Metal–ceramic bond strength

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Persson M, Bergman M. Metal–ceramic bond strength. Acta Odontol Scand 1996;54:160–165. Oslo. ISSN 0001-6357.

No currently available bond test for dental metal–ceramic systems has yet gained general acceptance. Such a bond test cannot be established without careful analysis of the stress distribution within the adherence region. The objective of the present study was to establish a mechanical shear stress test and combine the results with a finite element stress analysis of an idealized crown to enable comparisons of different metal–ceramic systems. The titanium–ceramic systems tested, both machined and cast titanium, showed higher shear strength values than the high gold–ceramic system used as a reference. Since the latter system has been used successfully in the clinic for many years, it seems reasonable to assume that the bond strength of titanium–ceramic systems is quite satisfactory for dental crown-and-bridge work. In conclusion, the interfacial bond strength test proposed in the present study should be applicable to all of the currently known material combinations in which a brittle fracture might be expected to occur. ☐ *Ceramics; dental crowns; dental materials; finite element analysis; shear stress*

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Since its introduction into clinical dentistry about 35 years ago, the metal–ceramic or porcelain-fused-to-metal technique has become increasingly popular. Several combinations of dental metallic and ceramic materials have been introduced onto the market, and most of these systems have proved successful. The longevity of the system concerned relies on the formation of a stable bond between metal and ceramic which can withstand stresses common in the mouth. It is therefore most unsatisfactory that no universally accepted bond test has as yet been established. Several metal–ceramic bond tests have, however, been suggested/used (see, for example, Refs. 1–13). Furthermore, in recent years the bond strength between unalloyed titanium and ceramics has been tested using a four-point bending test (14, 15) or a three-point bending test (16). Most of the bond tests concerned depend either on gross deformation of the metallic material or, as in most cases, on uncontrolled stress concentrations to introduce a load at the metal–ceramic interface.

Today it is generally accepted that a bond test applicable to all types of dental metal–ceramic systems cannot be established without a careful analysis of the stress distribution within the adherence region, thereby considering both test specimens and the clinical situation. By means of finite element analysis the interfacial stresses under various loading conditions have been studied by, for example, Farah & Craig (17), Anusavice et al. (18), De Hoff et al. (19), and Schwarz et al. (9). However, the general impression from what is known about currently available tests of metal–ceramic bonds is that there may be some doubt about whether it is really the interfacial bond strength per se that is being measured, without the tensile yield limit of the metal having any influence.

The aim of the present study therefore was to present a simple mechanical test to assess the strength of the material interface, in which the test itself does not introduce any uncontrolled stress concentrations or cause any gross deformation of the metallic material. Gross deformations of the metallic material makes the test dependent on the mechanical properties, mainly the yield strength, of the metal. Furthermore, uncontrolled stress concentrations will complicate the analysis of the results. In most cases the mean stress and not the actual fracture stress is calculated and presented. The actual fracture stress may vary depending on material combination or test setup even when the mean values are equal.

An approach combining the mechanical test results with finite element analysis of the clinical stress distribution is suggested to give the results a clinical meaning.

Materials and methods

Mechanical test

Three series of specimens were prepared, each consisting of 10 parallels (Table 1). Each specimen was a cylindrical rod, made from two metallic pieces fired together with the ceramic material concerned, in accordance with the manufacturer's instructions (Fig. 1). The metallic pieces were 25 and 5 mm long and had a diameter of 6.0 mm.

Table 1. Results from the mechanical test. The linear factor, ζ, and the estimated clinical load limit are presented. All material combinations show a good clinical performance.

Core material	Porcelain	Fracture shear stress, τ_{frac} (N/mm^2)	Linear factor ζ	Clinical load limit (N)
Machined titanium grade-2 ASTM B348; Permascand, Ljungaverk, Sweden	Procera, Ducera Gmbh, Rosbach, Germany	47.5 ± 6.1	1.95	391
Cast titanium grade-2 ASTM B348; Permascand, Ljungaverk, Sweden	Duceratin, Ducera Gmbh, Rosbach, Germany	48.9 ± 12.6	2.01	402
M2 gold alloy; J.S. Sjödings, Kista, Sweden	Vita VMK68, Vita Zahnfabrik, Säckingen, Germany	35.4 ± 7.1	1.45	290

The gold alloy specimens were cast in accordance with the manufacturer's instructions. The machined titanium specimens were cut into lengths from wrought titanium rods as delivered, and the cast titanium specimens were prepared using the Ohara system (Ohara Co., Ltd., Osaka). During the firing of the ceramic layer, which was about 1.5 mm thick, the two metallic pieces were centered in a special holder (Fig. 1), to create a center of rotation in common. This was of the utmost importance in facilitating the calculations and the evaluation of the test results. After the firing, the rod specimens were checked for absence of defects at the periphery of the interface region by visual inspection. The reason for firing both pieces together instead of glueing one metal piece to the ceramic fired on the other piece was that the glue would have constituted an unknown factor that would probably have absorbed an unknown part of the load. The values achieved would thus have been complex to analyze.

The test method used is a shear test based on a torsional load. After the samples had been manufactured, the end of the larger piece was fixed in a special holder. A torsional load, parallel to the interface, was transferred to the sample with the tip of a screwdriver connected to the shaft of an electric motor. By measuring the voltage over a resistor, connected in series to the motor, it was possible to plot the torque against time. The torque was increased linearly with 20 Ncm/sec.

Because of the cylindrical symmetry of the system (Fig. 1) it is possible to obtain an analytical relationship between torque and shear stress at the interface:

$$\tau_{max} = 2M/(\pi r_0^3)$$

where τ_{max} = maximal shear stress, M = torque, and r_0 = rod radius. The maximal shear stress appears at the periphery of the circular interface between ceramic and metallic material. Lenz et al. (20) state that the

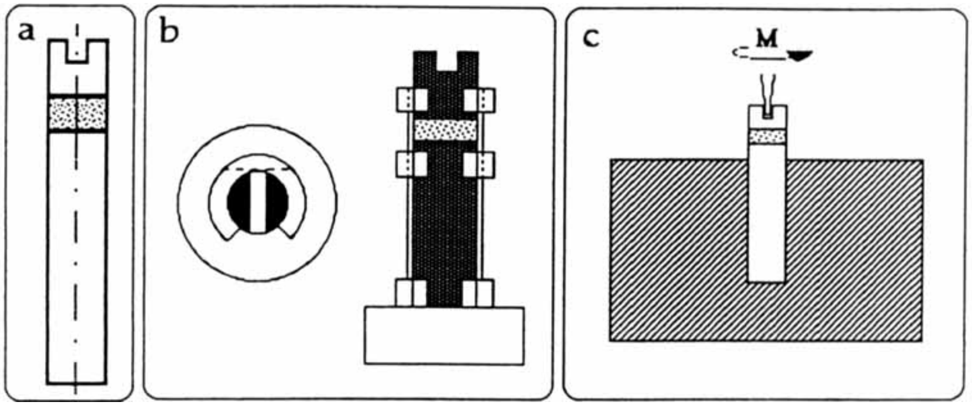

Fig. 1a. Sample for the shear stress test. The dotted region denotes the region of the ceramic. 1b. The samples are fired together in a special holder so they are exactly parallel and have a perfect center of rotation. 1c. Schematic description of the test method. The large piece is fixed in a holder. A torque is applied until the system fractures and the fracture torque is measured.

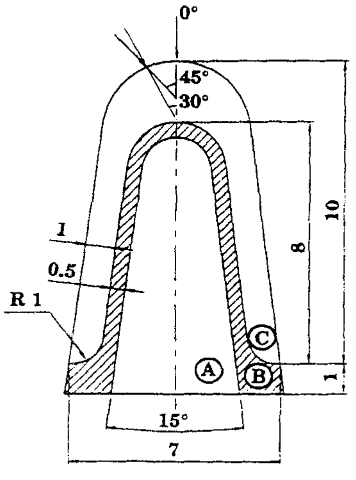

Fig. 2. Idealized circular symmetrical crown. All measures are in millimeters or degrees. The lines at the top denote different load angles. A = die region; B = metallic/Al_2O_3 region, referred to in the text as core; C = ceramic region.

specimen geometry and the loading of the specimen shall be chosen to create to the greatest possible extent a pure shear state of strain. According to the analyses made by these authors this is, in the ideal case, obtained by torsion of a metal–ceramic cylinder with circular cross-section. A pure state of strain enables analytical calculations of the stresses at the interface.

Another advantage of the chosen geometry is that the thickness of the ceramic layer is not critical. This made the manufacturing of the test pieces less time-consuming and less expensive.

Thus the shear strength τ_{frac} is obtained as the $\tau_{\max}$ value at torque M_{frac}, where fracture occurs.

Finite element stress analysis

A-three dimensional finite element (FE) analysis was performed on an idealized circular symmetrical crown (Fig. 2). The analysis program used was StressLab (Computer Vision Corp., Bedford, Mass., USA). As the model and load are symmetrical around the center plane, only half of the model had to be built up. The model was built up of 584 parabolic elements and 2855 nodes and fixed at the bottom of the die. The reason for using this kind of idealized model (Fig. 2) was that this crown is a well-defined body, and the calculations can be repeated by anyone using any material combination. The calculations were carried out to show the maximum tensile stress and the maximum shear stress, τ_{FEM}, at the interface between ceramic and core material. The materials investigated in this part of the work were various combinations of dental ceramics with a gold alloy, titanium, or dense sintered Al_2O_3 as core material and bovine dentin or stainless steel dies (Table 2). The reason for using dense sintered Al_2O_3 as core material in the calculations was that it is a very stiff material, in contrast to the titanium and high gold alloy used, which can be regarded as fairly soft materials.

Table 2. Elastic properties of materials considered in the finite element analysis. The properties are general values from the literature and are not connected to the real test materials

Material	Young's modulus (N/mm^2)	Poisson's ratio
Dentin	15×10^3	0.3
Titanium	110×10^3	0.375
Ceramic	70×10^3	0.25
Stainless steel	210×10^3	0.3
Al_2O_3	380×10^3	0.26
Gold alloy	110×10^3	0.3

Different load cases and material combinations were inspected (Table 3, Fig. 2). The questions posed were as follows: How does the core material affect the maximal stresses at the interface between core material and ceramic, and what effect would the choice of die material have on the calculated results? Stainless steel dies have been used in various tests described in the literature. It was therefore most interesting to analyze the dependence of die material on the calculated stresses. This is why stainless steel was used as a die material in one calculation.

Combining results from FE analysis with the shear strength values

An attempt to draw conclusions about the clinical situation was made. The results from the mechanical tests were combined with the theoretical results from the FE analysis. To combine these results, a linear factor Z was introduced. The linear factor, Z, is defined as the quotient $\tau_{\text{frac}}/\tau_{\text{FEM}}$, where τ_{frac} is the fracture shear stress in the mechanical test and τ_{FEM} is the maximum shear stress in the FE analysis. Thus, the actual fracture shear stress, τ_{frac}, is Z times greater than the maximum shear stress, τ_{FEM}, achieved in the FE analysis. As long as the mechanical test and the FE analysis are performed without any plastic deformations of the materials concerned, the mechanical properties are linearly dependent. Therefore, an expected fracture load of the ideal crown (Fig. 2) in the clinical situation can be calculated as

$$F_{\text{exp}} = F_{\text{FEM}} \cdot Z$$

In the present calculations $F_{\text{FEM}} = 200$ N.

It is of utmost importance to verify the estimations and calculations performed. Therefore, a real test, a so-called norm crown test (21) with circular symmetrical crowns (Fig. 3), was made. Ten titanium crowns with the circular symmetrical design were manufactured by means of the Procera technique (22), Procera porcelain was applied in accordance with the manufacturer's instructions, and the crowns were tested under loading

Table 3. Specification of and results from the finite element analysis. Calculated stresses at the ceramic–core interface. The area where the stresses appear is shown in parenthesis and refers to Fig. 3.

Loading condition	Material combination A/B/C*	Load (N)/angle(°)	Maximum shear stress, τ_{FEM} (N/mm^2)	Maximum tensile stress (N/mm^2)
1	Dentin/titanium/ceramic	200/45	24.3(II), 21.8(III)	32.9(I)
2	Dentin/titanium/ceramic	200/30	18.3(III)	23.9(I)
3	Dentin/titanium/ceramic	200/0	<5	<5
4	Stainless steel/titanium/ceramic	200/45	19.1(III)	17.5(I)
5	Dentin/Al_2O_3/ceramic	200/45	31.4(III)	19.8(I)
6	Stainless steel/Al_2O_3/ceramic	200/45	28.5(III)	14.0(I)
7	Dentin/gold alloy/ceramic	200/45	24.4(II), 21.9(III)	29.9(I)

* See Figs. 1–2.

condition 4 in Table 3. Zinc phosphate cement was used to fix the crowns to the stainless steel die. This is different from the FE model, in which the crown was perfectly fixed to the die by means of the nodes.

Statistical analysis

Student's t test ($p < 0.05$) was used for comparison of results.

Results

Mechanical test

In the center of rotation the shear stress is infinitely small. The maximum shear stress appears at the periphery of the rod. At the interfacial area of all the metallic rods no ceramic was observed at the peripheral region, whereas the ceramic remaining in the central part showed a significant vortex pattern at the center of rotation (Fig. 4). This is one proof of a pure torsional load.

The results from the mechanical test (Table 1) show that titanium, both cast and machined, shows significantly higher shear strength values than the gold alloy. No statistically significant difference was observed between cast and machined titanium.

Finite element stress analysis

Different stresses appear in different regions of the idealized crown (Fig. 3). Maximum shear stress appears in regions II and III of the crown, whereas maximum tensile and compressive stresses appear in regions I and II, respectively. In the first analyses all variables except the load angles were kept constant. In this manner the most unfavorable loading condition, a 45° load angle, was established. In the following calculations the other variables—that is, core and die materials—were varied, whereas the load angle 45° was kept constant. The results of the finite element analysis (Table 3) present some interesting findings. Variations in Young's modulus of the core material, condition 5 versus condition 1 in Table 3, give rise to dramatic variations in the maximum stresses at the interface between core material and ceramic. Therefore, the demands on the shear strength between a stiff core material and ceramic must be higher than those on the strength between less stiff core materials and ceramic.

The dependence on the die material is also remark-

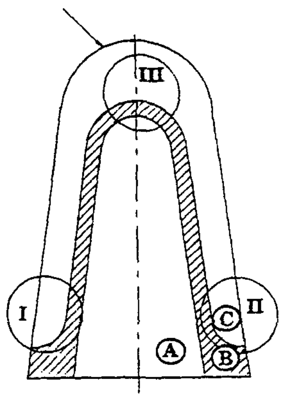

Fig. 3. A general load case applied to the idealized circular symmetrical crown. A = die region; B = metallic/Al_2O_3 region, referred to in the text as core; C = ceramic region. I, II, and III denote regions where the maximum stresses appear. The location and maximum stresses in different load cases are shown in Table 3.

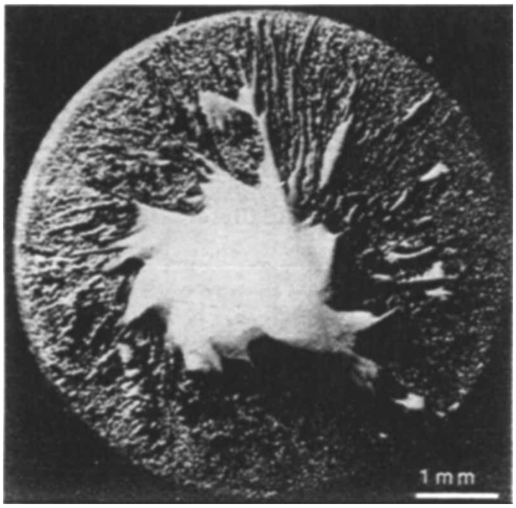

Fig. 4. Representative fracture surface. The remaining ceramic shows a significant vortex pattern in the center of rotation.

able (condition 4 versus condition 1 in Table 3). The tensile stress achieved with a titanium–ceramic crown on a stainless steel die is about half the stress achieved with a titanium–ceramic crown on a dentin die. The stresses are also reduced for the Al_2O_3 system. However, these variations are less dramatic than for the titanium system. It is also interesting that the area where the maximal shear stresses appear changes depending on the die material (Table 3).

Combining results from FE analysis with the shear strength values

Real tests with circular symmetric crowns—that is, norm crown tests (21)—were performed to verify the estimations and calculations made. In these tests the fracture load for the Procera titanium–ceramic crowns was 767 ± 67.4 N. The fracture occurred between the ceramic and metal in the direction of the load. The ceramic was sheared off from region III down to region II (Fig. 3).

From the results of the mechanical shear stress tests an estimated clinical shear fracture load was calculated, using data from Table 1 and load case 4 in Table 3. The calculated fracture load was as follows:

$$\mathcal{Z} = 47.5/19.1 = 2.49$$

$$2.49 \cdot 200 = 498\ \mathrm{N}$$

This value is lower than the value from the real test. This can be explained by the fact that the real crowns were cemented onto the die, whereas the simulated crown was perfectly fixed on the die by the nodes. That is, load can be absorbed by the cement, and the cement may yield in the real case, whereas no such effect can be obtained in the FE analysis since no cement is included in the FE model. Another possible explanation lies in the fact that it is very hard to manufacture a crown with exactly the same dimensions as the model.

The $\mathcal{Z}$ values of the two titanium systems are higher than the $\mathcal{Z}$ value for the gold alloy system. The estimated clinical fracture loads are therefore higher for titanium than for the gold alloy system.

Discussion

When studying the possibility of an interfacial crack occurring in a bimaterial appliance, such as a dental metal–ceramic construction, the stress pattern at the interface has to be considered. In the present study finite element analysis was used for this purpose since it is the most accurate method for studying the stress conditions in a three-dimensional body. One drawback of the calculations is that the material characteristics of dentin are poorly defined. This will of course affect the results of the calculations.

By combining the results of the finite element analysis of specimens of a shape relevant for that of a dental crown, thereby considering material combinations of topical interest in dentistry, it was possible to interpret the results of the mechanical shear test performed.

The differences in the results due to variations of die material show, for example, that so-called norm crown tests are not a direct test of the clinical performance. Furthermore, one has to be careful not to jump to conclusions about the clinical performance of a system when all the material properties are not properly considered. That is, to make statements about the clinical performance of different systems, one has to carry out some sort of analysis of the stress situation in the clinical case. Direct comparisons of test results are generally not valid.

One has to bear in mind that ceramics are brittle materials that are very sensitive to tensile stresses. The present approach to strength evaluation provides a determination of the shear strength at the interface between ceramic and metal. However, fracture may occur elsewhere in the crown before the interfacial fracture load is attained.

The interfacial bond strength between ceramic material and the high gold alloy used in the present study is in good agreement with the corresponding value for ceramic and a similar high gold alloy in a study by Lentz et al. (23). Their method relies on a deformation of the metallic material to introduce stresses at the interface. However, the load applied may exceed the yield limit for the metallic material. Some high gold alloys and commercially pure titanium of grade 2 have low yield limits, and the test method proposed by Lentz et al. (23) is therefore not applicable to these metal–ceramic systems.

An interesting result was that specimen combinations with titanium–ceramic showed a higher fracture shear stress—that is, interfacial bond strength—than the gold alloy–ceramic combination studied (Table 1). Since the latter combination has proved to be most successful for crown-and-bridge work during many years of clinical use, it seems reasonable to assume that the titanium–ceramic combinations studied have a satisfactory interfacial bond strength. However, it must be emphasized that the stiffness of the core material is most important for the overall strength of a metal–ceramic restoration under clinical conditions.

It is of interest that in the present study no difference in bond strength was observed between titanium/ceramic specimens, whether the titanium specimens had been cast or machined. However, the surface layer, the so-called α-case, of the cast titanium specimens had been removed in the same manner as in ordinary dental technical laboratory procedures. This indicates that no influence on bond strength of the unavoidable surface impurities emanating from investment and casting procedures needs to occur, provided there is proper technical handling. The results agree well with those of Dérand & Herø (15), who found no significant difference in bond capacity between thoroughly blasted titanium castings and wrought titanium.

In conclusion, the interfacial bond strength test used in the present study is applicable to all currently known material combinations in which a brittle fracture might be expected to occur. This is because no uncontrolled stress concentrations and no gross deformation of the metallic material are introduced by the test per se.

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Received for publication 22 June 1995
Accepted 20 September 1995