©Operative Dentistry, 2005, **31-2**, 180-187

Laboratory Research

The Effect of Base Materials with Different Elastic Moduli on the Fracture Loads of Machinable Ceramic Inlays

S Banditmahakun • W Kuphasuk
W Kanchanavasita • C Kuphasuk

Clinical Relevance

Based on the results of this study, the elastic moduli of base materials had an influence on fracture loads of machinable ceramic inlays. The use of a base material with a high elastic modulus to support a ceramic inlay is recommended.

SUMMARY

This study investigated the effect of two base materials with different elastic moduli (F2000 and Vitrebond) on the fracture load of machinable ceramic inlays. Standardized MOD cavities were prepared in 18 human maxillary first or second premolars. The teeth were randomly assigned to three groups of six premolars each; Group 1 (control: no base); Group 2 (base with a polyacid-modified resin composite: F2000); Group 3 (base with a resin-modified glass-ionomer cement: Vitrebond). The inlays were fabricated from Vitablocs Mark II using a Cerec II machine. After the inlays were cemented with Tetric Ceram and the Syntac adhesive system, using the Ultrasonic Insertion Technique (USI), they were stored in distilled water at 37°C for 24 hours prior to fracture testing in a universal testing machine using a crosshead speed of 0.5 mm/minute. The static transverse elastic moduli of base materials were measured using a three-point bending test. The mean fracture loads and standard deviations of the Cerec inlays in Groups 1, 2 and 3 were 1.15 ± 0.39 KN, 1.13 ± 0.36 KN and 0.58 ± 0.11 KN, respectively. Statistical analysis showed that the mean fracture load of Group 3 was significantly lower than that of Groups 1 and 2 (*p*<.05). There was no significant difference in fracture load between Groups 1 and 2. The means and standard deviations of the elastic moduli of F2000 and Vitrebond were 15.63 ± 0.32 and 2.16 ± 0.55 GPa, respectively. The results indicated that the fracture load increased significantly as the elastic modulus of a base material increased.

Sasithorn Banditmahakun, DDS, graduate diploma in clinical science, MSc, government officer, Health Promotion Center Region 11, Thailand

*Watcharaporn Kuphausuk, DDS, graduate diploma in clinical science, MSD, director of Graduate Operative Dentistry, Department of Operative Dentistry, Faculty of Dentistry, Mahidol University, Bangkok, Thailand

Widchaya Kanchanavasita, BSc, graduate diploma in clinical science, MSc, PhD, assistant professor, Department of Prosthodontics, Faculty of Dentistry, Mahidol University, Bangkok, Thailand

Chotiros Kuphasuk, DDS, graduate diploma in clinical science, MSD, lecturer, Department of Prosthodontics, Faculty of Dentistry, Mahidol University, Bangkok, Thailand

*Reprint request: 6 Yothi St, Phayathai, Bangkok 10400, Thailand; e-mail: dtwkp@mahidol.ac.th

DOI: 10.2341/05-3

INTRODUCTION

During the past decade, the use of all-ceramic restorations has expanded due to improvements in dental adhesives and resin cements, which consequently result in the improvement of fracture resistance of dental ceramics. However, fracture is still one of the most common reasons for ceramic inlay/onlay failure. Factors that may lead to ceramic inlay fractures include inadequate thickness, improper cavity design, deep fissures in the restorations, defects such as pores and cracks in ceramic restorations and the resiliency of subinlay materials (Bergman, 1999; Mörmann & Krejci, 1992; Martin & Jedynakiewicz, 1999; Milleding, Örtengren & Karlsson, 1995).

For all-ceramic crowns, their fracture resistance has been proven to be significantly influenced by the elastic moduli of the core materials. Increasing the elastic modulus of the supporting core structure has been suggested as a way to increase the fracture resistance of all-ceramic posterior crowns (Scherrer & de Rijk, 1993; Lee & Wilson, 2000). However, a number of materials with different elastic moduli are recommended for use as a base under ceramic inlays. These include conventional glass-ionomer cements (Milleding & others, 1995; Christensen, 1998; Dietschi, Magne & Holz, 1994), resin-modified glass-ionomer cements (Thordrup, Isidor & Hörsted-Bindslev, 1994; Dietschi & others, 1994), polyacid-modified resin composites (Cossu & others, 1997) and resin composites (Dietschi & others, 1994; Cossu & others, 1997; Moscovich & others, 1998). To prevent fracture of a ceramic restoration, some weak materials, such as calcium hydroxide or traditional type III glass-ionomer cements, should not be used as a base/liner (Cossu & others, 1997). However, to date, no studies regarding the effect of elastic moduli of base materials on fracture resistance of ceramic inlays/onlays have been published.

This study evaluated the effect of two base materials, a polyacid-modified resin composite and a resin-modified glass-ionomer cement, each with a different elastic modulus, on the fracture load of Class II ceramic inlays luted with a resin composite.

METHODS AND MATERIALS

Table 1 lists the base materials used in this study. Eighteen freshly extracted non-carious intact human permanent maxillary premolars were collected and stored in 0.1% thymol solution. After cleaning, the tooth dimensions were determined in a bucco-lingual width (across the height of contour), a mesio-distal width (along the central groove) and an occluso-gingival height (from the marginal ridge to the cementoenamel junction) with a micrometer (Absolute digimatic, Mitutoyo Co, Kanagawa, Tokyo, Japan). The roots of each tooth were embedded in plastic tubes filled with acrylic resin (Formatray, Kerr, Orange, CA, USA) to a level approximately 2 mm below the cementoenamel junction. The long axis of the tooth was perpendicular to the horizontal plane. The mounted teeth were then kept in distilled water at room temperature before commencement of the experiment.

A Class II MOD cavity was prepared on each tooth using a taper diamond bur (Inlay set No 8113R, Intensive SA, Swiss Dental Products, Lugano-Grancia, Switzerland), with a 3° angle of inclination. A high-speed handpiece was attached to the modified surveyor. With this paralleling device, the long axis of the diamond bur was positioned parallel to the long axis of the tooth. Water spray was used during the preparation procedure. Following 10 preparations, a new bur was used.

The cavity dimensions are shown in Figure 1. The cavity depth, 1.5 ± 0.1 mm, was measured from the

Table 1: *Base Materials Used in This Study*

Material	Type	Composition	Manufacturer	Lot #
F2000	Polyacid-modified resin composite	Paste: FAS Glass, Carboxylate dimethacrylate Glyceral dimethacrylate Poly (vinyl pyrrolidone) Camphorquinone Clicker: Side A: Vitrebond copolymer Hydroxyethyl methacrylate Water, Ethanol Photoinitiators Side B: Water, Maleic acid	3M ESPE St Paul, MN, USA	20010122
Vitrebond	Resin-modified glass-ionomer cement	Powder: fluoroaluminosilicate glass Camphorquinone Liquid: acrylic-itaconic acid copolymer with pendent methacrylate groups, 2 HEMA, water photo-activator	3M ESPE St Paul, MN, USA	20010221

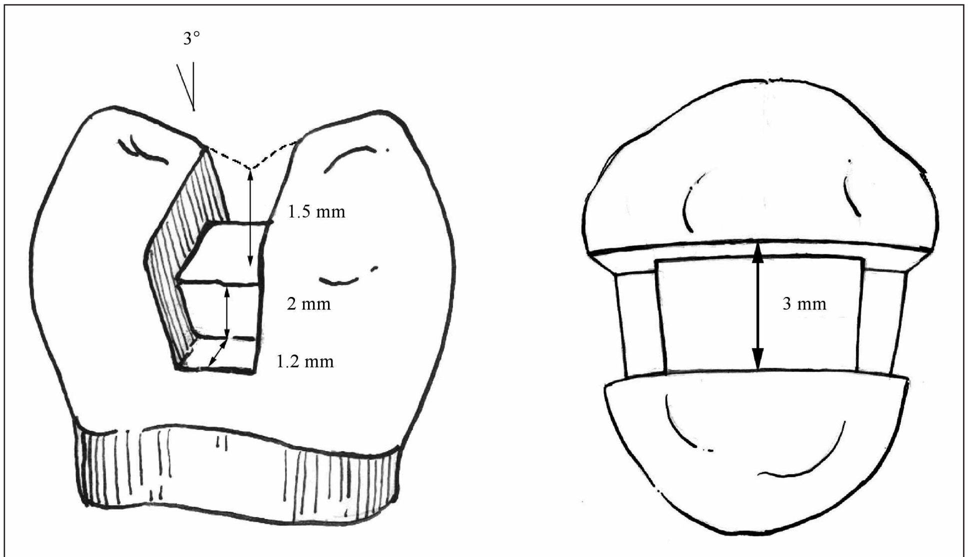

Figure 1: *Diagram illustrating the dimensions of a prepared MOD cavity.*

Table 2: *The Tooth Size (mm) Determined in Bucco-lingual (BL), Mesio-distal (MD) and Occluso-gingival (OG) Direction*

Treatment Group	Mean ± SD		
	BL	MD	OG
Group 1 (Control)	10.09 ± 0.48[a]	7.32 ± 0.24[b]	5.33 ± 0.22[c]
Group 2 (F 2000)	9.50 ± 0.44[a]	7.08 ± 0.32[b]	5.28 ± 0.47[c]
Group 3 (Vitrebond)	9.80 ± 0.40[a]	7.08 ± 0.24[b]	5.00 ± 0.32[c]

Same superscripts in the same column indicate no significant difference (p>.05)

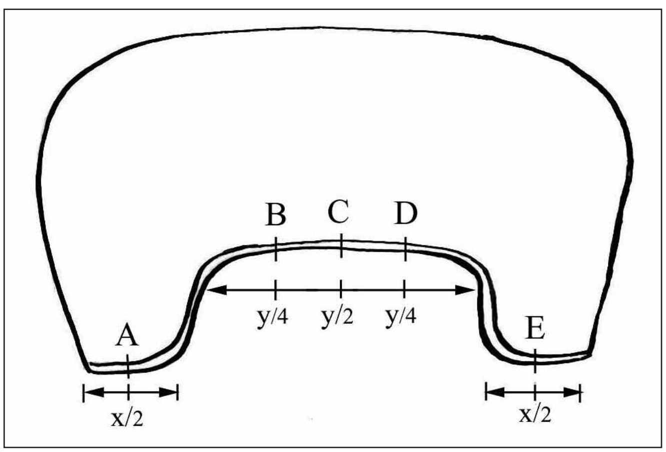

Figure 2: *Diagram of a replica for measuring the internal gap widths made from addition-reaction silicone. A-E refers to the selected points used for measurement of the internal gap widths. X = gingival wall Y = pulpal wall*

deepest point of the main fissure to the level of the pulpal wall. The bucco-lingual width, the width of the gingival wall and the height of the axial wall were 3.0 ± 0.1, 1.2 ± 0.1 and 2.0 ± 0.1 mm, respectively. All internal line angles and point angles were rounded. All cavity margins were placed within enamel without bevel. The preparation was finished with a fine diamond bur (Inlay set No 3113R, Intensive SA, Swiss Dental Products), an enamel hatchet (Hu-Friedy, Chicago, IL, USA) and an aluminum oxide abrasive flexible disc (Sof-Lex XT disc, 3M ESPE, St Paul, MN, USA).

The 18 prepared teeth were randomly assigned to three groups. A Kolmogorov-Smirnov test was used to verify whether a normal distribution of tooth dimensions existed in each group. The homogeneity of variance was determined using Levene's test. One-way ANOVA revealed that no statistically significant difference in tooth dimensions was found among the three experimental groups (p>.05) (Table 2).

In Group 1 (control), the prepared cavities were filled with a temporary material (Fermit N, Ivoclar-Vivadent, Schaan, Liechtenstein). In Groups 2 and 3, a silicone matrix fabricated from a putty-type addition-reaction silicone was used to replicate the original shape of the cavity. This matrix was used as a guide to facilitate base placement to the original shape. The pulpal wall was then reduced an additional 1.0 mm and replaced with either F2000 in Group 2 or Vitrebond in Group 3 to a thickness of 1 mm using the matrix. After curing, the excess base materials were removed with a fine diamond bur (Inlay set No 3113R) and an enamel hatchet. The prepared cavities were also filled with a temporary material. The teeth were stored in a closed chamber at 100% humidity and ambient temperature for 14 days.

The fabrication of ceramic inlays was performed using the machinable ceramics Vitablocs Mark II (Vita Zahnfabrik, Bad Säckingen, Germany, size I8, shade A2) and the Cerec II system (Siemens, Munich, Germany). After the temporary filling was removed, the cavity was cleaned with a slurry of pumice and polishing brush, rinsed with water spray and dried with oil-free air. The 1.5 mm inlay thickness was controlled from the projection and cross-section windows in the fissure line step. After fabrication, the inlays were examined for any cracks or fracture lines under a surgical microscope at 2.5x magnification (Microscope S21, Zeiss, Germany). Adjustment was performed using a fine diamond bur under water spray until the restorations were fully seated into the cavities. Before cementation, the interfacial gap widths of each specimen were determined by the multiple sectioning technique. A replica of the internal gap was made by coating the cavity surfaces with a thin layer of light-bodied addition-reaction silicone (Provil, Bayer, Leverkusen, Germany). The inlay was then delivered into the cavity, and the standard load (1.877 Kg) was applied until the impression material set. After removal of the inlays, the putty type of the same material (Provil) was placed onto the light-bodied film dressing the cavity. The replica was sectioned with a scalpel blade in a mesio-distal direction and measured at five points as shown in Figure 2.

For a replica of the marginal gap, the light-bodied addition-reaction silicone of a different color (Silagum, DMG chemisch Pharmaeutihe Fabrik GmbH, Hamburg, Germany) was used in the second layer in lieu of a putty-type silicone to prevent distortion of the first layer light-bodied silicone at the margin. The replica was sectioned and measured at 12 points (6 occlusal points, 6 proximal points) as shown in Figure 3A-3C.

Measurement of the gap widths was examined using a measuring microscope (Nikon, Tokyo, Japan) at 30x magnification. The data at each location were analyzed using one-way ANOVA at α=.05.

After determining the interfacial gap width, the Cerec inlays were then cemented with Syntac adhesive (Vivadent), Monobond-S (Vivadent) and Tetric Ceram (Vivadent) according to the manufacturer's instructions via the ultrasonic insertion technique. The inlays were pressed under a loading device (1.877 Kg) while light curing the resin composite. Before each curing, the intensity of light was calibrated with an Optilux Radiometer (Kerr) to ensure a constant value of at least 400 mW/cm^2 (580 mW/cm^2 in this study) of light intensity. Polymerization was activated for 40 seconds on each side (occlusal, mesial, distal, buccal and lingual). The cemented restoration was finished and polished with fine diamond burs (50, 15 µm Komet, Horico, Berlin, Germany), rubber polishers (Ceramiste, Shofu, Kyoto, Japan) and aluminum oxide abrasive flexible discs.

All test specimens were stored in distilled water at 37°C for 24 hours prior to testing. A stainless steel rod 3.1 mm in length and 1.1 mm in diameter was used to apply compression force along the central groove of the occlusal surface of the inlay. Each specimen was secured in a special device to prevent any movements during compression. The specimen was loaded to fracture with a universal testing machine (Instron 5566H 1612, Buckinghamshire, England) using a 0.5 mm/minute crosshead speed. The failure load in Newtons was obtained when the first discontinuity of the load-displacement curve appeared as a result of an early crack or catastrophic failure. The maximum load before failure was also recorded. The fracture load

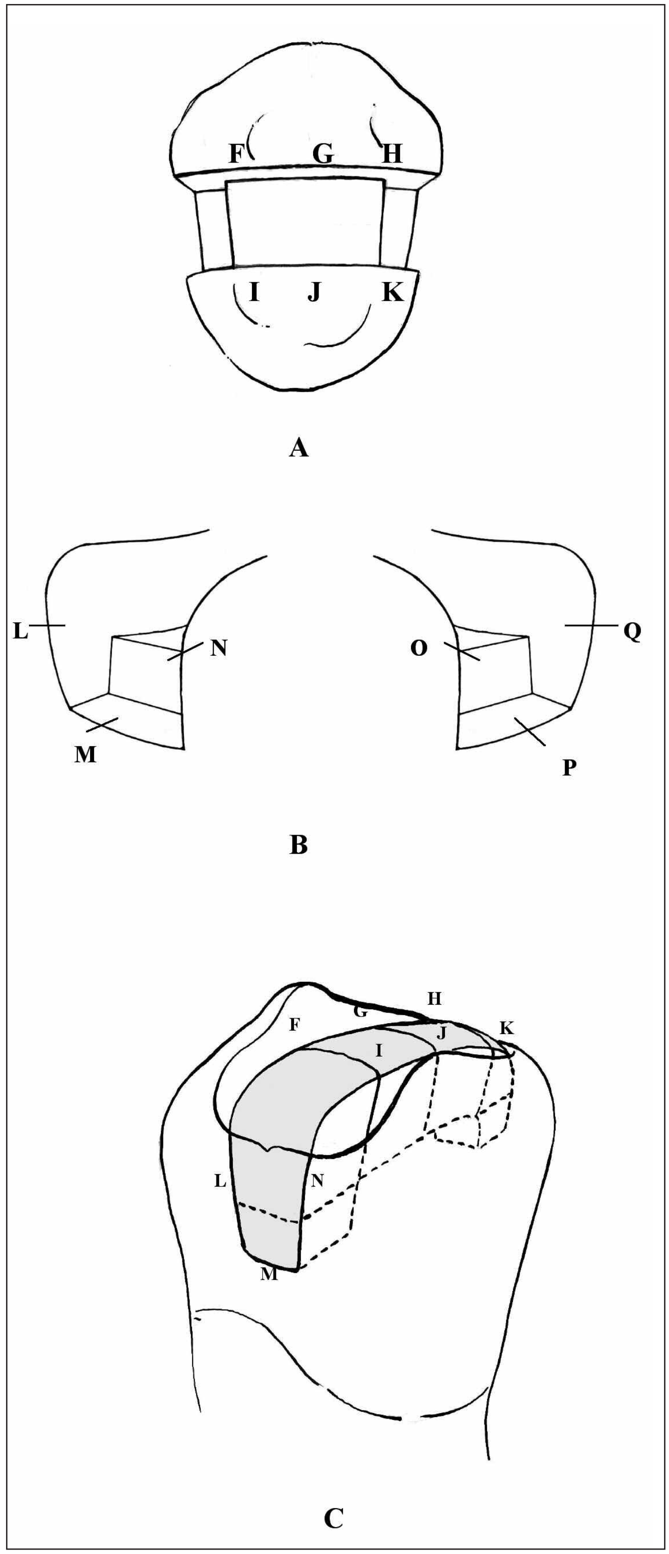

Figure 3: *Diagram of the six occlusal (F-K) and the six proximal marginal gap points (L-N, O-Q) selected for determination of the marginal gaps.*
A. Occlusal view
B. Proximal view
C. Replica in the cavity

data were analyzed using one-way ANOVA and Tamhane test at the .05 significance level.

The transverse elastic modulus of each base material was determined using the three-point flexure method. Five specimens (rectangular beams 2 mm x 2 mm in cross section and 25 mm long) of each material were prepared according to ISO/FDIS 4049:2000 and were stored in distilled water at 37 ± 1°C for 24 hours. The elastic modulus (E_b) was calculated from:

$$E_b = \frac{1}{48} \frac{P_1 L^3}{y_1 I} \text{ and } I = bd^3/12$$

where P is the load on the load-deflection curve, y is the deflection of the beam at load P_1 measured using a strain gage, L is the length of support, b=width and d=thickness of a specimen. The data were analyzed using t-test at α=.05.

RESULTS

Fracture Loads

The mean fracture loads and standard deviations of the Cerec inlays were 1.15 ± 0.39 KN in Group 1 (control), 1.13 ± 0.36 KN in Group 2 (F2000) and 0.58 ± 0.11 KN in Group 3 (Vitrebond). The results are presented in Table 3. Statistical analysis showed that fracture of the inlays supported by Vitrebond occurred at a significantly lower load than those supported by either F2000 or dentin (p<.05).

Elastic Moduli

The flexural modulus of F2000 (15.63 ± 0.32 GPa) was significantly higher than that of Vitrebond (2.16 ± 0.55 GPa) (p<.05) (Table 4).

Gap Widths

Table 5 shows gap dimensions between the internal surfaces of the inlays and the cavity walls measured from the walls depicted in Figures 2 and 3. The statistical analysis showed that there was no statistically significant difference (p>.05) in the values obtained among the groups.

DISCUSSION

A meaningful laboratory fracture test should be determined by performing carefully designed laboratory experiments that replicate clinical conditions as close as possible. The dimensions and shape of specimens used in the study should be similar to those used clinically (Kelly, 1999). In this study, the inlays were fabricated and bonded to the extracted human teeth to mimic clinical conditions. Additionally, the size of the extracted teeth and the cavity dimensions in each experimental group were controlled. Standardization of the cavity was achieved by using a paralleling device. This method could reduce operator errors and control occlusal divergence of the cavity.

Prefabricated ceramic blocks were used in this study, because they were processed by the manufacturers. The absence of voids or condensation defects in the machinable ceramic guarantees that the mechanical and physiochemical properties of the restorations are homogeneous throughout the piece. Therefore, confounding factors from crack initiation, due to small structural flaws such as voids, pores and cracks, which act as stress concentrators reducing ceramic strength, can be minimized (Jones, Jones & Wilson, 1972; Jones & Wilson, 1975). Although the milling process may lead to some flaws in inlays, the resin luting agent reduces the fracture incidence of restorations (Rosenstiel & others, 1993). Using machinable ceramic and a Cerec machine helped to maintain uniformity of the restorations in this study.

The marginal and internal gap widths of the Cerec inlays measured in this study (103.4 µm and 244.5 µm, respectively) were in the range reported by several

Table 3: *Mean and Standard Deviation of Fracture Load*

Group	Fracture Load (KN) Mean ± SD
Group 1 (Control)	1.15 ± 0.39[a]
Group 2 (F2000)	1.13 ± 0.36[a]
Group 3 (Vitrebond)	0.58 ± 0.11[b]

Different superscripts indicate significant difference (p<.05).

Table 4: *The Flexural Moduli of F 2000 and Vitrebond*

Material	Mean ± SD (GPa)
F2000	15.63 ± 0.32[a]
Vitrebond	2.16 ± 0.55[b]

Different superscripts indicate significant difference (p<.05).

Table 5: *Means and Standard Deviations of the Pooled Internal and Marginal Gap Widths (µm)*

Gap Width	Measurement Point	Treatment Groups			
		Group 1 (Control)	Group 2 (F2000)	Group 3 (Vitrebond)	Mean
Internal gap width	A-E (Figure 2)	239.4 ± 98.8[a]	242.6 ± 77.2[a]	251.4 ± 107.5[a]	244.5 ± 94.9
Marginal gap width	F-Q (Figure 3)	100.9 ± 81.1[b]	101.5 ± 73.6[b]	108.0 ± 92.7[b]	103.4 ± 82.7

Same superscripts along the same row indicate no significant difference (p>.05) other authors (Mörmann & Schug, 1997; Sturdevant, Bayne & Heymann, 1999; Denissen & others, 2000; Martin & Jedynakiewicz, 2000). This space must be occupied by a luting agent. Besek and others (1995) and Schmalz, Federlin and Reich (1995) suggested that light-cured resin composites should be used for the cementation of Cerec inlays, because the average marginal gap width of these inlays was more than 100 µm. Besek and others (1995) has shown that resin composites used as luting agents underneath Cerec inlays fabricated with Vitablocs Mark II can be adequately polymerized. As a result, a resin composite, Tetric Ceram, was used as a luting agent in this study.

Cement film thickness had some effects on the fracture load of ceramics. Scherrer and others (1994) found that the strength of all-ceramic restorations decreased significantly when cement film thickness exceeded 300 µm. However, below 300 µm, cement film thickness had negligible effects on their fracture resistance. The resin composite luting cement used in this study should not affect the fracture load of Cerec inlays, because gap widths were less than 300 µm.

The inlays supported by dentin and F2000 failed catastrophically; whereas those supported by Vitrebond had a non-linear load-displacement curve. This indicated that plastic deformation occurred in the latter group before failure occurred. Failure of ceramic restorations usually results from tensile stresses, because ceramic has a tensile strength much lower than its compressive strength (van Noort, 2002). Occlusal loading causes flexural deformation of the restoration, resulting in tensile stresses occurring on the internal surface, which is the origin of the fracture. With rigid support, flexion of the restoration and, therefore, plastic deformation, is reduced. The use of base materials with sufficiently high elastic moduli, for example F2000, can provide sufficient support to reduce tensile stress on the internal surface of ceramic inlays. This also gives rise to high resistance to occlusal load of the restoration. On the other hand, the use of base materials with low elastic moduli, for example, Vitrebond, allows for deflection of the restorations above them, thus producing a large amount of tensile stresses in the restoration, which leads to restoration fracture.

The transverse elastic modulus is a measure of the stiffness or resistance to deflection of a given material. A material with a higher elastic modulus can resist deformation from a given load better than one with a lower elastic modulus. In this study, the flexural moduli of F2000 and Vitrebond were 15.63 ± 0.32 GPa and 2.16 ± 0.55 GPa, respectively. These values were close to those reported in previous studies. The static and dynamic elastic moduli of F2000 were 11.3 GPa and 21.0-22.35 GPa (Sabbagh, Vreven & Leloup, 2002; Labella & others, 1999). The static elastic modulus of Vitrebond was 1.1 GPa (Tam, McComb & Pulver, 1991). Dentin has an elastic modulus in the range of 11-19.3 GPa (Renson & Braden, 1975; Bowen & Rodriguez, 1962; Sano & others, 1994; Lehman, 1967; Xu & others, 1998), which is similar to F2000. This might explain why the restorations supported by dentin and F2000 behaved similarly.

Scherrer and de Rijk (1993) reported the effect of the elastic modulus of supporting structure on the fracture load of all-ceramic crowns. They showed that fracture load increased when the elastic modulus of a core material or the supporting structure increased. Lee and Wilson (2000) also found the effect of different elastic moduli of cores on the fracture resistance of aluminous porcelain jacket crowns. They recommended that high elastic modulus materials be used for the core build-up of all-ceramic crowns. Farah, Hood and Craig (1975) reported that a base material should have a modulus of elasticity as high as possible to support a restoration from intermittent forces during mastication. Mean biting forces on premolars and molars were approximately 300 N and 400-800 N, respectively (Kohn, 2002) but six times greater in bruxer-clenchers (Gibbs & others, 1986). In this study, the inlays supported with a 1-mm layer of F2000 and Vitrebond could withstand loads up to 1,125 N and 576 N, respectively, before failure. These values are much higher than mean biting forces. However, clinically, the fracture loads of ceramic inlays have been found to be much lower than these values.

In this study, dry condition was performed instead of low cyclic intermittent loads under wet condition, in a similar fashion to what occurred in the clinical situation. Accumulation of microstructural damage during mastication may induce a catastrophic failure. With cyclic loading, these cracks fuse and become a growing fissure that insidiously weakens the restoration (Wiskott, Nicholls & Belser, 1995). This fatigue failure occurs in all-ceramic restorations and leads to clinical failure (Peterson & others, 1998). Under dry conditions, only low cyclic loads could produce no damage from accumulating at the cementing interface, and cyclic loads, alone, could not induce ceramic failures (Kelly, 1999). In contrast to wet conditions, water has been shown to affect the strength of ceramics (Fairhurst & others, 1993; Myers & others, 1994a,b; Kelly & others, 1998). Also, residual stresses under cyclic loading and the corrosive nature of oral fluids have possibly produced crack growth (Anusavice & Lee, 1989; Morena & others, 1986; Sherrill & O'Brien, 1974) and consequently decreased fracture resistance of the ceramics. This might be responsible for higher failure loads in this study, compared to those obtained clinically.

From the results of this study, the elastic moduli of base materials are suggested to have a significant influence on the fracture load of Cerec inlays. Therefore, this property must be considered when a base material is selected. Nevertheless, the fracture loads obtained from this study might not be directly clinically interpreted as the survival rate of the Cerec inlay, because only one of the variables contributing to inlay fracture was evaluated. Therefore, long-term clinical trials are necessary for a definitive conclusion.

CONCLUSIONS

Within the limitations of this study, it may be concluded that the fracture resistance of Cerec MOD inlays with dentin support or with foundation materials, such as F2000, which had high elastic moduli, was greater than that of inlays with less rigid support, such as Vitrebond.

(Received 4 January 2005)

References

Anusavice KJ & Lee RB (1989) Effect of firing temperature and water exposure on crack propagation in unglazed porcelain *Journal of Dental Research* **68(6)** 1075-1081.

Bergman MA (1999) The clinical performance of ceramic inlays: A review *Australian Dental Journal* **44(3)** 157-168.

Besek MJ, Mörmann WH, Persi C & Lutz F (1995) Curing of composites under Cerec inlays in Mörmann WH editors *CAD/CAM in Aesthetic Dentistry* Chicago Quintessence p 347-360.

Bowen RL & Rodriguez MS (1962) Tensile strength and modulus of elasticity of tooth structure and several restorative materials *Journal of the American Dental Association* **64** 378-387.

Christensen GJ (1998) Current use of tooth-colored inlays, onlays and direct-placement resins *Journal of Esthetic Dentistry* **10(6)** 290-295.

Cossu M, Frieder D, Magne M, Schönenberger A & Verdel G (1997) Indirect techniques in Dietschi D, Spreafico R, editors *Adhesive Metal-free Restorations: Current Concepts for the Esthetic Treatment of Posterior Teeth* Berlin Quintessence Publishing p 139-167.

Denissen H, Dozic A, van der Zel J & Van Waas M (2000) Marginal fit and short-term clinical performance of porcelain-veneered Cicero, Cerec, Procera onlays *Journal of Prosthetic Dentistry* **84(5)** 506-513.

Dietschi D, Magne P & Holz J (1994) Recent trends in esthetic restorations for posterior teeth *Quintessence International* **25(10)** 659-677.

Fairhurst CW, Lockwood PE, Ringle RD & Twiggs SW (1993) Dynamic fatigue of feldspathic porcelain *Dental Materials* **9(4)** 269-273.

Farah JW, Hood JA & Craig RG (1975) Effect of cement bases on the stresses in amalgam restorations *Journal of Dental Research* **54(1)** 10-15.

Gibbs CH, Mahan PE, Mauderli A, Lundeen HC & Walsh EK (1986) Limits of human bite strength *Journal of Prosthetic Dentistry* **56(2)** 226-229.

Jones DW, Jones PA & Wilson HJ (1972) The relationship between transverse strength and testing methods for dental ceramics *Journal of Dentistry* **1(2)** 85-91.

Jones DW & Wilson HJ (1975) Porosity in dental ceramics *British Dental Journal* **138(1)** 16-21.

Kelly JR (1999) Clinically relevant approach to failure testing of all-ceramic restorations *Journal of Prosthetic Dentistry* **81(6)** 652-661.

Kelly JR, Hunter BO, Brenyo MR & Peterson IM (1998) Simulating clinical failure during *in vitro* testing of all-ceramic crowns *Journal of Dental Research* **77** IADR Abstract #1175 p 778.

Kohn DH (2002) Mechanical properties in Craig RG, Powers JM, editors *Restorative Dental Materials* 11th Edition St Louis Mosby p 67-124.

Labella R, Lambrechts P, Van Meerbeek B & Vanherle G (1999) Polymerization shrinkage and elasticity of flowable composites and filled adhesives *Dental Materials* **15(2)** 128-137.

Lee SK & Wilson PR (2000) Fracture strength of all-ceramic crowns with varying core elastic moduli *Australian Dental Journal* **45(2)** 103-107.

Lehman ML (1967) Tensile strength of human dentin *Journal of Dental Research* **46(1)** 197-201.

Martin N & Jedynakiewicz NM (1999) Clinical performance of Cerec ceramic inlays: A systematic review *Dental Materials* **15(1)** 54-61.

Martin N & Jedynakiewicz NM (2000) Interface dimensions of Cerec-2 MOD inlays *Dental Materials* **16(1)** 68-74.

Milleding P, Örtengren U & Karlsson S (1995) Ceramic inlay systems: Some clinical aspects *Journal of Oral Rehabilitation* **22(8)** 571-580.

Morena R, Beaudreau GM, Lockwood PE, Evans AL & Fairhurst CW (1986) Fatigue of dental ceramics in a simulated oral environment *Journal of Dental Research* **65(7)** 993-997.

Mörmann W & Krejci I (1992) Computer-designed inlays after 5 years *in situ*: Clinical performance and scanning electron microscopic evaluation *Quintessence International* **23(2)** 109-115.

Mörmann WH & Schug J (1997) Grinding precision and accuracy of fit of Cerec 2 CAD/CIM inlays *Journal of the American Dental Association* **128(1)** 47-53.

Moscovich H, Roeters FJ, Verdonschot N, de Kanter RJ & Creuges NH (1998) Effect of composite basing on the resistance to bulk fracture in industrial porcelain inlays *Journal of Dentistry* **26(2)** 183-189.

Myers ML, Ergle JW, Fairhurst CW & Ringle RD (1994a) Fatigue failure parameters of IPS-Empress porcelain *International Journal of Prosthodontics* **7(6)** 549-553.

Myers ML, Ergle JW, Fairhurst CW & Ringle RD (1994b) Fatigue characteristics of a high-strength porcelain *International Journal of Prosthodontics* **7(3)** 253-257.

Peterson IM, Pajares A, Lawn BR, Thompson VP & Rekow ED (1998) Mechanical characterization of dental ceramics by hertzian contacts *Journal of Dental Research* **77(4)** 589-602.

Renson CE & Braden M (1975) Experimental determination of the rigidity modulus, Poisson's ratio and elastic limit in shear of human dentin *Archives of Oral Biology* **20(1)** 43-47.

Rosenstiel SF, Gupta PK, van der Sluys RA & Zimmerman MH (1993) Strength of a dental glass-ceramic after surface coating *Dental Materials* **9(4)** 274-279.

Sabbagh J, Vreven J & Leloup G (2002) Dynamic and static moduli of elasticity of resin-base materials *Dental Materials* **18(1)** 64-71.

Sano H, Ciucchi B, Matthews WG & Pashley DH (1994) Tensile properties of mineralized and demineralized human and bovine dentin *Journal of Dental Research* **73(6)** 1205-1211.

Scherrer SS & de Rijk WG (1993) The fracture resistance of all-ceramic crowns on supporting structures with different elastic moduli *International Journal of Prosthodontics* **6(5)** 462-467.

Scherrer SS, de Rijk WG, Belser UC & Meyer JM (1994) Effect of cement film thickness on the fracture resistance of a machinable glass-ceramic *Dental Materials* **10(3)** 172-177.

Schmalz G, Federlin M & Reich E (1995) Effect of dimension of luting space and luting composite on marginal adaptation of a Class II ceramic inlay *Journal of Prosthetic Dentistry* **73(4)** 392-399.

Sherrill CA & O'Brien WJ (1974) Transverse strength of aluminous and feldspathic porcelain *Journal of Dental Research* **54** 683-690.

Sturdevant JR, Bayne SC & Heymann HO (1999) Margin gap size of ceramic inlays using second-generation CAD/CAM equipment *Journal of Esthetic Dentistry* **11(4)** 206-214.

Tam LE, McComb D & Pulver F (1991) Physical properties of proprietary light-cured lining materials *Operative Dentistry* **16(6)** 210-217.

Thordrup M, Isidor F & Hörsted-Bindslev P (1994) A one-year clinical study of indirect and direct composite and ceramic inlays *Scandinavian Journal of Dental Research* **102(3)** 186-192.

van Noort R (2002) *Introduction to Dental Materials* 2nd edition London Mosby p 236.

Wiskott HW, Nicholls JI & Belser UC (1995) Stress fatigue: Basic principles and prosthodontic implications *International Journal of Prosthodontics* **8(2)** 105-116.

Xu HH, Smith DT, Jahanmir S, Romberg E, Kelly JR & Thompson VP (1998) Indentation damage and mechanical properties of human enamel and dentin *Journal of Dental Research* **77(3)** 472-480.