Chapter 5

The relationship between physical parameters and wear rates of dental composites
Abstract

Materials and Methods: A wear method was developed that mainly simulates attrition wear using a commercially available chewing simulator (Willytec, Germany). In this test, a standardized stylus made of pressable ceramic (Empress) hits flat specimens 120,000 times with a 5 kg weight, a descent speed of 60 mm/sec and a lateral movement of 0.7 mm with a speed of 40 mm/sec under constant exchange of water at different temperatures (325x 5°C/55°C). The volume loss was measured on plaster replicas with the Laserscan 3D (Willytec) and the Match 3D software. Twenty-four experimental and commercial composites (N=8) were tested with a volumetric wear range of between 5.5 and 147 10⁻² mm³. On standardized specimens made of the same composites, the Vickers hardness (H), elastic modulus (E), and fracture toughness (K₁c) were measured. The mean particle size (d) and volume content (v₁) of the inorganic filler were evaluated. Furthermore, a differentiation was made between the main filler with the largest mean size (d₁, v₁₁) and the total filler content (v₁tot).

Results: The best linear regression curve fit with an adjusted R² of 0.908 was found for:

wear index = \frac{d₁^{0.6}}{K₁c \cdot v₁ \cdot v₁₁ \cdot E \cdot \frac{E}{H}}.

Conclusions: The good mathematical fit of the formula may be an indication that the wear method is based on physical properties and that it provides a highly reproducible standard.
Introduction
In human beings wear occurs almost exclusively in the oral cavity of hard dental tissues whose surface cannot be regenerated (non-shedding teeth). The term “wear” applies to the net sum of loss of material whose etiology is composed of different factors, which occur almost simultaneously. In the oral cavity a lot of components contribute to the wear of enamel and dentin, such as the occlusal contacts to antagonist teeth (attrition), chewing on food items, toothbrushing with toothpaste or inhalation of dust (abrasion), acid attacks due to the consumption of acidic fruits and beverages, inhalation of industrial acids or vomiting and regurgitation of gastric juice, for instance, in bulimia and anorexia nervosa (erosion) cases [1]. Fatigue wear due to the cyclic character of the chewing process is a cause of degradation, too [2]. Salivary enzymes or acids may increase wear of restorative materials through corrosive processes [3]. The surface roughness and lubricants like saliva and the salivary pellicle have an effect on the friction coefficient [4]. Besides this, a high biting force and parafunctional habits such as bruxism can accelerate tooth wear. In general, the annual wear rate of molar enamel in non-bruxers is reportedly very low (29 µm) [5].

Wear resistance is a prerequisite for a dental material to be accepted by both dentists and patients. A high wear resistance may contribute to the longevity and durable aesthetics of a dental material [6]. Ideally, wear of a dental restorative material should be similar to enamel. Nonetheless, among the four different categories of dental materials (metal alloys, ceramics, amalgams, composites and unfilled polymers) only ceramics and special metal alloys may have this property [7]. The wear of amalgam restorations is higher than that of enamel but lower than that of composite resins. Although significant improvements have been made, composite resins still exhibit considerable wear in vivo in the long run [8].

Of all these materials the composite resins play a unique role as many variables that derive from their composition directly influence their wear-resistance. Dental composites can be classified as brittle materials: this is demonstrated by the low values of fracture toughness with a relatively high hardness, compared to unfilled polymers. The polymer matrix itself, which is constituted of a highly branched network of dimethacrylates, is fairly brittle, and the silica, glass and glass-ceramic particles included are too fine in order to substantially increase the toughness. Besides, dental composites show very limited plasticity before breaking. The size, shape and hardness of the fillers, the quality of the bond between the fillers and the polymer
matrix and the dynamics of polymerization of the polymer all have an influence on the wear characteristics. The variability of the composition, however, influences the physical parameters, such as flexural strength, fracture toughness, Vickers hardness, modulus of elasticity, curing depth, etc. [9]. Ceramic is a brittle material and due to its crystalline matrix it is less sensitive to attrition wear, however, more sensitive to fatigue resulting from flaws in the material and the materials composition [10].

As wear measurements in vivo are very time-consuming and complicated, wear is generally assessed in wear simulators such as chewing simulators, a pin-on-disc-machine, or rotating wheels where specimens are worn against a metal wheel and an abrasive medium (e.g. a device and method developed at the University of Amsterdam, ACTA). Most often the approaches are related to one or two wear mechanisms that occur in the mouth. Methods like the Ivoclar wear method or the one used at the University of Zurich focus on two-body wear (attrition) [11,12] while the wear simulator developed at the Oregon Health and Science University (OHSU) tries to combine both mechanisms by including different forces, a transversal movement of the sample as well as an abrasive medium [13]. Ivoclar Vivadent conducted a round robin test by preparing specimens of ten different restorative materials (eight composites for direct and indirect use, an amalgam and a ceramic) and sending them to five test centres, which all used different wear simulating methods (Ivoclar, Zurich, Munich, OHSU, ACTA) [14]. The test centres did not know which brands they were testing. After completing the wear tests, they sent the raw data to Ivoclar Vivadent for further analysis. When the relative ranking of the materials was calculated, the results varied tremendously between the individual test centres. The Zurich method showed the lowest discrimination power, ACTA and Ivoclar the highest; however with the ACTA method, 24 specimens per group had been used which resulted in a mean coefficient of variation of 15.3%, while for the Ivoclar method only 8 were sufficient to reach a mean coefficient of variation of 12.5% for vertical wear and 20.2% for volumetric wear.

Different approaches have been taken to relate physical properties, such as fracture toughness and flexural strength, to wear [11,15,16] but no attempt has been made so far to relate the physical parameters and filler characteristics of dental composites to the amount of wear of the same materials in a wear simulator. Generally, one may state that the wear behaviour is dependent on the following factors: modulus of elasticity, fracture toughness, hardness, size of filler, volume of filler within the
composite, hardness of filler, hardness of antagonist and other environmental conditions related to the wear simulation.

The aim of the present study was to search for a possible correlation between the amount of wear and the physical characteristics of particle-filled composites in a simulated environment. For that purpose 11 experimental and 11 commercial composites were subjected to the Ivoclar wear method and their physical properties (Vickers hardness, elastic modulus, fracture toughness, mean particle size and volume content of filler) were evaluated. One of the commercial composites was cured using two different modes.

**Material and methods**

The rationale for selecting the materials was based on the materials’ composition (microfillers, nano-scaled fillers and hybrid composites) and their significance on the dental market. The materials, their batch number, the type of filler, as well as the processing mode are listed in Table 1. As the specimens were fabricated according to the instructions for use of the manufacturer, the mode of fabrication, namely the form of polymerization, varied between the materials. However, the processing mode was identical for both the specimens that were subjected to the wear tests and those that were tested regarding their physical properties.

**Ivoclar wear method**

The materials were placed in round moulds (Ø 7 mm in diameter, 2 mm in depth) and polymerized according to the processing methods described in Table 1. The specimens were then bonded to the specimen holder (SEM-holder, Ø 8 mm in diameter, 2.5 mm in depth; No. 2455-202, Laubscher AG, Träuffelen, Switzerland) with the dual-curing composite luting resin Variolink II (Ivoclar Vivadent).

Before the specimens (n=8 for each material) were tested, they were kept dry at a temperature of 37 °C for 24 hours. After storage, the specimens were polished with 600 grit SiC, 1200 grit SiC and 2500 grit SiC grit by means of a polishing device (Phoenix 4000, Wirtz-Buehler, Düsseldorf, Germany). The specimens were mounted in a dual-axis chewing simulator (Willytec, SD Mechatronik, Feldkirchen-Westerham, Germany). In principle, the load is produced by weights, which are mounted on a bar. This bar is driven by a computer controlled stepper motor by means of programmable logic controllers (PLCs). After the specimens have been mounted in the test
chambers, the chewing simulator is calibrated and the reference points (point “zero”) are defined.

<table>
<thead>
<tr>
<th>Material</th>
<th>Manufacturer</th>
<th>Lot</th>
<th>Main filler type</th>
<th>Specimen polymerization method</th>
</tr>
</thead>
<tbody>
<tr>
<td>Chromasit S4</td>
<td>Ivoclar Vivadent</td>
<td>C15082</td>
<td>Fumed Silica</td>
<td>Ivomat 10 min at 120°C</td>
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<td>Targis Incisal S1</td>
<td>Ivoclar Vivadent</td>
<td>C05051</td>
<td>Ba-Al-Silicate</td>
<td>3 min Spectramat, Targis Power at 130°C</td>
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<td>Targis Incisal S1</td>
<td>Ivoclar Vivadent</td>
<td>C05051</td>
<td>Ba-Al-Silicate</td>
<td>3 min Spectramat, Targis Power at 95°C</td>
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<td>belleGlass enamel light</td>
<td>KerrLab</td>
<td>911422</td>
<td>B-Silicate</td>
<td>20 sec Heliolux DLX, 20 min belleGlass HP</td>
</tr>
<tr>
<td>Estenia Enamel E2</td>
<td>Kuraray</td>
<td>00202C</td>
<td>Glass Ceramic</td>
<td>3 min Spectramat, Targis Power at 130°C</td>
</tr>
<tr>
<td>SureFil</td>
<td>Dentsply</td>
<td>990615</td>
<td>Ba-Al-F-B-Silicate</td>
<td>2 x 3 min Spectramat</td>
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<td>Kerr</td>
<td>006671</td>
<td>Ba-Al-Silicate</td>
<td>2 x 3 min Spectramat</td>
</tr>
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<td>C16761</td>
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<td>D51683</td>
<td>Ba-Al-Silicate</td>
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<td>Ba-Al-F-Silicate</td>
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<td>EM 1</td>
<td>Ivoclar Vivadent (experimental)</td>
<td></td>
<td>Ba-Al-Silicate</td>
<td>2 x 3 min Spectramat</td>
</tr>
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<td>2 x 3 min Spectramat</td>
</tr>
<tr>
<td>EM 3</td>
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<td>2 x 3 min Spectramat</td>
</tr>
<tr>
<td>EM 4</td>
<td>Ivoclar Vivadent (experimental)</td>
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<td>Fumed Silica</td>
<td>3 min Spectramat, Targis Power at 110°C</td>
</tr>
<tr>
<td>EM 5</td>
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<td>Fumed Silica</td>
<td>3 min Spectramat, Targis Power at 110°C</td>
</tr>
<tr>
<td>EM 6</td>
<td>Ivoclar Vivadent (experimental)</td>
<td></td>
<td>Fumed Silica</td>
<td>3 min Spectramat, Targis Power at 110°C</td>
</tr>
<tr>
<td>EM 7</td>
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<td>Fumed Silica</td>
<td>3 min Spectramat, Targis Power at 110°C</td>
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<td>EM 8</td>
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</tr>
<tr>
<td>EM 9</td>
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<td>Ivomat 10 min at 120°C</td>
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<tr>
<td>EM 10</td>
<td>Ivoclar Vivadent (experimental)</td>
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<td>Fumed Silica</td>
<td>Ivomat 10 min at 120°C</td>
</tr>
<tr>
<td>EM 11</td>
<td>Ivoclar Vivadent (experimental)</td>
<td></td>
<td>Fumed Silica</td>
<td>3 min Spectramat, Targis Power at 110°C</td>
</tr>
</tbody>
</table>

Table 1: List of materials, manufacturers, lot numbers, main filler type and method of processing. EM = experimental material.

Ivomat, Spectramat, Targis Power, Heliolux DLX are polymerization devices produced by Ivoclar Vivadent. belleGlass HP is a polymerization device produced by KerrLab.

The chewing simulator contains eight test chambers and each test chamber has a bar and an individual weight. All the bars are linked by a transverse bar that is driven
by a stepper motor. When the stylus comes into contact with the specimen, the whole mass of the weight is released. By means of another stepper motor, lateral movements are also possible. In addition, the simulator includes a thermocycling system using magnetic valves in conjunction with a heating and cooling system controlled by PLCs. The antagonists for the wear simulation were made of pressed IPS Empress ceramic (Ivoclar Vivadent, Liechtenstein) according to the lost-wax technique and they were glazed two times at a temperature of 870°C. The radius of the conical shaped, spherically rounded antagonist was 1.18 mm at a height of 600 µm from the cuspal tip to the base. This geometry was chosen to mimic the curvature of the palatal cusp of the upper first molars of young adults (unpublished data). The antagonists were bonded to aluminium holders with dental resin cement (Dual Cement, Ivoclar Vivadent), which was light-cured for 40 s with an Astralis 5 curing light (560 mW/cm²), and then cured for an additional 10 minutes in a polymerization device (Spectramat, Ivoclar Vivadent).

Weights of 5 kg were put on all test chambers. The descent speed of the antagonist was 60 mm/sec and the speed of lateral movement was 40 mm/sec. The frequency of the antagonist movement was 1.6 Hz. The lateral movement had an amplitude of 0.7 mm. A total of 120,000 cycles of unidirectional antagonist movements were carried out. Thermocycling at a frequency of 325/120,000 load cycles and a temperature range between 5°C and 55°C was included in the wear testing. Each cycle lasted for 230 sec, 105 sec for each warm and cold phase and 10 sec evacuation time between the phases.

After completing the wear generating procedure, impressions of the material were made using a low viscosity vinyl polysiloxane material (President light, Coltène, Altstätten, Switzerland). After four hours, replicas of the impressions were fabricated with white super hard plaster (Fuji Superhard Rock, GC Corporation, Tokyo, Japan) using a vacuum, vibrator and pressure device (2 bar).

The plaster replicas were analysed by means of a commercially available laser scanning device (Laserscan 3D, WIllytec) and the appropriate Match 3D software [18]. In principle, a light beam, which has a width of 22µm and is created by a laser diode, is projected onto a surface and a CCD chip under a triangulation angle of 25°, thus encoding the height of every surface point within the lateral displacement of the light beam on the CCD chip. The specimen is moved along the y-axis by a microstepper motor. After each step, the CCD image of the light line is stored in a frame grabber. A digital signal processor allows the storage and measurement of
8.33 frames (580 lines) per second, which results in a scanning rate of approx. 5000 surface points per second. The precision of 3D data acquisition in flat specimens is calculated to be 2.9µm (±0.5) [18], while a precision of 5-8 µm can be assumed in specimens with a wear facet. The vertical resolution is 2-5 µm, 30 µm on the y-axis and less than 30 µm on the x-axis; objects with a vertical difference of up to 20 mm can be measured [19]. The time required to scan a flat specimen of 1 cm in diameter is about 20 seconds. However, up to eight specimens can be scanned in a row, thus facilitating the process. For this purpose the SEM holders are placed on a special plastic holder, which is fixed on the rotation table of the Laserscan 3D. All eight specimens can be stored in one file. The area around the wear facet was used as reference for the quantification of material loss and the procedures “fit plane”, “subtract plane” and “statistics” were used for the 3D calculation of the material loss. The volumetric loss was calculated by the software. The volumetric wear was calculated by dividing the volumetric loss by the sliding distance and the load and expressed as 10^{-2} \text{mm}^3/\text{N-mm}.

**Vickers hardness**

For each material a cylinder-shaped specimen (10 mm in diameter, 5 mm in thickness) was polymerized according to the processing methods described in Table 1. The surface was ground with 1000 and 4000 grit SiC abrasive paper and polished with 0.3 µm aluminium oxide paste. The specimens were stored in distilled water at 37 °C for 24 hours. The test was conducted in a hardness testing machine (Zwick Type ZHU 0.2, Zwick GmbH, Ulm, Germany) with a test load of 5 N. Six indentations were made on each specimen and the average Vickers hardness calculated.

**Modulus of elasticity**

The test was carried out according to the international standard ISO 10477 [20]. Eight specimens (25 x 2 x 2 mm) of each material were polymerized according to the processing methods described in Table 1 and ground at each side with 1000 grit abrasive paper. The specimens were stored in distilled water at 37 °C for 24 hours. The flexural test was conducted in a universal testing machine (Zwick, Type 1455, Zwick GmbH). The bar-shaped specimens were subjected to three-point bending (span 20 mm, 1 mm/min cross head speed) until fracture. The flexural modulus was calculated according to the standard ISO 10477 [20].
Fracture toughness

The fracture toughness (K\textsubscript{IC}) was determined using single-edge notched specimens in the three-point bending test (SENB), as described by Williams and Cawood [21]. Eight rectangular specimens (26 mm x 5 mm x 2.5 mm) of each material were polymerized according to the processing methods described in Table 1 and subsequently ground flat with 1000 grit abrasive paper. A first notch of approximately 2 mm depth was machined in the middle of the specimen and then sharpened using a razor blade. After storage in distilled water at 37\degree\textCelsius for 24 hours the specimens were subjected to three-point bending (span = 20 mm) at 0.25 mm/min cross head speed until fracture. The test was conducted with a universal testing machine (Zwick Type 1455). After testing, the notch lengths were measured perpendicularly to the fracture surface with the aid of an optical microscope (Olympus SZX12, Olympus GmbH, Hamburg, Germany) and a digital micrometer (Wild MMS235, Wild Heerbrugg Ltd., Heerbrugg, Switzerland) and the K\textsubscript{IC} was calculated.

Filler content

For the composite materials produced by Ivoclar Vivadent, the mean size and the volume fraction of the inorganic filler were known starting from the raw materials’ composition. A distinction was made between the total filler content (v\textsubscript{f,\text{tot}}) and size (d\textsubscript{1}) and the content of the main filler type (v\textsubscript{f,1}), that is, the filler type having the largest mean size. The reason for this distinction was the opinion that the surface roughness and wear debris particles would depend more on the largest filler than on the minor filler. Of course, if only one type of filler was present, v\textsubscript{f,\text{tot}} was equal to v\textsubscript{f,1}.

For composite materials which were not produced by Ivoclar and whose raw materials’ composition was unknown, the filler content was determined via the ignition test and density measurements [22]. The mean filler size was partially deduced from information contained in the product’s data sheet and partially determined by direct measurement of the particles on polished composite surfaces with a scanning electron microscope (Zeiss DSM 962, Zeiss Jena, Germany) with a magnification of 5,000 to 10,000; for the measurements of each material, the value of 15-20 measurements was averaged out.

For the ignition test six half-specimens left over from the flexural test were placed in a furnace for 4 hours at a constant temperature of 600\degree\textCelsius. Subsequently, the furnace
was turned off and left to cool down before the crucibles were taken out. The specimens were dried in a desiccator containing silica gel at room temperature for 30 minutes and subsequently weighed. The weight of the remaining inorganic powder, \( m_2 \), was measured and the filler weight fraction, \( w_f \), was calculated according to the following equation:

\[
w_f = \frac{m_2}{m_1}
\]

where \( m_1 \) is the initial weight of the half-specimen. The average filler weight fraction on the six specimens was then calculated.

Density measurements were carried out in a gas-pycnometer (AccuPyc 1330, Micrometries Instrument Corp., Norcross, GA, USA) on the residue left after the ignition test (\( \rho_i \), filler density) and on six other half-specimens left over from the flexural test (\( \rho_c \), composite density). These last six specimens were dried in a furnace at 80 °C overnight prior to measurement.

Once the two density values were obtained, the volume fraction filler was calculated according to the following formula:

\[
v_{f, \text{tot}} = w_f \frac{\rho_c}{\rho_f}.
\]

**Statistical methods**

For the descriptive analysis as well as the best fit of the mathematical formula and regression curve, the Excel software (Microsoft, USA) has been used. Best fit has been checked in an empirical way.

**Results**

The volumetric wear differed quite considerably ranging from between 5.5 and 147 \( 10^{-2} \) mm\(^3\) on average (Table 2). The coefficient of variation ranged from 11% to 27% with a mean of 17.7%.

The load-displacement curves of the fracture toughness test showed linearity until fracture, so that the linear elastic fracture mechanics assumptions for the validity of the test were fulfilled. The results of all the tests that evaluated the physical parameters as well as the filler content are presented in Table 2. For the sake of clarity, the standard deviations are only shown for volumetric wear and the wear index, as the ranges of variation for the other variables were quite similar: for \( K_{IC} \), \( E \)
and H in the range of ±5-10%, for $v_{t,\text{tot}}$ and $v_{t,1}$ in the range of ±3-8% and for $d_1$ in the range of ±10%.

Table 2: Physical properties and volumetric loss of material. EM = experimental material. Values in parentheses represent the standard deviation.

The best exponential regression curve fit with an adjusted $R^2$ of 0.908 was found for the following formula:

normalized Volume Wear = 11.747 [wear Index]^{1.159}

being:

\[
\text{wear index} = \frac{d_1^{0.6}}{K_{IC} \cdot v_{f,1} \cdot v_{f,\text{tot}} \cdot \frac{E}{H}}
\]

and normalized volume wear = volume loss / (load \cdot x sliding distance)

where the normalized volume wear is expressed in $10^{-2}$ mm$^3$/N·mm, $d_1$ is expressed in µm, $K_{IC}$ is expressed in MPa·m$^{1/2}$, E and H are expressed in MPa and $v_{t,1}$ and $v_{t,\text{tot}}$ are a fraction of unity.
Based on this equation, the uncertainty on computing the wear index was calculated, starting from the deviation standards of the single parameters according to the theory of error propagation. For a product of powers

\[ z = x^m \cdot y^n \cdot \ldots \]

the standard deviation is calculated as

\[
\frac{\Delta z}{z} = \sqrt{\left( m \frac{\Delta x}{x} \right)^2 + \left( m \frac{\Delta y}{y} \right)^2 + \ldots}
\]

The correlation as well as the regression curve is graphically shown in Figure 1.

![Figure 1: Wear index based on the physical parameters of 24 composite materials and correlated to the actual logarithmically transformed volumetric wear \( (10^{-2} \text{ mm}^3 / \text{N mm}) \) of these materials subjected to wear in the Willytec chewing simulator, applying the Ivoclar wear method.](image)

Discussion

A recent review by the first author about wear simulation methods applied in dentistry to evaluate dental materials with regard to their wear resistance revealed that almost all wear simulator devices lack control and regulation of force development during dynamic loading of the flat specimens, which may offer an explanation for the high coefficient of variation of the results in some wear simulators (28-40%) and the poor
reproducibility of wear results if dental databases are searched for wear results of specific dental materials (difference of 22-72% for the same material) [17]. If the rigid validation criteria of the U.S. Food and Drug Administration (FDA) are applied, the conclusion has to be drawn that only one wear simulation device and method can be given the attribute “qualified machine” and “validated method” and that is the Minnesota wear method. Two other methods (Munich, Ivoclar) partly fulfil the criteria, while the other methods are not validated and cannot be validated as the simulators used for these methods are not qualified or qualifiable.

The Ivoclar method is conducted with the Willytec chewing simulator that has been commercially available since 1997. Some 30 devices have been sold since its introduction. However, there are only few publications on the use of this simulator for evaluating the wear of dental materials. In most cases, the simulator is utilized to load crowns and bridges for fracture tests and to evaluate the deterioration of the marginal integrity of restorations placed in extracted teeth. The computer-controlled variables of the device are highly reproducible and have been tested by the manufacturer. Loading with weights is an efficient and cheap method for static loading but creates uncontrollable force impulses that are much higher than the actual weight during dynamic loading. As the weight is placed on the specimen with a descent speed, the force impulse created on the specimen is not equal to the mass of the weight. When the antagonist hits the flat specimen with a mass of 5 kg, a force of 150-200 N is generated within the first 25-30 ms, then fluctuating between 40 and 60 N for the following 100 ms and then the force varies between 20 and 100 N for about 50 ms [17]. The full contact time is 200 ms. As this variation occurs in all eight test chambers, the coefficient of variation is relatively low and has been calculated to be on average 12.5% for vertical wear and 20.2% for volumetric wear if 10 materials have been tested [14]. In the present study the mean coefficient of variation was 17.7% for the 24 tested materials. Therefore, the device can be regarded as being robust. Furthermore, the results are more or less reproducible and the machine is partly qualifiable for the purpose of wear testing [17]. As far as the loading of flat specimens is concerned, the variation in the force profile may be negligible and may level out during the simulation [17], but the consequences on the loading of crowns and bridges have not yet been systematically investigated.

The selected loading force of 50 N can be regarded as a mean value of the physiological biting forces of non-bruxist patients [23]. Higher forces during in vitro simulation lead to higher wear rates [24]. Sliding is an essential requirement in a
wear testing method for dental materials, since the chewing process itself is characterised by small lateral movements of the jaws. In the Ivoclar method, a lateral movement of 0.7 mm increases the wear of a specific composite material by about 8 times [17]. Another essential component is the elimination of worn particles, which is achieved by the constant exchange of water. Thermocycling (5°C/55°C) during wear simulation has a material-dependent effect on wear compared to the effect of constant temperature (20°C, 37°C). In some materials, it may reduce the wear rates while in other materials it may increase the wear rates and in others still, the wear is not affected by thermocycling [25-27].

It has been well documented in the literature that wear increases with increasing number of cycles. Most in vitro wear test methods demonstrate a running-in phase with a steep increase in wear in the initial phase and a flattening of the curve thereafter. From a certain point onwards, wear increases in an even linear pattern [28]. This wear pattern has also been confirmed for the Ivoclar wear method used in the present study. Tests on different composite materials had shown that 40-45% of the final wear is generated already after the completion of 8.3% of the total numbers of cycles (10,000 of 120,000 cycles) [17]. After the first 10,000 cycles there was a linear increase of wear.

There is no agreement in the literature as to which material should be used as the antagonist in wear simulation tests. Some of the wear simulation methods use enamel as the stylus. Enamel can be regarded as an optimal stylus material because of its relevance. However, the scarcity of extracted teeth and the impossibility to standardize natural substrate makes it difficult to use enamel and contributes to the scattering of the results [29]. The Ivoclar method uses IPS Empress ceramic material for wear testing. Recently, it was reported that antagonists made of Empress ceramic material produced a similar amount of wear on various composites as enamel antagonists [29,30].

The method to quantify the wear facets is a well established method and the accuracy has been evaluated elsewhere [18]. A recent analysis on the wear of 16 dental materials comparing the wear results obtained by means of a laser sensor, another optical sensor and the mechanical sensor of a profilometer came to the conclusion that all three sensors are suitable for the quantification of wear facets [31]. Due to its speed and simplicity, the laser sensor has greater advantages over the two other sensors.
Almost all wear simulation methods lack the scientific evidence to prove that the \textit{in vitro} simulation corresponds to the \textit{in vivo} situation, in spite of the fact that publications on three of the simulators tried to establish clinical correlations. This was already established by the Council on Dental Materials in 1989 [32] and not many useful insights have been gathered since then.

Wear is a mechanical process and it is controlled, among other things, by the material’s mechanical characteristics. In the literature, several analytical models of wear processes are mentioned, which are based mainly on observations of the wear behaviour of metal alloys or ceramics. The wear resulting from adhesive processes has been described by the well-known Archard equation [33], which contains hardness as the only material property. Another fatigue wear model also incorporates the strain to failure [34], while models for abrasive wear processes include the elastic modulus, fracture toughness, yield strength and shear strength as additional material properties [35-37]. Analyses of abrasion mechanisms for ceramic materials lead to models regulated by hardness, fracture toughness and the elastic modulus [33].

A few published investigations in the dental field tried to give an insight into the relationship between a material’s characteristics and its wear, but all of them failed to provide a conclusive answer, mainly because only a few parameters were considered [15,38-41].

In the Ivoclar method, wear is a consequence of two-body, direct contact between a material and its antagonist and can be described as mixed wear: adhesion, attrition and fatigue processes may take place simultaneously. In the case of composite materials, the wear process is characterized by the fact that the counterpart hardness is much higher than the composite hardness. The material used in this study, IPS Empress, possesses a Vickers hardness of about 6,500 MPa [29]. Estenia was the hardest composite tested and its Vickers hardness is still 4 times lower than that of the counterpart. The other composite materials are between 7 and 30 times softer than the antagonist material Empress. Even in other dental wear simulators where human enamel is used as the counterpart, high differences exist. The Vickers hardness of human enamel is reported to be in the range of 3,000 MPa and 3,600 MPa [30,42].

The works of Friedrich that describe the wear rate of polymers and composites lead to wear models in which hardness, fracture toughness, volume fraction and size of fillers play a simultaneous role [43]. Friedrich described a change in the wear mechanism, from adhesive to abrasive, above a certain critical pressure that is
proportional to $K_{IC}^{2}/H$, where an increase in hardness may have a detrimental effect on the wear resistance of the composite. A correlation of the formula

$$\text{wear rate} \propto F \frac{H^{1/2}E}{K_{IC}^{2}}$$

where $F$ is a function of critical pressure, volume fraction and size of filler, should lead to good agreements between experimental wear data and theoretical considerations [43]. For the materials tested in the present study, the critical pressure value is very low, compared for example to unfilled or low-filled polymers.

Other literature sources describe the influence of the size of hard particles in a soft matrix on the wear resistance of the composite structure [33]: a finely dispersed hard second phase causes an increase in the flow stress of the matrix which generally leads to increased wear resistance. But in a composite structure where the matrix itself is brittle, hard particles can act as internal stress concentrators, thus enhancing the abrasive wear [44].

Based on all these observations, one can assume that the wear process which took place in this study is mainly of an abrasive nature, dominated by micro-cracking and micro-cutting events, and that parameters like hardness, elastic modulus, fracture toughness, size and volume fraction of reinforcing particles would govern the model.

On the basis of the single material properties listed in Table 2 and taking into consideration the works of Friedrich [43], the formula reported in the present study was found to be the most reliable for predicting the wear behaviour of a dental composite in the Willytec dual-axis chewing simulator. In order to account for the geometry of the antagonist the volumetric wear has been used instead of the maximal vertical wear.

The correlation found, though empirical, may have the following material explanations:

- An increase of the filler size $d_1$ may lead to an increase of the friction coefficient and subsequently of the contact forces [44]. Increasing the filler size may also increase the dimensions of the wear debris. Moreover, it may increase the stress-concentration effect inside the polymer matrix.
- As for the Friedrich equation, hardness plays a negative role on the wear resistance, lowering the critical pressure for the onset of micro-cracking.
- Materials with high fracture toughness are more wear resistant. Increasing the fracture toughness by means of the use of larger filler particles may not have a positive effect (because of the direct proportion between wear and $d_1$).
Increasing the toughness of the polymer matrix should be more successful in decreasing the amount of wear.

- Highly filled composites (i.e. high values of $v_{t_{,tot}}$) provide higher resistance against attrition. The asperities of the antagonist hit a composite surface that is rich in hard particles, and therefore more difficult to plough into. At a constant total filler fraction, if a composite contains two types of filler, one with coarse grains, the other with finer grains, the strengthening effect is inferior.

- The ratio $E/H$ is proportional to the plasticity index [33], which is a measure of the capability of the material to flow. The higher the ratio, the better the composite can withstand micro-cracking through localized plastic flow.

The good mathematical fit of a formula based on physical parameters may be judged as an indication that the Ivoclar wear method is truly based on well defined physical properties and the method provides a high reproducible standard. Moreover, the correlation can aid the work in the development of new, highly-wear resistant composites, focussing the research efforts on the physical properties.

Of course this study presents some limitations. First of all, the conclusions are restricted to composites based on dimethacrylate-based polymer matrices only. The use of polymers based on non-methacrylic monomers may lead to wear data which do not fall within the confidence interval of the present equation. A second restriction regards the type of filler utilized in the examined composites. Only a certain class of fillers (silica, various glasses and some glass-ceramics) and a short range of grain size (from nano-scaled silica up to 1.8 μm glass particles) were included in the present composite materials. The question is whether larger particles or other types of fillers (for example, quartz, nano-scaled alumina and other ceramics) may confirm the theory of the present study. Indeed, the majority of today’s dental composites consist of a dimethacrylate-based matrix that is reinforced with micro and sub-micro glass particles. Therefore, it can be concluded that this investigation is representative of the present market situation.
References


