

## Influence of Surface Characteristics of Different Implant Abutment Materials on Growth of *Porphyromonas Gingivalis*

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The purpose of the present study was to determine the surface physicochemical properties of polished implant abutment materials and to investigate their relationship with the growth of *Porphyromonas gingivalis* *in vitro*. Four groups of the most popular prosthetic materials were used in this study: titanium alloy (Ti), yttria-stabilized zirconium oxide (3Y-TZP), polyether ether ketone (PEEK) composite, and poly(methyl methacrylate) (PMMA). The plate shape specimens (10 × 10 × 0.5 mm) were polished by applying sequential mechanical polishing. Measurements of water contact angle (WCA), surface free energy (SFE) and roughness were performed. Also, the growth of *P. gingivalis* was measured via counting colony-forming units to milliliter (CFUs/mL). The WCA means were significantly different among all groups, and the highest hydrophilicity was observed on the PEEK, whereas the lowest on PMMA surface. All measured surfaces had similarly low SFE values, but Ti, 3Y-TZP, and PEEK demonstrated more expressed polar parts. All means of roughness were beyond the micro-level and were lower than 0.2 μm. The highest CFUs/mL was assessed on the PMMA and it was significantly different from others, whereas the lowest was on 3Y-TZP. The surface roughness had a significant impact on CFUs/mL growth.

**Keywords:** abutment, biofilms, roughness, peri-implantitis, *Porphyromonas Gingivalis*, surface.

### 1. INTRODUCTION

A dental implant has two functional parts: the root and the coronary prosthetic part, and they differ in their properties. The root part is in direct contact with bone and essential for ingrown osteoblasts and successful osseointegration [1–3]. It has to be rough because such a surface enhances osseointegration. For the purpose of increasing the surface area, during the manufacturing process, it can be sandblasted, acid-etched, or plasma sprayed, or a laser can be applied to it to achieve roughness and increase the surface area [4–6]. The prosthetic part starts from the bone level, crosses the mucosa, and acts as an abutment for a dental prosthesis. Therefore, it makes contact with gingival cells and oral microbiota [7].

According to the Human Oral Microbiome Database (HOMD) (<http://www.homd.org/>) [8], more than 700 different bacteria species may exist in the oral cavity and some of them are pathogens. The mucosal barrier typically seals the intraosseous dental implant during the healing process and while it is being used [9], but the conventional transmucosal dental implant generally acts as a bridge between an aseptic tissue area and the colonised oral cavity. Only a firm soft tissue sealing can protect the implant from oral microbiota [10].

The surface of the prosthetic part has to be favorable for gingival cells [7, 11–13], but unfavorable for bacteria and resistant to plaque accumulation [7, 14, 15]. Many authors established that bacterial adhesion is related to the physicochemical properties of the material surface, such as hydrophilicity, surface free energy, and roughness [16, 17].

The peri-implant bone loss might be affected by overloading induced stress at supporting bone [18–21], but plaque-induced inflammation is the predominant factor [22, 23]. Bacterial plaque induces fibroblast response, causing the inflammation of peri-implant soft tissue – peri-implant mucositis [24, 25]. Because of the inflammation, the epithelial barrier is damaged; therefore, untreated peri-implant mucositis eventually reaches the bone and progresses into peri-implantitis [26]. The specific/significant promoter, *Porphyromonas gingivalis* and the presence of it may accelerate the progression of the disease due to the expression of several virulence factors [27–29]. Also, the amount of this pathogen in dental biofilm is associated with the contributing factors of Alzheimer's disease [30]. Thus, from a clinical perspective, it is desirable to avoid bacterial plaque formation around dental implants.

From the biological point of view, the choice of abutment materials, as well as the improvement of their surface physicochemical properties, might be essential for the peri-implant soft tissue health. Unfortunately, no clear evidence-based guidelines related to these choices across

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databases was found. Still, by changing surface properties, the possibilities of bacteria adhesion could be reduced and, at the same time, improved soft tissue sealing.

Polishing protocols of prosthetic materials are highly clinical relevant topics and various improvements are being suggested in the recent studies [31–33]. Across databases, various polishing strategies exist, but these have a lack of standardization. Mainly, commercially available rotary instruments are used for surface polishing in most of the present studies, but these have a lack of information about grain size and difficulties for establishing of identical conditions for all samples. For this study, an extended sequencing polishing strategy following ISO 6344:3 standards for abrasive materials was applied with a clearance of grinding materials and composition [34]. The present study aimed to determine the surface properties of different prosthetic materials, such as the water contact angle (WCA), surface free energy (SFE), and roughness, after sequential mechanical polishing and to investigate their relationship with the growth of *P. Gingivalis* *in vitro*.

## 2. EXPERIMENTAL DETAILS

### 2.1. Specimens preparation

According to the manufacturer's recommendations, the plate shaped samples (10 × 10 × 0.5 mm) were milled by using the computer-aided design and computer-aided manufacturing system (Dental Concept Systems DC1, Dental Concept Systems GmbH) of four commercially available prosthetic materials (Table 1). In total, four groups were prepared (n = 10 for each group): Ti, titanium alloy; 3Y-TZP, yttria-stabilized zirconium oxide; PEEK, polyether ether ketone composite; and PMMA, poly(methyl methacrylate). Samples of 3Y-TZP were additionally sintered in the furnace (Zubler Vario S400, Zubler USA) for two hours at 1450 °C temperature.

**Table 1.** Summary of used commercially available materials

| Groups | Materials  | wt. %  |
|--------|--|--|
| Ti     | Ti6Al4V, DC Titan 5, Dental Concept Systems GmbH, Ulm, Germany                         | Ti – 90 %<br>Al – 6 %<br>V – 4 %   |
| 3Y-TZP | ZrO <sub>2</sub> 3Y-TZP Nacera Pearl, Doceram Medical Ceramics GmbH, Dortmund, Germany | ZrO <sub>2</sub> – 92.5 %<br>Y <sub>2</sub> O <sub>3</sub> – 5.5 %<br>HfO <sub>2</sub> – 1.9 %<br>Others – 0.1 % |
| PEEK   | BioHPP, Bredent GmbH, Senden, Germany  | PEEK – 70 %<br>TiO <sub>2</sub> – 30 %   |
| PMMA   | Brecam Universal, Bredent GmbH, Senden, Germany  | PMMA – 100 %   |

### 2.2. Surface treatment

The surfaces were polished with the decreasing coarseness of water-resistant silicon carbide (SiC) abrasive paper P2000, P2500, P3000, P4000 (Starcke GmbH & Co.) under water cooling/washing. The grain size of SiC abrasive materials described by ISO 6344:3 [34]. The polishing was completed with a diamond polishing paste and a natural brush (Zirkopol, Feguramed GmbH). The polishing cycle of each coarseness was performed using a rotary machine Holzmann metal lathe ED3000ECO

(Maschinenhandel Gronau Inh.) for 60 seconds at 3000 RPM.

### 2.3. Contact angle and surface free energy

The WCA and SFE were measured with an optical tensiometer (CAM 200, KSV Instruments Ltd). Bi-distilled water was used in a heavy phase for measurements. Before measurement, all substrates were cleaned via sonication in hexane and bi-distilled water, and then, they were dried in a vacuum oven at 50 °C for two hours. The measurements were performed at room temperature by placing a small drop (4–7 µL) of a particular liquid on a tested substrate and measuring the WCA of the solvent droplet. The SFE was calculated by using the Owens-Wendt method [35], which provides the dispersive and polar components of a measured substrate's SFE.

### 2.4. Surface roughness

The roughness average ( $R_a$ ) was measured in a contactless way by using the atomic force microscope (AFM) (Agilent 5500 AFM/SPM, Agilent Technologies). Three 0.25 µm<sup>2</sup> areas were randomly scanned for each specimen, and data of the surface topography and roughness were collected.

### 2.5. Bacterial strain and culture media

*P. Gingivalis* ATCC 33277 (Microbiologics) was the strain selected for this study. Following provider recommendations, the lyophilized bacteria were spread on Schaedler agar with vitamin K<sub>1</sub> and 5 % sheep blood (Acumedia) under anaerobic conditions (10 % H<sub>2</sub>, 10 % CO<sub>2</sub>, and 80 % N<sub>2</sub>) at 35 ± 1 °C for 48 hours.

### 2.6. Bacterial formation

All samples were washed with a 70 % ethyl alcohol for 10 minutes, passively dried, packed in sterilization bags, and sterilized by autoclavation at 121 °C for 39 minutes (Hygoclave 90, Dürr Dental). Each sample was placed in a separate sterile tube and inoculated with a 0.5 McFarland standard suspension, which corresponds to a concentration of 1.5 · 10<sup>8</sup> bacteria per 1 mL. It was prepared from the sterile physiological saline and the two-day culture of bacteria. This suspension was supplemented with trypticase soy broth (Becton Dickinson and Company), hemin (10 µg/mL; Sigma Chemical Co.), and menadione (5 µg/mL; Sigma Chemical Co.). The tubes were left in an aerostat for 48 hours (10 % H<sub>2</sub>, 10 % CO<sub>2</sub>, and 80 % N<sub>2</sub>) at 35 ± 1 °C temperature. After cultivation, specimens were removed and transferred to other tubes with 1 mL phosphate-buffered saline (PBS) solution and 1 % proteinase K (50 mg/mL) (Sigma-Aldrich) to detach bacteria from surfaces chemically. Samples with proteinase K and PBS solutions were stored at 37 °C for 60 minutes. Vortexing for one minute was also applied to increase the detachment of bacteria mechanically. The samples were removed from the medium, and the density of the slurries was measured with the McFarland Densitometer (Biosan). McFarland's numerical value was calculated according to the proportion of colony-forming units per milliliter (CFUs/mL).

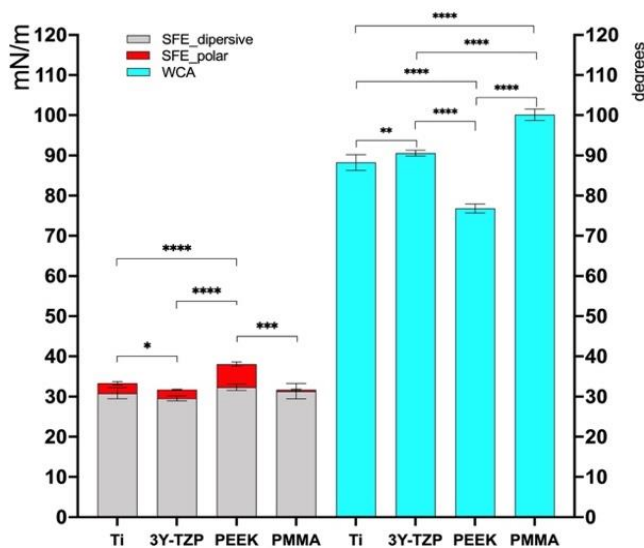
## 2.7. Statistical analysis

The data were analysed using GraphPad Prism 8 software package. The WCA, SFE, roughness and CFUs/mL measurements were analysed based on standard deviations and means. Variables were checked for Gaussian distribution based on D' Agostino-Pearson tests. Multiple comparisons of all measurements were performed among groups by using Anova analysis based on Tukey's tests of variance least significant difference. Also, regressions among measurements were checked. The significance level was 0.05 for all calculations.

## 3. RESULTS

### 3.1. Water contact angle and surface free energy

For the purpose of evaluating the wettability of surfaces, WCA and SFE measurements were performed (Fig. 1). The highest WCA values were observed on the PMMA substrate ( $100.1^\circ$ ) and the lowest on PEEK ( $76.79^\circ$ ), whereas Ti and 3Y-TZP demonstrated intermediate WCA means. A statistically significant difference at a significance level of  $p < 0.05$  was observed among all groups. According to these results of the WCA measurements, the PMMA surface was the most hydrophobic ( $p < 0.0001$ ).



**Fig. 1.** SFE and WCA of samples ( $n = 10$  for each group): titanium alloy (Ti), yttria-stabilized zirconium oxide (3Y-TZP), polyether ether ketone composite (PEEK) and poly(methyl methacrylate) (PMMA). Multiple comparisons by Tukey's test and significant difference at levels: ns –  $p > 0.05$  (non-marked groups), \* –  $p = 0.0243$ ; \*\* –  $p = 0.0031$ , \*\*\* –  $p = 0.0003$ , \*\*\*\* –  $p < 0.0001$

All measured surfaces had similarly low SFE\_total means, and no difference was found between 3Y-TZP and PMMA ( $p > 0.9999$ ). However, Ti, 3Y-TZP, and PEEK had more expressed polar parts. Differences in the SFE polar parts were detected among the polymeric materials, where the values were much higher on the PEEK surface ( $5.71$  mN/m) than on the PMMA surface ( $0.37$  mN/m) ( $p < 0.0001$ ).

However, our finding did not confirm any relation following linear regression between the WCA and SFE

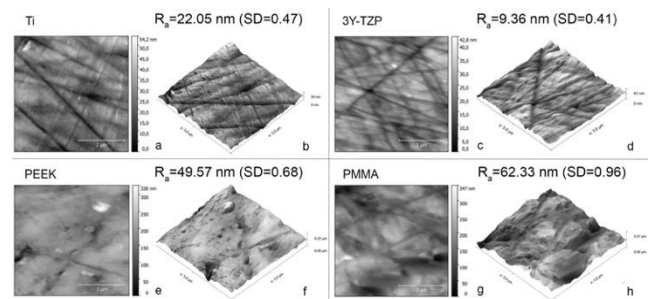
means either their relations with roughness or CFUs/mL.

### 3.2. Surface roughness

The  $R_a$  parameter represented the surface roughness. All means of surface roughness were beyond the micro level and significantly different among each other ( $p < 0.0001$ ) (Table 2). The highest roughness was detected for PMMA  $R_a = 62.33$  nm, whereas the lowest was for 3Y-TZP  $R_a = 9.36$  nm (Fig. 2).

**Table 2.** Multiple comparisons of surface roughness among groups ( $n = 10$  for each) by Tukey's test and significant difference of means at level \*\*\*\* –  $p < 0.0001$

| Tukey's test    | Mean 1 | Mean 2 | Diff.     |
|-----------------|--------|--------|-----------|
| Ti vs. 3Y-TZP   | 22.05  | 9.36   | 12.69**** |
| Ti vs. PEEK     | 22.05  | 49.57  | 27.52**** |
| Ti vs. PMMA     | 22.05  | 62.33  | 40.28**** |
| 3Y-TZP vs. PEEK | 9.36   | 49.57  | 40.21**** |
| 3Y-TZP vs. PMMA | 9.36   | 62.33  | 52.97**** |
| PEEK vs. PMMA   | 49.57  | 62.33  | 12.76**** |



**Fig. 2.** 2D (a, c, e, g) and 3D (b, d, f, h) surface scans, and roughness means with standard deviations (SD) of samples ( $n = 10$  for each group): Ti, 3Y-TZP, PEEK and PMMA. Multiple comparisons by Tukey's test and significant difference among all groups at level  $p < 0.0001$

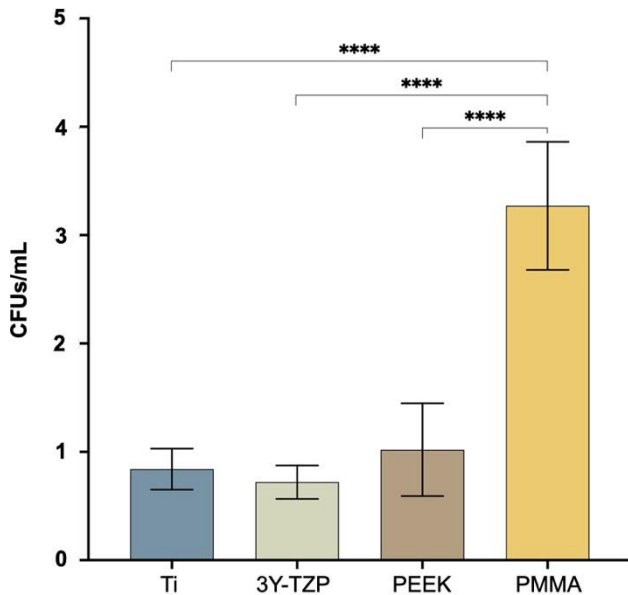
The qualitative analysis revealed that the polished surfaces of polymeric materials appeared to be nonhomogeneous (Fig. 2). Moreover, the difference in altitude for PEEK and PMMA was higher. Both of them had similar changes in the Z-axis: PEEK –  $0.33$   $\mu\text{m}$  and PMMA –  $0.37$   $\mu\text{m}$ . However, the Z-axis for Ti was  $0.54$  nm, and it was  $0.43$  nm for 3Y-TZP. Deeper and sharper grooves and scratches, as well as left prominences, were detected on the polished polymeric materials. No melting signs were detected on surfaces.

### 3.3. Bacterial growth

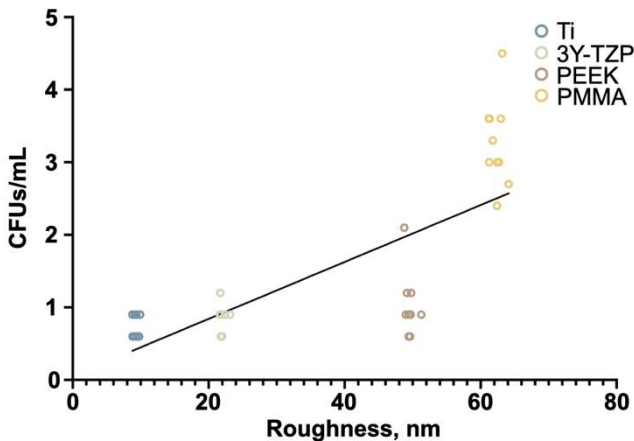
The qualitative bacteriological verification confirmed *P. Gingivalis* colonies in all groups. Following quantitative results, the highest formation of colonies was assessed on the PMMA, and it was significantly different from other groups ( $p < 0.0001$ ) (Fig. 3). The CFUs/mL mean was  $3.27 \cdot 10^8$ . The lowest CFUs/mL mean was on the 3Y-TZP, but no significant differences between Ti, 3Y-TZP and PEEK were detected.

Based on non-parametric Spearman correlation, linear regression showed that the growth of CFUs/mL increased with increasing surface roughness ( $R = 0.5554$ ) (Fig. 4). The significance level between CFUs/mL vs. roughness was  $p < 0.0001$ . The influence of WCA and SFE on the growth of CFUs/mL was not confirmed in this study.





**Fig. 3.** CFUs/mL of *P. Gingivalis* on polished surfaces of abutment materials (n = 10 for each group): Ti, 3Y-TZP, PEEK and PMMA. Multiple comparisons by Tukey's test and significant difference at levels: ns  $-p > 0.05$  (non-marked groups), \*\*\*\*  $-p < 0.0001$



**Fig. 4.** Non-parametric Spearman correlation between CFUs/mL of *P. Gingivalis* and roughness of surfaces, and linear regression:  $p < 0.0001$ ,  $R = 0.5554$

#### 4. DISCUSSION

To prevent biofilm attachment, it is necessary to improve the surface properties of abutments [36, 37]. It is important to ensure biocompatibility for soft tissues as well as to be antibacterial and limit the formation of biofilm. In this way, the traditional method of surface improvement is to perform polishing [37, 38], which was applied in this study.

According to the results of this study, the suggested novel polishing protocol with extended sequencing polishing resulted in especially low roughness of the abutment materials. The linear correlation between  $R_a$  and CFUs/ml revealed that this surface improvement decreased bacteria formation.

This study was conducted evaluating surface physicochemical properties of abutment materials as WCA, SFE and roughness. The WCA parameter reveals

hydrophilicity/hydrophobicity rate, which might influence surface attraction for bacteria [39, 40]. Also, the performed systematic review revealed that SFE and roughness of transmucosal prosthetic parts are mainly related to the biofilm formation [16]. The higher SFE is more favorable for bacteria [41, 42] and roughness of the surface increase biofilm adhesion [37, 40].

In theory, the Ti should exhibit much lower WCA and higher SFE due to the oxide layer that instantly forms at room temperature compared with our data [43]. However, hydrocarbons or other molecules in the atmosphere might contaminate such surfaces too, resulting in higher hydrophobicity [44, 45]. Other researchers also obtained similar WCA and SFE values of the titanium [43, 46].

Due to the chemical composition based on the polymeric chain, both PMMA and PEEK should exhibit a hydrophobic character. However, in this study, the WCA values of PEEK were the lowest compared with other substrates. The ceramic filler of the PEEK could influence this. It is possible that during the polishing procedure part, inorganic filler was uncovered, resulting in a decreased overall surface hydrophobicity.

The findings of this study did not confirm relations between the WCA and strain formation and coincided with the results of another study [47]. Otherwise, existing data from other published studies provide us with ambiguous results – that extra-hydrophilic and extra-hydrophobic surfaces were unfavorable for bacteria – whereas moderate wettability with a WCA of about 90 degrees is the most favorable for bacterial adhesion [39, 48]. However, the findings of other studies established that the WCA was not as influential as the roughness of the surface [49].

With a higher SFE, the surface incorporates a higher quantity of active ions, and such a surface becomes much more favourable for biofilm formation [41]. In this study SFE measurements were very close to previously published data with similar origin groups [38]. The highest SFE was detected on the PEEK surface. These results could also be due to the hybrid composition, and the exposed filler may promote unique surface properties as the oxygen-containing compounds increase the SFE. Also, composite polymers demonstrated a more expressed polar part, which could lead to a higher biocompatibility.

Data on correlations between the increased SFE of the material and the growth of bacteria have been published [42]. Results of this study SFE measurements were very similar for all of the tested groups, but correlations between SFE and CFUs/mL did not exist. However, studies on the interaction between *P. Gingivalis* and the SFE of various materials must be developed more in the future.

To prevent bacterial adhesion, the roughness ( $R_a$ ) must be less than  $0.2 \mu\text{m}$  [16], but also contrary studies showed that bacteria could accumulate on even a smoother surface [50]. However, the accumulation is directly related to the roughness [49, 51]. All findings of this study on surface roughness were below the level of  $0.2 \mu\text{m}$ , which was mentioned in previous studies [16], and the surface smoothness was higher compared with other findings [52]. The same surface polishing changed the surface roughness differently of different materials, and it could be associated with the origin of the materials. It proves the possibility to

achieve the nano-level of a glossy surface only by applying the sequential mechanical polishing of prosthetic materials. Moreover, the high smoothness of surfaces may also influence the previously mentioned surface properties, such as the WCA and SFE.

Changes in surface roughness affect the growth of bacteria on it. The linear regression was detected between the increment of surface roughness on the investigated prosthetic materials and the growth of *P. Gingivalis* ( $R = 0.5554$ ). Also, a positive correlation between subgingival plaque retention and the roughness of the surface was established via *in vivo* findings [53, 54]. The results of this study showed that CFUs/mL of *P. Gingivalis* on the PMMA was higher than on other prosthetic materials, and inorganic materials, such as Ti and 3Y-TZP, as well as PEEK, which were less attractive to *P. Gingivalis*. Different materials showed different surface roughness after polishing, and bacterial growth depended on the roughness. The results of this study revealed an especially low surface roughness of titanium and zirconium oxide compare to other studies [33, 55]. From the clinical point of view, the choice of accurately polished titanium or zirconium could be promising surface at the gingival level. The relevance of surface treatment was established in other studies with gingival fibroblasts [56].

The findings of this study provide a valuable information for following *in vivo* and clinical studies. The late colonizer *P. Gingivalis* was used for this study as the main pathogen of peri-implant disease [57]. Also, the results of this study correlate with the results of the systematic review, which concluded that the majority of the studies (78.9 %) found that surface roughness (with varying degrees of roughness) influenced the bacterial biofilm attachment [58]. Still, the limitation of this study exists because of the monoculture biofilm model. Otherwise, this model is widespread for *in vitro* models, because of reliable implementation under anaerobic conditions [50].

## 5. CONCLUSIONS

The surface roughness was detected as a critical role factor, influencing the growth of bacteria, and fewer CFUs/mL of *P. Gingivalis* was detected on smoother surfaces versus rough surfaces. The highest CFUs/mL of *P. Gingivalis* was assessed on PMMA and was significantly different from others, whereas the lowest was on 3Y-TZP. This study confirmed the relation between changes in surface roughness and CFUs/mL, while other investigated surface physicochemical properties did not. Further studies are needed to investigate surface physicochemical properties of prosthetic materials on the biofilm formation *in vivo* and their influence on clinical outcomes.

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