Influence of Resilient Support of Abutment Teeth on Fracture Resistance of All-Ceramic Fixed Partial Dentures: An In Vitro Study

Martin Rosentritt, BSc, PhD[a]/Michael Behr, DDS, PhD[a]/Peter Scharnagl, DDS[b]/
Gerhard Handel, DDS, PhD[c]/Carola Kolbeck, DDS, PhD[a]

Purpose: The influence of resilient support of abutment teeth on the fracture resistance of all-ceramic three-unit fixed partial dentures (FPDs) was tested in this study. ***Materials and Methods:*** Three groups (n = 8) of glass-infiltrated, alumina-based, all-ceramic FPDs that were adhesively bonded to human molar teeth were investigated. In control group A, teeth that were rigidly inserted in polymethyl methacrylate (PMMA) resin were used for thermocycling and mechanical loading (TCML), as well as for fracture testing. In group B, TCML was conducted on teeth that had their roots covered with a polyether layer. After TCML, the polyether layer was removed entirely, and the teeth were rigidly fixed in PMMA for fracture testing. In group C, teeth roots remained covered with a polyether layer during TCML as well as during fracture testing. Using a resilient attachment, tooth mobility was determined in axial, buccal, and oral directions. ***Results:*** Mean tooth mobility was 76 ± 4 μm in the axial direction, 278 ± 41 μm in the buccal direction, and 128 ± 17 μm in the oral direction. Group C showed the lowest mean fracture strength (polyether during both TCML and fracture testing) of 523 N. For group B (polyether during TCML but not during fracture testing), a fracture strength of 676 N was found, and for control group A (rigidly embedded teeth), it was 919 N. ***Conclusion:*** These results confirmed the influence of resilient attachments on the aging process and fracture strength testing of FPDs. Fracture resistance was significantly reduced when this particular interface was used in both TCML and fracture testing. *Int J Prosthodont 2011;24:465–468.*

In vitro studies support the evaluation of dental materials and new types of fixed partial dentures (FPDs). Thus, such studies enable scientists to conduct more efficient and less time-consuming clinical trials.[1] The large amount of data on laboratory simulations available underlines the demand for these aging tests. Different appliances for simulating the oral environment have been described,[2–7] and such devices are available commercially (EGO, EnduraTEC, SD Mechatronik, SDE). However, the validity of laboratory tests and simulations is restricted, and data correlating in vitro results with in vivo experiences are rare.[1,8,9] Different simulation parameters have been used to imitate clinical conditions, but the influence of particular parameters (occlusal force, chewing frequency, thermal loading, moisture, lateral jaw motion, type of abutments, or antagonistic dentures) has been described only rarely.[5]

The periodontal ligament in particular represents a complex system with highly individual characteristics that strongly depend on the age, sex, general health, and dental treatment of the individual patient.[10] Therefore, determining and calculating general tooth mobility is rather difficult.[11–14] Clinical loading situations physiologically varying between 12 and 70 N[15,16] result in compression of the periodontium (0.03 to 0.15 mm[10]). At the same time, torque and bending moments are applied to existing FPDs depending on the location of load application and the anatomical shape of the tooth root. Low occlusal forces may be balanced by periodontal compression, but higher loading may result in tooth deflection and thus in stress on the FPD. Chewing generates tensile stress

[a]*Assistant Professor, Department of Prosthodontics, Regensburg University Medical Center, Regensburg, Germany.*
[b]*Private Practice, Beilngries, Germany.*
[c]*Head, Department of Prosthodontics, Regensburg University Medical Center, Regensburg, Germany.*

Correspondence to: *Dr Martin Rosentritt, Department of Prosthodontics, Regensburg University Medical Center, 93042 Regensburg, Germany. Fax: +49 941 944 6171. Email: martin.rosentritt@klinik.uni-regensburg.de*

mainly on the bottom side and the bottom connector areas of the FPD. Unilateral vestibular or oral loading may cause additional torsion moments. All-ceramic materials especially show limited resistance against bending, tension, or torsion.[17–20] Therefore, the mobility of abutment teeth may consequentially influence the behavior and fracture resistance of all-ceramic three-unit reconstructions, as seen in an earlier investigation of resin-bonded FPDs.[21] In cases without any catastrophic failures (fracture of the restoration) during oral application, aging and deterioration effects (subcritical crack growth, wear, cracks) may nevertheless occur, and stress-induced defect areas may weaken the ceramic structure.[22–24]

The background for this investigation was the question of whether in vitro studies should consider the impact of resilient support of abutment teeth during thermocycling and mechanical loading (TCML) and fracture testing. Therefore, the hypothesis of this study was that resiliently attached abutment teeth would affect the fracture resistance of all-ceramic three-unit FPDs during TCML but not during fracture testing itself.

Materials and Methods

Forty-eight human molars were allocated into three groups to simulate different abutment situations for all-ceramic three-unit FPDs (n = 8 per group). Two teeth were fixed pairwise at a distance of 10 mm to represent a molar gap. Three groups were formed to investigate the influence of resiliently supported abutment teeth on FPD performance during TCML and fracture testing:

- Group A (control): The abutment teeth were rigidly inserted in polymethyl methacrylate (PMMA) resin (Palapress Vario, Kulzer) and underwent TCML as well as fracture testing without simulation of a resilient attachment.
- Group B: The roots of the abutment teeth were covered with a polyether layer (Impregum, 3M ESPE). After TCML, the polyether layer was entirely removed, and the teeth were rigidly fixed in PMMA for fracture testing.
- Group C: Teeth roots remained covered with a polyether layer during TCML as well as during fracture testing.

Resilient Attachment

For simulating periodontal mobility, the roots of the abutment teeth were dipped in a wax bath (2 × 1 seconds at 180°C; Micro Dura Dip, Girrbach). The teeth roots were then fixed in PMMA resin up to 1.5 mm to the dentinoenamel junction. The crown of each individual tooth was enclosed with plaster (Moldano blue, Heraeus Kulzer) up to the PMMA top surface. This plaster cast was used to extract the tooth root from the PMMA socket. All wax was removed, and the inner sides of the socket and the teeth roots were sandblasted (aluminum oxide; 250 µm, 2 bar) and painted with polyether adhesive (3M ESPE). The space between the root and PMMA (previously covered by wax) was filled with polyether impression material. A groove on the cast and the PMMA socket allowed exact repositioning of the tooth into the PMMA socket. The plaster cast was broken, plaster and excess polyether were removed, and the tooth crowns were cleaned. The resulting thickness of the polyether layer was 0.75 ± 0.1 mm. This procedure was developed previously using an artificial tooth (maxillary right second molar; Morita) with different polyether layer thicknesses. The tooth deflection was determined up to a load of 75 N (v = 1.5 mm/min; UTM Zwick 1446) in axial, buccal, and oral directions (n = 10 per direction). Teeth were sliced, and the polyether thickness was measured and compared with tooth deflection parameters for human teeth as described in the literature.[12]

Manufacturing of the FPDs

The pairwise embedded teeth were circularly prepared with diamond burs (Intensiv) for full crown–retained FPDs with 1-mm chamfer finishing lines. Polyether impressions (Permadyne, 3M ESPE) were taken, and dye casts were made. The all-ceramic FPDs were manufactured using a glass-infiltrated, alumina-based, all-ceramic material (Inceram Alumina, Vita Zahnfabrik). The dimensions of the connector areas measured 4.0 mm both in height and width. The insides of the abutment crowns were thinly covered with silane (Monobond-S, Ivoclar Vivadent). All FPDs were adhesively bonded onto the abutment teeth with a dual-curing resin cement (Variolink II, Ivoclar Vivadent) of low viscosity combined with a dentin adhesive system (Syntac classic, Ivoclar Vivadent).

Twenty-four hours after cementation, the FPDs underwent TCML. The thermal loadings were applied by rinsing the dentures with distilled water (5°C and 55°C, 2 minutes each cycle, 6,000 times). At the same time, the mechanical loadings were conducted with a human molar antagonist and a load of 50 N (1.2×10^6 times, 1.66 Hz). The fracture force of the FPDs was determined (UTM Zwick 1446; v = 1 mm/min) by applying the load with a steel ball (d = 12.5 mm) that was placed in the center of the pontic. To avoid local force peaks, tin foil measuring 0.4 mm in thickness was inserted between the steel ball and the FPD. Failure of the dentures was defined as a 10% loss of the actual loading force.

Calculations and statistical analysis were carried out using SPSS 17.0 for Windows (IBM). Means and standard deviations were calculated and analyzed using one-way analysis of variance and the Bonferroni multiple comparison test for post hoc analysis. The level of significance was set at $\alpha = .05$.

Results

The Morita teeth with the 0.75 ± 0.1 mm polyether layer showed mobility values (mean and standard deviation) of 76 ± 4 µm in the axial direction and 128 ± 17 µm in the oral direction, and 278 ± 41 µm in the buccal direction. The standardization procedure for the polyether layer was easy to handle and showed reproducible thicknesses for resilient attachments to human abutment teeth.

All FPDs survived TCML without failure independent of examination group (A, B, or C). The lowest mean fracture strength was determined for group C (polyether during both TCML and fracture testing) with 523 N. In group B (polyether during TCML, fracture testing without polyether), the mean fracture strength was 676 N. A comparison of groups C and B showed no statistically significant differences ($P = .961$). The highest fracture strength value of 919 N was found for control group A (no polyether). No significant difference could be determined between groups A and B ($P = .364$) but could be determined between groups A and C ($P = .047$) (Fig 1). In each case, the fracture occurred at the connector area between the pontic and the crown abutment.

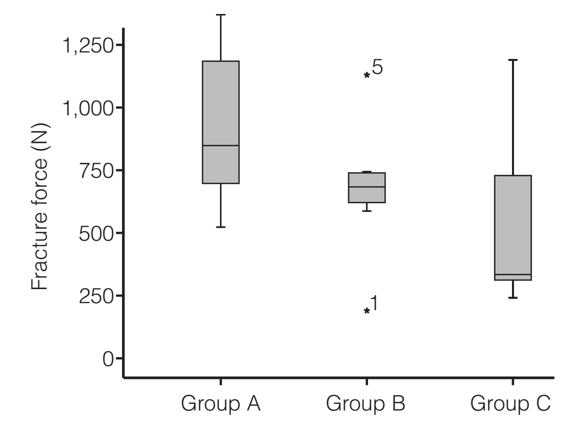

Fig 1 Fracture strength after TCML. *Extreme outliers. Numbers represent specimens corresponding to outlier values.

Discussion

Chewing simulation is a sufficient way to test new materials or new indications before being introduced to the clinical user. Many parameters should be testable and adapted so that the in vitro results can be compared to clinical situations. The chosen parameters in this study were considered to represent 5 years of oral stress.[1,5,9]

A different method for the simulation of accelerated aging on dental restorations would be the cyclic loading technique, as introduced by Wiskott et al.[24] Mechanical loading during chewing simulation is comparable to the repeated small load applications in cyclic loading tests. A single load would not result in complete destruction of a restoration. Nevertheless, it can introduce subcritical crack growth or crack initiating points. Thus, the brittle ceramic structure is weakened during TCML. This effect is even greater if resilient support of the abutment teeth is used because of additionally occurring bending and torsion effects on the dentures.

The resilient interface had a significant influence on the fracture strength of the tested FPDs when used during TCML as well as fracture testing. The simulated tooth mobility seemed to be responsible for an aging effect during TCML. However, fracture resistance was significantly reduced if the polyether interface was used during fracture testing as well. The results indicate the influence of resiliently attached abutment teeth with a fracture force reduction of the dentures of approximately 40% to 50%. Omitting the resilient attachment for the TCML or fracture force testing would therefore lead to an overestimation of the fracture resistance of brittle all-ceramic FPDs in clinical applications.

The observed type of fracture is in accordance with finite element analysis studies and clinical observations[17–19] showing failures in the connecting areas resulting from tensile and torsion stress, partly in combination with insufficient FPD design. Some studies on clinical behavior of all-ceramic FPDs found additional failures in occlusal and marginal areas caused by pressure and wear.[17,25,26]

Constant clinical occlusal forces of 12 to 70 N and occasional maximum forces up to 909 N in posterior areas can be assumed depending on the type of measurement, sex, restoration, food, and other parameters. Therefore, failures of the tested posterior FPDs may occur because loading capabilities of approximately 500 N are required in molar regions. Thus, the dentures tested in this study design would not withstand clinical applications without restrictions.

It is obvious that further studies on the examined topic are required because there are a number of limitations to the actual investigation. First, identical FPDs on artificial teeth would avoid individual tooth and FPD morphologies and dimensions. However, the abutment material (modulus, bonding capacity) strongly influences the fracture resistance results in in vitro tests.[27,28] Second, the simple design of the polyether interface does not stay abreast of the complex human periodontal ligament, with desmodontal movements providing a nonlinear compression.[11–13] A sufficient exact polyether layer thickness provided reproducible tooth mobility, which was in the range described in the literature[10]; however, an easier handling and fabrication process for the resilient interface would be desirable. Further investigations are necessary to develop a reliable in vitro test model for dental restorations. However, the concept of using natural teeth with resilient attachment as shown in this study seems to be a promising base for future work.

Conclusion

In vitro simulations should be carried out with defined resiliently supported abutments to ensure significant results. The periodontal mobility should be simulated at least during TCML to avoid overestimation of the fracture strength, especially of brittle materials.

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