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Wear of Ceramic-Based Dental Materials

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Abstract

An investigation is made of wear mechanisms in a suite of dental materials with a ceramic component and tooth enamel using a laboratory test that simulates clinically observable wear facets. A ball-on-3-specimen wear tester in a tetrahedral configuration with a rotating hard antagonist zirconia sphere is used to produce circular wear scars on polished surfaces of dental materials in artificial saliva. Images of the wear scars enable interpretation of wear mechanisms, and measurements of scar dimensions quantify wear rates. Rates are lowest for zirconia ceramics, highest for lithium disilicate, with feldspathic ceramic and ceramic-polymer composite intermediate. Examination of wear scars reveals surface debris, indicative of a mechanism of material removal at the microstructural level. Microplasticity and microcracking models account for mild and severe wear regions. Wear models are used to evaluate potential longevity for each dental material. It is demonstrated that controlled laboratory testing can identify and quantify wear susceptibility under conditions that reflect the essence of basic occlusal contact. In addition to causing severe material loss, wear damage can lead to premature tooth or prosthetic failure.

Keywords

dental ceramics; prostheses; enamel; wear; micromechanics; microstructure; fracture

1. Introduction

Occlusal wear occurs to a greater or lesser extent in natural teeth and in the crowns and fixed dental prostheses used to replace them. Incidental wear facets are not usually a primary concern to dentists, but pathological wear from clenching and grinding (bruxing) can lead to sensitivity, loss of function and esthetics, and infection, especially once the dentin is exposed. Wear rates depend on material microstructures and bite force (Borrero-Lopez et al., 2014), among several other tribological and clinical factors (Lambrechts et al., 1989; Kelly, 1997; Lewis and Dwyer-Joyce, 2005; Kraemer et al., 2006; Heintze et al., 2008; Zhou and

Zheng, 2008; d'Incau et al., 2012; Lee et al., 2012; Mitov et al., 2012; Janyavula et al., 2013; Preis et al., 2013; Stober et al., 2014; Dupriez et al., 2015; Stober et al., 2016; Matzinger et al., 2018; Santos et al., 2018). Dental materials with a ceramic component tend to be relatively hard, leading in extreme cases to accelerated antagonistic wear of opposing dentition. Ceramic-based materials also tend to be brittle, so wear facets may act as sources of crack initiation and propagation (Keown et al., 2012; Zhang et al., 2013c; Scherrer et al., 2017) (Fig. 1). Knowledge of the manner in which wear damage progresses in such cases at the microstructural level is fundamental to the optimization of dental materials for prosthetic applications.

Wear is also of great interest to paleodontologists, in special relation to inference of diet and hence evolution. Wear rates in the enamel of various animal classes depend on the food source, and are dramatically accelerated by the presence of extraneous (dust) or intrinsic (phytolith) particulates (Janis and Fortelius, 1988; Ungar et al., 1995; Ungar, 1998; Fortelius and Solounias, 2000). Excessive tooth wear in herbivores can be life-limiting. Microwear patterns on fossil enamel surfaces are fingerprints of dietary history (Grine and Kay, 1988; Ungar and Sponheimer, 2011; Lucas et al., 2013; Constantino et al., 2016). Studies in the evolutionary biology literature have afforded a useful basis for developing models of tooth wear (Borrero-Lopez et al., 2014; Borrero-Lopez et al., 2015; Borrero-Lopez et al., 2018). Transitions from 'mild' (microplasticity-controlled) to 'severe' (microcrack-controlled) wear can be quantified by a threshold 'wear coefficient', dependent on material properties as well as chewing conditions.

The advent of monolithic zirconia-based dental prostheses (Preis et al., 2013; Zhang and Lawn, 2018) has brought the issue of wear into closer focus. Zirconia ceramics are among the hardest and stiffest of dental materials, with consequent increase in vulnerability of opposing dentition (Stober et al., 2014; Stober et al., 2016). While much attention has been on the wear of zirconia on enamel, relatively little has been devoted to zirconia on other restorative dental materials (Albashaireh et al., 2010). In the present study we examine the relative wear susceptibilities of a representative suite of ceramic-containing dental materials, using a well-established testing protocol with a hard zirconia spherical contact. We demonstrate that the wear responses fall into groups, dependent on underlying microstructures, some with rates higher than that of enamel itself. Results are discussed in relation to competing micromechanisms of deformation and fracture.

2. Materials and Testing

2.1 Materials and microstructures

Representative dental materials studied in this work, along with typical values of their basic properties, are listed and sourced in Table 1. Typical microstructures are shown in Fig. 2 (Zhang et al., 2013b). The zirconias have fine-scale biphasic tetragonal/cubic microstructures (3Y-TZP, 71%t/29%c; 5Y-PSZ, 31%t/69%c) with strong grain boundaries. The graded 3Y-TZP has an infiltrated feldspathic glass component with continuous gradation in properties from porcelain-like at the outer surface to bulk zirconia at depth ~ 150 μm (Zhang and Kim, 2009). The remaining materials in Table 1, including tooth enamel, consist of crystalline phases embedded in a glassy (lithium disilicate, feldspathic ceramic) or

polymeric (composite) matrix. Enamel and lithium disilicate are distinguished by their crystalline phases in elongate rod form, in the former case oriented closely normal to the outer tooth surface.

Specimens of the ceramic-based dental materials in Table 1 were cut and ground from blocks of each dental material into plates 1 mm thick, diamond-polished to 1 μm surface finish (Phoenix 4000, Buehler, Lake Bluff, IL), and stored in distilled water at room temperature prior to testing. Enamel specimens were cut from extracted human molars parallel to the occlusal surface and given the same preparation treatment (Borrero-Lopez et al., 2014).

2.2 Test methods

Specimens for any one material were mounted in a tetrahedral configuration into a rotating ball-on-3-flat tribometer testing machine (Falex Multispecimen, Faville-Le Valley Corp, Bellwood, IL). Full photographic and partial schematic views of the specimen holder and ball are shown in Fig. 3. This test has been widely employed to characterize the sliding wear behavior of ceramics (Hsu and Shen, 1996; Thompson et al., 2004; Borrero-Lopez et al., 2007; Borrero-Lopez et al., 2012). In relation to dental materials, the configuration offers an advantage over others, such as commonly used pin-on-disk or chewing simulators (Lewis and Dwyer-Joyce, 2005; Heintze et al., 2012), in that the ensuing near-circular wear scars reproduce the basic topographical features of clinically observed wear facets while retaining an essential sliding component at the contacts. It also affords a relatively simple evaluation of steady-state wear rates (Hsu and Shen, 1996). A fully dense 3Y-TZP sphere (Luis Aparicio S.L., Barcelona, Spain) of radius $R = 6.35 \text{ mm}$ and surface polish to $< 0.1 \mu\text{m}$ rotating about a vertical axis at a fixed rotational frequency $f = 25 \text{ rpm}$ delivered a normal load $P = 30 \text{ N}$ onto each specimen surface (Borrero-Lopez et al., 2014). This chosen frequency and load combination was at the lower end of the operational specifications for this tester, to conform to typical chewing conditions and to avoid auxiliary damage in the specimens from heating. Artificial saliva solution (LACER S.A., Barcelona Spain) was used as lubricant medium. The tests were interrupted at regular intervals for examination over cumulative durations up to one hour, with provision for precision relocation of the configuration for test continuation.

Near-circular wear scars on the plates were examined by optical microscopy (Epiphot 300, Nikon, Tokyo, Japan) and scanning electron microscopy (FE-SEM; Quanta 3D FEG, FEI, The Netherlands). Worn specimens were washed and cleaned gently in ethanol prior to examination, in order not to dislodge tell-tale adherent surface debris. The sliding distance L across each scar at cumulative test duration t was determined from tetrahedral geometry as $L = 2\pi Rft \sin\theta$, where $\theta = 37^\circ$ is the angle between rotation axis and line from sphere center to specimen contact. Scar radii r were measured optically both normal and perpendicular to the sliding direction, and averaged over the 3 specimens in each test run, i.e. 6 measurements per run. After allowing for an initial 'run in' period for the scar depression to attain the curvature of the sphere, scar volumes V were computed from the simple conformational relation $V = \pi r^4/4R$ (Hsu and Shen, 1996; Borrero-Lopez et al., 2014). A condition that the

test has indeed attained ‘steady state’ is that the data should obey Archard’s law (Hutchings, 1992; Borrero-Lopez et al., 2014)

$$VH/PL = K \quad (1)$$

with V/PL the ‘specific wear rate’ and K a dimensionless ‘wear coefficient’.

The surfaces of the antagonist zirconia sphere were also examined at the end of each run and wear scars evaluated as above. Since this damage is distributed over the circumference of the rotating sphere and is therefore less severe than in comparable chewing simulators (where the contacting agent is held fixed), the wear scars were more superficial and therefore less amenable to comparative quantification.

3. Results

3.1 Images

Images of wear scars in representative materials from Table 1 are shown in Fig. 4. The images were all taken at the end of each test ($t = 60$ min, $L = 37$ m), and are displayed at the same magnification. Sliding direction is left to right. The widely different wear patterns of the selected materials are palpable, from 3Y-TZP (Fig. 4a) to lithium disilicate (Fig. 4d), with tooth enamel (Fig. 4b) and ceramic–polymers (Fig. 4c) intermediate. Since all the zirconia materials in Table 1 revealed indistinguishable wear scars, only the Zpex material is shown in Fig. 4a. Despite the original smoothness of the rotating zirconia sphere, there are discernable scratch track marks across the scar widths, indicative of nascent third-body particles in the test medium.

Higher magnification images of the central scar regions in Fig. 4 are shown in Fig. 5. Individual track widths are on the μm scale, but there is clear evidence of larger-scale surface debris around some of the wider tracks, again increasing in severity from (a) through (d). The tracks in zirconia (Fig. 5a) are comparatively smooth, typical of plastic grooves from microcontacts, with just a few wider tracks showing minor signs of incipient microcracking. Again, all the zirconias in Table 1 showed the same wear track characteristics as those in Fig. 5a. Enamel (Fig. 5b) also show plastic grooving, but now with considerably more evidence of material dislodgement along some tracks (Borrero-Lopez et al., 2014). In the ceramic–polymer composites (Fig. 5c) there is widespread disruption of the subsurface, with smearing of the polymeric matrix material. Finally, the lithium disilicate (Fig. 5d) shows copious particulate ejections from inter-grain microcracking. The buildup of surface debris from (a) through (d) is a classic manifestation of a progressive transition from ‘mild’ to ‘severe’ wear (Borrero-Lopez et al., 2014).

3.2 Wear data

The evolution of the wear scars over a test duration of 60 min is plotted in Fig. 6 as scar radius r versus sliding distance L for each material in Table 1. The large differences in wear rates so evident from the scar sizes in Fig. 4 are reflected by the divergence in data sets in Fig. 6. Note that the data for each material initially rise rapidly up to $L \sim 10$ m as the scars

undergo their run-in phase, before leveling out. Standard deviations in the measured values of r are comparable with the size of the plotted data points. Systematic errors from material variation are not anticipated to be significantly higher than that (except perhaps for enamel, which is notoriously variable from tooth to tooth, orientation to orientation).

Figure 7 replots the steady state data beyond the run-in phase in Fig. 6 as wear volume V versus sliding distance L . The data lie on best-fit straight lines of slope unity in logarithmic coordinates, in accord with eqn. 1. From the contact load $P = 30$ N, it is straightforward to evaluate specific wear rates V/PL from these fits and thence, in conjunction with hardness H in Table 1, wear coefficients K . These evaluations are listed in Table 2, with materials ranked in order of increasing wear rate. The zirconias show the lowest wear rates, lithium disilicate the highest. It is noteworthy that tooth enamel, while not as wear resistant as zirconia, is nevertheless less susceptible than either the feldspathic ceramic or ceramic–polymer composites.

Surface examination of the antagonist zirconia sphere also revealed wear scars. However, the wear volumes were much smaller than those measured on any of the flat specimens, because the damage is distributed around the sphere circumference and not over a static contact. Interestingly, visual observations revealed that scratch density on the sphere was noticeably greater for contact against the softer dental materials. This is attributable to the presence of a greater density of third-party particulate debris in these latter materials, which has been shown to greatly enhance roughness and wear rates (Borrero-Lopez et al., 2014).

4. Discussion

4.1 Analysis of the wear test

The ball-on-3-specimen laboratory wear test configuration used in this study has demonstrable virtues. Loading with a sliding sphere in an artificial oral solution simulates the essential conditions of *in vivo* occlusal contact, and produces damage scars that resemble the basic geometric features of clinically observable wear facets in tooth enamel and prosthetic restorations (Fig. 1). Use of a zirconia sphere represents a worst-case antagonist. The test enables quantification of relative wear susceptibilities against a hard contact, not only on enamel but also on other restorative ceramic-based materials. A perceived limitation might be that well-controlled test configurations of this kind do not accurately embody all the complex elements of real-life mouth motion (DeLong and Douglas, 1983), and that alternative laboratory wear tests of any kind are notoriously subject to variability (Heintze et al., 2005; Lewis and Dwyer-Joyce, 2005; Heintze et al., 2008; Heintze et al., 2012). While this may be so, more ‘realistic’ test arrangements may not have the same amenability to fundamental materials analysis, an important focus of the present study.

The scar images in Figs. 4 and 5, and the removal-rate plots in Figs. 6 and 7, visually and graphically demonstrate the differences in wear resistance between the various dental materials against a hard antagonist. In accordance with current clinical understanding, the zirconias have the highest wear resistance, with the feldspathic ceramic and ceramic–polymer composites markedly lower. It is noted that this latter group of materials has even lower resistance than tooth enamel. The high wear rate of the feldspathic ceramic may be

exacerbated by the presence of thermal expansion anisotropy stresses within individual grains (Cho et al., 1992), evidenced by the appearance of preexisting microcracks in Fig. 2e. Of special interest is the comparatively low wear resistance for lithium disilicate, this despite the fact that it has markedly superior basic properties (E , H , T , S) relative to any material below it in Table 1. This reinforces the contention made elsewhere that stand-alone measurements of such material properties in ‘standardized’ test protocols can be poor proxies for indicating prospective wear and fatigue behavior (Zhang et al., 2013c; Zhang and Lawn, 2019).

It is noteworthy that the wear data for the three zirconia overlap in Figs. 6 and 7. This overlap reveals a relative insensitivity of structural integrity to compositional variants within this material. Special mention may be made of graded Zpex in Table 1, which has low hardness and modulus at the material surface relative to values at the infiltration depth, due to ~ 40 vol% glass content at the outer surface. The modulus gradients lead to a redistribution of subsurface stresses in such a way as to partially negate any potential diminishment in the basic mechanical properties of the material (Zhang, 2012).

There is a question as to how deep the scar damage in Figs. 4 and 5 extends below the surface. Indentation studies have shown that sphere contacts can cause otherwise brittle materials to undergo ‘quasiplastic’ damage deep below the contact surface, where shear stresses attain their maximum values within the Hertzian field (Hamilton and Goodman, 1966; Hamilton, 1983; Lawn, 1998; Peterson et al., 1998), and that this deformation is cumulative under repeat (contact fatigue) loading (Guiberteau et al., 1993; Cai et al., 1994; Zhang et al., 2013c). Generally, high contact loads and pressures are required to generate this kind of deformation, prompting some to dismiss quasiplasticity as a viable damage mechanism (Kelly, 1999). The contact pressures below the sliding sphere, evaluated by dividing the test load P by the scar area (πr^2), never exceed 100 MPa in any of the materials in the present tests, and generally fall well below that level. This is in the range of contact pressures ~ 40 MPa measured directly on patients during heavy clenching in the mouth (Hidaka et al., 1999). The condition for subsurface quasiplasticity within the Hertzian field is that the contact pressure should exceed the yield stress, which may be approximated as $H/3$ (Tabor, 1951; Rhee et al., 2001). It is evident from the relatively high values of H in Table 1 that this condition will not be met in the current tests, confirming that the damage is surface-localized, as it is in ground (Dalladay, 1922; Frank et al., 1967) or sandblasted (Zhang et al., 2004) surfaces. Nevertheless, the micromechanisms by which wear occurs is much the same as those responsible for quasiplastic deformation and microfracture, so that wear, machinability (Xu and Jahanmir, 1994; Padture et al., 1995) and fatigue (Padture and Lawn, 1995) are inextricably interrelated at the microscale.

4.2 Micromechanics

Wear is associated with the cumulation of multiple microcontacts at the sliding interface. Such microcontacts may be the product of surface roughness of an antagonist body, extraneous particles in the food source or, in the present case in the severe wear regime, debris from the abraded surface itself. (Of course, if an initially smooth antagonist undergoes any wear, ensuing surface roughness will augment the wear process.) On the

other hand, in the mouth abrasion debris may be continually washed away, so laboratory testing where particulates aggregate may once more be considered as a worst case. The contact pressure on any individual microscopic asperity can greatly exceed the macroscopic value over the Hertzian contact area, readily generating grain-localized microplasticity and microfracture. The force generated from a single asperity i , say, can be estimated from track half-widths a_i in Fig. 5 using the hardness relation $H = P_i / \pi a_i^2$ (Borrero-Lopez et al., 2015): for example, for a nominal $a_i \sim 1 \mu\text{m}$ and hardness $H \sim 10 \text{ GPa}$ we have $P_i \sim 0.03 \text{ N}$, i.e. 3 orders of magnitude below the net contact force delivered by the zirconia sphere. This value of P_i is on the order of static threshold loads for microcrack initiation below sharp fixed-profile indenters in brittle solids (see Table 1 in (Lawn, 1993)). P_i for any given material depends sensitively on the parameters in Table 1 as well as on indenter sharpness (Borrero-Lopez et al., 2015), and is likely to be significantly lower in repeat translations in aqueous environments (Lawn et al., 1983). Such thresholds appear to be well exceeded in all materials here, except possibly the zirconias, accounting for the cumulative microcracking and subsequent debris ejection.

Figure 2 reveals grain sizes on the μm scale. All materials except the zirconias consist of grains bound by a relatively weak glassy or polymeric matrix. Deformation of these bonding phases and associated interfacial microcracking facilitates detachment and dislodgement of the crystallites. In lithium disilicate this process is readily enabled by microcrack propagation along weak crystal–glass interfaces, a weakness that accounts for the machinability of this material and the consequent proliferation of ejected debris in Fig. 5d. The ceramic–polymer composites may absorb more energy within the bonding phase, affording some resistance to ensuing detachment. In the case of tooth enamel, the micromechanism is slippage of protein sheaths between the relatively long hydroxyapatite rods (He and Swain, 2007); the high degree of orientation of these rods near the enamel surface arguably makes it more difficult to effect dislodgement. The zirconias with their tightly bound grain structures are much more resistant to surface degradation, the flip side of which is the difficulty of chairside adjustment and polishing of zirconia prosthetics in a clinical setting.

4.3 Wear lifetimes

Wear rates in Table 2 lie just above the nominal boundary values $\text{SWR} \sim 1 \times 10^{-6} \text{ mm}^3/\text{Nm}$ and $K \sim 1 \times 10^{-5}$ delineating a transition from ‘mild’ and ‘severe’ wear (Hutchings, 1992; Kato and Adachi, 2002; Borrero-Lopez et al., 2014), i.e. between microplasticity-controlled and microfracture-controlled contact (Borrero-Lopez et al., 2015). This is consistent with the increasing density of surface debris in the micrographs of Figs. 5a through 5d. It is arguable that occlusal contacts in the mouth are generally likely to be less intense than those in the current laboratory test arrangement, except perhaps in bruxers. Again, in this context the data may be taken as representative of a worst case, thus serving as a guide to conservative materials design.

The evaluations of wear coefficients in Archard’s law affords a basis for quantitative prediction of wear-limiting lifetimes of a tooth prosthesis in occlusal contact with a hard zirconia antagonist, especially on molar and premolar teeth where the brunt of chewing is

borne (Hidaka et al., 1999; Borrero-Lopez et al., 2014). Consider a severely worn tooth of width W subjected to N cyclic contacts. Then the net sliding distance is $L = NW$. For the abraded surface to wear through a cylindrical cap of thickness d to the dentin, a volume $V = \pi W^2 d/4$ has to be removed. Inserting these expressions for L and V into Eqn. 1 yields the critical number of tooth contacts

$$N_c = (\pi/4K)HWd/P \quad (2)$$

Evaluation of this quantity for a crown or tooth of width $W = 10$ mm and cap thickness $d = 1.5$ mm at bite force $P = 30$ N in conjunction with the data in Tables 1 and 2 is plotted for each material in Fig. 8. The dashed horizontal line at $N_c = 10^6$ represents a nominal number of contacts per year (Lewis and Dwyer-Joyce, 2005; Lambrechts et al., 2006; Lee et al., 2012), from which long-term survival lifetimes may be inferred. Of course, values of N_c will vary depending on chewing conditions, notably on tooth occlusal load and particulars of tooth geometry, and will be higher for antagonists with lower hardness than zirconia. However, while absolute values estimated from the present tests in Fig. 8 may be open to some uncertainty, the plotted data trends may be taken as a useful guide to relative longevity.

Another way wear can be life-limiting is by increasing vulnerability to fracture (Fig. 1) (Lawn and Lee, 2009). Edge chipping in natural teeth (Constantino et al., 2010) and ceramic-based prostheses is not uncommon, and in severe cases gross tooth splitting can occur (Barani et al., 2015). Not only can wear provide surface flaws for crack initiation, a severely worn occlusion can lower the critical loads required to generate macroscopic cracks (Keown et al., 2012). Mouth motion tests with frictional sliding spheres on dental ceramics, albeit at somewhat higher normal loads than in the present tests, can produce macroscopic partial cone cracks (Ren and Zhang, 2014). Hydraulic pumping of fluids within a cyclic contact can drive surface flaws within the wear scar deep into the sublayer (Zhang et al., 2005). Even if such cracks do not lead to immediate failure, they can severely degrade the remaining strength of a dental prosthesis (Zhang and Lawn, 2019).

5. Conclusions

- i. Wear scars in controlled laboratory tests with sliding zirconia spheres can simulate clinically observed wear facets on tooth enamel and prosthetic replacements from antagonistic contacts.
- ii. Wear rates measured for a suite of ceramic-based dental materials, including zirconias, lithium disilicate, feldspathic ceramic, ceramic-polymer composites and tooth enamel, against a hard, smooth antagonist show zirconias to have the lowest removal rate, lithium disilicate the highest.
- iii. The wear volume is proportional to sliding distance in the steady state region, in accordance with Archard's law.
- iv. High magnification of the scars reveals material detachment debris at individual wear tracks, least noticeable in the zirconias and most in the lithium disilicate. The nature of the debris is determined at the microstructural level.

- v. Wear models based on microplasticity and microfracture beneath multiple asperities within the contact zone account for the relative wear susceptibilities of the different materials.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Fig. 1. Scanning electron micrograph of epoxy replica showing lingual edge fracture (white arrows) from an occlusal wear facet (black arrows) in a veneered 4-unit fixed dental prosthesis. The framework is zirconia (Cercon zirconia, Dentsply Sirona) with fired porcelain (Cercon Ceram S, Dentsply Sirona). The fracture occurred from occlusal contact after 3 years. Reproduced from Ref (Pang et al., 2015).

Fig. 2.

Scanning electron micrographs of microstructures for dental materials listed in Table 1: (a) 3Y-TZP (Zpex, thermally etched), (b) 5Y-PSZ (Zpex Smile, thermally etched), (c) infiltrated zirconia (Zpex graded, acid etched), (d) lithium disilicate (IPS e.max CAD, acid etched), (e) feldspathic ceramic (Vitablocs, polished), (f) ceramic-polymer composite (Enamic, thermally etched), (g) ceramic-polymer composite (Lava Ultimate, polished), (h) tooth enamel (human molar, polished). Figs. (c) and (d) reproduced from Ref. (Zhang et al., 2013a), (h) from (Lawn et al., 2010).

Fig. 3.

View of rotating ball-on-3-flat tribotester. A zirconia sphere of radius 6.35 mm is loaded normally on each of three symmetrically placed specimen plates and rotated at a frequency of 25 rpm. (a) Full photo view, (b) partial schematic showing contact with one of three specimens. The contacts produce well-defined, near-circular wear scars on the flat specimens, from which wear volumes may be evaluated (Hsu and Shen, 1996).

Fig. 4.

Optical microscope images of wear scars in (a) 3Y-TZP (Zpex), (b) tooth enamel (human molar), (c) ceramic-polymer composite (Enamic), (d) lithium disilicate (IPS e.max CAD). Tests with zirconia sphere of radius 6.35 mm in ball-on-3-specimen configuration in artificial saliva, normal load 30 N on each specimen. Sliding direction left to right. Scars at end of test run, i.e. at time $t = 60$ min, sliding distance $L = 37$ m. Scale marker same for all images.

Fig. 5. Scanning electron micrographs of central regions of wear scars in Fig. 4. (a) 3Y-TZP (Zpex), (b) tooth enamel (human molar), (c) ceramic-polymer composite (Enamic), (d) lithium disilicate (IPS e.max CAD). Note increasingly high density of debris in (b), (c) and (d).

Fig. 6.

Radius r of wear scars versus sliding distance L for the materials listed in Table 1.

Fig. 7. Scar volume V versus sliding distance L in the steady-state region for the materials listed in Table 1.

Fig. 8.

Critical number of occlusal contacts N_c for each material with zirconia antagonist to wear through prosthesis cap 1.5 mm thick. Horizontal dashed line at $N_c = 1 \times 10^6$ is indicator of approximate number of annual tooth-tooth contacts. Note logarithmic scale.

Table 1.**Ceramic-based dental materials used in this study ***

Material	Manufacturer	Modulus <i>E</i> (GPa)	Hardness <i>H</i> (GPa)	Toughness <i>T</i> (MPa.m ^{1/2})	Strength <i>S</i> (MPa)
Zirconia					
Zpex (3Y-TZP)	Tosoh	210	14	4.0	1100
Zpex Smile (5Y-PSZ)	Tosoh	210	14	2.5	740
Zpex (graded)	Tosoh	68–210	6.0–14	1.1–4.0	1300
Lithium disilicate					
IPS e.max CAD	Ivoclar Vivadent	105	5.8	2.1	488
Feldspathic ceramic					
Vitablocs	Vita Zahnfabrik	72	5.8	1.1	120
Ceramic-polymer composites					
Enamic	Vita Zahnfabrik	38	2.3	1.1	140
Lava Ultimate	3M ESPE	13	1.1	1.2	225
Enamel					
Molar	Human patient	70	5.0	0.75	40

* Vickers indentation hardness *H*

Piston-on-3-ball biaxial strength *S*

Graded Zpex, values listed outer surface to interior

Tooth enamel, properties also graded, only outer surface values shown

Table 2.

Wear parameters for ceramic-based materials

Material	Specific wear rate (mm ³ /Nm)	Wear coefficient <i>K</i>
Zpex (3Y-TZP)	2.7×10^{-6}	3.8×10^{-5}
Zpex Smile (5Y-PSZ)	3.1×10^{-6}	4.3×10^{-5}
Zpex (graded) *	3.3×10^{-6}	2.0×10^{-5}
Tooth enamel	1.7×10^{-5}	8.6×10^{-5}
Enamic	3.7×10^{-5}	8.4×10^{-5}
Lava Ultimate	7.7×10^{-5}	8.4×10^{-5}
Vitablocs	5.5×10^{-5}	3.2×10^{-4}
IPS e.max CAD	1.2×10^{-4}	6.8×10^{-4}

* For Zpex (graded), value of *H* used to compute *K* is that at outer surface