Clin Oral Invest (2012) 16:1659–1667
DOI 10.1007/s00784-011-0653-5

ORIGINAL ARTICLE

In vitro investigation of the performance of different restorative materials under cast circumferential clasps for removable dental prostheses

Peter Pospiech · Florian Nagel · Florian Gebhart · Frank P. Nothdurft · Gergo Mitov

Received: 20 May 2011 / Accepted: 25 November 2011 / Published online: 17 December 2011

Abstract
Objectives The objective of this in vitro study was to investigate the behavior of different composite restorative materials under the load of cast circumferential clasps for removable dental prostheses (RDPs).
Methods In 60 human molars, standardized mesial–occlusal–distal cavities were prepared. The cavities were restored with the following materials: Definite, Tetric Ceram, SureFil, Heliomolar RO, Ariston pHc, and Oralloy, and provided with a rest seat. The rest seats were subjected to 5,000 cycles of thermal cycling and 1,200,000 masticatory cycles in a mastication simulator via cobalt–chromium circumferential clasps cast to standardized frameworks in a laboratory model designed to simulate the biomechanics of a free-end denture base. Fracture analysis of the restorations was performed by light microscopy. Before and after loading, material wear was measured with a 3D-laser scanner, and an analysis of the marginal quality was performed in an SEM at ×200 applying the replica technique.
Results No significant differences in the fracture behavior among the composite materials were found; the amalgam control group showed a significantly higher fracture resistance. Regarding the wear of the materials, the composites Definite and SureFil exhibited a behavior similar to that of amalgam. The other composites demonstrated higher wear rates. The initial marginal quality was significantly worse for Ariston pHc. The marginal adaptation decreased significantly after thermal and mechanical loading for Definite and Ariston pHc.
Conclusions In terms of the investigated aspects of mechanical performance, the tested composites seemed to be inferior to amalgam. Further clinical studies are needed to evaluate the ability of composite restorations to provide support for RDP clasps.
Clinical relevance The use of composites as direct restoration materials should be avoided in teeth, which serve as abutments for clasp-retained RDPs.

Keywords Prosthodontics · Thermal and mechanical stress · Biomaterials · Removable dental prosthesis (RDP) · Restoration · Clasp · Fracture

P. Pospiech · G. Mitov (✉)
Department of Prosthetic Dentistry, Danube Private University,
Dr.-Karl-Dorrek-Str. 23,
Campus West 3500 Krems, Austria
e-mail: gergo.mitov@dp-uni.ac.at

F. Nagel
Private practice,
Ottobeuren, Germany

F. Gebhart
Private practice,
Munich, Germany

F. P. Nothdurft
Clinic of Prosthetic Dentistry and Dental Materials Sciences,
Saarland University,
Homburg, Saarland, Germany

Introduction

Since their introduction in the 1950s, clasp-retained removable dental prostheses (RDPs) have been well established as a method to restore the function of the masticatory apparatus. Despite disadvantages in terms of aesthetics, function, and tertiary prophylaxis, they are still regarded as the simplest and most straightforward treatment option for partial edentulism [1] with a well-documented clinical long-term prognosis of over 50% after 10 years [2–5]. Support is one of the main features of RDPs from a biomechanical point of view. In the posterior region, occlusal rests usually provide vertical support and allow occlusal forces to be transmitted along the long axis of the abutment tooth to the bone. They also provide indirect retention for the denture [6, 7].

For decades, amalgam has been the restorative material of choice for large class II restorations. Discussions on its biocompatibility and higher aesthetic demands have led to the increasing use of tooth-colored composites, even in occlusion-bearing posterior teeth [8]. To date, research has delivered contradictory findings about the clinical performance of both material groups. While some studies have shown comparable annual failure rates for both materials [9] or even higher survival rates for large composite restorations [4], other authors have reported better longevity of amalgam restorations compared with composite restorations [10–12]. The main reasons for failure of the restorations discussed in the literature are secondary caries, marginal deficiencies, fracture, and material wear [11, 13].

Finite element analyses of posterior RDP abutment teeth have demonstrated that, in simulated function, the highest maximum equivalent stress appears on the occlusal rest seat surface [14]. In clinical practice, the posterior abutment teeth for an RDP are often restored with large direct restorations, so that the occlusal rest of an RDP clasp may be positioned in the restorative material. Amalgam restorations in posterior abutment teeth have been described as suitable for rest seats [15, 16]. As there are almost no data available on the performance of composites under the increased load of RDP clasps, the aim of this in vitro study was to evaluate the fracture behavior, substance wear, and marginal integrity of different composite restorative materials under a clasp rest in functional loading. The null hypothesis tested in this study was that, in the present experimental setup, no difference between the composite materials and amalgam would be observed in terms of those parameters.

Method and materials

Sixty extracted caries-free maxillary and mandibular human molars of similar coronal size were embedded 2 mm apical to the cement–enamel junction (CEJ) in autopolymerizing resin blocks (PalaXpress, Heraeus-Kulzer, Wehrheim, Germany) with dimensions of 2.5 cm×1.4 cm×1.5 cm. Both the extracted teeth and the prepared samples were stored continuously in 0.1% thymol solution.

A single operator prepared standardized mesial–occlusal–distal (MOD) cavities according to the preparation instructions for amalgam restorations with proximal boxes limited within enamel (Fig. 1). The dimensions of the cavities were as follows: cavity depth and bucco-lingual width, 3±0.1 mm; and width of the gingival wall and height of the axial wall, 1.5±0.1 mm. To achieve divergence angles between opposing walls of 6°, cavities were prepared using coarse diamond burs with a corresponding taper (ISO 806 314234534 012, Komet, Lemgo, Germany) in a high-speed dental handpiece (Contact-Air 632D, Kavo, Biberach, Germany) at 200,000 rpm under water cooling. Fine-grained diamond burs of the same shape (8855.314, Komet) were used for finishing the preparations at 100,000 rpm. The cavity dimensions ensured that the proximal boxes ended above the CEJ and the cavity was limited by enamel. In the area of the subsequent rest seat, the bucco-lingual width was increased to 4 mm to ensure sufficient thickness of the restorative material around the occlusal seat. The internal point and line angles were rounded, and enamel margins were not beveled but prepared in butt-joint configuration. After visual inspection of the cavities for imperfect finish lines and for correct dimensions (Dial Caliper, Kori Seiki, Tokyo, Japan), the 60 prepared teeth were randomly assigned to five experimental groups with eight teeth each (Table 1).

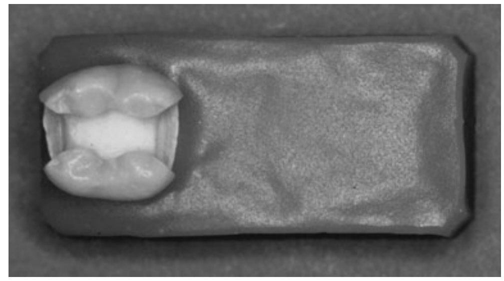

Fig. 1 MOD cavity preparation

The restorative materials and the bonding systems were used with strict adherence to the manufacturers' instructions. The composite restorations were placed using an incremental technique, and each increment was light-cured for 40 s (400 mW/cm^2, Elipar II, 3 M ESPE, St. Paul, MN, USA). Under the amalgam restorations, a 0.5-mm base of zinc-phosphate cement (Harvard Cement, Richter and Hoffman Harvard Dental, Berlin, Germany) was placed. The restorations were polished with finishing discs (Soft-Lex XT, 3 M ESPE), disk-shaped aluminum-oxide-impregnated silicone points (Enhance, Dentsply DeTrey, Konstanz, Germany), and an aluminum oxide polishing paste (Prisma Gloss, Dentsply DeTrey).

Spoon-shaped occlusal rest seats (1.5 mm depth×2 mm bucco-lingual×2 mm mesio-distal width) (Fig. 2) were prepared in the restorations by the use of a round-headed diamond bur (ISO 801001016, Komet), sharp edges were removed, and the occlusal seats were polished with aluminum-oxide-impregnated silicone points (Enhance, Dentsply DeTrey).

Impressions of the specimens, including their resin bases, were made with an A-silicone (Adisil Blau, Siladent Dr. Böhme und Schöps, Goslar, Germany), and casts were produced of type 4 dental stone (Resin Rock, Frankonia Dental, Erlangen, Germany).

The prosthetic equator was marked with pencil lead, and the deepest position of the retention arm in the retentive section

Table 1 Restorative materials used in the study

	Restorative material	Type of material	Bonding system	Matrix	Filler type	Filler particle content weight (%)
1.	Definite (Degussa)	Ormocer	Etch & Prime 3.0	Polysiloxane and photo-polymerizable methacrylate-groups	Ba-glass, pyrogenic SiO_2, modified apatite	76 %
2.	Tetric Ceram (Ivoclar Vivadent)	Micro-filler hybrid composite	Syntac classic	Bis-GMA, UDMA, TEGDMA	Ba-glass, YbF_3, pyrogenic SiO_2, BaAlF-silicate glass	80 %
3.	SureFil (Dentsply)	Packable composite	Prime & Bond NT	Urethane-modified BIS-GMA	BaAlFB- silicate glass, pyrogenes SiO_2,	82 %
4.	Heliomolar RO (Ivoclar Vivadent)	Inhomogenous micro-filler composite	Syntac classic	BIS-GMA, UDMA, decandiol-dimethacrylate	Pyrogenic SiO_2, YbF_3	76 %
5.	Ariston pHc (Ivoclar Vivadent)	Ion-releasing composite	–	BIS-GMA, UDMA, dimethacrylate	Alkal. glass fillers, BaAlF-silicate glass, YbF_3, pyrogenic SiO_2	79 %
6.,	Oralloy (Coltène)	Spheroidal non-gamma-amalgam	–	–	–	–

was determined using a depth gauge (Scribtometer, Degussa, Hanau, Germany). The clasp design for the circumferential clasps was determined according to the Bios/Rapid-Flex System (Degussa). For every clasp, one-third of the retentive arm was placed below the prosthetic equator, one-third on it, and one-third above it; the guiding arm was placed on the prosthetic equator. The undercuts (except for the areas used for retention) were then blocked out with blockout wax (Block out wax, Dentsply DeTrey), and the modified models were doubled with A-silicone. Investment casts were then produced (Rema Dynamik, Dentaurum, Pforzheim, Germany), and the prefabricated wax patterns were adapted along the ledges formed with the blockout material.

The wax patterns were @#@sprued, invested, and cast in Co–Cr alloy (Remanium 6 M 800, Dentaurum) with a casting machine (Globucast, Krupp, Essen, Germany) according to the manufacturer's instructions. After being cast, they were divested, cleaned, and subjected to airborne-particle abrasion with 50-µm aluminum oxide using 0.4-MPa air pressure. The restorations were finished under water-cooling with fine and superfine finishing stones (Gebr. Brasseler, Lemgo, Germany)

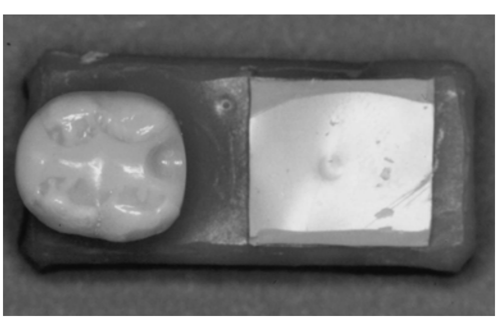

Fig. 2 Heliomolar restoration with an occlusal rest seat. On the *right side*, silicone layer for the simulation of the oral mucosa

and rubber polishing kits (Eveflex Polisher, EVE Ernst Vetter GmbH, Pforzheim, Germany).

For simulation of a free-end denture base, a horizontal bar with dimensions 10 mm×5 mm×2 mm, ending with a hemisphere with a diameter of 8 mm, was positioned on the clasp shoulder (Fig. 3). To simulate the biomechanics of a free-end denture base, a 3-mm layer of an A-silicone with Shore hardness 60 (Dimension Penta H Quick, ESPE, Seefeld, Germany) was placed in a cavity measuring 10 mm by 10 mm in the resin block under the hemisphere [17]. A load-deflection specification of the material was established using a standardized loading device Z1445 (Zwick, Ulm, Germany). Static load from 1–100 N was applied with an 8-mm-diameter stainless steel ball, and the sinking depths were recorded, resulting in a sinking depth of 0.5 mm at a load of 20 N.

Before and after the completion of cyclic loading in the mastication simulator, 2 polysiloxane (Adisil Blau, Siladent Dr. Böhme und Schöps) impressions of the test samples and the antagonist were taken, and replicas of the impressions were fabricated. The replicas were cast in type 4 die stone (Fuji Superhard Rock, GC Corporation, Tokyo, Japan) for the wear analysis and in an epoxy resin (Epoxy-Die, Ivoclar, Schaan, Liechtenstein) for the evaluation of marginal adaptation.

Thermal cycling and mechanical loading (TCML)

The test specimens were mounted in a commercially available dual-axis mastication simulator (Willytech, Munich, Germany) and were subjected to 1,200,000 masticatory cycles of unidirectional antagonist movements with a frequency of 1.2 Hz, descending cross-speed of 10 mm/s, and an applied force of 88 N. Using lever mechanics calculations (Fig. 3), the position of the loading stylus was defined to be 2.3 mm from the occlusal rest, so that the loading

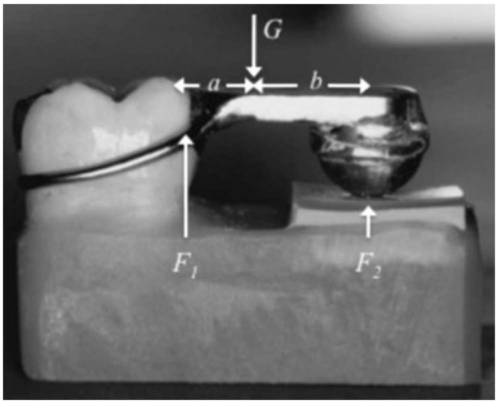

Fig. 3 Fabricated framework placed on the abutment tooth before mastication simulation. According to the lever mechanics in a state of balance, $F_1*a = F_2*b$ and $F_1 + F_2 = G$, where F_1 and F_2 are the forces applied to the rest seat and the hemisphere, respectively, a and b are the distances from the rest seat and the hemisphere to the loading point. The loading force G applied by the chewing simulator was chosen to be 88.3 N (9 kg), and for F_2=20 N, the sinking depth of the hemisphere was determined to be 0.5 mm. Under these conditions, F_1=68.3 N, and a=2.3 mm

forces at the occlusal rest and the hemisphere were 68.3 and 20 N, respectively. Simultaneously, the samples were subjected to 5,000 cycles of thermal loading at temperatures between 5°C and 55°C. The loading force of 88 N (9 kg) was chosen according to literature values for the maximum biting forces on single teeth in partial dentures [18, 19]. In a similar experimental setting, *Borchers* [20] estimated that the clasp arms transferred 19 N of the applied load. Thus, the loading force applied on the restorations by the clasp rests would be 49 N. Under these conditions (TCML: 1,200,000 cycles between 1 and 49 N; 5,000 thermal cycles 5/55°C in distilled water), a simulation of five years of clinical performance of the restorations was performed [21].

Fracture analysis

All specimens were investigated for fractures and cracks by stereomicroscopy (Leica Wild M420, Leica, Bensheim, Germany) at ×30 magnification. Each specimen was evaluated according to the following criteria: "no fracture"/"hairline crack"/"complete fracture" (Figs. 4 and 5). For further fractographic analysis, two specimens of each group were embedded in epoxy resin (Citofix, Struers, Willich, Germany), and additional cross-sections were prepared by initial cutting with a diamond saw (Accutom, Struers) and polishing with a commercial preparation system (Abramin, Struers).

Wear analysis

The surfaces of the type 4 die stone replicas were analyzed by means of the optical 3D surface profilometer Laserscan 3D (Willytech, Munich, Germany) [22]. The mean volumetric wear loss was determined by the appropriate software, Match 3D, Version 2.3 (Willytech).

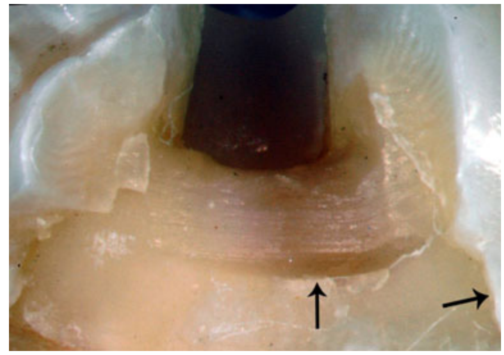

Fig. 4 Total fracture for a Heliomolar specimen

Margin analysis

The marginal adaptation of the class II restorations was evaluated by quantitative marginal analysis [23]. The epoxy resin replica models, both before and after in vitro thermomechanical loading, were viewed by scanning electron microscopy (Leitz AMR 1200, Leica, Wetzlar, Germany) at a ×200 magnification. The proximal and occlusal margin areas up to the middle of the mesio-distal circumference of the restorations on the side of the occlusal rest seat were then categorized by means of a continuous series of images according to the criteria "perfect margin" and "marginal gap/irregularity" (Figs. 6 and 7). The percentage distribution of the various qualities of marginal adaptation was calculated with image analysis software (Quanti-Gap, Küppers/Kunzelmann GmbH, Erlangen, Germany).

Specimens whose restorations were fractured during the mastication simulation were not used for the wear and quantitative/qualitative margin analysis. Therefore, the number of specimens that could be examined decreased from initially 60 to 48 (8 in the "amalgam" group, 6 in the groups "Definite" and "Tetric Ceram," 7 in the groups "SureFil" and "Heliomolar RO," and 5 in the group "Ariston pHc").

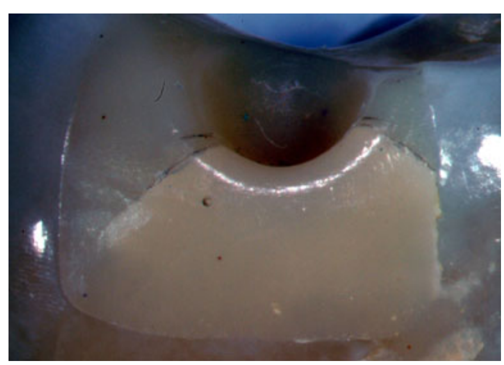

Fig. 5 Hairline crack in a Definite specimen

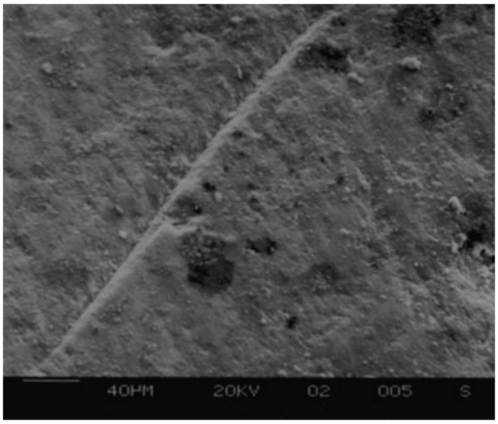

Fig. 6 Margin analysis under SEM: gap-free margin (original magnification 1:200)

Statistical evaluation

Statistical evaluation was performed with SPSS for Windows, Release 17.1 (SPSS Inc., Chicago, IL, USA). Non-parametric tests (Mann–Whitney U-test for unpaired sample groups; Wilcoxon test for paired sample groups) were used to determine significant differences ($p=0.05$) as data were not normally distributed (Shapiro-Wilks test, $p<0.05$). The Levene test was used for testing the equality of variances ($p<0.05$). Any differences in fracture mode were calculated using Fisher's exact probability test.

Results

Fracture analysis

The results of the fracture type analysis are shown in Fig. 8. Amalgam showed a significantly better fracture behavior (Fisher exact test: $p=0.004$), compared with 80% intact restorations after mastication simulation. The composite restorations had a high range of damage, with 80–100% of hairline cracks and total fractures. No statistically significant difference among the composites could be found.

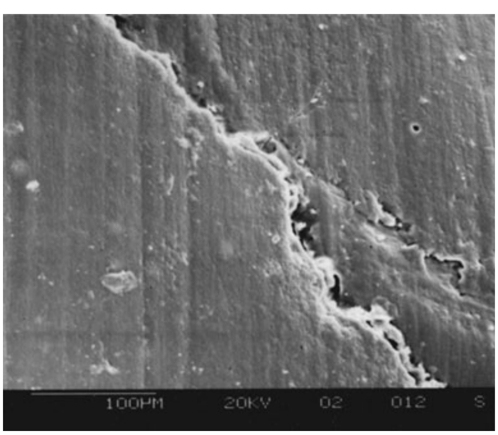

Fig. 7 Margin analysis under SEM: gap formation (original magnification 1:200)

Wear measurement

An overview of the results of the wear measurements is shown in Fig. 9. According to the mean wear volumes, the investigated materials can be divided into two groups: Definite (0.055 mm^3, standard deviation 0.028), amalgam (0.056±0.018), and SureFil (0.058±0.025), with lower wear volumes; and Ariston (0.00954±0.0552) Tetric Ceram (0.011±0.031), and Heliomolar (0.0145±0.085), showing higher wear volumes. Statistically significant differences could be found between Heliomolar and amalgam ($p=0.013$), between Heliomolar and SureFil ($p=0.024$), and between Heliomolar and Definite ($p=0.024$).

Margin analysis

The results of the marginal assessment for the criterion "gap-free margin" are shown in Fig. 10. In the Definite and Ariston groups, a significant increase in the percentage of gap-free margins could be found after cyclic loading in the mastication simulator. The deterioration in marginal quality for Definite was statistically higher than that for Tetric Ceram, SureFil, and Heliomolar RO.

Discussion

The present study examined the mechanical behavior of direct restorative materials under vertical loading via occlusal rests of tooth and mucosa-borne dentures. For approximation of the clinical situation as much as possible, natural molars of similar dimensions were selected, and the oral mucosa was simulated. However, all laboratory studies present limitations, and the results should thus be interpreted carefully. The use of 3-mm layer of A-silicone for the simulation of the oral mucosa resilience has already been documented [17], and the desired sinking depth of 0.5 mm was determined for static loading. However, the dynamic load applied in the present study in a wet environment might result in a different sinking depth due to the mechanical properties of the A-silicone. A further limitation of the study is the vertical direction of the stress application during the mastication simulation. Although the main purpose of rest seats is to provide vertical support, their movements during the mastication process are quite complex as the biomechanics of tooth and mucosa-borne RDPs is influenced by a great variety of parameters [14, 24]. The experimental setting did not consider a possible lateral denture displacement due to the horizontal components of the masticatory forces.

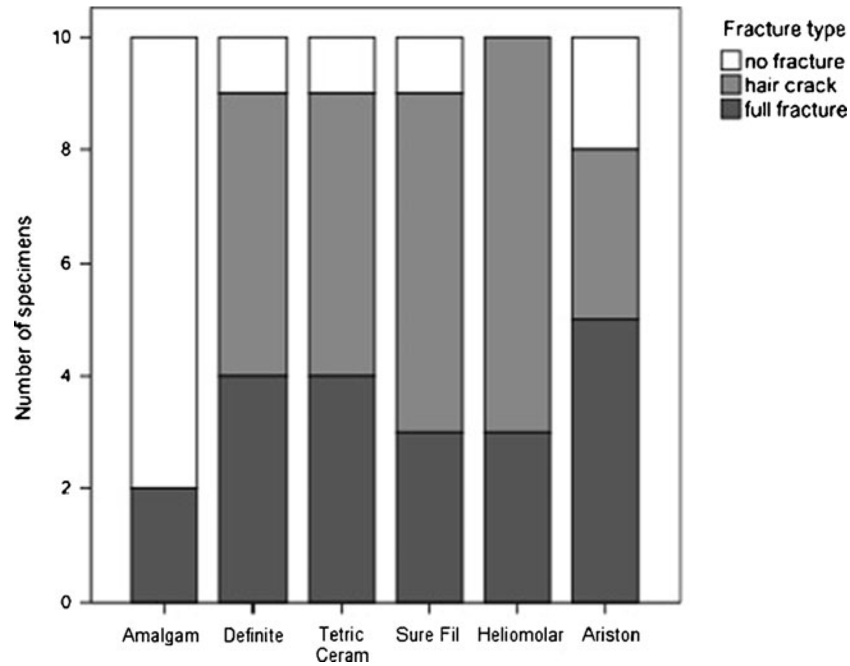

Fig. 8 Fracture behavior of the restorative materials

The used weight-controlled chewing simulator has proved a reliable method for long-term fatigue testing. However, *Steiner et al.* could show that the reliability of the loading forces is strongly influenced by the descending speed and loading weight [25]. Using a slow descending speed of 10 mm/s, an initial overloading peak at load contact can be avoided, but the cycling frequency is lower than the physiologic masticatory frequency [26]. The calculations for the load distribution were performed for the assumption that no deformation of the clasp rest would occur under loading. Additionally, the point transmission of force in the chewing simulator could hardly be adjusted with an accuracy of 0.1 mm. These limitations of the experimental setting might also have an effect on load distribution.

The restorations were placed in natural maxillary and mandibular molars. Despite our efforts to standardize the preparation of the MOD cavities, the forms and dimensions of the direct restorations differed in certain respects, mainly due to the different morphology of the teeth. Thus, the present in vitro simulation can provide only clues to the clinical behavior of the restorative materials used under RDPs.

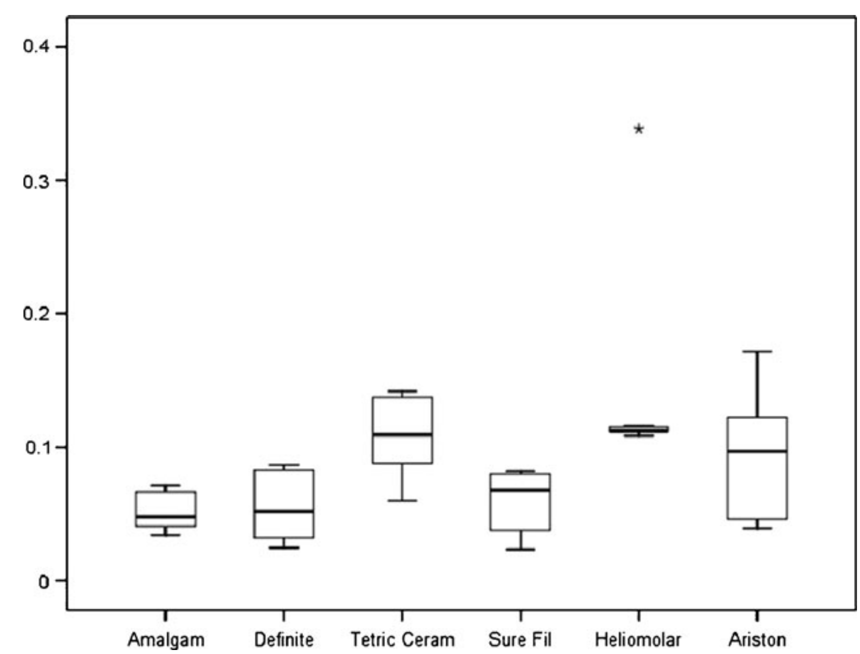

Fig. 9 Volume loss (mm^3) after mastication simulation

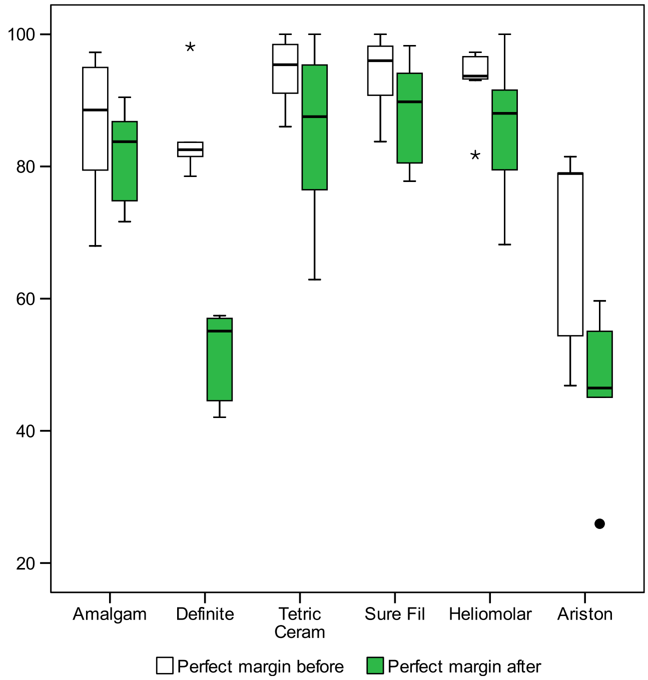

Fig. 10 Box-plot diagram of the margin analysis of the proximal and occlusal margin area

Fracture analysis

In the present study, the amalgam Oralloy showed a fracture resistance significantly higher than that of the composite materials. Therefore, the null hypothesis was rejected. Among the composite materials, no significant differences could be found with regard to their fracture behavior, in spite of different filler and matrix qualities.

For brittle materials, such as amalgam and composites, fracture resistance is dictated by the materials' ability to retard crack initiation and propagation. An overview of the mechanical properties of the materials used is given in Table 2, showing equal or even superior values for the fracture toughness and flexural strength for the composite materials. For the modulus of elasticity, a greater scattering can be observed, with values ranging from 6 to 11.4 GPa for the composites and from 25 to 60 GPa for dental amalgams. However, a higher modulus of elasticity alone can hardly explain the different fracture behavior of both materials. A possible explanation may be seen in the findings reported by Beatty and Pidaparti that, for composites, the elastic modulus was nearly twice as great in tension as in compression, whereas for dental amalgam, the elastic modulus is more than 3.5 times greater in tension than in compression [27]. This phenomenon results in a reduction of the amount of tensile stress produced within the materials when they are subjected to bending. For amalgams, this would lead to an effectively smaller zone of tensile stress, in which cracks can originate, compared with composites [28].

The fact that amalgam and composites are fundamentally different material groups has to be taken into consideration. Whereas composites show much brittleness and little elasticity, the metal alloy amalgam can reduce tensions through plastic deformation [27, 29, 30]. The material evades mechanical stress through plastic deformation. Composites do not show the ductile capacity of metals, so these are always exposed to full occlusal stress.

Wear

For the optical 3D scanner used in this study, the accuracy and the precision of the 3D data acquisition depend on the surface inclination [22]. Up to an angle of 60°, the accuracy is better than 6 µm, and the precision is better than 3 µm. With greater inclination angles, the accuracy and precision decrease as the inclination increases. Therefore, for the vestibular and lingual cavity areas, the accuracy of the wear measurement was reduced. With an area of the occlusal rest of about 10 mm^2, the vertical loss would be about 5 µm (e.g., Amalgam). This value would be within the matching accuracy of the scanning device. A further limitation of the wear measurements was the fact that, after mastication simulation, only a reduced number of samples could be used for analysis (n=8 for amalgam and

Table 2 Mechanical properties of the tested materials [40]

Restorative material	Flexural strength (MPa)	Compressive strength (MPa)	Modulus of elasticity (GPa)	Fracture toughness (MN*m-3/2)
Oralloy	110–150	518	25–60	1.0–1.5*
Definite	128	400	7.3	1.6
Tetric Ceram	130	230	9.4	2.0
SureFil	125	330	11.4	2.0
Heliomolar RO	100	330	6.0	0.84**
Ariston pHc	125	280	11.0	1.9

*Value according to Lloyd [48]

**Value according to Choi and co-workers [49]

n=5–7 for the composite groups) as the fully fractured restorations were excluded. Thus, the results of the wear analysis should be treated very cautiously.

In this study, the micro-filler composite Heliomolar RO showed significantly higher wear rates compared with SureFil, Tetric Ceram, and Ariston pHc. In a wear investigation of ten dental restorative materials in five wear simulators, Heliomolar showed better wear performance in the two-body wear test than SureFil and Tetric Ceram and higher wear values for the three-body wear simulators [31]. At first sight, the test arrangement chosen in this study could be categorized as a two-body wear test, so the results concerning the wear behavior of Heliomolar might seem surprising.

A possible explanation of the significant wear of Heliomolar RO in the present study was reported by Condon and Ferracane [32]. With a specific wear simulator, they found that, compared with hybrid composites, Heliomolar RO shows superior wear behavior at a relatively low cyclic loading of 20 N and higher wear rates at a cyclic loading of 70 N. The authors attribute the results to fatigue processes at higher loading forces. For micro-filled composites, the low modulus of elasticity [30] and low filler content [33, 34], as well as the quality of the filler/matrix interface [35], have been discussed as factors influencing their fatigue resistance. In this study, the high fatigue susceptibility of Heliomolar was affirmed by the fracture analysis results, with all Heliomolar restorations showing fracture after loading.

Margin analysis

The evaluation of the margin quality between composite materials and tooth structures is a realistic and valid test for adhesive restorations [36]. Before mechanical and thermal testing, the adhesively bonded composites Tetric Ceram, SureFil, and Heliomolar RO showed good margin quality. The mean values were obtained with Tetric Ceram (94.4% "perfect margin"), followed by SureFil (93.9%) and Heliomolar RO (93.2%). After mechanical and thermal stress, a slight, insignificant increase of gap-free margins could be found for the materials used with the two-step etch-and-rinse adhesive: SureFil (87.8%), Heliomolar RO (85.4%), and Tetric Ceram (84.9%). The stable margin quality of these materials after artificial aging has also been observed in numerous other studies [37, 38].

After thermomechanical loading, the largest decrease of marginal adaptation of 40.0% was detected for Definite (from 81.4% to 51.8%). These findings are in accordance with further studies on the marginal adaptation of ormocers. In an evaluation of the marginal and internal adaptation of class II ormocer and hybrid resin composite restorations, the enamel adaptation of Definite/Prime & Bond NT restorations was clearly worse compared with that of hybrid composite restorations before, as well as after, load cycling [39]. The unfavorable physical properties of Definite [40, 41], as well as the weak performance of Etch & Prime 3.0 with respect to micro-leakage and bond strength [42, 43], are proposed to be responsible for the deterioration of marginal quality after thermomechanical loading. In a clinical study with class II restorations, Oberlander and co-workers [44] determined a significantly worse marginal adaptation for Definite/Etch & Prime after one year in function than with Solitaire/Solid Bond and thus rated the material as unacceptable according to the ADA acceptance criteria for restorative materials [45].

Prior to loading, only the ion-releasing composite Ariston pHc showed a significantly worse ratio of "perfect margin" (68.1%). This result was to be expected, since the material was inserted into the cavity without the use of the acid-etching technique and dentine bonding. Clinical studies could show that the presence of only a liner (Ariston liner) cannot provide sufficient durability for marginal integrity [46]. As a result, Ariston pHc restorations in class I and II cavities showed failure rates of over 50% after 2 years of clinical service, mainly due to tooth fractures and gap formations [47]. In the meantime, the manufacturer has taken Ariston pHc off the market.

Acknowledgement The authors gratefully thank Mr. StD D. Gebhart for his help with the translation into English.

Declaration of Conflict of Interest The authors declare that they have no conflict of interest.

References

1. Rice JA, Lynch CD, McAndrew R, Milward PJ (2011) Tooth preparation for rest seats for cobalt–chromium removable partial dentures completed by general dental practitioners. J Oral Rehabil 38:72–78
2. Bergman B, Hugoson A, Olsson C-O (1982) Caries, periodontal and prosthetic findings in patients with removable partial dentures: a ten-year longitudinal study. J Prosthet Dent 48:506–514
3. Carlsson GE, Hedegård B, Koivumaa KK (1976) Late results of treatment with partial dentures. J Oral Rehabil 3:267–272
4. Chandler JA, Brudvik JS (1984) Clinical evaluation of patients eight to nine years after placement of removable partial dentures. J Prosthet Dent 51:736–743
5. Wagner B, Kern M (2000) Clinical evaluation of removable partial dentures 10 years after insertion: success rates, hygienic problems, and technical failures. Clin Oral Investig 4:74–80
6. Davenport JC, Basker RM, Heath JR, Ralph JP, Glantz PO (2000) A clinical guide to removable partial dentures. BDJ Books, London
7. Tyson KW, Scott BJ, Yemm R (2006) Understanding partial denture design. Oxford University Press, Oxford
8. Opdam NJ, Bronkhorst EM, Loomans BA, Huysmans MC (2010) 12-year survival of composite vs. amalgam restorations. J Dent Res 89:1063–1067

9. Manhart J, Chen H, Hamm G, Hickel R (2004) Buonocore memorial lecture. Review of the clinical survival of direct and indirect restorations in posterior teeth of the permanent dentition. Oper Dent 29:481–508
10. Van Nieuwenhuysen JP, D'Hoore W, Carvalho J, Qvist V (2003) Long-term evaluation of extensive restorations in permanent teeth. J Dent 31:395–405
11. Bernardo M, Luis H, Martin MD, Leroux BG, Rue T, Leitão J, DeRouen TA (2007) Survival and reasons for failure of amalgam versus composite posterior restorations placed in a randomized clinical trial. J Am Dent Assoc 138:775–783
12. Soncini JA, Maserejian NN, Trachtenberg F, Tavares M, Hayes C (2007) The longevity of amalgam versus compomer/composite restorations in posterior primary and permanent teeth: findings from the New England Children's Amalgam Trial. J Am Dent Assoc 138:763–772
13. Hickel R, Kaaden C, Paschos E, Buerkle V, García-Godoy F, Manhart J (2005) Longevity of occlusally-stressed restorations in posterior primary teeth. Am J Dent 18:198–211
14. Muraki H, Wakabayashi N, Park I, Ohyama T (2004) Finite element contact stress analysis of the RPD abutment tooth and periodontal ligament. J Dent 32:659–665
15. Firtell DN, Jacobson TE (1983) Removable partial dentures with rotational paths of insertion: problem analysis. J Prosthet Dent 50:8–15
16. Beaumont AJ Jr (1990) A clinical view of mandibular premolars in removable partial denture design. Quintessence Int 21:21–26
17. Wegmann U, Seebauer H, Mauksch J (1984) In vitro study of the dynamics of free-end saddle dentures with different supports. Dtsch Zahnarztl Z 41:210–214
18. Craig RG, Peyton FA (1967) Strain on the framework of a mandibular free-end saddle partial denture under load. J Biomed Mater Res 1:263–274
19. Lassila V, Holmlund I, Koivumaa KK (1985) Bite force and its correlations in different denture types. Acta Odontol Scand 43:127–132
20. Borchers L, Jung T, West M (1989) Creep of amalgam fillings under clasp rests. Dtsch Zahnarztl Z 44:802–805
21. Krejci I, Lutz F (1990) In-vitro test results of the evaluation of dental restoration systems. Correlation with in-vivo results. Schweiz Monatsschr Zahnmed 100:1445–1449
22. Mehl A, Gloger W, Kunzelmann KH, Hickel R (1997) A new optical 3-D device for the detection of wear. J Dent Res 76:1799–1807
23. Roulet JF (1987) A materials scientist's view: assessment of wear and marginal integrity. Quintessence Int 18:543–552
24. Aoda K, Shimamura I, Tahara Y, Sakurai K (2010) Retainer design for unilateral extension base partial removable dental prosthesis by three-dimensional finite element analysis. J Prosthodont Res 54:84–91
25. Steiner M, Mitsias ME, Ludwig K, Kern M (2009) In vitro evaluation of a mechanical testing chewing simulator. Dent Mater 25:494–499
26. Peyron MA, Mishellany A, Woda A (2004) Particle size distribution of food boluses after mastication of six natural foods. J Dent Res 83:578–582
27. Beatty MW, Pidaparti RM (1993) Elastic and fracture properties of dental direct filling materials. Biomaterials 14:999–1002
28. Asgar K, Sultfin L (1965) Brittle fracture of dental amalgam. J Dent Res 44:977–988
29. Indrani DJ, Cook WD, Televantos F, Tyas MJ, Harcourt JK (1995) Fracture toughness of water-aged resin composite restorative materials. Dent Mater 11:201–207
30. Braem M, Lambrechts P, Vanherle G (1994) Clinical relevance of laboratory fatigue studies. J Dent 22:97–102
31. Heintze SD, Zappini G, Rousson V (2005) Wear of ten dental restorative materials in five wear simulators—results of a round robin test. Dent Mater 21:304–317
32. Condon JR, Ferracane JL (1996) Evaluation of composite wear with a new multi-mode oral wear simulator. Dent Mater 12:218–226
33. Htang A, Ohsawa M, Matsumoto H (1995) Fatigue resistance of composite restorations: effect of filler content. Dent Mater 11:7–13
34. Lohbauer U, von der Horst T, Frankenberger R, Krämer N, Petschelt A (2003) Flexural fatigue behavior of resin composite dental materials. Dent Mater 19:435–440
35. Turssi CP, Ferracane JL, Vogel K (2005) Filler features and their effects on wear and degree of conversion of particulate dental resin composites. Biomaterials 26:4932–4937
36. Frankenberger R, Tay FR (2005) Self-etch vs etch-and-rinse adhesives: effect of thermo-mechanical fatigue loading on marginal quality of bonded resin composite restorations. Dent Mater 21:397–412
37. Dietschi D, Herzfeld D (1998) In vitro evaluation of marginal and internal adaptation of class II resin composite restorations after thermal and occlusal stressing. Eur J Oral Sci 106:1033–1042
38. Krejci I, Besek M, Lutz F (1994) Clinical and SEM study of Tetric resin composite in posterior teeth: 12-month results. Am J Dent 7:27–30
39. Kournetas N, Chakmakchi M, Kakaboura A, Rahiotis C, Geis-Gerstorfer J (2004) Marginal and internal adaptation of class II ormocer and hybrid resin composite restorations before and after load cycling. Clin Oral Investig 8:123–129
40. Manhart J, Kunzelmann K-H, Chen HY, Hickel R (2000) Mechanical properties and wear behavior of light-cured packable composite resins. Dent Mater 16:33–40
41. Chen HY, Kunzelmann K-H, Manhart J, Hickel R (2000) Setting stress of light-cured packable composite resins (abstract). J Dent Res 79(Spec Iss):246
42. Burmann P, Cardoso PEC, Silveira B, Casagrande L, Anziliero L (2000) The influence of composite resin polymerization techniques on microleakage (abstract). J Dent Res 79(Spec Iss):183–4
43. Cardoso PE, Placido E, Francci CE, Perdigão J (1999) Microleakage of class V resin-based composite restorations using five simplified adhesive systems. Am J Dent 12:291–294
44. Oberlander H, Hiller KA, Thonemann B, Schmalz G (2001) Clinical evaluation of packable composite resins in class-II restorations. Clin Oral Investig 5:102–107
45. American Dental Association (1996) Acceptance program guidelines—restorative materials. http://www.ada.org/prof/prac/stands/restmat.pdf
46. Braun AR, Frankenberger R, Kramer N (2001) Clinical performance and margin analysis of Ariston pHc versus Solitaire I as posterior restorations after 1 year. Clin Oral Investig 5:139–147
47. Krämer N, Garcia-Godoy F, Frankenberger R (2005) Evaluation of resin composite materials. Part II: In vivo investigations. Am J Dent 18:75–81
48. Lloyd CH, Adamson M (1985) The fracture toughness (KIC) of amalgam. J Oral Rehabil 12:59–68
49. Choi KK, Ferracane JL, Hilton TJ, Charlton D (2000) Properties of packable dental composites. J Esthet Dent 12:216–226