugers, Kayser and Van’t Hof (1992) found a survival rate of 28% for perforated RBFPDs and 64% for electrolytically-etched RBFPDs during a seven-and-a half-year study; and Berekally and Smales (1993) reported a debonding rate of 70% for Rochette RBFPDs as opposed to 20% for Maryland RBFPDs.

The main disadvantages of the electrolytic etching are that it requires special equipment, and the procedure varies according to the metal alloy used as well as the area to be etched. Also, because the result is a surface that is microscopically roughened, it is not possible to assess the quality of the etching on the framework without microscopic examination, and the dentist must rely on the technician’s judgement for this purpose. Moreover, the etched surfaces can be easily contaminated resulting in a weak mechanical bond between the composite cement and the metal framework.

2.2. Alternatives to the electrolytic etching technique.

In 1985, Love and Breitman described a different technique to etch the alloy surface. The metal was placed in an immersion etchant prepared using equal volumes of nitric acid and a solution composed of 1 part hydrochloric acid and 1 part methanol. After 5 minutes of etching, the metal was placed in an ultrasonic cleaning
with 18% hydrochloric acid for 10 minutes. One of the problems was that the solution was not chemically stable, and it was hazardous to store because of pressure build-up within a closed container. Later, Doukoudakis, Cohen and Tsoutsos (1987) described the same procedure using a gel commercialised by Met-Etch Gresco Products Inc. According to the manufacturer, the formula was a stable gel and was capable of chemically etching metal retainers without an electrolytic procedure.

Comparisons between chemical etching and electrolytic etching of nonprecious alloys showed no statistical differences when shear or tensile forces were applied to specimens bonded together (Love & Breitman, 1985; Krueger, Diaz-Arnold, Aquilino & Scandrett, 1990; Kuyino, Levine, Grisius & Fenster, 1990). However, in one study, the bond strength between enamel and chemically treated retainers was found to be higher when compared to electrolytically etched retainers (El-Sherif, Shillingburg & Duncanson, 1989).

Shen, Forbes, Boettcher, Dvidevi and Morrow (1983) described a metal bonding technique based on the characteristics of some orthodontic brackets. It consisted of incorporating a prefabricated mesh casting pattern on the fitted surface of the metal framework in order to create surface roughness. They compared the bond strength of cast mesh surface with that of acid-etched surface using bovine enamel and the “peel/shear” bond strength test. The bond force for the etched surface was 84 lb., while for the cast mesh surface it was 143 lb. The advantages advocated for this technique were the elimination of the necessity of surface etching and the improvement of the retentive capabilities. Besides, it could be used with any dental alloy and it was easily verified by the naked eye. The difficulty was to adapt the mesh to posterior retainers (LaBarre & Ward, 1984). The major problem with this technique was that it resulted in bulky wings, which were not periodontally acceptable.

In another attempt to overcome the restrictions of the electrolytic etching system, Moon and Knap (1983) developed a new procedure that consisted of incorporating roughness into the cast fitted surface using salt particles. Cubic crystals of sodium chloride were embedded into the fitted surface of the wax pattern and dissolved in water after casting, creating microcubic pores. Particle sizes ranging between 149 μm to 250 μm achieved the strongest bond when compared to other
sizes. The bond strength achieved by this lost salt procedure was less than that achieved with etched-metal procedure (Barnes, Moon, Eshelman & Button, 1986). However, bond failure tended to occur at the enamel-resin interface with the lost salt specimens (Hudgins, Moon & Knap, 1985). For this reason, Moon (1987) considered the bond strength to the metal framework treated with the lost salt procedure “more than adequate”.

Based on the work of Posti, Nakki and Siirila (1980), who described a system in which resin retention beads were placed on the internal surface of cast inlays, LaBarre and Ward (1984) suggested the application of such retention concept in RBFPDs. The advantages of the resin bead design included the possibility of visual evaluation of the number and distribution of the beads by the naked eye, and the elimination of contamination as a critical factor. However, they found two problems during fabrication of this kind of prosthesis: overcontouring, caused by the diameter of the retention beads and the thickness of the metal retainer; and failure of the beads to adapt to small grooves and rests of the preparation, resulting in substandard fit to the prepared teeth.

Stokes and Tidmarsh (1986) recommended a microretentive system that was able to create retention in the metal surface without the problem of overcontour. It consisted of fusing thin, porous layers to precious and nonprecious alloys. They compared the bond strength of enamel bonded to a nonprecious alloy treated with porous metal coating system, to a nonprecious alloy treated with electrolytic etching system, and to welded-mesh brackets. They also included in the study a precious alloy treated with porous metal coating system and bonded to enamel. The results showed no difference in bond strength among the enamel/precious alloy treated with porous coating system, enamel/nonprecious alloy treated with electrolytic etching system, and enamel/mesh pattern. However, the specimens made using nonprecious alloy treated with porous coating system bonded to enamel had significantly lower bond strength. Although the system was not efficient in promoting the adhesion of nonprecious alloy to enamel, the advantage was that it added only 20 to 30 μm to the wing thickness, decreasing the problem of overcontouring.
Following some indication in the literature that simple sandblasting of the metal surface could improve the retention between metal and resin (Renk & Hartmann, as cited in Wiltshire, 1986), some researchers started to test the effect of this technique and particle size on bond strength. Wiltshire (1986) reported a significant increase in tensile bond strength when the alloy was air abraded with 250 μm aluminum oxide particles compared to a control where the metal was submitted only to a simulation of porcelain firing conditions (4 cycles at 920 °C in a porcelain oven). The results achieved with sandblasted specimens were similar to the ones submitted to electrolytic etching. However, Wiltshire did not find any improvement in bond strength when 50 μm and 110 μm aluminum oxide particles were used for sandblasting. It is interesting to note that in the control group and in the groups treated with 50 μm and 110 μm particles, the fracture occurred at the alloy/resin interface, while for the group treated with 250 μm particle and those electrolytically-etched, failure occurred cohesively in the resin composite cement.

Improvement in the retention between the sandblasted metal surface and resin, particularly with Co-Cr and Ni-Cr alloys was also reported by Bronsdijk, van der Veen and van de Poel (1986), and Creugers, Welle and Vrijhoef (1988). Contrary to Wiltshire’s findings, better results were achieved through sandblasting the metal with 50 μm Al₂O₃ than with 280 μm Al₂O₃ (Bronsdijk et al.). One of the advantages of sandblasting is that it is less technique-sensitive than electrolytic etching. Besides, it is less costly and eliminates many of the problems associated with alloy etching. It can be evaluated by naked eye and it can be done at the chair side using special apparatus built for this purpose. In a study by Hussey et al. (1991), the authors reported no difference in survival rates between RBFPDs that were electrolytically-etched and RBFPDs that were sandblasted.

Silicacoating, another kind of metal treatment, was developed in Germany. It was first reported in 1984 by Tiller et al. (as cited in Bronsdijk et al., 1986) to improve the retention of resin veneers on castings. The procedure consisted of sandblasting the retainers with aluminum oxide powder and then, using a silicacoating apparatus, applying a thin layer of silica to the alloy surface. The intermediate ceramic layer bonds to the alloy at one interface while providing chemically-reactive groups for
silane coupling to resin composite at the other interface. This system could be used with either a noble or a base alloy.

*In vitro* studies have demonstrated that the bond strength between resin composite and non-noble alloy treated with silicacoating is similar to the bond strength between resin composite and sandblasted non-noble alloys (Creugers et al., 1988; Imbery, Burgess & Patrick Naylor, 1992; Kolodney, Puckett, Breazeale, Patterson & Lentz, 1992). However, the bond strength can be improved by applying one layer of unfilled resin over the silicacoated metal surface prior to application of resin composite (Kolodney et al.). In addition, Creugers et al. and Caeg, Leinfelder, Lacefield and Bell (1990) found that the system resin composite/silicacoated non-noble alloy have higher bond strength that the system resin composite/electrolytically etched non-noble alloy.

Disadvantages of silicacoating include the expenses to purchase a silicoating machine and decreased bond strength if cementation of the prosthesis is delayed (Imbery & Eshelman, 1996).

For gold alloys a special procedure called tin-plating was described in 1984 by van der Veen and Bronsdijk (as cited in Bronsdijk et al., 1986). It consisted of deposition of tin oxide onto the metal surface forming an arrangement of surface crystals. The resin could penetrate between these crystals and thus result in micromechanical retention (Bronsdijk et al.). The electroplating procedure had to be performed after the RBFPD was tried in the mouth, because it had the same disadvantage of the electrolytic etching procedure in that the surface could be easily contaminated by moisture or from operator's fingers resulting in reduced bond strength. Nevertheless, tin-plating had advantages over etching: the procedure was very simple and fast, and could be used with any type of alloy.

Tin-plated noble and high noble metal alloys bonded to resin exhibited mean bond strength values similar to those achieved with sandblasted-base metal alloys (Gates, Diaz-Arnold, Aquilino & Ryther, 1993; Breeding & Dixon, 1996) and higher than those achieved with silicacoated non-noble alloy (Imbery et al., 1992). However, when only base metals were compared, silicacoating and sandblasting provided better results than tin-electroplating technique (Creugers et al., 1988).
Clinical evaluation of 85 RBFPDs cast in either silver-free palladium alloy or gold followed by tin electroplating showed a one-year success rate of 98% (van der Veen, Krajenbrink, Bronsdijk & van de Poel, 1986).

In conclusion, the techniques that seem to provide the best retention between nonprecious alloys and enamel are electrolytic etching, sandblasting and silicacoating. Electrolytic etching had its use discontinued by a number of dentists because of its inherent problems. It has to be done by a technician who must select the appropriate acid concentration, electric current, and duration for the procedure. Different alloys require different concentrations of acid, and even within alloys of similar composition, different brands require different approaches (Simonsen, Thompson & Barrack, 1983). Besides, the surface can be easily contaminated and can not be evaluated by the dentist with the naked eye.

The *in vitro* results achieved using nonprecious alloy and silicacoating also have been very promising. Nevertheless, the surface treated with the silicacoating procedure is also very susceptible to contamination. The major problem, however, is the need for an apparatus that is fairly expensive and is not available in the majority of the laboratories.

Sandblasting seems to be the more reasonable technique to prepare the fitted surface of the RBFPD for cementation. It provides good *in vitro* bond strength to composite, as well as good clinical results. It can be easily evaluated by the naked eye and the apparatus necessary to do sandblasting is fairly inexpensive. If some adjustment has to be made to the inner surface of the metal, the dentist can redo the sandblasting chairside. The only problem is the dust associated with the procedure, which can be easily controlled by using specially-designed cab units. For these reasons, sandblasting has been the most used technique for metal treatment of RBFPDs.

3. Development of Special Adhesive Resins

The early materials used to cement RBFPDs consisted of unfilled resins made with polymethyl methacrylate. The fixation of resin to dental alloys relied on the creation of mechanical retention devices on the metal surface without chemical
reaction between the resin and the metal. Low bond strength of the resin often resulted in its fatigue and subsequent fracturing under repeated occlusal stresses.

The first resin cement introduced specifically for use with etched retainer was a filled bisphenol glycidyltrimethacrylate (bis-GMA) cement with an unfilled bonding agent (Comspan - L.D.Caulk). Imbery and Eshelman (1996) stated “The bond between this cement and electrolytically-etched base metal alloy is the standard to which the newer generation of adhesive resins has been compared”.

In 1981, Tanaka et al. described a new adhesive monomer called 4-META (4-methacryloxyethyl trimellitate anhydride) which when combined with tri-butyl borane oxide (TBB-O), produced a resin cement (Super-Bond C&B) capable of chemical-adhering to base metal alloys. Bond strength tests of a 4-META opaque resin to a nickel-chromium alloy that received different surface treatments showed that the 4-META component significantly improved the adhesion of the resin to the alloy and that the durability of the adhesion was promoted by both oxidation and roughening of the metal surface. The oxidation, in Tanaka’s et al. study, was induced with 61% HNO₃ for 15 minutes at room temperature. Their results suggested that the 4-META opaque resin had the capability of adhering to the oxide film more strongly than to the metal itself. In addition, the resin tested was considered easier to manipulate and was also highly effective in concealing the color of the underlying metal. Furthermore, it did not contain glass fillers which caused the resin to be smoother than conventional opaque resins, resulting in less gingival irritation. Their conclusion was that the 4-META opaque resin could eliminate the necessity for retention devices on metal castings. However, the problem was that the bond would gradually weaken when specimens were stored in water (Wada, 1986). Nevertheless, a good clinical result was reported by Samama (1996), who followed 243 bonded prosthetic devices cemented with Superbond (Sun-Medical) for 10 years and reported only 26 failures (11%). The mean age of the restorations at evaluation was 5.7 years. He concluded that Superbond is capable of assuring the durability of bonded prostheses. The bonding of 4-META to gold alloys, however, is not as strong, as reported by Hansson (1994) who observed a median survival time of only 9 months for anterior RBFPDs treated with silicacoating and cemented with Super-Bond C&B.
Omura, Yamauchi, Harada and Wada (1984), developed a new dental adhesive in the form of a filled Bis-GMA resin cement with a phosphate ester monomer known as 10-methacryloyloxydecyl dihydrogen phosphate, or MDP. It promotes chemical adhesion to alloys as a result of MDP's bonding to oxides of nickel, chromium, cobalt or tin (Imbery & Eshelman, 1996). Omura et al. tested the adhesion of this resin cement (commercially called Panavia Ex, Kuraray Co, LTD., Osaka, Japan) to various dental alloys and demonstrated that it bonded very well to a sandblasted Ni-Cr alloy and a Co-Cr alloy. However, its bond to precious alloys was a little lower, and it gradually weakened when it was stored in water. The authors also tested the durability of the bond by varying the length of storage time in water at 37 °C up to 12 months. There was no deterioration in bond strength between the Panavia Ex and bovine enamel or between Panavia Ex and sandblasted Ni-Cr alloy. Their conclusions were: (1) Phosphonate resin had excellent adhesive property to dentin, enamel, various dental alloys and porcelains; (2) its bond exhibited good resistance to water; and (3) its mechanical properties were suitable for wide application to cementing uses.

Hussey et al. (1991) found in a clinical trial study that only 16% of prostheses luted with Panavia Ex debonded compared with 45% of those luted with other cements (Comspan - LD.Caulk, Delaware, MI, USA.; ABC Cement - Vivadent, Schann, Liechtenstein; and others).

In vitro studies have demonstrated that the bond strength between Panavia Ex and non-noble alloy is higher than the bond strength between Super-Bond C&B (4-META resin) and non-noble alloy (Diaz-Arnold, Williams & Aquilino, 1989; Osama Atta, Smith & Brown, 1990; Imbery et al., 1992; Reilly, Davis, Joynt & Quevedo, 1992). However, one study (Diaz-Arnold et al.) reported it to be the same as the one achieved with the resin Comspan.

The nature of the chemical interaction between these special resins and metal is not defined yet. It has been claimed that the new generation of resin composite cements are capable of chemical bonding to an oxide layer present on the nonprecious alloy surface, and both the composition and integrity of the oxide layer are considered critical for proper bonding (Simonsen et al. 1983). The fact that
precious alloys do not have an oxide layer would explain inferior bond strength between such alloys and resin cements. However little work has been done to demonstrate the existence of this oxide layer and the thickness necessary to ensure a good bond strength. Diaz-Arnold, Keller, Wightman and Williams (1996) investigated the changes in predominant surface oxides and oxide layer depth in a NiCrBe alloy when exposed to different laboratory procedures prior to cementation using X-ray photoelectron spectroscopy and Auger electron spectroscopy. In addition, tensile tests were used to verify the bond strength achieved with the different treatments. Oxide layer formation was induced by air firing the metal in a calibrated porcelain furnace or under 71 cm Hg vacuum. The oxidation was followed by air abrasion with 280 μm Al₂O₃, 50 μm Al₂O₃, or no air abrasion. The groups were compared to a control in which no treatment was done and to air-abraded groups with the mentioned particle sizes. Tensile tests showed no significant differences among the groups with the exception of the control, which had the lowest values. However, the group treated only with air abrasion with 50 μm Al₂O₃, and the one treated with air firing under vacuum alone had results similar to the control. In the control group, the oxide layer was found to be of approximately 50 nm and for the groups treated with air firing sequence and vacuum firing sequence, the layer was 500 nm and 310 nm thick, respectively. For the other groups, the oxide layer could not be measured due to the presence of residual Al₂O₃ particles used to abrade the surface. However, because Ni and Be were present in both oxide and elemental states, the oxide layer was assumed to be less than 10 nm thick. It is interesting to note that the thickness of the oxide layer did not have a major effect on bond strength. Further investigations are necessary to verify this aspect.

Recently, Jenkins, Golledge and Thompson (1998) examined the surface of a Ni-Cr alloy (Rexillium III) before and following application of MDP. Prior to coating, half of the specimens were sandblasted with 50 μm aluminum oxide powder. After the application of the MDP, specimens were rinsed in organic and acid solutions with different pH. The results showed that application of the MDP to the as-received alloy resulted in poor wetting of the surface and monomer removal with rinse solutions. For the abraded samples, however, MDP was detectable following all rinses, although it was removed to a greater degree with the two highest pH's solutions (11 and 13).
They were able to demonstrate that the MDP monomer chemically bonds to the alumina blasted alloy surface and that the residual alumina on the surface could be a prerequisite for chemical bonding. The nature of the chemical interaction, however, was not explained.

4. Developments in Design

The first resin-bonded fixed partial denture described in the literature was for the anterior segment. According to Howe and Denehy (1977), no tooth reduction was necessary and the lingual coverage should be maximised in order to improve retention. The metal framework was constructed with nonprecious alloy, with feathered margins, and as many retentive holes as possible. Hosseini (1994) found a survival rate of 82.2% for this design in a 10-year period of evaluation, while Berekally and Smales (1993) reported a debonding rate of 70% for Rochette RBFPDs over a period of 7 years with a median cumulative survival period of only 2.14 years.

4.1. The need for tooth reduction.

The need for tooth reduction for posterior abutments of RBFPDs was described by Livaditis (1980) and was directly related to the development of a path of insertion. The modification was originally limited to the proximal surfaces and was eventually extended to the lingual surfaces of the mandibular posterior teeth because of the lingual inclination. Livaditis also described the inclusion of occlusal rests similar to that for a removable partial denture to prevent displacement of the restoration in a gingival direction during trial insertion and under occlusal loading.

At the present time, the lingual reduction is done in all abutment teeth in order to ensure a single path of insertion and to create parallel guiding planes. A convergence angle of up to 10 degrees is clinically acceptable since it does not decrease the retention of the metal framework significantly (Sarafianou & Kafandaris, 1997). However, excessive tapering can substantially reduce retention.

Tooth preparation also reduces the problem of overcontouring that predisposes to periodontal problems (Hansson & Bergstrom, 1996). Creugers, Snoek, Vogels and Van't Hof (1986), however, found that cervical overcontouring is only a minor factor in
plaque accumulation and gingival reaction to resin-bonded retainers. The use of nonprecious alloys, which are stronger, also contributes to a thinner framework, reducing overcontouring.

Tooth reduction helps to improve retention of retainers of RBFPDs, and is one of the crucial factors that affects strength and durability of bonding (Berekally & Smales, 1993; Besimo, 1993; Thayer et al., 1993; Shakal, Pfeiffer & Hilgers, 1997). It was reported that surface preparation increases the bond strength of composite resin to the enamel by removing the aprismatic layer which can be resistant to acid etching (Schneider, Messer & Douglas, 1981). However, Samama (1996) and Pröbster and Henrich (1997) did not find any difference in survival rates for RBFPDs cemented to prepared or to unprepared abutments during 10-year and 11-year follow-up studies, respectively.

Livaditis described the basic principles in retainer design for both posterior and anterior RBFPDs in 1983. According to him, each retainer consists of three components: (1) the lingual segment, (2) the proximal segment, and (3) the occlusal rest. The lingual segment has the primary function of increasing the overall bond strength of the prosthesis by increasing the amount of surface area for bonding, and it also contributes to the dissipation of laterally directed forces.

The proximal segment provides the connector area for the pontic, and also contributes additional surface area for enamel bonding. Its significance, however, lies in the buccolinguval bracing that it provides against lateral forces. The labial wrap technique requires the framework to be extended labially to just beyond the proximal/labial line angle. The proximal and lingual segments together, described as a ‘wrap-around’ design, increase the stability of the prosthesis in a faciolingual direction. The use of circumferential reduction greater than 180 degrees seems to be important for good long-term stability (Crispin, 1991).

The occlusal rests provide, as their primary function, a positive stop for the restoration during trial insertion and cementation. It will also resist the occlusogingivally directed forces throughout the life of the restoration. There is no evidence to support need for use of two occlusal rests in each abutment since in an in vitro study (Pegoraro & Barrack, 1987), no difference were reported in retention of
resin-bonded frameworks with one or two occlusal rests. However, the use of both mesial and distal rests on each abutment is recommended because it prevents flexing and debonding of the lingual arm (Barrack, 1993).

With the development of the electrolytic etching by Thompson et al. 1981, the incorporation of perforations in the metal framework became unnecessary. It was shown that the etched metal would provide better retention than the holes suggested by Rochette (Creugers et al. 1992; Berekally & Smales, 1993) and that the retainers were less susceptible to fracture (Berekally & Smales, 1993). Nevertheless, the wrap-around design with one occlusal rest for posterior abutments was not enough to resist occlusal forces in the long-term (Hansson & Bergstrom, 1996).

4.2. The use of grooves.

In 1981, McLaughlin, who described a technique for electrolytic etching, suggested that vertical grooves placed on the mesial and distal surfaces of the abutment teeth would create a single path of insertion and would resist lateral displacement of the framework improving retention.

Yamashita and Yamami (1986) found that anterior and posterior prostheses cemented with Super-Bond C&B tended to dislodge after a short period of time. The failures happened even when the preparation was extended to include a large portion of the lingual area, despite the fact that a number of studies demonstrated that an increased surface area would provide better retention (Murakami & Barrack, 1986; Creugers et al. 1992; Thayer et al., 1993). However, the inclusion of two vertical grooves in each abutment tooth: one in the proximal surface adjacent to the edentulous area and the other in the opposite proximal surface, provided better retention, and a higher level of success was achieved (Yamashita & Yamami, 1986).

In addition to the increase in retention, the use of vertical grooves placed lingual to the proximal line angle eliminates the need for extension onto the labial surface (Meiers & Meetz, 1985), providing a better aesthetic result, particularly in the anterior region of the mouth.

A clinical study by Simon, Gartrell and Grogono (1992) in which 77 anterior and posterior RBFPDs were evaluated for four years showed that the presence of grooves
markedly elevated the longevity of posterior prostheses. Only 5% of the RBFPDs made with the conventional design (Livaditis, 1980) and grooves failed while 41% of the RBFPDs made without grooves were dislodged. No difference was found between the two groups tested in the anterior segment of the mouth.

In 1993, Rammelsberg, Pospiech and Gernet found, in a clinical trial study, that not only grooves, but also the incorporation of at least 1 mm deep channel on each abutment tooth would reduce substantially the risk of failure. The survival rate was increased from 37% for the design with parallel guide planes and shallow occlusal rests to 96% for the design with parallel grooves/channels during the 6-year evaluation period.

The use of grooves was also considered to improve the retention of RBFPDs in a recent clinical trial study by De Kanter, Creugers, Verzijden and Van’t Hof (in press). The 5-year study demonstrated 46% survival rate for conventional (wrap-around design and occlusal rest) and 62% for modified preparation with the inclusion of grooves.

4.3. Alternative designs.

In some cases, the abutment teeth are moderately restored requiring special attention when they are to be included in a RBFPD design. If a posterior abutment tooth has an occlusal amalgam restoration it is suggested that the restoration should be avoided or covered to prevent damage to the prosthesis if the existing amalgam fails and must be removed. In this case an inlay occlusal rest could be used (Meiers & Meetz, 1985). In this procedure, the occlusal surface of the amalgam is reduced by approximately 0.5 to 1.0 mm so that it acts as a base for the “cavity” preparation. The resin-bonded FPD, in this case, has the proximal segment, the lingual segment, proximal grooves and an inlay occlusal rest that also provides added lingual resistance for the framework.

Another option is to remove the restoration and replace it with inlays (two- or three-surface restorations) as retainers. This technique is more conservative because it eliminates the need for the lingual segment (wing) as the inlay retainers provide enough retention and resistance (Isidor & Stokholm, 1992). A clinical study by Isidor
and Stokholm (1992) in which 23 RBFPDs were made using inlay retainers and no lingual segment showed that none of the prostheses lost retention during 4 years of observation. However, an in vitro study (Pegoraro & Barrack, 1987) comparing the inlay design with proximal and lingual segments to the conventional design with one or two occlusal rests and to a full metal crown reported that the lowest retention values were obtained with the inlay/wraparound design. Interestingly, this group presented highest number of tooth fracture before separation of the retainer. Serdar Çörtert and Öztürk (1997) compared the conventional design for posterior abutments to the inlay design without proximal and lingual segments in a clinical study of 60 RBFPDs for the period of 6 years (median time 37.7 months). At three years, no difference between them was found (survival rate was 69.2% for inlay design and 73.5% for conventional design).

There are some situations when one of the abutments is intact, but the other has unacceptable surfaces for bonding, or have large, multisurface restorations and require full coverage. Bushfield, Shiu and Blakeslee (1984) described the use of attachments similar to that found in sectional fixed bridgework to overcome this problem. A full crown is made with a recessed lingual shelf and a well-defined occlusal rest section. The female attachment is positioned in the crown to follow the path of insertion of the metal retainer and a resin-bonded partial denture framework incorporates the male attachment. In the event of failure of the acid etched metal retainer, it would be possible to remake the RBFPD part only, without involving the full-coverage crown. No long-term prognosis for this design was found in the literature. However, Hussey et al. (1991) observed that this “hybrid bridge” performed well in a short-term clinical trial study (mean time of 2.7 years).

To ensure a single path of insertion, undercuts are carefully eliminated. However, Shi, Chen and Yuan (1992) described a new preparation design that uses the natural undercuts of abutments by changing the path of insertion of the RBFPD. The prosthesis consists of two parts, with two different paths of insertion: one lingual and one facial, which are inserted simultaneously. Both parts have two wings: in one part the wings cover the facial surface and in the other, the lingual surface. When the wings are bonded to the etched enamel, the two parts of the pontic are bonded to
each other. The authors recommend this technique when the undercuts of abutments are excessive and the patient cannot endure tooth preparations. The chief disadvantage of this design is the complex formulation of the wax up and a demanding precision casting technique.

Debonding can present problems if it affects only one abutment in a fixed-fixed design. Caries has been reported under an abutment following the debonding of a resin-bonded wing (Berekally & Smale, 1993). Unilaterally debonded cases have been converted into cantilevered designs by cutting off the debonded unit (Shaw & Tay, 1982; Olin, Hill & Donahue, 1991). The use of non-perforated metal framework makes the repair, removal and re-cementation of partially de-bonded RBFPDs more difficult and less successful (Dunne & Millar, 1993). For these reasons, some authors are suggesting the use of a single abutment, single pontic, cantilever RBFPD. Cantilevering a pontic from a single abutment may reduce shear peel forces which result from the differential movements of the abutments in fixed-fixed designs which, in certain situations, would make debonding less likely (Hussey & Linden, 1996).

Hussey et al. (1991) suggested the efficacy of cantilevered RBFPD in an interim report of a study involving 400 adhesive RBFPDs with a mean duration of clinical service of 2.7 years. Seventy cantilevered prostheses were included in the study and at the evaluation 12 were reported debonded (17%). Briggs, Dunne and Bishop (1996), who evaluated 54 posterior cantilever RBFPDs for a mean period of 27 months, reported that 11 (20%) prostheses failed, with an average service time of only 8 months. These prostheses were re-cemented, and only 3 (6%) were then deemed failed at the end of the evaluation period. Hussey and Linden (1996), who studied 142 cantilevered RBFPDs, reported that 88% remained bonded in a mean period of 36.8 months. The debonded RBFPDs were recemented and, by the end of the study, 6% (9 prostheses) were judged to have failed.

An in vitro research by Shakal et al. (1997) showed that both the surface area available for bonding and the design of the tooth preparation have substantial effects on strength and durability of adhesive bonding joints. The original design for posterior abutments (Livaditis, 1980), that consisted of one occlusal rest, proximal and lingual segments recorded the lowest bond strength values. The inclusion of proximal
grooves as well as lingual cusp coverage improved the resistance to debonding. Their work demonstrated that the mechanical retention created with a larger surface area and the incorporation of grooves is necessary to resist complex stresses. The potential rotation of the retainer around the occlusal rest axis, flexure, and bending of the arms can cause fatigue failure of the resinous cements at the bonding interface.

Recently, a new design for posterior RBFPD was introduced by El-Mowafy (1996). The new prosthesis is indicated for the replacement of missing single posterior tooth with intact or minimally restored abutment teeth. It consists of lingual segment, proximal segment, occlusal rests and one proximal slot preparation made in the proximal area adjacent to the edentulous space. The metal framework is made with two small projections, one in each side of the pontic, designed to fit in the centre of the slot preparation and to rest on the gingival seat. The prosthesis is cemented and the slots are restored with composite resin creating mechanical interlocking of the cast projections into the slot cavities. Two factors may lead to an improved retention with this design: the increased surface area for bonding, and the interlocking of the projections within the composite resin. However, no clinical or in vitro evidence is yet available to support it. Further studies are necessary to verify the efficacy of this design in the clinical situation.

In summary, the evolution of the design for RBFPDs has been following a very clear direction that is to increase the surface area for bonding as well as to create new means of retention in the preparation design by inclusion of grooves, channels and slot preparations. It has been fully demonstrated that the ideal situation where no tooth preparation is done simply do not provide a definitive restoration and failures do occur. However, the idea of replacing a missing tooth without involving the facial surfaces of the abutments and without extension to the gingival crevices is still very attractive. If such design can provide clinical results comparable to the ones achieved with conventional fixed partial dentures, dentists will be more confident in prescribing this conservative option to replace a missing tooth for those patients who are not suitable for implant prosthesis.
5. Factors Affecting Longevity of RBFPDs

It is now established that there are several factors that can affect the longevity of conventional fixed partial dentures. However, although there has been many clinical studies published, only a few investigated the correlation between factors that affect the success rate of RBFPDs. This is due mainly to the fact that the majority of the studies are retrospective with little or no control over factors such as patient selection criteria, materials used, conditions under which the RBFPDs were placed and the designs used. Despite these difficulties, it is possible to draw some conclusions from these studies.

Patient selection is a very important issue and is highly associated with the prognosis of RBFPDs (Olin et al. 1991). Although it is known that masticatory forces are stronger in men than in women (Helkimo, Carlsson & Helkimo, 1977; Kiliaridis, Johansson, Haraldson, Omar & Carlsson, 1995), gender does not seem to affect the longevity of RBFPDs. The numbers of failures reported have been similar for both males and females (Hussey et al., 1991; Dunne & Millar, 1993; Hosseini, 1994; Hussey & Linden, 1996). Contrary to this, two studies attributed a higher debonding rate of RBFPDs to the stronger force of mastication associated with males (Olin et al., 1991; Thayer et al., 1993).

Controversy is also present when the issue is age. Although some studies associated a higher debonding rate with younger patients (less than 30 years old) (Dunne & Millar, 1993; Hussey & Linden, 1996), no relationship between age and longevity of RBFPDs was reported in other studies (Hussey et al., 1991; Thayer et al., 1993).

Parafunional habits and occlusal interference have been associated with higher debonding rates. Bruxism is considered a stress factor that causes premature failure of RBFPDs (Berekally & Smales, 1993; Hansson & Bergstrom, 1996). Berekally and Smales (1993) also observed that debonding occurred in 40% of prostheses that had premature occlusal contacts in centric occlusion and in 61% of prostheses that had occlusal interference on the pontic or metal frame in lateral or protrusive mandibular excursions, indicating that occlusal interference was a significant cause of failures. Pröbster and Henrich (1997) who evaluated 325 RBFPDs
for 11 years, observed that the clinical outcome for cases initially estimated as being of higher risk because of abutment mobility, bruxism or clenching habits or cleft palate, were worse than for those assessed as having a "normal" risk.

RBFPDs should be limited to single tooth replacement since the failure rate of prostheses with multiple tooth pontics is higher (Dunne & Millar, 1993; Pröbster & Henrich, 1997). Failures are also associated with the use of three or more abutments (Olin et al., 1991; Dunne & Millar, 1993 and Hosseini, 1994). Although surface area for bonding is increased, differential movements in the abutments result in shear peel forces and in greater stress in the bonding area.

Location of the RBFPD is thought to be a factor in determining longevity, although some researches did not find differences in survival among prostheses placed in the anterior or posterior quadrants, or among prostheses placed in the maxilla or in the mandible (Creugers & Van’t Hof, 1991; Berekally & Smears, 1993; Dunne & Millar, 1993; Rammelsberg et al., 1993; Thayer et al., 1993; Pröbster & Henrich, 1997). Olin et al. (1991) observed that mandibular RBFPDs failed nearly twice as often as maxillary RBFPDs. Mandibular posterior RBFPDs are thought to be associated with the poorest prognosis (Olin et al., 1991; Boening, 1996; DeKanter et al., in press). Several reasons have been suggested for this observation:

1. The clinical crowns of maxillary premolars and molars are usually longer than those of mandibular teeth resulting in larger bonding areas,

2. Direction of occlusal forces is different. In the maxilla, the point of impact of the occlusal load is located lingually, pushing the framework favourably against the abutment tooth. In the mandible, this point is located on the buccal cusps resulting in forces that push the framework away from the abutment teeth (Brabant, 1997),

3. The isolation methods might be more effective in the maxilla because it is more easily controlled, and

4. The deformation of the mandible during mastication results in stresses at the site of the RBFPD (Olin et al., 1991; Boening, 1996).

Anterior prostheses are, also, more retentive than posterior prostheses (Creugers et al., 1992 and De Rijk, Wood & Thompson, 1996) and a predicted lifetime
of 26.2 years for the anterior RBFPDs was calculated by De Rijk et al. as opposed to 17 years for posterior. The better prognosis could be related to better adaptation found for anterior abutments in a 10-year clinical study by Wood, Thompson, Romberg and Morrison (1996).

Some studies attempted to correlate higher success rates with operators’ skill. However, Hussey and Linden (1996), and Hussey et al. (1991) did not find any correlation. On the other hand, Crispin (1991) concluded that the high level of clinical expertise needed to ensure a successful conventional fixed prosthesis is also necessary to optimise the prognosis for a bonded prosthesis, which is in agreement with Berekally and Smales (1993) findings.

Another factor that affects the longevity of RBFPDs is the alloy used to cast the metal framework. Because of the high modulus of elasticity, base metal alloys such as Nickel-Chromium or Cobalt-Chromium are preferred to gold alloys (Besimo, 1993). In order to prevent deformation, a framework made out of a gold alloy must be thicker and, therefore, the preparation must be more aggressive to avoid overcontouring. Besides the lower modulus of elasticity, the bond strength between high-gold alloys and resin cement was found to be insufficient to resist fatigue fractures on the long-term (Hansson & Bergstrom, 1996).

A recent in vitro study reported that the bond strength in enamel/resin/metal specimens is significantly higher with sandblasted non-noble alloys when compared to sandblasted noble alloys (Rubo & Pegoraro, 1995). However, the bond strength of noble alloys can be improved by using tin electroplating (Imbery et al., 1992; Gates et al., 1993; Breeding & Dixon, 1996), although this improvement was not substantiated in other studies (Eder & Wickens, 1996; Rubo, Pegoraro & Ferreira, 1996).

The use of rubber dam when cementing RBFPDs is also important to prevent contamination of the site, which could lead to a weaker bond between the cement, tooth structure and metal. Many clinical studies have demonstrated that a high number of failures occurred within the first year of the restoration (Crispin, 1991; Hussey et al., 1991; Berekally & Smales, 1993; Dunne & Millar, 1993 and Hussey & Linden, 1996). Dunne and Millar (1993) and Hussey and Linden (1996) observed that a high percentage of the failures occurred in the first six months (41% and 46%,
respectively). The higher debond rate soon after insertion is due, presumably, to faulty bonding procedure, and in particular to contamination (Dunne & Millar, 1993). However, Verzijden, Creugers and Mulder (1994) found that the success rate was the same in the mandible when rubber dam was used or not used, although it appeared to be effective in the maxilla.

After cementation, it is important to keep the patient in a re-call program. A high number of failures with the fixed-fixed design were noticed by the clinician and not by the patient during re-call examination (Olin et al., 1991; Verzijden et al., 1994). This could lead to the development of caries without the patient noticing it.

Up to this point it is possible to conclude that in order to achieve a good result with RBFPDs the following criteria have to be met. Abutment preparation as well as inclusion of retention grooves is important to improve bond strength to enamel and to create a well defined path of insertion facilitating cementation. It is important to remember that parafunctional habits and occlusal interference may decrease the longevity of the restoration, and that a better prognosis is associated with prosthesis located in the maxilla and in the anterior region of the mouth. RBFPDs should be used to replace single teeth and no more than two abutments should be involved in the preparation. A base metal alloy is preferred both because of its high modulus of elasticity resulting in a thinner framework, and because it is easily microetched through sandblasting. And, finally, it is important to have rubber dam placed during cementation to avoid contamination with subsequent early failures.

6. Longevity of RBFPDs

A number of clinical trial studies evaluated the longevity of RBFPDs. Some selected ones are discussed below and are all summarised in Table 1 at the end of this section.

Van der Veen et al. (1986) reported a one-year clinical evaluation of RBFPDs made with either palladium or gold alloys. Only one of the eighty five prostheses originally placed debonded corresponding to a one-year success rate of 98.8%. The metal surface was treated with tin electroplating and debonding occurred at the resin-metal interface on 80% of the surface.
In 1991, Creugers and Van't Hof reported the results of a meta-analysis of 11 different clinical studies in which a total of 1598 RBFPDs were analysed. The longest follow-up period was ten years, and in about 51% of the cases, the follow-up period was at least five years. The overall survival rates were: one year, 89%; two years, 84%; three years, 80% and four years, 74%.

Crispin (1991) reported results of a 5-year prospective study of 71 RBFPDs placed under controlled clinical conditions. The prostheses ranged in service from 13 to 65 months. Twenty six RBFPDs (36%) debonded and most failures occurred within the first year. Most of the failures occurred due to operator’s errors or design errors which led the authors to conclude that RBFPDs are technique-sensitive and a high level of clinical expertise is needed to ensure their success.

Hussey et al. (1991) reported the performance of 400 adhesive RBFPDs with variable designs including Rochette’s, with a mean duration of clinical service of 2.7 years (S.D. = 1.6). One hundred of the prostheses dislodged (25% failure rate). However, if the Rochette RBFPDs, of which 73% debonded, were not taken into consideration, the success rate increases to 78%.

In 1991, Olin et al. reported results of a retrospective study of 103 RBFPDs placed by practitioners at the University of Minnesota School of Dentistry. After an average service of 3.25 years, 12.6% of the RBFPDs had debonded. It is interesting to note that three of the patients who presented with loose prostheses were completely unaware of any problems with their prostheses. A healthy gingiva was noted in the majority of the patients and a 98.8% patient satisfaction was reported.

Creugers et al. (1992) reported a 75% survival rate for anterior RBFPDs and 44% survival rate for posterior RBFPDs after 7.5 years. One hundred and sixty-six anterior and 37 posterior prostheses were placed and no tooth preparations were carried out, except in premolars and molars, where guiding planes and occlusal rests were made. Ninety-two prostheses were of the perforated type and 111 prostheses were electrolytically etched. Perforated-type RBFPDs had a risk of failure twice as high as electrolytically etched ones.

Barrack (1993) presented an 11-year prospective clinical longitudinal study of RBFPDs with a mean life at evaluation of 5 years and 8 months (range of 1 to 11
years). Nine of the 127 prostheses dislodged representing a success rate of 92.9%. However, a higher success rate was observed when a retentive design, including additional grooves, two occlusal rests and the wings, was used (100%) as opposed to when a non-retentive design was used (80.0%).

A retrospective clinical evaluation by Berekally and Smales (1993) at the Adelaide Dental Hospital showed a success rate of only 21% for Rochette RBFPDs over 6 years (median survival of 2.14 years), and 49% for electrolytically-etched RBFPDs over 5 years (median survival of 2.6 years). RBFPD failures were due to debonding in 89% of the cases for Rochette design and in only 40% of the cases for the electrolytically etched design. A great number of failures was attributed to pontic fracture (56%) for the electrolytically-etched RBFPDs and to fractured retainers (7%) for the Rochette RBFPDs. If pontic fractures are not included in the failures, the success rate for Rochette design is 24% and for the electrolytically-etched design is 78%. Eighty percent of the failures occurred in the enamel/resin interface, 10% had a cohesive failure within the resin cement layer and 8% had a combination of resin failure patterns.

Dunne and Millar (1993) evaluated 382 RBFPDs and splints over a period ranging from 5 months to 8 years. During the evaluation period, 125 restorations (33%) debonded of which 69 (55%) were rebonded. Rochette prostheses had significantly higher debonding rates (80%) than that of the electroetching Crystal-bond design and the sandblasting design (26% and 29%, respectively). However, the authors reported a high rebond rate and the RBFPDs continued to perform well over the study period. In overall, the systems together had a 60% survival rate at 4 years.

Rammelsberg et al. (1993) reported the results of a 6-year prospective study of 141 RBFPDs inserted under controlled conditions. A survival rate of 82.9% over the study period was reported. Different methods of retainer conditioning were tested (sandblasting/acid etching/silane-coating) and no difference in retention rates was found among them. The prostheses were cast in a Co-Cr alloy and cemented with a dual-curing cement (Micropont/Kulzer Co., Wehrheim, Germany), and all failures displayed adhesive debonding between the metal framework and the resin cement.
Thayer et al. (1993) reported a 61% success rate of 85 RBFPDs after 5 to 15 years. The perforated and electrolytically etched RBFPDs had nearly equal rates of debonding. A statistically significant difference was reported for the mean coverage area of the retainers. A smaller coverage area was associated with more failures.

Hansson (1994) reported results of a prospective study of 13 prostheses made with gold alloy (Esteticor Swiss - Metaux Precieux SA. Metalor) and treated with sandblasting. After one year, seven of the 13 prostheses had debonded and the study was discontinued because of unsatisfactory clinical results. The other six prostheses debonded after 13, 21, 26, 43, 46 and 75 months. All fractures occurred at the resin-metal interface.

Priest (1995) reported on the condition of 77 RBFPDs placed over 10 years. Four percent of the evaluated prostheses had perforated retainers, 35% were electrolytically etched and 60% were chemically etched. The prostheses were evaluated for a mean period of 4.2 years (range 1 to 11 years). Thirty (39%) became dislodged or were removed, however, nine of them were successfully rebonded. The design was modified along the study to improve resistance form by including proximal grooves and additional rests, besides the lingual and proximal wraparounds and the occlusal or cingulum rest that was being used since the beginning of the study. The modification significantly improved the success rate of the RBFPDs.

Boening (1996) reported survival of 46 RBFPDs for periods from 1 to 45 months, with a mean evaluation time of 26 months. Tooth preparation included the lingual and proximal surfaces, 2 occlusal rests for posterior abutments and 1 horizontal groove for anterior abutments. No grooves were prepared for posterior abutments. The framework were cast in a Ni-Cr alloy and tribochemically-coated with silica (Rocatec System, ESPE, Seefeld, Germany). Ten of the prostheses had failed (21.7%) and most of the failures occurred at the metal/composite interface. Two failures were due to porcelain fracture. The survival rates of RBFPDs replacing incisors, canines, or premolars decreased from 90% after 1 year to 84% after 2 years and 81% after 3 years. Prostheses that replaced molars recorded a significantly lower level of survival (only 30% after 2 years). The probabilities of survival for RBFPDs in all locations were 82% and 80% after 2 and 3 years, respectively.
Hansson and Bergstrom (1996) reported an 82.7% success rate of 29 RBFPDs after 8.5 years (mean observation time 6.1 years) and a total debonding rate of 2.8% per year. The frameworks were made with high gold or cobalt-chromium alloys and silicacoating was used as an adhesion promoter between composite and metal. Tooth preparation did not include grooves for posterior abutments. The prostheses made with gold alloy detached earlier (3.1 to 5.0 years) than those constructed with the cobalt-chromium alloy (7.4 and 8.3 years). The authors concluded that preparations with one groove for anterior teeth and one occlusal rest for posterior teeth were not enough to resist occlusal forces on the long-term.

Kerschbaum et al. (1996) reported results of a study that involved 1637 three-unit RBFPDs which were placed at 25 different clinical centres. One thousand and sixty nine RBFPDs were inserted in the maxilla (65.3%) and 568 (34.7%) in the mandible. The majority of the frameworks was cast using nonprecious alloys and was conditioned by air abrasion and silicoating. A total of 333 of 1637 prostheses debonded. After 1 year, 92% (±1.4%), and after 5 years, 66.1% (±3.7%) of the prostheses were still in service. The overall success rates regardless of the number of recementations were 97.2% after 1 year and 82% after 5 years. The authors concluded that rebonded three-unit RBFPDs have a great risk of debonding and that RBFPDs are an efficient method of treatment for missing tooth.

De Rijk et al. (1996) made an estimate for the longevity of resin-bonded prostheses. Mathematical equations were applied to the results of a retrospective clinical investigation. Of the 164 prostheses evaluated for a mean time of 123 months (10.3 years), 47 (29%) failed with a median time in service of 74 months (6.2 years). The maximum likelihood estimate of the lifetime of the prostheses was 255 months (21.3 years), a prediction made with the oldest RBFPD existing for 171 months. Differences in the estimated lifetime were observed between prostheses placed anteriorly, 338 months (28 years), and posteriorly, 207 months (17 years). The authors considered resin-bonded prostheses as long-term treatment option when compared with conventional prostheses.

Samama (1996) reported a 10-year follow-up study of 145 prostheses. There were only 11 failures or 7.6%, and the mean age of the restorations at the time of the
last recall was 5.7 years. Statistical analyses estimated a survival rate of 83% for a period of 10 years. No difference in survival was found between prostheses that were cemented to prepared abutments and ones that were cemented to unprepared abutments. The author attributed this result to patient-selection process. Implant treatment possibilities were also evaluated by examining available preoperative radiographs. It was assumed that if the available space was enough to receive a 13 mm long Branemark’s implant, the patient could be considered as a potential candidate for implant treatment. The results showed that the implant treatment would have been possible in only 14 of 98 cases evaluated. The author concluded that RBFPDs are appropriate and reliable treatment options for patients in whom only one or a few teeth are missing.

Besimo, Gächter, Jahn and Hassell (1997) reported results of a medium-term clinical follow-up investigation of 127 RBFPDs made with nickel-chromium and cobalt-chromium alloys and electrolytic etching in a well-controlled situation. Six failures occurred (4.7%) in a mean observation time of 3.4 years. According to the authors, the high success rate was due to (1) patient follow-up (6 months recall system); (2) patient selection based on stringent requirements; (3) clinical and laboratory procedures performed carefully; and (4) proper preparation design (guiding planes, proximal grooves and occlusal rests). The clinical success rate per patient calculated with life-table analysis by Kaplan and Meier was 97% at 3 years and 94% at 5 years.

Pröster and Henrich (1997) conducted an 11-year follow-up study of 325 RBFPDs. Several factors were taken into consideration when the statistical analysis was performed, and base metal alloys, silicacoating, mesh retention, as well as immobility of abutments were considered as positive factors. Retentive abutment preparation represented by placement of grooves and pinholes, and occlusal rests of partial crown preparation did not result in a higher survival rate compared to unprepared abutments. Including the rebonded restorations, overall survival rate calculated was 76% after 5 years and 60% after 10 years. If rebonding was not taken into consideration, the survival rate was 64% after 5 years and 50% after 10 years.

A five-year multicenter clinical trial study was reported recently by DeKanter et al. (in press). Two hundred and one posterior resin-bonded FPDs made with Ni-Cr
alloy were evaluated for a period of at least 5 years. Two designs: the conventional suggested by Livaditis (1981) and the conventional with grooves, were used in the study. The authors defined survival at two levels: (1) complete-survival (survival without any debonding), and (2) functional-survival (survival including loss of retention one time and successful rebonding of the original prosthesis without further debonding). The overall complete-survival was 53% at the end of the follow-up period (5 years). When the prostheses were rebonded once, the functional-survival increased to 79%. The complete-survival at 5 years for the conventional design (46%) was significantly lower than the complete-survival for the design with grooves (62%). Therefore, the authors recommended the use of extensive preparations in order to improve the survival rates.

It is a fairly difficult task to compare all these clinical trial studies and to attempt to draw a definitive conclusion about longevity of RBFPDs. The number of potential factors that can play an important role in the survival of these prostheses such as design preparation, alloy used, metal treatment, cement, etc. is too large and varies too much from one study to another. However, there is one factor common to many studies that as experience was gained and the designs were modified by including grooves or extending the preparation area, the survival rate was improved as well (Barrack, 1993; Priest, 1995; Pröbster and Henrich, 1997). Pröbster and Henrich (1997) reported that although no difference was found among designs, three years of experience with the RBFPD technique led to a 25% better survival probability when the 5-year results were compared. Barrack (1993) reported a success rate of 92.2% after follow-up of 127 RBFPDs for 11 years (mean 5 yr. and 8 months). However, in the last 9 years of his study, a more retentive design including grooves, incorporation of two rests and the concept of maximum coverage of the enamel was used in 82 prostheses and a success rate of 100% was obtained.
Table 1. Success rates for RBFPDs according to different studies.

<table>
<thead>
<tr>
<th>Reference</th>
<th>Mechanism of retention</th>
<th>Metal</th>
<th>No. of RBFPDs</th>
<th>Observation period</th>
<th>Success rate*</th>
<th>Survival rate*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Livaditis, 1981</td>
<td>Electrolytic etching</td>
<td>Not stated</td>
<td>66</td>
<td>1-12 m</td>
<td>96.9%</td>
<td>Not stated</td>
</tr>
<tr>
<td>Van der Veen et al., 1986</td>
<td>Tin electroplating</td>
<td>Pd/ Au</td>
<td>85</td>
<td>Mean 1 yr</td>
<td>98.8%</td>
<td>Not stated</td>
</tr>
<tr>
<td>Crispin, 1991</td>
<td>Electrolytic etching</td>
<td>NiCr</td>
<td>71</td>
<td>1-5.5 m</td>
<td>64%</td>
<td>Not stated</td>
</tr>
<tr>
<td>Hussey et al., 1991</td>
<td>Perforation</td>
<td>NiCrMo</td>
<td>22</td>
<td>Mean 2.7 yr</td>
<td>27%</td>
<td>Not stated</td>
</tr>
<tr>
<td></td>
<td>Electrolytic etching</td>
<td>AuPd</td>
<td>378</td>
<td></td>
<td>78%</td>
<td></td>
</tr>
<tr>
<td>Olin et al., 1991</td>
<td>Not stated</td>
<td>Not stated</td>
<td>103</td>
<td>Mean 3.25 yr</td>
<td>87.4%</td>
<td>Not stated</td>
</tr>
<tr>
<td>Creugers et al., 1992</td>
<td>Perforation</td>
<td>Nonprecious</td>
<td>92</td>
<td>7.5 yr</td>
<td>Not stated</td>
<td>Ant 75%-7.5yr</td>
</tr>
<tr>
<td></td>
<td>Electrolytic etching</td>
<td></td>
<td>111</td>
<td></td>
<td></td>
<td>Post 44%-7.5yr</td>
</tr>
<tr>
<td>Barrack, 1993</td>
<td>Sandblasting</td>
<td>Not stated</td>
<td>127</td>
<td>1-11yr (mean 5.6 yr)</td>
<td>92.9%</td>
<td>Not stated</td>
</tr>
<tr>
<td>Study</td>
<td>Steel Type</td>
<td>Coating</td>
<td>Method</td>
<td>Life</td>
<td></td>
<td></td>
</tr>
<tr>
<td>------------------------------</td>
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<td>-----------------</td>
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<td></td>
<td></td>
</tr>
<tr>
<td>Berkaly and Smale, 1993</td>
<td>NiCrBe</td>
<td>Not stated</td>
<td>Electrolytic etching</td>
<td>6 yr (mean 2.1 yr)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Dunne and Millar, 1993</td>
<td>NiCrBe</td>
<td>Not stated</td>
<td>Electrolytic etching</td>
<td>5 yr (mean 2.6 yr)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Rammelsberg et al., 1993</td>
<td>CoCr</td>
<td>64%</td>
<td>Sandblasting</td>
<td>5 m-8 yr</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Thayer et al., 1993</td>
<td>NiCrBe</td>
<td>59.6%</td>
<td>Silica coating</td>
<td>6 yr</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hansson, 1994</td>
<td>NiCr</td>
<td>90.9%</td>
<td>Electrolytic etching</td>
<td>4.5-15 yr</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Priest, 1995</td>
<td>NiCrBe</td>
<td>Not stated</td>
<td>Electrolytic etching</td>
<td>1 yr</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Boening, 1996</td>
<td>NiCr</td>
<td>Not stated</td>
<td>Chemical etching</td>
<td>1-1.1 yr (mean 3.4 yr)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>hansson and Bergstrom, 1996</td>
<td>Au</td>
<td>59%</td>
<td>Silica coating</td>
<td>4.5 m (mean 2 yr)</td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>CoCr</td>
<td>50%</td>
<td>Sandblasting</td>
<td>82.7%</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Life is given in years.
<table>
<thead>
<tr>
<th>Study</th>
<th>Method/Coating</th>
<th>Retention</th>
<th>Mean Duration</th>
<th>Survival Rate</th>
<th>Failure Reason</th>
</tr>
</thead>
<tbody>
<tr>
<td>De Rijk et al., 1996</td>
<td>Not stated</td>
<td>Not stated</td>
<td>164</td>
<td>Mean 10.3 yr</td>
<td>71%</td>
</tr>
<tr>
<td>Kerschbaum et al., 1996</td>
<td>Sandblasting (27.3%)</td>
<td>nonprecious</td>
<td>1637</td>
<td>8 yr</td>
<td>79.6%</td>
</tr>
<tr>
<td></td>
<td>Silica coating (25%)</td>
<td>(74%)</td>
<td></td>
<td></td>
<td>92%-1yr</td>
</tr>
<tr>
<td></td>
<td>Etching (19.5%)</td>
<td>Pd (15.8%)</td>
<td></td>
<td></td>
<td>66%-5yr</td>
</tr>
<tr>
<td></td>
<td>Net (8.6%)</td>
<td>precious (8.2%)</td>
<td></td>
<td></td>
<td>53%-7yr</td>
</tr>
<tr>
<td></td>
<td>Others (19.6%)</td>
<td>others (2%)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Samama, 1996</td>
<td>Electrolytic etching NiCr</td>
<td>145</td>
<td>1-10 yr (mean 5.7 yr)</td>
<td>92.4%</td>
<td>83%-10yr</td>
</tr>
<tr>
<td>Besimo et al., 1997</td>
<td>Electrolytic etching NiCr</td>
<td>127</td>
<td>Mean 3.4 yr</td>
<td>95.3%</td>
<td>Not stated</td>
</tr>
<tr>
<td></td>
<td>Electrolytic etching CoCr</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pröbster and Henrich, 1997</td>
<td>Silicacoating (70.8%)</td>
<td>precious (17.2%)</td>
<td>325</td>
<td>10 yr</td>
<td>63%</td>
</tr>
<tr>
<td></td>
<td>Net retention (16.3%)</td>
<td>nonprecious</td>
<td></td>
<td></td>
<td>64%-5yr</td>
</tr>
<tr>
<td></td>
<td>Sandblasting (6.8%)</td>
<td>(26%)</td>
<td></td>
<td></td>
<td>50%-10yr</td>
</tr>
<tr>
<td></td>
<td>Electrolytic etching</td>
<td>Pd (56.7%)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>(6%)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>DeKanter et al., in press</td>
<td>Electrolytic etching NiCr</td>
<td>201</td>
<td>5 yr</td>
<td>Not stated</td>
<td>53%-5yr</td>
</tr>
<tr>
<td></td>
<td>Sandblasting</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td></td>
<td>Silicacoating</td>
<td></td>
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</tbody>
</table>

*Reasons for failure include debonding and fractured retainer
7. Conclusion

Resin bonded fixed partial dentures have gone through many modifications and developments since the concept was first proposed by Rochette (1973). The use of specially-made resin cements as well as certain metal conditioning techniques had a positive effect on the clinical longevity of this kind of prostheses. However, the modification in design played the most important role as a positive predictor of their longevity.

It is unfortunate that a prosthesis that gained initial popularity for being extremely conservative is nowadays becoming less so. It is now established that cementation of this type of prosthesis to unprepared enamel fails to provide long-term clinical longevity. The original design, which does not involve cutting of tooth structure, simply does not work as it fails over a short period of time. Therefore, it is now accepted that lingual reduction as well as placement of retention grooves and occlusal rests are of primary importance for improved retention of RBFPDs. Also, it has been demonstrated that RBFPDs placed under well-controlled environment and in carefully selected patients have better clinical success (Barrack, 1993). However, there is still some caution about using RBFPDs for the replacement of mandibular posterior teeth (Brabant, 1997), which indicates that the current design still fails to provide adequate long-term retention in certain situations. Hence, developing designs that will provide superior retention is important and necessary. Although the modifications in design typically led to more involving preparations, RBFPDs still have the advantage of being more conservative than conventional FPD. Besides, their supragingival margins contribute to the maintenance of periodontal health, and render the clinical technique more simple.

Modified designs such as the one suggested by El-Mowafy (1996) may lead to a broader indication of RBFPDs in the posterior area by improving the clinical longevity in situations where performance of RBFPDs is still unpredictable. The increase in surface area for bonding as well as the mechanical interlocking created by the metal projections into the resin composite restorations may result in an increased resistance to dislodgement. However, before making clinical assumptions about this design, it is imperative to have it tested in vitro under conditions that approximate
those of the oral environment. Positive results in such an in vitro study would substantiate the need for a well-controlled clinical trial study.

8. Aim

Recently a modified design for posterior RBFPD was reported (El-Mowafy, 1996) and the goal of this research project was to study in vitro the resistance of this modified resin-bonded prosthesis to detachment in comparison with that of the currently used design. The factor of mechanical load-cycling to simulate the clinical situation was a major component of this investigation. Effect of different resin cements was also tested. The hypothesis was that the new design would provide superior retention even under prolonged effects of mechanical load-cycling compared to the currently used design, independent of the resin cement used.
Materials and Methods

This study was preceded with a pilot study to identify problem areas of the proposed method of testing. The results of the pilot study were analysed and modifications were made when necessary. Details of the pilot and main studies are given below.

1. Pilot Study

Eight recently extracted sound teeth, four lower premolars and four lower molars, were selected. Doctors in private practices, who collected the teeth, were provided with jars containing distilled water, and were asked to keep the teeth in the fridge. All teeth were employed in the study within a period of three months following extraction.

After cleaning with periodontal curettes, the teeth were sterilised with gamma radiation. The glass jar containing the teeth in water was placed in a Cobalt 60 radiation chamber and exposed for 3 1/2 hours. The total radiation dose used was 2.5 Mrad.

1.1. Preparing the specimens.

Four bridge specimens, each made of one premolar and one molar tooth with a space between them equal to the size of a first molar, were prepared. Care was taken to select teeth of comparable size for each specimen, and therefore control the surface area involved in the preparation. Each tooth received three or four self threading pins (TMS Minim, Whaledent Int., New York, NY 10001, USA) that were inserted into the root surface to ensure good retention within the resin base which was later made (Figure 1). To facilitate positioning of the teeth, they were first mounted in a metallic split mold with six hexagonal holes (Figure 2).
Figure 1. Premolar and molar teeth with the pins inserted into the root surface.
One premolar and one molar tooth were mounted in two adjacent holes with tray acrylic resin (SR-Ivolen, Ivoclar AG, FL-9494- Schaan, Liechtenstein). Petroleum gelly was used to lubricate the metal surfaces of the mold for easy removal of the resin bases. A molar denture tooth placed between the teeth was used to maintain a space for a molar and to help in positioning of the teeth in a continuous occlusal plane. Boxing Wax (The Hygenic Co./ Akron, Ohio 44310, USA) was placed at the base of the denture tooth in order to maintain it in the desired occlusal position (Figure 3). Following polymerisation of the resin, the teeth were attached together with sticky wax (Kerr®, Sybron, Emeryville, CA 94608, USA) and then removed from the mold (Figure 4). The aim of this procedure was to maintain the desired space between the two teeth and their alignment until they were remounted in a second mold.
Figure 3. Premolar and molar teeth positioned in the mold with a molar denture tooth between them.

Figure 4. Premolar and molar teeth attached to a denture tooth with sticky wax and removed from the mold.
A mold made with polyvinylsiloxane impression material (Extrude™ Extra and Extrude™ Wash, Kerr® Manufacturing Co., Romulus, MI 48174, USA) by taking an impression of a rectangular box 2.0 cm high, 2.5 cm wide, and 7.4 cm long, was used for a second mounting. The two mounted teeth were inserted into the box mold and one spiral steel nail 5 cm long was placed at each side of the resin bases in order to enforce the resin. The mold was then filled with tray acrylic resin (Figure 5).

Figure 5. Mold made with impression material with the two teeth mounted in a larger resin base with a space of one molar tooth separating them.

After complete polymerisation, the bridge specimen was removed from the mold (Figure 6) and stored in the fridge in distilled water until tooth preparation.
1.2. Preparing the teeth.

One type of prostheses was assigned to each specimen (SP) as follow:

SP 1: current design for RBFPD,
SP 2: modified design for RBFPD,
SP 3: conventional fixed partial denture,
SP 4: RBFPD with only proximal slot preparations.

1.2.1. SP 1 - current design

For SP 1, the preparation followed the design described by Thompson and Wood (1986). The preparation for each abutment tooth included proximal and lingual reductions, occlusal rests and one central proximal groove at the edentulous space side (Figure 7). This design is going to be referred to throughout the text as design 1.
Figure 7. Current design employing multiple rests, wrapping of the retainer wings and proximal grooves. The framework included one groove underneath the pontic running in a facio-lingual direction for test purposes. (Diagram adapted from Thompson VP & Wood M, Proceedings of the International Symposium on Adhesive Prosthodontics, page 104, 1986)

The proximal and lingual reductions were made with round-end tapered diamond bur mounted in high-speed handpiece with air/water spray coolant. The total area for these two segments was extended for more than 180 degrees of tooth circumference. The proximal segment was extended beyond the proximal contact point. The lingual reduction was extended to the far proximal area as much as possible without involving the contact area. The handpiece was manoeuvred in such a manner that the long axis of the bur remained parallel to the plane of the preparation creating a single path of insertion. The reduction, which was terminated in enamel, was just enough to eliminate the natural undercut of the tooth. The gingival margin was placed at least 1 mm above the cemento enamel junction. Care was taken to ensure that the area covered by the preparation would allow fabrication of cast wings with at least 4 mm height.

Occlusal rests were prepared in the mesial and distal fossae of occlusal surfaces of the premolar and molar teeth. Because of the anatomic characteristics, lower molars received a third occlusal rest at the lingual groove area. Rest
preparations were made with high-speed carbide bur # 245 without penetration into dentin with near parallel walls.

A tapered fissure carbide bur # 169 was used to prepare the proximal grooves. The bur was aligned parallel to the path of insertion and the grooves were placed in the proximal wall at the middle of the occlusal rest. Width and depth of the groove was maintained the same as the diameter of the bur and height was dictated by the height of the proximal segment.

1.2.2. SP 2 - modified design for RBFPD

The modified design (El-Mowafy, 1996), referred to in this study as design 2, consisted of proximal and lingual segments, occlusal rests and a proximal slot with two retentive grooves for each abutment tooth (Figure 8).

Figure 8. Modified RBFPD design employing occlusal rests, wrap-around of retainer arms and proximal slots. The framework includes metal projections that seat on the gingival floor of the slot cavities.

The proximal and lingual segments, as well as the occlusal rests were made as described in design 1 (above). However, the occlusal rests adjacent to the edentulous space were replaced by proximal slot cavity preparations.

A carbide bur # 57 was used to prepare the slots. The walls were parallel to the guiding planes to ensure a single path of insertion. Dimensions for the slots were
approximately: 2 to 2.5 mm in the facial-lingual direction, 1.5 to 2.0 mm in the mesial-distal direction and 2.5 to 3.0 mm in the gingival-occlusal direction. These measurements were made using a digimatic caliper model CD-6"BS (Mitutoyo, Co., Japan).

Following impression taking and fabrication of the castings, and before cementation, vertical grooves were placed along the bucco-axial and linguo-axial line angles of the slot cavities. A 1/4 round bur at low speed was used for this purpose.

1.2.3. SP 3 - conventional fixed partial denture

Both abutments were prepared to receive full metal retainers (Figure 9) according to the principles described by Shillingburg et al. (1997a).

![Figure 9. Conventional fixed partial denture with near parallel walls.](image)

The walls were reduced using a chamfer diamond bur in a high-speed handpiece under air/water spray coolant. The same bur was also used to reduce the occlusal surfaces. Care was taken to obtain a smooth preparation with a well defined and continuous finish line that was located 1 mm above the cemento-enamel junction.

1.2.4. SP 4 - RBFPD with only proximal slot preparations

For SP 4, each abutment tooth received one proximal slot cavity preparation with two retentive grooves similar to the ones made in design 2 (Figure 10).
Figure 10. Proximal slots made adjacent to the edentulous space in both abutment teeth. The framework included metal projections that seated in the gingival floor of the slots.

Following preparation, all specimens were carefully examined to ensure proper wall alignment. The specimens were then stored in the fridge in distilled water until impression taking.

1.3. Impression.

Materials used for impression taking, as well as for casting, cementation and restoration are listed in Table 2.
Table 2. Materials used for impression, metal framework, cementation and restoration in order of use.

<table>
<thead>
<tr>
<th>Brand Name</th>
<th>Manufacturer</th>
<th>Description</th>
<th>Batch no</th>
</tr>
</thead>
<tbody>
<tr>
<td>Permadyne®</td>
<td>ESPE, D82229, Seefeld, Germany</td>
<td>Polyether impression material-high</td>
<td>base 532 25706</td>
</tr>
<tr>
<td></td>
<td></td>
<td>viscosity</td>
<td>catalyst 979 25477</td>
</tr>
<tr>
<td>Permadyne® Garant 2:1</td>
<td>ESPE, D82229, Seefeld, Germany</td>
<td>Polyether impression material-low</td>
<td>B039 C007</td>
</tr>
<tr>
<td></td>
<td></td>
<td>viscosity</td>
<td></td>
</tr>
<tr>
<td>Litecast® B</td>
<td>Williams/Ivoclar North America, Inc., St. Catherines, ON L2W 1A3 Canada</td>
<td>Nickel-Chromium-Beryllium alloy</td>
<td>91592F111896</td>
</tr>
<tr>
<td>Aluminum Oxide</td>
<td>Danville Engineering Inc., Danville, CA 94526, USA</td>
<td>Aluminum oxide powder - 50 μm</td>
<td>502913</td>
</tr>
<tr>
<td>Powder</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Bond-It!™</td>
<td>Jeneric®/Pentron® Inc., Wallingford, CT 06492, USA</td>
<td>Enamel/Dentin bonding system</td>
<td>091796</td>
</tr>
<tr>
<td>Cement-It! C&amp;B</td>
<td>Jeneric®/Pentron® Inc., Wallingford, CT 06492, USA</td>
<td>Chemical-cure resin cement</td>
<td>100996</td>
</tr>
<tr>
<td>Fleck's Cement</td>
<td>Mizzy, Inc. Cherry Hill, NJ 08002, USA</td>
<td>Zinc phosphate cement</td>
<td>powder P24 020596 liquid H85 040788</td>
</tr>
<tr>
<td>TPH™ Spectrum</td>
<td>Dentsply/Caulk, Milford DE 19963-0359 USA</td>
<td>Hybrid composite</td>
<td>9503271</td>
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</tbody>
</table>
Before impression taking, exposed root surfaces of teeth with severe undercuts were covered with Boxing Wax (The Hygenic Co./ Akron, Ohio 44310, USA) for easy removal of the impression. Impression was taken using a double mix technique with Permadyne® and Permadyne® Garant 2:1-syringe (ESPE, Seefeld D-82229, Germany). Permadyne® was proportioned in ratio of 1:1, mixed for 30 seconds and loaded in a sectional tray (Disposable Spacer Trays™, COE-ICI Dental, Coe Laboratories, Inc., Chicago, IL 60658, USA). Meanwhile, Permadyne® Garant 2:1 was applied to the teeth surfaces and the loaded tray was immediately positioned over the specimen, pressed and held in place with minimal pressure until complete setting. The impression was then removed, examined for accuracy and sent to the technician for fabrication of casting.

All specimens were stored in the fridge in distilled water until castings were ready for try-in and cementation.

1.4. Casting.

The same technician did all laboratory work. A working cast was obtained by pouring the impression in die stone (Die-Keen®, Bayer Corporation Dental Products, South Bend, IN 46614, USA) according to the manufacturer’s instructions. The wax pattern was made with Green Inlay Casting Wax (Kerr®, Sybron, Emeryville, CA 94 608, USA). The pontic was designed in such a way so that a distance of at least 5 mm was maintained between the acrylic resin base of the specimen and the bottom surface of the pontic to allow space for mechanical testing. An orientation groove was placed at the centre of the bottom surface of the pontic running in a facial-lingual direction to assist in positioning the loading device during testing and hence ensure even distribution of forces.

For the specimens that contained proximal slots (SP2 and SP4) projections of the framework were waxed up into the boxes on the gingival floor of the cavity (Figure 11). The projections were at least 1 mm thick and were built with margin wax (Pro-Art Margin Wax - Bordeaux, Williams, St. Catherines, ON L2W 1A3, Canada). This provided sufficient space on top of the projections for resin composite restorations.
Figure 11. Diagram showing the slot cavities in cross-section with the metal projections of the framework extending into them. Notice the groove placed underneath the pontic for test purposes. (Diagram adapted from El-Mowafy OM., J Can Dent Assoc 1996;62, page 864)

The wax pattern was invested using phosphate bonded carbon-free investment for high fusing alloys (Hi-Temp, Whip Mix® Co., Louisville, Kentucky 40217-0183, USA). The investment was burnt out and the casting was made using a centrifugal casting machine, a gas-air blowpipe, and a Nickel-Chromium-Beryllium alloy (Litecast® B, Ivoclar/Williams, St. Catherines, ON L2W 1A3, Canada). Following casting, the prosthesis was removed from the investment, finished and polished following standard techniques.

1.5. Metal surface treatment.

Finished castings were tried-in and if a casting did not seat all the way, a thin coat of water soluble indicator solution was used to detect the presence of interference. Areas preventing seating were relieved with a # 2 round bur. The procedure was repeated until adequate seating was obtained. If a casting was found to have a problem that could not be corrected by simple adjustments such as rocking, it was sent back to the technician to be remade.

Before cementation, the inner surfaces of the casting were air-abraded with 50 μm aluminum oxide powder (Danville Engineering Inc., Danville, CA 94526, USA) at a distance of less than 1 cm between the nozzle and the metal surface for 5 to 10 seconds. Air abrasion was performed using a Micro Etcher™- Model ERC (Danville
Engineering Inc., Danville, CA 94526, USA) with an air pressure of 90 psi inside a dust-collecting cabinet (Micro Cab™ - Model MC - Danville Engineering Inc., Danville, CA 94526, USA).

After sandblasting, the surface was cleaned with water in an ultrasonic bath (Dual Action Ultrasonic Cleaning System, model FS5, Fisher Scientific, Nepean, ON K2E 7L6, Canada) for 2 minutes, washed with tap water and dried.

1.6. Cementation.

Before cementation, all teeth were cleaned with pomes and prophy cups, washed and dried. For all specimens, except SP 3, the cementation was done using a dentin bonding agent (Bond-It!™, Jeneric®/Pentron® Inc., Wallingford, CT 06492, USA) and a chemical-cure resin cement (Cement-It! C&B, Jeneric®/Pentron® Inc., Wallingford, CT 06492, USA). The slots for SP2 and SP4 were restored with TPH™ Spectrum (Dentsply/Caulk, Milford DE 19963-0359, USA). The conventional fixed partial denture (SP 3) was cemented with Fleck's cement (Mizzy Inc., Cherry Hill, NJ 08002, USA).

1.6.1. Cementation procedures for SP 1

Enamel was acid-etched with 37% phosphoric acid for 30 seconds, washed for 20 seconds and dried. Bond-It!™ primers A and B were mixed and applied to the enamel surface in five consecutive coats and to the inner surface of the casting in two coats. Immediately after application they were dried with a gentle stream of compressed air. Primed surfaces were examined to ensure the presence of a glossy surface.

Cement-It! C&B was dispensed in catalyst:base ratio of 1:1, mixed until a uniform mix was obtained, and applied to the fitted casting surface. After the casting was seated using finger pressure for five minutes, the pressure was released and excess cement was removed using a hollenback instrument. The specimen was then stored in distilled water at 37 °C until testing.

1.6.2. Cementation procedures for SP 2

Before cementation and cleaning with pomes and prophy cup, vertical grooves were placed along the bucco-axial and linguo-axial lineangles of the slot cavities using
a ¼ round bur in a slow-speed handpiece. Enamel and dentin were conditioned with 37% phosphoric acid. The acid was first applied to the enamel, left for 15 seconds and then extended to the dentinal surfaces of the slot cavities for another 15 seconds. The cavities were washed and dried with a gentle stream of air in order to prevent desiccation. If the dentin surface was overdried, it was rewet with wet cotton pellets. Bond-It!™ primers A and B were mixed and applied to the enamel and dentin surfaces in five coats and to the inner metal surface in two coats.

Cement-It! C&B was mixed as above and applied to the casting wings and the bottom surface of the metal projections, taking care to leave the upper part of the projections clean. The framework was seated and maintained under finger pressure for five minutes. The pressure was then released and the excess cement was removed using a hollenback instrument.

If the cement could not be removed from the retentive grooves of the proximal slots, a low speed ¼ round bur was used to clean the grooves. The enamel and dentin surfaces of the slots were acid etched with 37% phosphoric acid for 15 seconds, washed for 20 seconds and dried with a gentle stream of compressed air. Bond-It!™ primers A and B were mixed and applied to enamel and dentin in 5 coats. The surfaces were then dried and a single coat of Bond-It!™ resin was applied and light cured for 20 seconds with Visilux™ 2, model 5520/AA (3M Dental Products Division, St. Paul, MN 55-144-1000, USA). Slot cavities were restored using TPH™ Spectrum (shade B3) in 3 increments to minimise the effect of polymerisation shrinkage. The first one was inserted obliquely to the facial and gingival walls, the second to the lingual and gingival walls, and the third filled the cavity. Each increment was light cured for 60 seconds with Visilux™ 2. The specimen was then stored in distilled water at 37 °C until testing.

1.6.3. Cementation procedures for SP 3

For cementation of SP 3, the enamel was cleaned with pomes and prophy cup, washed and dried and the inner metal surfaces of the casting was sandblasted as described before.

Fleck's® cement was dispensed on a glass mixing slab, divided and mixed according to the manufacturer's instruction. The cement was applied to the casting
fitted surface and seated with finger pressure for five minutes. The pressure was released, excess cement removed and the specimen was stored in distilled water at 37 °C until testing.

1.6.4. Cementation procedures for SP 4

Cementation procedures for SP 4 were the same as described for SP2.

1.7. Fatigue test.

Seven days after cementation, the specimens were subjected to compressive fatigue loading in a uniaxial servohydraulic testing machine (Instron Model 8501 test frame with Model 8500 electronic controller, Instron Corporation, Canton, Mass 020210, USA) capable of operating in both position- and force-control. The test machine was equipped with a built-in LVDT (Linearly Variable Differential Transformer) position sensor and interfaced to a 1 kN load cell (Model TH-LP1B, T-Hydronics Inc., Bunburry, Ohio 43074, USA) for force measurements.

A specially-made rounded-end stylus was attached to the load cell. The ball in the end had 7.6 mm diameter. Below the load cell, a Plexiglas water tank equipped with a specimen holder was secured to the machine's platform. A specimen was clamped in the specimen holder and water (at room temperature) was poured into the Plexiglas tank until the specimen was completely submerged. The tank was adjusted on the platform until the stylus's tip was centred above the central fossa of the occlusal surface of the pontic, then secured in position with C-clamps (Figure 12 and Figure 13).
Figure 12. A specimen in place during fatigue testing. Notice the stylus head touching the occlusal surface of the pontic.
Each bridge specimen was subjected to fatigue cycling by applying a sinusoidal compressive force ranging between 50 N and 800 N at a frequency of 4 Hz for a total of 25,000 cycles. Initially, with the test machine in position-control, the actuator was moved upward until the stylus engaged the pontic’s occlusal surface. A compressive force of 50 N was applied by further moving the actuator upward, then the test machine’s control was switched from the position-control to force-control. The machine’s waveform generator was then set to a Haver-Sine waveform with amplitude of 750N and a frequency of 4 Hz.

Electronic event switches on the machine’s controller were set to halt the test after 25,000 cycles, or if the specimen failed before the number of cycles were completed. Specimen failure was defined as a displacement of 1 mm or more of the mean position of the actuator and hence, the pontic during cycling. When a specimen failed, the machine automatically stored the number of cycles at which failure took place.
1.8. Tensile test.

Specimens that sustained the fatigue test were subsequently subjected to a tensile test on the next day. Tensile tests were performed using a uniaxial servomechanical testing machine (Instron Model 4301 test frame). The test machine was equipped with a built-in position encoder and a 1 KN load cell (Instron) for force measurements. The test machine was interfaced to a Strip Chart Recorder (Instron) that plotted force vs. time.

The load cell was secured to the movable crosshead of test machine. Below the load cell, a specimen was placed on the platform using a specimen holder. One end of a 20 cm long steel chain was attached to the load cell and in its other end, an S-shaped hook was placed. With the hook hanging steady in a vertical position, the specimen was positioned on the platform such that the hook was directly above the pontic. The specimen was then fixed to the platform with a pair of C-clamps.

The movable crosshead was then lowered, and the hook was positioned to engage the groove underneath the pontic (Figures 14 and 15). The force signal was electronically zeroed to compensate for the weight of the chain and hook. The chart recorder’s speed was set to 100 mm/min and set in motion. The test machine’s crosshead speed was set to 1 mm/min and switched on. Load was applied until failure of the specimen took place.
Figure 14. Specimen in position in the uniaxial servomechanical testing machine for tensile test.
The peak tensile force was recorded using the machine's peak capture feature and stress/strain curves were retrieved from chart recorder.

2. Main Study

Following analysis of the pilot study, 70 sound extracted teeth: 35 premolars and 35 molars, were collected, cleaned and sterilised with gamma radiation as described previously (Chapter 2, Section 1).

Thirty five bridge specimens; 20 with lower molars and premolars and 15 with upper molars and premolars, were prepared as described before (Chapter 2, Section 1.1). The specimens were divided in five equal groups in such a way that each group had three specimens with upper teeth and four specimens with lower teeth. Care was taken to ensure that teeth of similar sizes were assigned evenly among the groups.

The groups were randomly assigned to the following preparation designs:
Group 1 - current design for RBFPD

The specimens were prepared and cementation was done in the same way as for SP 1 of the pilot study. The preparation of each abutment tooth included proximal and lingual reductions, occlusal rests and one proximal groove (Design 1). Cementation of the castings was performed using Bond-It!™ and Cement-It! C&B.

Group 2 - modified design for RBFPD cemented with Cement-It!C&B

The specimens were prepared and cemented in the same way as for SP 2 of the pilot study. The preparation of each abutment tooth included proximal and lingual reduction, occlusal rests and one proximal slot with two retention grooves (Design 2). Cementation of the castings was performed using Bond-It!™ and Cement-It! C&B and the slots were restored with TPH™ Spectrum.

Group 3 - modified design for RBFPD cemented with Panavia® 21

The preparation of the abutment tooth was the same as for Group 2 (Design 2). However, the cementation was performed using Panavia® 21 (Kuraray Co., LTD., Osaka 530, Japan). The slots were also restored with TPH™ Spectrum.

Group 4 - RBFPD with only proximal slot preparations

The preparation of each abutment tooth included one proximal slot with two retention grooves with bevel of buccal, lingual and gingival cavosurface margins (Design 3). Cementation of the framework was performed using Bond-It!™ and Cement-It! C&B and the slots were restored with TPH™ Spectrum.

Group 5 - inlay design for RBFPD

The preparation of each abutment tooth included proximal and lingual reductions, occlusal rests and one proximal slot with slightly divergent walls (Design 4). Cementation of the framework was done using Bond-It!™ and Cement-It! C&B.

2.1. Preparing the teeth.

Tooth preparations for Groups 1, 2 and 3 was done as described previously (Chapter 2, Sections 1.2.1. and 1.2.2.).
For Group 4, the preparation of the slot cavities was done as described for SP 4 in the pilot study (Chapter 2, Section 1.2.4.). However, bevels were added to the buccal and lingual walls using a round-end tapered diamond. The same bur was used to bevel the gingival cavosurface margin of the box continuous with the buccal and lingual bevels (Figure 16).

Figure 16. Proximal slots with buccal and lingual bevels continuous with the gingival bevel. The framework covered the region corresponding to the bevel and had metal projections that seated in the gingival floor of the slots.

Following impression taking, retentive grooves were made along the bucco-axial and linguo-axial lineangles of the slot cavities. A 1/4 round bur was used for this purpose.

For Group 5, the proximal and lingual reductions as well as the occlusal rests were prepared in the same manner as for Groups 1, 2 and 3. However, the proximal slots were prepared with divergent walls. A tapered fissure carbide bur # 170 was used for this purpose. The handpiece was manoeuvred in such a manner that the long axis of the bur was maintained parallel to the long axis of the tooth giving the cavity walls the desired inclination (Figure 17). Dimensions of the slot cavities were approximately 2 to 2.5 mm in the facio-lingual direction, 1.5 to 2.0 mm in the mesio-
distal direction and 2.5 to 3.0 mm in the cervico-occlusal direction. No grooves were added to the slots.

Figure 17. Inlay design employing occlusal rest, wraparound wings and proximal slots with divergent walls. The framework was extended to completely fill the slots as metallic inlays.

Following preparation, all specimens were carefully examined for accuracy of work and adjusted if necessary.

2.2. Impression.

Impression was taken using the same technique described previously (Chapter 2, Section 1.3).

2.3. Casting.

Castings were done as described before (Chapter 2, Section 1.4.).

For Group 4, besides the metal projections into the slot cavities, the casting extended to cover the bevels of the cavosurface margins of the cavity. For group 5, the casting was extended to completely fill the slot cavities as a metallic inlay.


2.4. Cementation.

Preparation of the metal surface and cementation for groups 1, 2, 4 and 5 was conducted in the same manner as for the pilot study (Chapter 2, Sections 1.5. and 1.6., respectively). Only one modification was done to the technique: Bond-It!™ primers A and B were not applied to the inner surfaces of the castings and it was replaced by two coats of Cement-It! C&B Metalbond. For group 4, the cement was also applied to the region of the bevels. The cementation and restoration was then executed as described before.

For Group 3, the cementation of the castings was done using Panavia® 21 (Kuraray Co., LTD., Osaka 530, Japan, Batch n°.61162) and slot cavities were restored with TPH™ Spectrum (shade B3). Enamel and dentin were conditioned with 40% phosphoric acid (supplied gel). The acid was first applied to the enamel, left for 15 seconds and then extended to dentin and left for another 15 seconds. The preparation was washed for 20 seconds and dried with a gentle stream of compressed air. If dentin was judged to have been overdried, surface wetness was restored with wet cotton pellets.

Panavia® 21's primers A and B were mixed together and applied to enamel and dentin surfaces in one coat. After 60 seconds, the primer was dried with a gentle stream of air.

No treatment, besides sandblasting, was performed on the metal surfaces. Panavia® 21's paste and catalyst were dispensed on a pad and mixed until a uniform mix was obtained and applied to the inner surfaces of the castings. Care was taken to leave the upper part of the metal projections free of cement. After the framework was seated with finger pressure for 60 seconds, the pressure was released and the excess of cement was removed using cotton pellets. All tooth/metal margins were covered with a thin layer of Panavia® 21’s Oxyguard II. After three minutes, Oxyguard II was removed with cotton pellets and teeth were washed and dried. Slot cavities were then restored with TPH™ Spectrum as described in the pilot study.

Following cementation and restoration, the specimens were stored in distilled water at 37 °C until testing.
2.5. Fatigue test.

The same technique and instruments used for fatigue testing of the specimens of the pilot study (Chapter 2, Section 1.7.) were used in the main study. However, a different load cycle was applied here. The specimens were fatigued by applying a sinusoidal compressive force ranging between 10N (minimum) and 450N (maximum) at a frequency of 4 cycles per second for a total of 230,000 load cycles. The amount of time spent to load-cycle each specimen was 16 hours.

The testing machine was set so that if displacement of 1 mm or more of the mean position of the pontic was detected during load-cycling, the test was terminated and the machine automatically stored the number of cycles at which failure occurred.

2.6. Tensile test.

Specimens that sustained the fatigue test were further subjected to tensile testing until failure (detachment) of the prostheses took place, in the same manner as for the pilot study (Chapter 2, Section 1.8.).

During tensile testing, a video camera recorder with 12X variable zoom (Handycam Video 8, model no. CCD-TR64, Sony® Corporation, Japan) was used to record the entire test. Each recorded test was subsequently analysed carefully by viewing at slow speed in a VCR machine with capacity to run the tape frame by frame (Video 8 Digital Stereo, model no. EV-S550, Sony® Corporation, Japan). Collected information was correlated to the stress/strain chart recordings.

If the stress/strain curve was a continuous line throughout, the tensile force to dislodge the prosthesis was the force registered at the peak of the line. If there was evidence of a break or blip in the line before peak force was achieved, the video film recording was carefully examined to see if one of the two wings separated from one of the abutment teeth before complete dislodgement of the prosthesis. If this was the case, the tensile force necessary to dislodge one side of the prosthesis was recorded as the tensile force responsible for causing failure of the prosthesis. If no movement of the framework was detected, the peak point of the line was considered as the force that caused the failure.
2.7. Analysing the mode of failure

Following tensile testing, each abutment was examined carefully to determine mode of failure. Failures were classified as either cohesive if occurred within the tooth structure or within the cement, or adhesive if occurred at the enamel/cement interface or at the metal/cement interface. Therefore, each abutment tooth was classified as having either adhesive, cohesive, or adhesive/cohesive failures if both types of failures were present.

Fractures of the tooth structure were classified into three types:
Minor: small fracture involving enamel only or enamel and some dentin, but without cuspal fracture;
Medium: fracture of one or two cusps;
Major: horizontal fracture at the cervical level of the tooth.

2.8. Measurement of the preparation area

Preparation areas for all specimens were calculated using tin foil (Tin Foil .001, Buffalo Dental Mfg. Co., Inc., Syosset, NY 11791, USA) and a digimatic caliper (Model CD-6”BS, Mitutoyo, Co., Japan). The foil was carefully adapted to the area corresponding to the proximal and lingual segments of the prostheses on the stone model. A black permanent marker with ultra fine point was used to mark wing margins on the tin foil. A surgical blade was then used to cut out the tin foil covering the wing area.

Each tin foil piece was straightened and placed in a slide mold together with a square plastic sheet with marked squares measuring 1 mm² each. A slide scanner (Microtek Scanmaker 35T, Microtek Lab. Inc., Redondo Beach, CA 90278, USA) with a Macintosh Quadra 800 computer (Apple Canada Inc., Markham, ON L3R 5G2, Canada) and a 16 inch 8-bit color monitor were used to digitise the images of the tin-foil cut outs.

Scanning was conducted with a resolution of 968 dots per inch, which translates into approximately 38 pixels per mm and a magnification of 13x. The result
was a black and white image with a square background with the black corresponding to the image of the tin foil.

The public domain NIH Image program (U.S. National Institute of Health, available from the Internet by anonymous FTP from zippy.nimh.nih.gov or on floppy disk from the National Technical Information Service, Springfield, Virginia, USA, part number PB93-504868) was used to digitally process the image and perform the measurements correspondent to the area of the wings. The known measurement of the square in the background was used to calibrate the program. The periphery of the image was outlined and the area inside was calculated in square millimeters.

A caliper was used to measure the area corresponding to the slot cavities and occlusal rests. For the area corresponding to the marginal walls of the slot cavities, the height of both lingual and buccal walls, as well as the width of the gingival floor were measured. The three measurements were added together and multiplied by the thickness of the marginal walls (mesio-distal direction) measured in the gingival floor. Height and width of the axial wall was measured, multiplied and added to the area corresponding to the marginal walls to obtain the total area of the slot.

Occlusal rests had a shape that resembled a triangle and the area was estimated based on this shape. The width and depth of the rest was measured, multiplied and divided by two. The number obtained corresponded to the area of the floor of the rest. Both marginal walls were measured in length, added together and multiplied by the height. The area corresponding to the floor plus the area corresponding to the walls was considered as the estimated area of the rest.

The areas corresponding to the rests, slots and wings of each abutment tooth were added together and the total surface area involved in the prosthesis was finally calculated by adding the numbers obtained for the premolar and molar teeth.

2.9. Statistical analysis

Tensile forces recorded for dislodgement of the prostheses were analysed using one factor analysis of variance and Duncan’s multiple range test. SAS®
program (SAS Institute Inc., Cary NC 27512-8000, USA) was used to perform the statistical analysis.
1. Pilot Study

Failure during the fatigue test occurred only with SP 4. The gingival seat of the slot cavity of the premolar tooth collapsed under loading after 2,250 cycles (Figure 18).
All remaining specimens sustained the fatigue test and were further subjected to the tensile test to detach the prostheses. Separation force in N for each specimen is recorded in Table 3.

Table 3. Separation force in N for each specimen.

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Separation force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>SP 1</td>
<td>267.6</td>
</tr>
<tr>
<td>SP 2</td>
<td>502.9</td>
</tr>
<tr>
<td>SP 3</td>
<td>254.9</td>
</tr>
<tr>
<td>SP 4</td>
<td>_</td>
</tr>
</tbody>
</table>

A higher tensile force was necessary to separate the prosthesis from the specimen prepared with design 2. Fracture of the mesio-lingual cusp of the molar tooth and chipping of the enamel close to the slot cavity of the premolar tooth occurred with this specimen. All resin cement remained attached to the enamel except for the tooth parts that fractured.

Despite of a lower force, a more severe fracture occurred on prosthesis separation with the specimen prepared for conventional FPD. Horizontal fracture at the cervical level of the premolar tooth occurred without fracture of the molar tooth. Most of the zinc phosphate cement remained attached to the preparation surface of the molar tooth.

The specimen with design 1 recorded separation force similar to that of the specimen with the conventional FPD. No tooth fracture occurred with this specimen and all cement remained attached to the enamel.

2. Main Study

Following load cycling, all specimens were subjected to the tensile test since no failure occurred during fatigue load cycling.
2.1. Tensile force.

The separation force for each specimen as well as the mean separation force and standard deviation for each group is recorded in Table 4. Results for specimens with teeth of similar size and shape are displayed in the same row.

Table 4. Separation forces (N) for each specimen as well as means and standard deviations (SD) for each group.

<table>
<thead>
<tr>
<th>Specimen number</th>
<th>Group 1</th>
<th>Group 2</th>
<th>Group 3</th>
<th>Group 4</th>
<th>Group 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sp 1</td>
<td>279.1</td>
<td>526.2*</td>
<td>529.4*</td>
<td>446.8*</td>
<td>341.1</td>
</tr>
<tr>
<td>Sp 2</td>
<td>310.1</td>
<td>373.8*</td>
<td>575*</td>
<td>506.8</td>
<td>414.2</td>
</tr>
<tr>
<td>Sp 3</td>
<td>209</td>
<td>627.6*</td>
<td>641.9*</td>
<td>537.2*</td>
<td>495.8</td>
</tr>
<tr>
<td>Sp 4</td>
<td>346.6</td>
<td>619.1*</td>
<td>678.4*</td>
<td>442</td>
<td>435</td>
</tr>
<tr>
<td>Sp 5</td>
<td>479</td>
<td>404*</td>
<td>600.3</td>
<td>438.3</td>
<td>-</td>
</tr>
<tr>
<td>Sp 6</td>
<td>466</td>
<td>538.5</td>
<td>597.6*</td>
<td>414*</td>
<td>416.6</td>
</tr>
<tr>
<td>Sp 7</td>
<td>437</td>
<td>588.7*</td>
<td>312.6*</td>
<td>355.4</td>
<td>399.9</td>
</tr>
<tr>
<td>Mean</td>
<td>360.9</td>
<td>525.4</td>
<td>562.2</td>
<td>448.6</td>
<td>417.1</td>
</tr>
<tr>
<td>SD</td>
<td>102.7</td>
<td>100.9</td>
<td>119.8</td>
<td>59.5</td>
<td>50.2</td>
</tr>
</tbody>
</table>

*Specimen had horizontal fracture of the premolar tooth.

The lowest mean value (360.9 N) occurred with Group 1 while the highest mean value (562.2 N) occurred with Group 3.

One specimen from Group 5 was discarded as root fracture at pin level occurred during the tensile test in both abutment teeth. Failure occurred at 321 N indicating that the force necessary to dislodge the prosthesis was higher than this value. This specimen was not included in the study.

One factor analysis of variance (ANOVA) indicated that there was a significant difference among the groups (p=0.0022). Duncan’s Multiple Range Test indicated that there was no difference in mean separation force among groups 4, 5 and 1, and
between groups 3 and 2. Also, mean separation force for groups 2 and 4 were not significantly different. Duncan's grouping is presented in Table 5.

Table 5. Statistical analysis results obtained with Duncan Multiple Range test. Means with the same letter are not significantly different.

<table>
<thead>
<tr>
<th>Duncan grouping</th>
<th>Mean</th>
<th>N</th>
<th>Group</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>562.17</td>
<td>7</td>
<td>Group 3</td>
</tr>
<tr>
<td>B</td>
<td>525.41</td>
<td>7</td>
<td>Group 2</td>
</tr>
<tr>
<td>B</td>
<td>448.64</td>
<td>7</td>
<td>Group 4</td>
</tr>
<tr>
<td>C</td>
<td>417.1</td>
<td>6</td>
<td>Group 5</td>
</tr>
<tr>
<td>C</td>
<td>360.97</td>
<td>7</td>
<td>Group 1</td>
</tr>
</tbody>
</table>

2.2. Mode of failure.

Classification of the mode of failure at enamel/cement/metal interface for premolar and molar teeth is presented in Tables 6 and 7, respectively. Some of the prostheses did not detach from the molar teeth after separation from the premolars. It happened with one specimen from Group 1 and one specimen from Group 4, as well as with two specimens from Group 5. Therefore, no failure in the molar tooth was recorded.

Classification of fracture patterns for premolar and molar teeth is presented in Tables 8 and 9, respectively.
Table 6. Classification of failures at the enamel/cement/metal interface for premolar teeth. Number of abutments in each category is indicated.

<table>
<thead>
<tr>
<th>Type of failure</th>
<th>Group 1</th>
<th>Group 2</th>
<th>Group 3</th>
<th>Group 4</th>
<th>Group 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adhesive</td>
<td>6</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>3</td>
</tr>
<tr>
<td>Cohesive</td>
<td>-</td>
<td>6</td>
<td>7</td>
<td>3</td>
<td>-</td>
</tr>
<tr>
<td>Adhesive/Cohesive</td>
<td>1</td>
<td>1</td>
<td>-</td>
<td>4</td>
<td>3</td>
</tr>
<tr>
<td>Total</td>
<td>7</td>
<td>7</td>
<td>7</td>
<td>7</td>
<td>6</td>
</tr>
</tbody>
</table>

Table 7. Classification of failures at the enamel/resin/metal interface for molar teeth. Number of abutment in each category is indicated.

<table>
<thead>
<tr>
<th>Type of failure</th>
<th>Group 1</th>
<th>Group 2</th>
<th>Group 3</th>
<th>Group 4</th>
<th>Group 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adhesive</td>
<td>3</td>
<td>1</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Cohesive</td>
<td>-</td>
<td>3</td>
<td>5</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Adhesive/Cohesive</td>
<td>3</td>
<td>3</td>
<td>2</td>
<td>6</td>
<td>4</td>
</tr>
<tr>
<td>Total</td>
<td>6</td>
<td>7</td>
<td>7</td>
<td>6</td>
<td>4</td>
</tr>
</tbody>
</table>
Table 8. Classification of tooth fracture of premolar teeth. Number of abutments in each category is indicated.

<table>
<thead>
<tr>
<th>Type of fracture</th>
<th>Group 1</th>
<th>Group 2</th>
<th>Group 3</th>
<th>Group 4</th>
<th>Group 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Minor</td>
<td>-</td>
<td>1</td>
<td>-</td>
<td>1</td>
<td>2</td>
</tr>
<tr>
<td>Medium</td>
<td>1</td>
<td>-</td>
<td>1</td>
<td>3</td>
<td>1</td>
</tr>
<tr>
<td>Major</td>
<td>-</td>
<td>6</td>
<td>6</td>
<td>3</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 9. Classification of tooth fracture of molar teeth. Number of abutments in each category is indicated.

<table>
<thead>
<tr>
<th>Type of fracture</th>
<th>Group 1</th>
<th>Group 2</th>
<th>Group 3</th>
<th>Group 4</th>
<th>Group 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Minor</td>
<td>2</td>
<td>1</td>
<td>-</td>
<td>4</td>
<td>-</td>
</tr>
<tr>
<td>Medium</td>
<td>-</td>
<td>4</td>
<td>7</td>
<td>1</td>
<td>4</td>
</tr>
<tr>
<td>Major</td>
<td>-</td>
<td>1</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>
For Group 1, most of the failures occurred adhesively at the cement/metal interface and the cement remained attached to the abutment teeth (Figure 19). With only one abutment adhesive failure at the cement/enamel interface occurred with the cement remaining partly attached to the metal wing (Figure 20) and partly to the tooth structure. Only three abutments had tooth fractures and they were classified as minor (2 molar teeth) and medium fractures (1 premolar tooth). In one specimen, the prosthesis dislodged from the premolar tooth but remained attached to the molar tooth (Figure 21). Therefore, no failure in the molar tooth was recorded.

For Group 2, there was only one abutment tooth (molar) in which failure was classified as adhesive as tooth and restoration remained intact after dislodgement of the prosthesis (Figure 22). Tooth and restoration fractures occurred with all remaining abutments. Horizontal fracture at the cervical level occurred in six premolars (Figures 22 and 23) and one molar tooth (Figure 23). Most of the molar teeth (4) had failures classified as medium (Figure 24) and only one had failure classified as minor failure.

Group 3 had 12 abutments having cohesive failures within tooth structure (Figures 25 and 27). Two molar teeth had failures classified as adhesive/cohesive (Figure 26). Failures at the cervical level of teeth occurred with six premolars (Figures 26 and 27). All molar teeth and one premolar tooth had cusp fractures (Figures 26 and 27).

For Group 4 all failures were classified as either cohesive or cohesive/adhesive. For one specimen, the framework remained attached to the molar tooth and, therefore, no failure was recorded for this abutment. One molar tooth had failure only within the restorative material. Horizontal fractures at the cervical level of the teeth occurred with three premolars (Figure 28). The remaining abutments had tooth fractures classified as either minor fractures (one premolar and three molar teeth) or medium (three premolar and one molar tooth) (Figures 28 and 29).

As mentioned earlier, one specimen of Group 5 was disregarded because tooth fracture occurred at the level or the pins used to attach the tooth to the acrylic resin base (Figure 30). In addition, with two other specimens the prostheses remained attached to the molar teeth after dislodgement of the wings of the premolar teeth. Three premolar teeth had adhesive failures at the cement/metal interface (Figure 31).
The remaining abutments had failures classified as adhesive/cohesive. Cusp fractures occurred in four molar teeth (Figures 31 and 32) and in one premolar tooth. Minor fractures occurred in two premolar teeth (Figure 32).
Figure 19. Specimen from Group 1 showing adhesive failure at the cement/metal interface without tooth fracture.

Figure 20. Fitted surface of metal framework following dislodgement of the prosthesis (Group 1). This is the only specimen that had adhesive failure at the cement/enamel interface.
Figure 21. Specimen from Group 1 in which the prosthesis remained attached to the molar tooth. No tooth fracture occurred in the premolar abutment.

Figure 22. Specimen from Group 2 showing adhesive failure at the cement/metal restoration/metal interfaces with no fracture of the molar abutment. Premolar showing major fracture.
Figure 23. Specimen from Group 2 showing major fractures in both premolar and molar abutment teeth.

Figure 24. Specimen from Group 2 showing medium tooth fracture of the molar abutment and minor tooth fracture of the premolar abutment.
Figure 25. Specimen from Group 3 showing cohesive failure within tooth structure.

Figure 26. Specimen from Group 3 showing cohesive failure of the premolar abutment classified as major, and cohesive/adhesive failure of the molar abutment with the fracture being classified as medium.
Figure 27. Specimen from Group 3 showing cohesive failure within tooth structure of both premolar and molar abutments. Fracture of premolar was classified as major and that of the molar as medium.

Figure 28. Specimen from Group 4 showing major fracture of the premolar abutment and cohesive/adhesive failure of the molar abutment with medium fracture.
Figure 29. Specimen from Group 4 showing minor tooth fracture of the molar abutment and medium tooth fracture of the premolar abutment.

Figure 30. Specimen from Group 5 that had fracture at the pin level. This specimen was not included in the study.
Figure 31. Specimen from Group 5 showing adhesive failure at the cement/metal interface for the premolar abutment and cohesive/adhesive failure of the molar abutment with the fracture classified as medium.

Figure 32. Specimen from Group 5 showing minor fracture of the premolar abutment and medium fracture of the molar abutment.
2.3. Preparation area.

Areas corresponding to the preparation for each specimen are displayed in Table 10. The least amount of preparation area was associated with Group 4 (77.41 mm²), which had only the proximal slots. The most preparation area was associated with group 3 (188.08 mm²). However, this value was very similar to mean values for Groups 2 and 5 and a little higher than the mean value for Group 1 (167.23 mm²).

There was some variation in the area prepared for each specimen especially for Sp 7, which had the greatest surface area prepared in all five groups. The preparation of the slot cavities increased the surface area in 17.8 mm² for Group 2, 20.8 mm² for Group 3 and 14.6 mm² for Group 5, when compared to Group 1. These increases in surface area correspond to 9.6%, 11% and 8% of the total preparation areas for Groups 2, 3 and 5, respectively.

Table 10. Total preparation areas for each bridge specimen in mm². Mean areas and standard deviation (SD) for each group is given at the bottom.

<table>
<thead>
<tr>
<th>Bridge specimen</th>
<th>Group 1</th>
<th>Group 2</th>
<th>Group 3</th>
<th>Group 4</th>
<th>Group 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sp 1</td>
<td>152.88</td>
<td>198.13</td>
<td>188.17</td>
<td>90.61</td>
<td>159.65</td>
</tr>
<tr>
<td>Sp 2</td>
<td>147.01</td>
<td>190.13</td>
<td>181.30</td>
<td>79.16</td>
<td>177.46</td>
</tr>
<tr>
<td>Sp 3</td>
<td>186.22</td>
<td>177.71</td>
<td>172.96</td>
<td>79.14</td>
<td>183.68</td>
</tr>
<tr>
<td>Sp 4</td>
<td>164.71</td>
<td>187.77</td>
<td>189.11</td>
<td>71.98</td>
<td>188.10</td>
</tr>
<tr>
<td>Sp 5</td>
<td>156.18</td>
<td>157.04</td>
<td>163.10</td>
<td>58.69</td>
<td>-</td>
</tr>
<tr>
<td>Sp 6</td>
<td>140.25</td>
<td>146.44</td>
<td>162.85</td>
<td>67.18</td>
<td>189.97</td>
</tr>
<tr>
<td>Sp 7</td>
<td>223.35</td>
<td>238.49</td>
<td>259.09</td>
<td>95.12</td>
<td>191.97</td>
</tr>
<tr>
<td>Mean</td>
<td>167.23</td>
<td>185.1</td>
<td>188.08</td>
<td>77.41</td>
<td>181.80</td>
</tr>
<tr>
<td>SD</td>
<td>28.81</td>
<td>29.97</td>
<td>33.11</td>
<td>12.77</td>
<td>12.02</td>
</tr>
</tbody>
</table>
Freshly extracted teeth are by nature a potential source for cross-contamination to laboratory equipment and personnel and, therefore, must be disinfected or sterilised. Ideally, the sterilising system must not only render the teeth sterile, but also should not modify in any way the structure of the enamel or dentin. Presently, methods of sterilising extracted teeth include steam (autoclave), liquid chemicals (solutions or substances that inhibit bacterial growth) and gaseous chemicals (ethylene oxide). However, while these methods are efficient, they might permanently alter the structure of the enamel and dentin (White, Goodis, Marshall & Marshall, 1994). Alternative methods such as YAG laser at 60W and silver cations were tried unsuccessfully. The fact that ionising radiation can kill microorganisms was recognised in 1896, shortly after the discovery of x-rays. Nevertheless, the development of radiation sterilisation commenced only in the 1950s, when large sources of ionising radiation became available (Woods & Pikaev, 1994). In 1956, it began to be used to sterilise sutures by Ethicon, Inc. Currently, radiation sterilisation is applied to a broad range of disposable medical products such as syringes and needles, surgical sutures and utensils, implant materials and tissues (artificial heart valves, bone grafts, dental implants, etc), blood-handling and transfusion equipment, dressings, gloves, masks, etc. It is also used for some pharmaceuticals and in the packaged foods industry. Ionising radiation, because it achieves sterilisation without high temperature, high pressure or use of chemicals or gases, do not alter or damage the nutritional quality of food.

Recently White et al. (1994) evaluated the effects of gamma radiation on the structure of dentin and compared it to the effects caused by steam autoclave, ethylene oxide and dry heat. They concluded that gamma radiation sterilisation alters
the specular reflectance FTIR and optical properties of dentin less than the other procedures. A variation in dentin permeability values was detected after gamma radiation procedure. Nevertheless, the differences were similar to others observed in other studies and were attributed to the inherent variability normally associated with permeability. FTIR Spectral comparison revealed no alteration in both amine peak (associated with the collagen component) and phosphate peak (associated with the mineral component) for gamma radiated specimens. Changes in the collagen component were observed with the dry heat and steam autoclave methods. Changes in mineral contents were observed with steam autoclave, ethylene oxide and dry heat. The least change in the absorbance of dentin was associated with gamma radiation. For the other methods of sterilisation tested, the changes were considered significant.

White et al. (1994) investigated radiation dose and concluded that the total dose necessary to achieve sterilisation of teeth was 173 Krad. However, in their study each tooth was placed separately in one tissue culture well and only 24 teeth were placed in the radiation chamber at each time. The dose rates within the chamber at different positions ranged from 320 to 450 rads per minute demonstrating that a larger number of teeth may need a higher dose to achieve optimum sterilisation since the dose is not uniform inside the chamber.

In the current study, gamma radiation was the method chosen for sterilisation of the teeth. Each jar was sent for sterilisation with approximately 70 teeth and, therefore, we chose to use the recommended dose for sterilisation of medical products that is 2.5 Mrad (White et al., 1994; Woods & Pikaev, 1994).

Scientists have made a constant effort to build artificial oral environments in which dental materials or techniques can be tested. This is because clinical trials, which should be the final test for any new dental material or technique, are time-consuming and are usually fairly expensive to conduct. A great share of research done with restorative dental materials and duplication of masticatory forces and movements has been concerned with wear of materials. Experimental methods have varied from the very simple to the rather sophisticated, including mimicking of forces and motions occurring during mastication, and also effects of thermal fluctuations.
Recently there has been an attempt to generate forces and movements similar to those found during mastication with anatomically-corrected samples. In 1983, DeLong and Douglas reported the development of a system that would reproduce the movements and forces of mastication. According to them, to reproduce the masticatory system movements, the equipment should be able to generate straight-line movements in the horizontal plane and follow the anatomy of the test sample in the frontal plane. A force between 9 and 180 N with a duration of 1/4 to 1/3 of a second and a shape of half-sine wave-function, which approximate the force pattern of mastication (Ahlgren & Öwall, 1970), should be generated. To accomplish these requirements, it is necessary to have movement in the horizontal and frontal planes (vertical plane) which can be achieved by two integrated servo-hydraulic units. However, to produce such sophisticated apparatus can be a fairly expensive venture.

Alternatively, in 1993, Gundler, Lockowandt and Erhardson reported a study in which retention parameters such as convergence angle, roughness, and surface area of crown preparations were examined by cementing cast gold crowns to extracted human teeth. However, instead of using a tensile force to determine the retention of the crowns, the authors used compressive fatigue testing. Each cemented crown was subjected to a compressive loading oscillating between 50 N and 700 N with each loading cycle repeated twice a second (2 Hz) until the crown either loosened or a maximum of 1 million cycles was achieved. During this procedure only the vertical component of masticatory forces was simulated.

Also, in 1993, Kern, Douglas, Fechtig, Strub and DeLong tested the retention of all-porcelain resin-bonded prostheses bonded to natural anterior teeth in a study model. Before dislodging the prosthesis by applying compressive load to the lingual surface of the pontic, each prosthesis was subjected to a fatigue test. Using a stainless steel ball, an occlusal (compressive) load ranging between 22 N and 2 N was applied to the lingual surface of the pontic for a total of 1,250,000 cycles. The occlusal force profile was in the form of a half-sine wave.

In the current study, a combination of fatigue load-cycling (compressive) and tensile loading to failure was used for the purpose of practicality.
When conducting fatigue testing it is important to consider factors such as the forces generated during mastication and the duration of the forces. There is a substantial amount of disagreement in the literature concerning the magnitude of the forces generated during mastication. The reported values for chewing force of molar teeth vary between 55 N (Killiaridis et al., 1995) and 532 N (Helkimo & Ingervall, 1978), and for maximum biting force between 98 N (Helkimo et al., 1977) and 1308 N (Pruim, DeJongh & Ten Bosch, 1980). These differences are partly due to the different measuring techniques employed. Most of the studies used strain gauges attached to a metal-fork like device with different thickness, which could cause differences in the readings. Fields, Proffit, Case and Vig (1986) found greater vertical occlusal forces at 6.0 mm than at 2.5 mm opening during swallowing and chewing (but not for maximum biting) for normal children. However, no difference in chewing and maximum biting forces between 2.5 and 6.0 mm openings for adolescents and adults was found. Increased age for both men and women (Helkimo et al., 1977; Killiaridis et al., 1995), as well as patients with clinical symptoms of dysfunction (Helkimo & Ingervall) have been found to be significantly correlated with lower maximum biting forces. Higher maximum biting forces, in contrast, have been correlated with patients with tooth grinding or clenching habits (Helkimo & Ingervall) and also with practice, simulated in one study by asking the patient to chew paraffin for a certain amount of time for several days and then measuring the biting force at different points in time (Brekhus, Armstrong & Simon, 1941).

Variation in the chewing force values has also been associated with the type of food used (Howell & Brudevold, 1950; Craig, 1993). The maximum chewing force reported for celery, carrot, apple and most meat were 84.3, 75.5, 62.1, and 44.4 N, respectively.

Table 11 shows the results obtained from some selected studies for biting forces and chewing forces in the molar area.
Table 11. Mean and range of variation of biting forces and chewing forces for molar teeth in N.

<table>
<thead>
<tr>
<th>Reference</th>
<th>Biting Force</th>
<th></th>
<th>Chewing Force</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Mean</td>
<td>Range</td>
<td>Mean</td>
<td>Range</td>
</tr>
<tr>
<td>Howel &amp; Manly, 1948</td>
<td>564</td>
<td>405-880</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Helkimo et al., 1977</td>
<td>444</td>
<td>98-715</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Helkimo &amp; Ingervall, 1978</td>
<td>471</td>
<td>191-802</td>
<td>246</td>
<td>67-532</td>
</tr>
<tr>
<td>Pruim et al., 1980</td>
<td>965</td>
<td>609-1308</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Fields et al., 1986</td>
<td>349</td>
<td>-</td>
<td>160</td>
<td>-</td>
</tr>
<tr>
<td>Craig, 1993</td>
<td>565</td>
<td>390-800</td>
<td>129</td>
<td>-</td>
</tr>
<tr>
<td>Killiaridis et al., 1995</td>
<td>651</td>
<td>270-980</td>
<td>140</td>
<td>55-340</td>
</tr>
</tbody>
</table>

DeLong and Douglas (1983) suggested that chewing forces should be used when doing fatigue testing (9 to 180 N). However, a new material or technique must be able to withstand the highest forces found in the oral environment. The maximum biting force reported by Pruim et al. (1980) of 1308 N was found to be too high when compared to the other studies and, therefore, 800 N, as reported by Craig (1993) and Helkimo and Ingervall (1978) was chosen for the pilot study. However, after only 2,250 cycles the specimen with the proximal slot cavities only failed and the force was considered to be a little too high.

After reevaluation, it was decided to use a force closer to reported mean biting force, and to increase the number of fatigue cycles. Thus, the test was extended to 230,000 load cycles with the force oscillating between 10 and 450 N. Changes in the fatigue test were made because the specimen that failed was the one that was going to provide us with an indication of retention values from the added device in the experimental design for RBFPD, without interference from the wings and occlusal rests. In an effort to maintain this group in the study, the marginal enamel was reinforced by adding a bevel to the cavosurface margins of the buccal, lingual and gingival walls of the slots.
Another important aspect of the fatigue test is cycling time. It is desirable to complete the testing as rapidly as possible, and this, of course, can be accomplished by cycling at faster rates. It is estimated that the duration of force during chewing is between 0.25 and 0.33 seconds (DeLong & Douglas, 1983; Graf, 1969). Therefore, a maximum cycling rate of 3 - 4 Hz (3 to 4 cycles/second) can be used. However, it is important to remember that a reasonable estimate for the chewing rate is 80 strokes per minute (Bates, Stafford & Harrison, 1975), which translates into 1.3 stroke per second. Therefore, in practical terms, 4 strokes per second is an extremely fast rate for a person to sustain without muscular fatigue (Bates et al.).

The Instron machine that was available to carry out the fatigue test for the current study was capable of operating in load-control mode, generating a movement in the vertical plane, and generating a force curve having the shape of a positive half of a sine wave at a chewing rate of 4 Hz. However, because there was only one actuator, the horizontal component of the force could not be generated. A stylus with a round-end with 7.6 mm diameter was used to apply the uniaxial compressive load. This rested on the inclines of both labial and lingual cusps and above the occlusal central fossa of the metallic pontic.

To further approximate the experimental design to what actually happens in the clinical situation it would have been better to continue fatigue load cycling until the moment of failure, which would have been similar to what Gundler et al. (1993) did when they tested crown retention with different preparation designs. There were, however, two potential problems that could arise if this kind of testing was followed. First, the testing machine needs to be fed with specific parameters to recognise that failure has occurred and stop the test. Parameters such as that the test would terminate once 1 mm of deformation from the mean position occurs. However, one is only speculating that detachment of the prosthesis will cause 1mm deformation in the pontic area. It is possible that it may not, and in such case, the test will continue, despite of the failure. Second, dislodgement of the prosthesis under compressive fatigue loading may take a long time and can prove to be a very expensive way for testing. Gundler et al. (1993) in their study applied a maximum force of 700 N for a total of 1 million cycles with each loading cycle being repeated twice a second (2 Hz).
This corresponds to 5.8 days of testing for each specimen. Despite the long period of test, none of the 20 crowns made with 20 degrees convergence angle loosened, while 4 out of 20 crowns made with 40 degrees convergence angle did not fail. While one would agree that 40 degrees is a very exaggerated convergence angle and that the mechanical retention of this design would be expected to be fairly poor, some of the crowns made with this kind of preparation survived the 1 M cycles.

In the current study, although the prostheses were not made with the full crown design, the abutments received preparations with walls that were almost parallel. Therefore it was expected that a high number of cycles would be needed before failure, which would consume a lot of time and expense.

When testing a material or technique in an in vitro situation it is important to know how the results can be correlated to an in vivo situation. Correlation between fatigue testing and in vivo situation was reported in three experiments that compared the occlusal contact wear produced in the artificial environment with the corresponding wear found in clinical trials. The first of these experiments (DeLong, Sakagushi, Douglas & Pintado, 1985) used polished amalgam disks 12 mm in diameter and 3 mm thick occluding against a maxillary molar. Both elements were attached to a cycling machine and the following parameters were maintained: the lateral excursion at 0.82 mm; occlusal force at 13.35 N with a force profile in the form of half sine wave; time of cuspal contact 0.23 seconds; and a chewing rate at 4 cycles/s. The total number of masticatory cycles was 500,000 and occlusal mapping of both elements (maxillary and mandibular) took place at 0; 30,000; 100,000; 200,000 and 500,000 cycles. Occlusal wear was recorded in terms of depth and volume. The experimental wear was compared with the clinical data and it was found that by equating 250,000 cycles with one year of clinical use, the correlation coefficient between the in vitro wear and the in vivo wear was 0.94. The greatest difference between the two situations occurred at two years: at 500,000 cycles the wear was much less in vitro than in the in vivo situation.

The second study (Coffey, Gookind, DeLong & Douglas, 1985) examined in vitro wear of denture teeth made of acrylic resin occluding against denture teeth and against extracted maxillary molar tooth. The fatigue test was performed with a lateral
excursion of 0.5 mm; maximum occlusal force of 22 N, cuspal contact time of 0.23 s, a total number of chewing cycles of 300,000 and a waveform of the shape of a half-sine wave. The amount of wear found in this study after 300,000 chewing cycles collected during a 21-hour period appeared to correspond to approximately 18 months of clinical wear.

The methodology of the third study (Sakagushi, Douglas, DeLong & Pintado, 1986) was similar to the first one except that a posterior composite was used in place of the amalgam and the study ran for 300,000 cycles. The findings indicated that the correlation coefficient between the clinical data and the data from the artificial mouth was 0.84. Similar to the first study, 250,000 cycles were equated to one year of clinical use.

The fatigue testing used in the current study was done in a manner very similar to the ones described above except that no lateral excursion was performed and the maximum load was 450 N.

It can be concluded from the above fatigue-testing studies that 250,000 cycles with a maximum load of 13 N is equivalent to about one year of clinical use. If the load is increased to 22 N (40% higher), one year clinical use is obtained after only 200,000 cycles. Therefore, it is reasonable to state that the 230,000 cycles used in this study at a maximum load of 450 N would be indeed equivalent to more than one-year of clinical service. However, the precise number of years that this represents cannot be determined and further work will be necessary to establish this figure.

The results of the pilot study indicated that the force necessary to dislodge the prosthesis from the specimen prepared with design 2 was almost two times the force recorded with the specimen with the currently used design for RBFPD. This increase in resistance can be attributed to the increase in surface area for bonding, to the mechanical interlocking of the metal framework within the resin composite restorations, and perhaps to variations in the preparation such as inclination of the walls. Since effort was made to keep wall inclinations similar in all preparations, it is more likely that the higher resistance to dislodgement was associated with the increased surface area and the mechanical interlocking. Since one of the goals of this study was to examine the effect of the retentive device alone, the group with only the
proximal slot preparations was maintained in the main study. To determine the resistance to dislodgement achieved with the increase in surface area, a new group with approximately the same amount of prepared area, but with the framework filling the slot preparation as a metallic inlay, was added.

In the pilot study, the specimen prepared with the full coverage restoration had a separation force value similar to that of the specimen with the currently used design for RBFPD. However, failure was associated with horizontal fracture of the premolar tooth. The important question is: was the force recorded sufficient to dislodge the full-coverage prosthesis from the molar tooth and, as a result of torsion, it fractured the premolar at the cervical level? Or did the force first fractured the premolar tooth and then, as a result of torsional movement, separated the prosthesis from the molar? According to the appearance of the specimen after fracture, the first hypothesis is more likely to be true and this conclusion is based on two observations.

First, the molar tooth was intact after dislodgement of the prosthesis. This is more likely to occur if the prosthesis was removed almost parallel to the long axis of the preparation. In the case of torsion, oblique forces would be applied to the molar abutment in a direction other than parallel to its walls. In such situation the prosthesis could only be removed by either distortion of the framework or fracture of the tooth, which was not the case.

Second, the value obtained with Sp2 was substantially higher and, even though, it did not result in horizontal fracture of the abutment tooth. If 254.9 N was the force necessary to fracture the premolar tooth (under tension) at the cervical level, one would expect that below 502.9 N (value obtained with Sp2) Sp2 would also fail as a result of horizontal fracture of the premolar, which did not happen. Some variation should be allowed for the force, of course, as teeth were not exactly of the same size. This can be easily observed in Table 4, which shows the results of the main study. Fifteen premolar fractures were observed with forces varying from 312.6N to 678.4N. However, 86.6% (13) of the failures occurred with values above 404 N and 66.6% (10) occurred with values above 526.2 N. Also, 12 specimens had separation forces above 400 N and showed no horizontal fractures of the premolar, which led us to believe that the force necessary to break a premolar abutment tooth under tension
must be much above the 254.9N observed in Sp 3. Therefore, it is more likely that the fracture was caused by an oblique force resultant of the torsion of the prosthesis after detachment of the wing from the molar tooth.

Also, according to Shillinburg et al. (1997b) the diameter of the abutment tooth is related to the arc of displacement of the crown/retainer. Preparations on smaller teeth have a short rotational radius for the arc of displacement, and the occlusal portion of the axial wall resists displacement better. The longer rotational radius on larger preparations allows for a more gradual arc of displacement, and the axial wall does not resist displacement well. This theory, according to him, applies more to short crown preparations. In the current study, the preparations were not considered short although they were made in lower molar and premolar teeth. Lower teeth usually have shorter crowns than upper teeth. Therefore, this could explain, at least in part, why the retainer separated from the molar abutment before separation from the premolar abutment.

The second question that arises here is: if both current design for RBFPD and full coverage FPD had similar separation forces, why in a clinical situation the longevity of the second type of prostheses is much superior? It is known that to improve the longevity of RBFPDs it is necessary to use a circumferential reduction greater than 180 degrees in the abutment tooth (Crispin, 1991). This is done to dissipate the complex forces that act on the prosthesis, resulting in an improved stability in the facio-lingual direction. However, if 360 degrees of the tooth circumference is covered, it is expected that the prosthesis will be even more stable. Therefore, the answer for the question is that this study used only uniaxial load cycling and did not replicate the complex multidirectional occlusal forces that occur in the oral environment during mastication. This is the reason why this study is considered only a preliminary step for an appropriate clinical trial study. The inclusion of the full coverage FPD design in the main study was niled out, as it was not going to lead us to meaningful conclusions. Only designs that involved partial coverage of the abutments were included.

All materials in this study were used according to the manufacturer’s instructions except in two instances. The manufacturer of Panavia® 21 clearly states
that it is not necessary to etch prepared enamel surfaces prior to application of the primer and resin cement. However, Dixon and Breeding (1997) showed that failure to etch the enamel surface significantly reduces the shear bond strength of enamel specimens bonded to metal with Panavia® 21. Accordingly, in the current study, the prepared enamel surface was etched with the supplied etching gel (40% phosphoric acid) prior to application of the primer and bonding resin.

Another deviation from the manufacturer's instructions occurred with the use of the adhesive/luting agent system Bond-It!™ and Cement-It! C&B. The manufacturer stated that for short span FPDs a mix of primers A and B should be used to condition the metal surface. However, it was noticed that in the practical situation it was almost impossible to use the primer because as soon as the cement would come in contact with it, it caused snap setting and it was not possible to seat the prosthesis completely. Therefore, the primer had to be substituted with the MetalBond agent indicated for long span FPDs (4 or more units).

In the current study both Panavia® 21 cement and Cement-It! C&B were used. With regards to the time necessary to apply each of these two cementation systems, Panavia® 21 has an extra step that is the application of Oxyguard II at the margins of the restoration after cementation. This procedure is necessary since the material is an anaerobic self-cure cement that does not set in the presence of oxygen. The function of the Oxyguard II is to isolate the superficial margins of the cement from oxygen in the atmosphere to ensure optimum polymerisation. While many dentists consider this extra step a drawback of the system, it was felt in this study that it really helped the cementation of the prostheses because Panavia® 21 had an extended working time when compared to Cement-It! C&B. In addition, after seating the prosthesis, the margins can be cleaned while the cement is still unset before applying the Oxyguard II. With Cement-It! C&B, it is recommended by the manufacturer that finger pressure be applied to the seated restoration for five minutes and then removal of excess cement from the margins is started. At this point, the cement is already set and very hard to remove from the occlusal surface areas and margins of the prosthesis. Consequently, overhangs that form might be difficult to remove leading to deleterious effects on the restoration and periodontal tissues over time. In conclusion, although
Panavia® 21 has an extra step in the cementation procedure of prostheses, it was noticed that it was relatively easier to use and less time was spent after cementation to clean the hard cement from the margins of the restorations compared to when Cement-It! C&B was used.

Most of the adhesive failures in this study occurred at the metal/cement interface indicating that this was the weakest link in the system, particularly with Group 1. This observation is in agreement with others of several clinical trial studies (Hansson & Moberg, 1992; Rammelsberg et al. 1993; Boening, 1996; Hansson & Bergstrom, 1996) and of in vitro studies (Brady et al., 1985). It is not known if this is also true for Panavia® 21 (phosphonate resin) because the group in which this material was used had a very different failure pattern. However, it was noticed that the luting cement remained attached to the tooth surfaces that did not fracture indicating that if this cement was used in Group 1, the same type of failure would be expected to happen.

For Groups 2 and 3 most of the failures occurred within tooth structure. Six specimens from each group had horizontal fractures at the cervical level of the premolar tooth, which is a fairly consistent result. It is clear from these groups that the modification in the design led to an increase in resistance to dislodgement of the prostheses. The application of a continuous tensile force to the specimen until detachment results in failure at the weakest link of the system. When the modified design was used, the weakest point was the cervical area of the premolar tooth. As horizontal fracture occurred at the premolar, a torsional force was applied to the molar teeth, which resulted in fracture of the lingual cusps in the majority of the cases. If one would only take the teeth into consideration, it would be expected that the failure would occur in the premolar, which is the smaller of the two abutments, and typically has a very accentuated constriction at the cervical area. However, it was not expected that the prostheses made with the modified design would have a higher resistance to dislodgement than the cohesive strength of the premolar tooth. These results suggest that the maximum bond strength for this design is actually higher than the numbers recorded. In addition, no comparison can be made between Cement-It! C&B and Panavia® 21 used with the same design since the force recorded is only the force.
necessary to break the premolar abutment under tension and not the force necessary
to dislodge the prosthesis.

In group 4, a less severe fracture pattern was observed. Horizontal fracture of
the premolar occurred with only 3 specimens. However, the fact that horizontal
fractures did occur with this group is yet another indication that the higher resistance
to dislodge the prosthesis with the modified design is more related to the added
retention device rather than to the increased surface area for bonding.

This can be easily concluded if one was to consider the mean separation
forces and compare them with the total preparation areas of the groups. Groups 2, 3
and 5 had very similar preparation areas (185.1, 188.08 and 181.8 mm², respectively)
and yet the mean separation forces were significantly different between the first two
groups and the third (525.4, 562.2 and 417.1 N, respectively). Another indication that
most of the retention of the modified design is attributed to the mechanical interlocking
of the prosthesis within the resin composite restoration is the result obtained with
Group 4. This group had the smallest mean preparation area (77.41 mm²) and yet
had a high mean separation force (448.6 N), similar to the mean separation force
achieved with Group 2, which had a mean surface area of 185.1 mm².

To further illustrate the significance of the retention device on separation force,
when the separation force is divided by the area of the preparation, the following
mean force/area for each group is obtained:

Table 12. Mean force per area for each group in N/mm².

<table>
<thead>
<tr>
<th>GROUPS</th>
<th>FORCE PER AREA</th>
</tr>
</thead>
<tbody>
<tr>
<td>Group 1</td>
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</tr>
<tr>
<td>Group 2</td>
<td>2.84</td>
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<tr>
<td>Group 3</td>
<td>2.99</td>
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<tr>
<td>Group 4</td>
<td>5.80</td>
</tr>
<tr>
<td>Group 5</td>
<td>2.30</td>
</tr>
</tbody>
</table>
It is now very easy to see how high the value is for group 4. Groups 1 and 5 had very similar values while group 2 and 3 had higher values. It is interesting to note that whether mean separation force or force per area is considered, Group 1, which had the currently used design for RBFPDs, had the lowest values.

While *in vitro* testing attempts to simplify *in vivo* events, the practical interpretation of the results of this study, which is intricate, has to be made with caution. The prosthesis design that presented the best results in this *in vitro* study may not, necessarily, present the best performance in the clinical situation. Failure of prosthesis in the clinical situation occurs under a combined effect of a number of challenging variables encountered in the oral environment. Other factors that were not simulated in this *in vitro* study such as thermal fluctuations, acidic conditions of the mouth, ageing of the restoration, and lateral forces not represented here, may play an important role on determination of longevity of these prostheses.

The importance of this study, however, lies in the demonstration that the modification in design offers a great potential in increasing the resistance to dislodgement of RBFPDs. The higher resistance encountered with removal of the prostheses indicates that it may be prudent to try to use this modification in a clinical study setting.
Conclusions

Within the limits of the findings of this investigation it may be concluded that:

1. The RBFPD with the modified design had significantly more potential for better retention than the one with the currently used design as indicated by a higher magnitude of tensile force needed for its separation associated with more severe tooth fractures on separation.

2. Increased resistance of the RBFPD with the modified design to dislodgement does not appear to be directly related to the surface area of the fitted surface.

3. The increase in resistance to dislodgement of the modified design may be attributed primarily to the mechanical interlocking of the retentive device (the frame metal projection) within the composite resin restorations.

4. The type of resin cement used did not appear to influence on the separation force of the RBFPDs made with the modified design.

5. These observations appear to underscore the merit of the new design. A clinical trial study is needed to follow up on these findings.
References


Appendices

Appendix 1. Radiopacity of Dual-Cure and Chemical-Cure Resin-Based Cements.

MHM Rubo & OM El-Mowafy

- Presented at the Research Day at the Faculty of Dentistry, University of Toronto – February, 1997
- Presented at the IADR/AADR Meeting in Orlando, Florida – March 1997.
- Published in The International Journal of Prosthodontics, volume 11, Number 1, pages 70-74, 1998.

Abstract

Purpose: The objective of this study was to determine radiopacity values of a group of resin-based cements and to compare them to those of enamel and dentin.

Materials and methods: Three specimens 2.5 mm thick were prepared from each of 18 cements (nine chemical-cured and nine dual-cured). Three tooth sections 2.5 mm thick were cut from extracted molars. Specimens and tooth sections were divided into three matching groups. Following standard radiographic techniques, images of each group of specimens, along with an aluminum step wedge were obtained. Optical density readings for each material were determined with a transmission densitometer. Radiopacity values were subsequently calculated as equivalents of aluminum thickness. Results: Analysis of variance indicated significant differences in radiopacity values among the materials (p < 0.0001). Four dual-cured cements (Variolink, Geristore, Enforce and Nexus) had radiopacity values significantly greater than that of enamel, while two (Choice and Adherence) had radiopacity values similar to enamel. For the remaining three dual-cured cements, Duolink had a radiopacity value significantly lower than that of enamel but higher than that of dentin, while both Lute-lt and Resinomer had values similar to dentin. For the chemical-cured cements, five materials had radiopacity values significantly higher than enamel (Sealbond, Advance, Scotchbond, Cement-lt, and Dyract-Cem), while two had values significantly lower than enamel but higher than dentin (Biomer and Panavia 21). The remaining
two cements had radiopacity values significantly lower than dentin. **Conclusion:** For a group of 18 resin-based cements, some had radiopacity values similar to or significantly lower than that of dentin.

**Introduction**

The applications of resin-based cements have increased considerably over the last few years for a number of reasons. With increasing public concern about mercury in amalgam restorations, patients are demanding that their dentists use alternative restorative materials, particularly aesthetic ones. These include ceramic inlays and onlays (such as IPS Impress, Ivoclar/Vivadent) as well as indirect restorations made from resin composites (such as Concept, Ivoclar/Vivadent). Restorations made of ceramic or resin composite materials are typically cemented using dual-cure resin cements. It is essential that such resin cements be sufficiently radiopaque to permit detection of marginal overhangs, open gingival margins, as well as recurrent caries in the gingival areas.\(^1\)\(^-3\) In addition to their use in the cementation of non-metallic inlays and onlays, dual-cured cements are also used for cementation of all-ceramic crowns and fixed partial dentures. Chemical-cured resin cements are currently used for a number of clinical applications including cementation of posts, particularly the prefabricated posts, due to their superior retentive properties when used in conjunction with a dentin bonding agent.\(^4\) Chemical-cured resin cements are the cement of choice for cementation of the new nonmetallic carbon-fiber posts (such as the Composipost, RDT). Because these carbon-fiber posts are radiolucent, it is imperative that the cement used for their cementation be sufficiently radiopaque to afford verification of complete seating of the post(s) and their presence in the root canal should the need for retreatment arise. Chemical-cured resin cements are also used for cementation of metallic crowns and fixed partial dentures as well as gold inlay/onlay restorations. A recent study indicated the superiority of chemically cured cements to nonpolymeric cement when used in conjunction with dentin bonding in providing supplementary retention for crowns with less than ideal preparation design.\(^5\)

Resin-modified glass-ionomer cements are also used for the same purposes, and have the additional advantage of improved resistance to recurrent caries through
their ability to release fluoride. For all these cements, it is also important to have sufficient radiopacity to permit detection of cement marginal overhangs in interproximal areas.

In a previous study, the radiopacity of some of a group of resin-based inlay cements was found to be less than that of enamel. It is desirable that resin cements have radiopacity values similar to or higher than that of enamel. The purpose of this study was to determine radiopacity values of a group of resin-based cements, including new dual-cured and chemical-cured materials, and to compare the radiopacity values to those of enamel and dentin.

**Materials and methods**

Specimens measuring 7.5 x 7.5 mm with a thickness of 2.5 mm were made in silicone molds from a group of nine dual-cure resin-based cements. One of these was a polyacid-modified material (Table 1). Specimens were similarly made from a group of nine chemical-cure resin-based cements, two of which were polyacid-modified materials (Table 2). Three specimens of each material were made. Three longitudinal sections 2.5 mm thick were cut from recently extracted permanent molar teeth using a microslicing machine (Accutom, Struers). The specimens and the tooth sections were randomly divided into three equal groups, each group consisting of one specimen of each material and one tooth section. Each group of specimens was placed on a Kodak occlusal x-ray film, Ultraspeed D (Eastman Kodak) together with an aluminum step (Dent-X) with a thickness increasing by 1 mm for each step to a maximum of 13 mm. This was used to correlate the optical density of the images of the specimens to that of aluminum. The films were exposed using a dental x-ray machine (model 46-181121G1, General Electric) at 65 KV with a focus-object distance of 40 cm and 48 impulses. The films were developed following standard techniques in an automatic x-ray processor (Dent-X 9000, Dent-X) at a temperature of 27.5°C ± 1°C using a 4.5-minute cycle. A transmission densitometer (Macbeth TD-504) was used to record the readings of the radiographic images of the specimens. A minimum of four optical density readings was obtained from each image of each material, from the images of the enamel-dentin specimen and those of each step of the step wedge. The data
obtained from the step-wedge images were used to draw a curve for aluminum thickness equivalency for optical density readings. This curve was used to calculate aluminum thickness equivalency values for the mean optical density readings of each cement specimen as well as those of enamel and dentin. Data of each group of cements were analysed statistically using one-way analysis of variance (ANOVA) and Tukey’s test.

To illustrate the clinical significance of the findings of this study, three extracted permanent molar teeth were prepared with MOD inlay cavities. Inlay restorations were made for these teeth in a computer-aided design-computer-aided machining (CAD-CAM) inlay machine (Cerec, Model 2996510 D 3255, Siemens) from Cerec Vita Blocks (Vita Zahnfabrik). One inlay restoration was cemented with one of three dual-cured cements (Enforce, Adherence and Resinomer) following standard bonding and cementation techniques. Following finish and polish, a radiographic was made of these three teeth. Three extracted permanent incisor teeth were prepared to receive carbon fiber posts (Composipost). Each of the three Composiposts was cemented with one of three chemical-cured cements (Advance, Panavia, or C & B Metabond). Following cementation of the posts, a radiographic was made of the three teeth.
Table 1. Dual-Cured Cements Examined in this Study.

<table>
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<tr>
<th>Brand Name</th>
<th>Manufacturer</th>
<th>Shade</th>
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<th>Material batch nos.</th>
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All materials are resin cements except for Resinomer, which is classified as a polyacid-modified resin material.
Table 2. Chemical-Cured Cements Examined in this Study.

<table>
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<th>Brand Name</th>
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<th>Material batch nos.</th>
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<td>Not stated</td>
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<tr>
<td>Biomer</td>
<td>Dentsply/Caulk</td>
<td>One provided</td>
<td>950420</td>
<td>Not stated</td>
</tr>
<tr>
<td>Cement-It</td>
<td>Jeneric/Pentron</td>
<td>One provided</td>
<td>100996</td>
<td>Base:690972 Catalyst:690952</td>
</tr>
<tr>
<td>Dyract-Cem</td>
<td>Dentsply/Caulk</td>
<td>Opaque</td>
<td>960126</td>
<td>Powder:9511132</td>
</tr>
<tr>
<td>C&amp;B Metabond</td>
<td>Parkell</td>
<td>Tooth shade</td>
<td>203011</td>
<td>Powder:20102 Catalyst:203011 Base:20303</td>
</tr>
<tr>
<td>Adhesive Luting Cement</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Panavia 21</td>
<td>Kuraray</td>
<td>One provided</td>
<td>61162</td>
<td>base:0236 catalyst:0229</td>
</tr>
<tr>
<td>Scothbond Resin Cement</td>
<td>3M Dental Products</td>
<td>One provided</td>
<td>141295</td>
<td>Not stated</td>
</tr>
<tr>
<td>Sealbond Cement</td>
<td>RTD</td>
<td>One provided</td>
<td>A993</td>
<td>Not stated</td>
</tr>
<tr>
<td>Universal Post</td>
<td>Brasseler</td>
<td>One provided</td>
<td>Not stated</td>
<td>Part B:RA062</td>
</tr>
</tbody>
</table>

All materials are resin cements except Advance and Dyract-cem, which are classified as a polyacid-modified resin material.

Results

Table 3 gives means and standard deviations of optical density values expressed as equivalents of aluminum thickness for the dual-cured cement group. Significant differences in these mean values were found (p < 0.0001). Variolink, Geristore, Enforce and Nexus had the highest radiopacity values. While Choice and Adherence had radiopacity values that were lower, they were not significantly different from the radiopacity value of enamel. Duolink had a radiopacity value significantly lower than that of enamel but significantly higher than that of dentin. Lute-It and Resinomer had radiopacity values not significantly different from that of dentin.

Table 4 gives means and standard deviations of optical density values expressed as equivalents of aluminum thickness for the group of chemical-cured cements. Significant differences in these mean values were found (p < 0.0001). Sealbond, Advance, Scotchbond, Cement-It, and Dyract-Cem had radiopacity values significantly higher than that of enamel. Biomer and Panavia 21 had radiopacity values significantly higher than that of dentin but significantly lower than that of
enamel, while C & B Metabond and Universal Post Cement had radiopacity values significantly lower than that of dentin.

Table 3: Radiopacity of Dual-Cured Cements as well as Enamel and Dentin Expressed as Equivalents of Aluminum Thickness (mm Al/mm of material).

<table>
<thead>
<tr>
<th>Material</th>
<th>Mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Variolink</td>
<td>3.42</td>
<td>0.16</td>
</tr>
<tr>
<td>Geristore</td>
<td>2.62</td>
<td>0.11</td>
</tr>
<tr>
<td>Enforce</td>
<td>1.95</td>
<td>0.12</td>
</tr>
<tr>
<td>Nexus</td>
<td>1.69</td>
<td>0.11</td>
</tr>
<tr>
<td>Choice</td>
<td>1.53</td>
<td>0.21</td>
</tr>
<tr>
<td>Adherence M5</td>
<td>1.49</td>
<td>0.09</td>
</tr>
<tr>
<td>Enamel</td>
<td>1.47</td>
<td>0.08</td>
</tr>
<tr>
<td>Duolink</td>
<td>1.25</td>
<td>0.05</td>
</tr>
<tr>
<td>Lute-It</td>
<td>0.94</td>
<td>0.04</td>
</tr>
<tr>
<td>Resinomer</td>
<td>0.87</td>
<td>0.08</td>
</tr>
<tr>
<td>Dentin</td>
<td>0.81</td>
<td>0.08</td>
</tr>
</tbody>
</table>

Vertical bars indicated that values are not statistically different (p>0.05).
Table 4: Radiopacity of Chemical-Cure Cements as well as Enamel and Dentin Expressed as Equivalents of Aluminum Thickness (mm Al/mm of material).

<table>
<thead>
<tr>
<th>Material</th>
<th>Mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sealbond</td>
<td>2.62</td>
<td>0.09</td>
</tr>
<tr>
<td>Advance</td>
<td>2.35</td>
<td>0.14</td>
</tr>
<tr>
<td>Scotchbond</td>
<td>2.18</td>
<td>0.08</td>
</tr>
<tr>
<td>Cement-It</td>
<td>2.03</td>
<td>0.06</td>
</tr>
<tr>
<td>Dyract-Cem</td>
<td>1.88</td>
<td>0.05</td>
</tr>
<tr>
<td>Enamel</td>
<td>1.46</td>
<td>0.05</td>
</tr>
<tr>
<td>Biomer</td>
<td>1.29</td>
<td>0.05</td>
</tr>
<tr>
<td>Panavia 21</td>
<td>1.29</td>
<td>0.18</td>
</tr>
<tr>
<td>Dentin</td>
<td>0.82</td>
<td>0.05</td>
</tr>
<tr>
<td>C&amp;B Metabond</td>
<td>0.29</td>
<td>0.03</td>
</tr>
<tr>
<td>Universal Post</td>
<td>0.10</td>
<td>0.02</td>
</tr>
</tbody>
</table>

Vertical bars indicated that values are not statistically different (p>0.05).
Figure 1: Radiographic images of three molar teeth restored with MOD Cerec inlays. Each inlay restoration was cemented with different cements as indicated. Cement lines of both Enforce and Adherence can be clearly seen, while it is difficult to detect in the other cement (Resinomer), which has a radiopacity value similar to that of dentin.

Figure 2: Radiographic images of three incisor teeth that were restored with Composi posts. Each post was cemented with different cements as indicated. The outline of the post is visible in the teeth in which Advance or Panavia 21 were used, while the post cemented with C & B Metabond, which is less radiopaque than dentin, is not visible.
Discussion

The method of determining radiopacity of the resin cements used in this study had been previously used to determine radiopacity values for resin composite restoratives. The radiopacity values reported in this study were 1.5 mm Al/1.0 and 0.8 mm Al/1.0 for enamel and dentin, respectively. While slightly lower than the values reported by Abou-Tabl et al. (1.6 mm Al/1.0 mm enamel and 1.0 mm Al/1.0 mm dentin) and by others, these findings are nevertheless comparable. Variability in radiopacity values reported in different studies can be attributed to many factors, as discussed elsewhere.

Resin cements are typically composed of inorganic fillers dispersed in a resin matrix. While resin matrices such as bisphenol glycidyl methacrylate (Bis-GMA) or urethane dimethacrylate contribute little to the radiopacity of the material, it is typically the inorganic filler component that contributes mostly to the radiopacity of resin cements. However, not all inorganic fillers are radiopaque. While barium, strontium and zirconium fillers are radiopaque, quartz and silica fillers are not. Manufacturers add variable amounts of radiopaque fillers to enhance the radiopacity of resin composite restoratives. Resin cements are treated in a similar manner, and, depending on the amount of radiopaque fillers added, the radiopacity of the cement will vary.

Figure 1 shows radiographic images of three permanent molar teeth that were restored with MOD Cerec inlays luted with three cements with different radiopacity values. As can be seen, inlays that were cemented using Enforce or Adherence, both cements with radiopacity values similar to or higher than that of enamel, demonstrated clearly visible cement films. For the inlay cemented with Resinomer, which had a radiopacity value similar to that of dentin, the cement line is difficult to detect. Figure 2 similarly shows radiographic images of anterior permanent teeth that had received Composiposts. For those cemented with Advance and Panavia 21, the outline as well as serrations of the cemented posts can be clearly seen, while it is difficult to confirm the presence of the radiolucent post in the canal for the post cemented with C & B Metabond. While Advance has a radiopacity value higher than that of enamel, Panavia 21 has a radiopacity value higher than that of dentin but lower
than that of enamel. In contrast, C & B Metabond had a radiopacity value significantly lower than that of dentin.

In a previous study in which the radiographic images of different resin composite restoratives were compared with enamel and dentin, no consensus on a minimal degree of radiopacity for contrast was reached. All 11 examined materials had radiopacity values higher than that of dentin but not necessarily higher than that of enamel. In the present study, it was difficult to detect the cement line on the radiograph when a cement material with a radiopacity value not significantly different from that of dentin was used to cement a Cerec inlay, which is not a highly radiopaque restoration. For this reason it is important that resin cements have radiopacity values higher than that of dentin and perhaps similar to or higher than that of enamel.

In a recent study in which the radiopacities of three resin-modified glass-ionomer cement liners and bases were determined and compared with those of nonpolymeric cements, the authors concluded that resin-modified glass-ionomer cements may be insufficiently radiopaque. In contrast, in the present study, of the three such cements examined, two (Geristore and Dyract-Cem) were found to have radiopacity values higher than that of enamel, which is considered to be sufficiently radiopaque for the various cement applications.

Conclusions
1. The radiopacity of a group of nine dual-cure resin-based cements was determined, and seven were found to have radiopacity values significantly higher than that of dentin.

2. The radiopacity of a group of nine chemical-cure resin-based cements was determined, and seven were also found to have radiopacity values significantly higher than that of dentin.

3. For optimum radiographic image contrast, the radiopacity of resin cements should be higher than that of dentin and perhaps ideally similar to or higher than that of enamel.
Acknowledgement
The authors would like to express their appreciation to the dental manufacturers (Bisco, Dentsply/Caulk, Jeneric/Pentron, Confi-Dental Products, Kerr, Kuraray, RTD, 3M, and Vivadent) that donated sample cement material packages.

References
Appendix 2. Hardening of New Resin Cements Cured through Ceramic Inlay.

El-Mowafy OM, Rubo MHM, El-Badrawy WA

-Presented at the IADR/AADR Meeting in Orlando, Florida – March 1997.
-Accepted for publication in Operative Dentistry in November 4, 1997.

Abstract

This study investigated the degree of hardening achieved through self-curing only and through dual curing of a group of eight new resin-based cements. In addition, the effect of ceramic inlay thickness on cement hardness was determined. Eight disc specimens measuring 6 mm in diameter and 2.5 mm thick were prepared from eight cements: Adherence, Choice, Duolink, Enforce, Lute-lt, Nexus, Resinomer and Variolink. Half were self-cured while the remainders were dual-cured. Knoop hardness measurements were then made at one-hour, one-day and one-week intervals. In addition 12 specimens of the same dimensions were prepared from each cement and were dual-cured through ceramic spacers of varying thickness (1 - 6 mm). Hardness measurements were made as above. ANOVA showed significant differences in hardness of self-cured versus dual-cured specimens for all cements (p < 0.0001). Significant differences were also found in the hardness of specimens dual-cured through ceramic spacers 2-3 mm in thickness or more compared with those which were dual-cured without spacer. It is concluded that for some materials self-curing alone was not sufficient to achieve sufficient hardening; and cement hardness was significantly reduced when ceramic inlay thickness was 2-3 mm or more.

Clinical significance

Among a group of eight dual-cure resin cements, some had hardness values obtained through self-curing less than 50% of those obtained through dual curing. Ceramic inlay thickness of more than 2 mm inhibited maximum hardening of a number of the cements.
Introduction

Applications of resin cements have increased considerably over the last few years. They are the materials of choice for cementation of ceramic inlay and onlay restorations, indirect resin composite restorations, ceramic crowns as well as porcelain veneers. They can also be used for cementation of posts; both cast and prefabricated, as well as cast metal restorations and crown and bridgework. According to their method of polymerisation, resin cements can be classified into three main categories: self-cured cements, used mostly for cementation of metallic restorations and posts; light-cured cements, used for cementation of porcelain veneers; and dual-cured cements used for cementation of ceramic inlay and onlay restorations, indirect resin composite restorations and ceramic crowns.

With increasing public concern about mercury-containing amalgam restorations, dentists are now using alternative restorative materials particularly ones that are aesthetic. These include ceramic and resin composite inlays and onlays. Dual-cure cements, which are typically used for cementation of such restorations, polymerise chemically upon mixing of base and catalyst components (self-curing) and also when subjected to light from a light-curing unit. During cementation of an inlay restoration, the peripheral parts of the cement interface, mainly at the occlusal aspect, are about the only parts that can benefit largely from both self- and light curing, as they are readily accessible to the curing light. Polymerisation of the more remote parts of the cement, for example at the gingival floor of the cavity, relies more extensively on the self-curing component of the polymerisation system. Studies have indicated that self-curing alone was insufficient for dual-cure cements to achieve maximum hardening (Darr & Jacobsen, 1995; El-Badrawy & El-Mowafy, 1995; Linden & others, 1991). In one recent study which investigated seven different dual-cured cements (El-Badrawy & El-Mowafy, 1995) one of the cements completely failed to harden when self-curing only was used. Other studies reported adverse effect of increasing ceramic inlay thickness on hardening of light-cured and dual-cured resin cements (Blackman, Barghi & Duke, 1990; Chan & Boyer, 1991; El-Badrawy & El-Mowafy, 1995; Uctasli, Hasanreisoglu & Wilson, 1994). Furthermore, Hasegawa, Boyer and Chan (1991), who studied the hardening of three dual-cured cements under resin composite inlays,
reported that self-curing alone did not completely harden the cements when light was attenuated by tooth and restoration material. Thus, it is important for newly manufactured dual-cured cements to be formulated in such a way so that they are capable of achieving a sufficient degree of hardening with and without light curing.

This study evaluated the hardening of a group of eight new dual-cure inlay resin cements when self-cured only and when dual-cured. Also, the effect of ceramic inlay thickness on the hardening of the cements was investigated.

Methods and materials

Eight dual-cured resin-based cements were examined in this study (Table 1) one of which is a resin/ionomer cement (Resinomer). Following manufacturers’ instructions for proportioning and mixing, four disc specimens measuring 2.5 mm in thickness and 6 mm in diameter were prepared from each cement using steel rings. When a selection of cement shades was available, middle range shades were selected (Table 1). Each ring was placed on a glass plate lined with a Mylar strip, filled with the mixed cement and covered with another Mylar strip-lined glass plate. The two glass plates were pressed together with a clamp and were maintained in darkness in a box at 37°C without light curing. Another set of four specimens was prepared from each cement in the same manner as above, however, these were subjected to light from a light-curing unit for 60 seconds from one surface only. Prepared specimens were stored at 37°C until testing. Using a Tukon 300 microhardness tester (Acco Industries Inc., Wilson Instrument Division, Bridgeport - Connecticut 06602) with a Knoop indenter and 30 g weight, the surface microhardness of each specimen was determined at 1 hour, 1 day and 1 week. Five readings were obtained from each specimen at each test interval. Mean Knoop hardness numbers (KHNs) were calculated for each material at the three test intervals. Data were analysed statistically using three-way ANOVA and Duncan’s tests.

The effect of inlay thickness on the degree of cement hardening was assessed in a second part of this investigation. This procedure was carried out by applying the light through ceramic spacers of varying thickness (1 through 6 mm) during specimen
preparation. A set of six ceramic spacers was cut from Cerec Vita Blocks (Vita Zahnfabrik, Bad Sackingen, Germany) shade A2C. These were used during the preparation of twelve specimens from each cement material in the manner described above, however, prior to light application a ceramic spacer was placed between the top surface of the specimen and the curing light source. For each cement material two specimens were prepared using one spacer thickness at a time. Specimens were stored at 37 °C until testing. Knoop hardness measurements were made in the same manner as above. Mean KHNs were calculated and data analysed statistically using three-way ANOVA and Duncan’s tests.

Using a light radiometer (Cure Rite, model # 8000, EFOS Inc., Williamsville, New York 14221), the curing light intensity was measured directly and through the six ceramic spacers in order to determine the degree of light attenuation as it passed through the different spacers.

Table 1. The eight dual-cure resin-based cements examined in this study.

<table>
<thead>
<tr>
<th>Material</th>
<th>Manufacturer</th>
<th>Shade(s) used</th>
</tr>
</thead>
<tbody>
<tr>
<td>Adherence M5</td>
<td>Confidental Products CO., Louisville, CO 8007</td>
<td>Light yellow / light gray</td>
</tr>
<tr>
<td>Choice</td>
<td>Bisco, Inc., Itasca, IL 60143</td>
<td>A1 / B1</td>
</tr>
<tr>
<td>Duolink</td>
<td>Bisco, Inc., Itasca, IL 60143</td>
<td>One shade provided</td>
</tr>
<tr>
<td>Enforce</td>
<td>Dentsply/Caulk, Milford, DE 19963-0359</td>
<td>A2 / C2</td>
</tr>
<tr>
<td>Lute-It</td>
<td>Jeneric/Pentron Inc., Wallingford, CT 06492</td>
<td>Light / dark</td>
</tr>
<tr>
<td>Nexus</td>
<td>Kerr U.S.A., Orange, CA 92667</td>
<td>Neutral / dark</td>
</tr>
<tr>
<td>Resinomer</td>
<td>Bisco, Inc., Itasca, IL 60143</td>
<td>One shade provided</td>
</tr>
<tr>
<td>Variolink</td>
<td>Vivadent, FL-9494 Schaan, Liechtenstein</td>
<td>Yellow / brown</td>
</tr>
</tbody>
</table>
Results

Tables 2 and 3 record mean KHNs and standard deviation values of self-cured and dual-cured specimens of all examined cements at the three test intervals. Generally, KHNs were lower when self-curing alone was used compared to dual curing. At 1 hour, Variolink had a mean KHN obtained from self-cured specimens less than 25% of that obtained from dual-cured ones; while for Adherence, Duolink and Resinomer, mean KHNs obtained from specimens self-cured were less than 50% of those obtained from dual-cured specimens. All cements, however, showed increases in hardness with time for both curing methods. For one cement (Resinomer), the mean KHN obtained from self-cured specimens increased to 68% of the value obtained from dual-cured specimens at 1 day and to 72% at 1 week. ANOVA revealed significant differences in KHNs among the cements (p < 0.0001) and also significant difference in KHNs between self- and dual-curing (p < 0.0001). At the three test intervals, Lute-It and Duolink had the significantly highest KHNs when the specimens were dual-cured (Table 3), while Enforce had the significantly highest KHN when the specimens were only self-cured (Table 2). In contrast, Resinomer had the significantly lowest KHN (33.1) when the specimens were dual-cured 1 hour and Resinomer and Adherence had the significantly lowest KHNs when the specimens were dual-cured at 1 day and 1 week. Variolink had the significantly lowest KHNs when the specimens were only self-cured at the three test intervals. For four cements there was no significant difference in KHNs obtained at 1 day and 1 week (Adherence, Duolink, Enforce and Nexus).
Table 2: Means and standard deviations, in parentheses, of KHNs obtained from self-cured specimens of the examined cements at the three test intervals. Vertical bars indicate mean KHNs that were not significantly different (p>0.05).

<table>
<thead>
<tr>
<th>Material</th>
<th>One-hour</th>
<th>One-day</th>
<th>One-week</th>
</tr>
</thead>
<tbody>
<tr>
<td>Enforce</td>
<td>32.1 (1.3)</td>
<td>42 (1.8)</td>
<td>42.9 (2.9)</td>
</tr>
<tr>
<td>Nexus</td>
<td>29.2 (1.9)</td>
<td>38.5 (1.8)</td>
<td>38.5 (1.3)</td>
</tr>
<tr>
<td>Choice</td>
<td>26.1 (2.2)</td>
<td>39.2 (3.8)</td>
<td>39.7 (2.8)</td>
</tr>
<tr>
<td>Lute-Lt</td>
<td>25.6 (4.6)</td>
<td>32.4 (6.4)</td>
<td>34.6 (6.4)</td>
</tr>
<tr>
<td>Duolink</td>
<td>15.9 (1.5)</td>
<td>20.7 (1.8)</td>
<td>21.2 (2.3)</td>
</tr>
<tr>
<td>Resinomer</td>
<td>15.6 (1.2)</td>
<td>30 (2.1)</td>
<td>32.3 (2.5)</td>
</tr>
<tr>
<td>Adherence M5</td>
<td>11.5 (3.4)</td>
<td>16 (4)</td>
<td>16.2 (4.7)</td>
</tr>
<tr>
<td>Variolink</td>
<td>7.7 (1.1)</td>
<td>9.3 (0.8)</td>
<td>10 (1.7)</td>
</tr>
</tbody>
</table>

Table 3: Means and standard deviations, in parentheses, of KHNs obtained from dual-cured specimens of the examined cements at the three test intervals. Vertical bars indicate mean KHNs that were not significantly different (p>0.05).

<table>
<thead>
<tr>
<th>Material</th>
<th>One-hour</th>
<th>One-day</th>
<th>One-week</th>
</tr>
</thead>
<tbody>
<tr>
<td>Duolink</td>
<td>47.4 (2.2)</td>
<td>55.9 (2.6)</td>
<td>57.1 (3)</td>
</tr>
<tr>
<td>Lute-Lt</td>
<td>47.4 (2.2)</td>
<td>57 (1.9)</td>
<td>57.5 (2.6)</td>
</tr>
<tr>
<td>Nexus</td>
<td>45.4 (1.9)</td>
<td>50.8 (2.1)</td>
<td>52.1 (1.8)</td>
</tr>
<tr>
<td>Enforce</td>
<td>44.1 (2)</td>
<td>51.4 (2.5)</td>
<td>52.1 (2.4)</td>
</tr>
<tr>
<td>Variolink</td>
<td>44.5 (2.6)</td>
<td>50.1 (2.4)</td>
<td>53.8 (2.4)</td>
</tr>
<tr>
<td>Resinomer</td>
<td>33.1 (1.7)</td>
<td>44.3 (1.1)</td>
<td>44.6 (1.8)</td>
</tr>
<tr>
<td>Adherence M5</td>
<td>36.5 (1.6)</td>
<td>44.1 (2.2)</td>
<td>44.8 (2.4)</td>
</tr>
<tr>
<td>Choice</td>
<td>36.2 (1.9)</td>
<td>45.8 (2.6)</td>
<td>48.4 (3.8)</td>
</tr>
</tbody>
</table>
In the second part of this study, when specimens were cured through ceramic spacers, there was a tendency for hardness to decrease gradually with increasing thickness of the spacer. However, the degree of decrease varied among the eight cements (Figures 1-8). ANOVA revealed significant difference in KHNs among the materials ($p < 0.0001$) and between different spacer thickness ($p < 0.0001$). For Adherence, decreases in KHN ranging from 69% to 74% occurred when spacer thickness was increased from 1 mm to 6 mm at the three test intervals (Figure 1). Mean KHNs for Adherence with the six spacers were significantly different at the three test intervals with the exception of KHNs with 5 and 6 mm spacers, while for Choice, decreases in KHN ranging from only 16 to 29 % occurred when spacer thickness increased from 1 to 6 mm at the three test intervals (Figure 2). Significant decreases in KHNs of Choice occurred when the spacer thickness was more than 2 mm for the three test intervals. For Duolink, KHN decreases ranging from 54% to 59% occurred when spacer thickness increased from 1 to 6 mm at the three test intervals (Figure 3). Significant decreases in KHNs of Duolink occurred when the spacer thickness was 2 mm or more for the three test intervals. In contrast, Enforce had decreases in KHNs ranging from only 9% to 14% when the spacer thickness increased from 1 to 6 mm at the three test intervals (Figure 4). Enforce’s KHNs decreased significantly when the spacer thickness was 3 mm or more at 1 day and 1 week. For Lute-lt, decreases in KHNs ranging from 66% to 69% occurred when spacer thickness increased from 1 to 6 mm at the three test intervals (Figure 5). All KHNs obtained for this cement were significantly different at the three test intervals. In contrast, Nexus had decreases in KHNs ranging from only 18% to 29% when the spacer thickness increased from 1 to 6 mm at the three test intervals (Figure 6). KHNs for Nexus decreased significantly when the spacer thickness was 3 mm or more. For Resinomer, mean KHNs decreased when the spacer thickness increased from 1 to 6 mm by 29% to 45% at the three test intervals (Figure 7). KHNs for Resinomer decreased significantly when the spacer thickness was more than 2 mm at the 1 hour and 1 week. Variolink’s mean KHNs deceased by 61% to 66% when the spacer thickness increased from 1 to 6 mm (Figure 8). Significant decreases in KHNs of Variolink occurred when the spacer thickness was more than 2 mm.
Figure (9) gives radiometer readings of light intensity of the light-curing unit when measured with and without spacers. Through only 1 mm ceramic spacer there was a decrease in light intensity of about 75%. Beyond 1 mm, light intensity continued to decrease gradually with increasing thickness of the ceramic spacer until the light was totally obstructed at 6 mm.

![Graph showing mean KHNs for Adherence obtained with six ceramic spacers at three test intervals.](image)

Figure 1: Mean KHNs for Adherence obtained with the six ceramic spacers at the three test intervals. Mean KHNs obtained without spacer were included for comparison.
Figure 2: Mean KHNs for Choice obtained with the six ceramic spacers at the three test intervals. Mean KHNs obtained without spacer were included for comparison.

Figure 3: Mean KHNs for Duolink obtained with the six ceramic spacers at the three test intervals. Mean KHNs obtained without spacer were included for comparison.
Figure 4: Mean KHNs for Enforce obtained with the six ceramic spacers at the three test intervals. Mean KHNs obtained without spacer were included for comparison.

Figure 5: Mean KHNs for Lute-It obtained with the six ceramic spacers at the three test intervals. Mean KHNs obtained without spacer were included for comparison.
Figure 6: Mean KHNs for Nexus obtained with the six ceramic spacers at the three test intervals. Mean KHNs obtained without spacer were included for comparison.

Figure 7: Mean KHNs for Resinomer obtained with the six ceramic spacers at the three test intervals. Mean KHNs obtained without spacer were included for comparison.
Figure 8: Mean KHNs for Variolink obtained with the six ceramic spacers at the three test intervals. Mean KHNs obtained without spacer were included for comparison.

Figure 9: Curing-light intensity measurements made with and without ceramic spacers. Note the significant drop in light intensity with merely 1 mm ceramic spacer.
Discussion

Ideally, dual-cure resin cements should be capable of achieving a degree of hardening through self-curing similar to or not much lower than that achieved through dual curing. This is to ensure adequate polymerisation of the cement in those areas underneath inlay/onlay restorations that are inaccessible to the curing-light. Despite the fact that the majority of the cements examined were recently made available on the market, their degree of hardening, which is an indicator of the extent of polymerisation, when only self-curing was used continued to be significantly lower than their degree of hardening when dual-curing was used. This finding is in agreement with previous reports (Hasegawa & others, 1991; Rueggeberg & Caughman, 1993; Watts & others, 1994; El-Badrawy & El-Mowafy, 1995). However, there was variability among the cements with regard to the amount of hardening achieved through self-curing alone. For one material (Enforce), a mean KHN of 42 was achieved through self-curing compared with a mean KHN of 51.4 achieved through dual curing at 1 day. For another (Variolink), a mean KHN of only 9.3 was achieved through self-curing compared with a mean KHN of 50.1 achieved through dual curing, also at 1 day. Mean KHNs of the remaining materials fell somewhere between these two extremes. The reason for this variability is most likely due to the way these cements were formulated. When a sufficient amount of self-curing chemical is incorporated in the material, this allows the cement to achieve an adequate amount of hardening in the absence of light activation, while if the self-curing element is deficient, maximum cement hardness cannot be achieved. Insufficient hardening of the cement may predispose to post-operative sensitivity due to wash out of the unset cement material with subsequent microleakage and recurrent caries.

The significant increase in KHN of Resinomer with time (Table 2) must be attributed to a slow polymerisation reaction. The mean KHN for this material increased from 15.6 at 1 hour to 32.3 at 1 week for the self-cured specimens, and from 33.1 at 1 hour to 44.6 at 1 week for the dual-cured ones.

According to Ash (1984), the cervico-occlusal length of premolar teeth ranges from 8 to 8.5 mm, and from 7 to 7.5 mm for permanent molars. Thus, the range of spacer thickness used in this study (1 - 6 mm) realistically represents the clinical
situation assuming that the gingival margin of a bonded inlay/onlay restoration should ideally remain on enamel, i.e., above the cemento-enamel junction; and bearing in mind that the tip of the light-curing rod remains at or above the level of the cusp tip during curing.

For the majority of cements examined, there was little difference in KHNs obtained with the 5 and 6 mm spacers. This can be easily explained by reference to Figure 9, which indicates little difference in light attenuation between these two spacers. Basically at this depth the cements were dependent mostly on their self-curing capabilities for polymerisation, as penetration of light was almost totally obstructed. For spacers 1 - 4 mm thick, the effect of light attenuation varied among the cements. For some of them the decrease in KHNs was small (such as Enforce, Nexus and Resinomer), while for others more abrupt decreases in KHNs took place between 1 and 4 mm spacing (e.g., Adherence, Duolink). In the former case, the presence of a potent self-curing component compensated for light attenuation even with the thicker spacer, while in the latter case a less potent self-curing component can be blamed.

In a clinical situation where an inlay/onlay restoration with deep gingival seat(s) is being cemented, the operator should probably apply the curing light from the buccal and lingual aspects of the restoration as well as from the occlusal aspect in order to maximise the amount of curing light that reaches the cement in the gingival seat areas. In the meantime manufacturers should modify their dual-cured resin cement formulations to optimise the efficiency of the self-curing component. This has to be done with great care in order to avoid incorporation of an excessive amount of the chemical-curing component which can lead to significant shortening of the working time of the cement with subsequent problems in insertion of inlay/onlay restorations.

Conclusions
1. For three of the cements examined (Adherence, Duolink and Variolink), self-curing alone was found to result in hardness values less than 50% of those obtained when dual-curing was used, even after 1 week of storage.
2. For many of the examined cements, there were significantly lower hardness values beyond 2-3 mm of ceramic inlay thickness.

3. Enforce cement exhibited the highest values of hardness when specimens were self-cured at the three test intervals (32, 42 & 43 KHN). This was sustained best through up to 6 mm of ceramic inlay material among the eight examined cements, with hardness ranging from 50 KHN at 1 mm to 45 KHN at 6 mm for the one-day test interval.

Acknowledgement

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References


MHM Rubo & OM El-Mowafy

-Presented at the AADR/CADR Meeting in Minneapolis, Minnesota – March 1998.

Abstract
The increasing demand for esthetic restorations in the last decade led to the development of inlays and onlays made of resin composite and ceramic. This was possible because of recent developments in bonding technique which allowed resin cements to bond to both dentin and enamel as well as inlay surfaces. Dual-cure resin cements have light-curing and chemical-curing components and therefore, should achieve maximum polymerization even in areas not readily reached by light. The aim of this study was to determine the efficiency of the chemical-cure component under varying inlay thickness. Discs 6 x 2.5 mm were prepared from cements: Adherence M5 (ADH), Choice (CHO), Duolink (DUO), Enforce (ENF), Lute-lt (LUT), Nexus (NEX), Resinomer (RES), and Variolink (VAR). Sixteen specimens were made from each cement. Twelve were dual-cured through resin composite spacers of varying thickness (1-6 mm) and four were dual-cured with no spacer. Knoop hardness measurements were then made at 1h, 1d and 1w intervals. ANOVA showed significant differences in mean hardness values among materials (p<0.0001), spacer thickness (p<0.0001) and time (p<0.0001). The hardness value achieved with no spacer after 1d was: LUT - 57.0, DUO - 55.9, ENF - 51.4, NEX - 50.8, VAR - 50.1, CHO - 45.8, RES - 44.3 and ADH - 44.1. All materials had significant decrease in hardness with 3 mm spacer or more. For 1d interval, the value dropped 48% for LUT and DUO, 42% for ADH, 21% for VAR and CHO, 16% for NEX and 12% for ENF and RES. For some cements further reductions in hardness occurred with thicker spacers. There was no difference in hardness values between 1d and 1w for most materials, except CHO and VAR. It is concluded that for some dual-cure cements a weak
chemical-curing component fails to enable the cement to reach optimum hardness with thick spacers.

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Appendix 4. Retention of a Posterior Resin-Bonded Fixed Partial Denture with a Modified Design.

MHM Rubo & OM El-Mowafy

- Presented in the Research Day at the Faculty of Dentistry, University of Toronto – February, 1998.
- To be presented at the IADR Meeting in Nice, France – June 1998.

Abstract

Recent studies indicated a lower rate of clinical longevity of posterior resin-bonded fixed partial dentures (RBFPD) compared to anterior ones. The aim of this study was to examine in vitro the effect of adding a retentive device to the design of posterior RBFPDs on their retention. Freshly extracted premolar and molar teeth were used in this study. Four groups of RBFPD specimens were prepared. For each specimen one premolar and one molar tooth with a space for one missing molar in between were mounted in a resin base. Group 1 (n=7) had RBFPD preparations with proximal grooves (PG), occlusal rests (OR) and lingual wings (LW). In group 2(n=7) PG were replaced with retentive slot preparations (SP) and prostheses were made with projections to fit into the bottom of these slots. Group 3 (n=7) had only slot preparations without LW and OR. Group 4 (n=6) was similar to group 2 but with non-retentive SP and the castings were made to fit the whole slot. Prostheses were cast in a non-precious alloy. Following microetching of the fitted surface of the prostheses with aluminum oxide powder, they were cemented with Cement-It! C&B (Jeneric/Pentron). Slot cavities of groups 2 and 3 were restored with a resin composite. Following water storage for 1 week, each prosthesis was subjected to compressive load-cycling in a Instron machine between 10 and 450 N at 4 Hz for a total of 230,000 load cycles. Prostheses were then separated using tensile force in an Instron machine and the failure loads were recorded. Data were analysed statistically using one factor ANOVA and Duncan’s test. Mean separation force (N) and standard deviations were:
Group 1: 361 (102.8) - Group 2: 525.4 (101) - Group 3: 448.6 (59) - Group 4: 417.1 (50.2)

Group 2 had significantly higher failure force than groups 1 and 4. Mean separation forces for groups 2 and 3 were not significantly different. It is concluded that the added means of retention (retentive slot preparations) significantly increased prosthesis resistance to dislodgement. A clinical trial is needed to follow up on these important findings.

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IMAGE EVALUATION
TEST TARGET (QA-3)

1.0
1.1
1.25

1.6
1.4
1.8

150mm
6"

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