Evaluation of surface characteristics of titanium and cobalt chromium implant abutment materials

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D Herbst[1], H Dullabh[2], L Sykes[3], C Vorster[4]

ABSTRACT

Introduction: Micro-organism adhesion and plaque formation is affected by surface free energy (SFE), surface roughness, hydrophilicity, surface chemistry, surface charge and the presence of proteins.

Aims: The aim of this study was to assess and compare surface characteristics of surgical grade cobalt chromium alloy (CCM) and of commercially pure titanium (cpTi).

Method: Nine metallic cylinders were machined to precise standards from each material. Surface roughness was measured at four different points on each sample and the average Ra value was calculated for each material. Contact angles were obtained using the sessile-drop method and applied in calculating the SFE. Surface hardness was evaluated by means of a Vickers hardness micro-indentation.

Results: Surface roughness was similar for both metals,but total SFE values and Vickers surface hardness scores showed significant differences ($p<0.0001$).

Conclusion: SFE analysis showed CCM to be more hydrophobic and that oral bacteria might therefore be less adherent than to cpTi. The mean Vickers Hardness scores of the cpTi were significantly lower ($p<0.0001$), suggesting that CCM may be more resistant to surface modifications and surface roughening, thus remaining smoother with less plaque accumulation than cpTi. This study demonstrated that CCM might be a suitable alternative implant abutment material.

ACRONYMS

ASTM F67-95: International Standards for Testing and Materials
CCM: cobalt chromium alloys
cpTi: commercially pure titanium
SFE: surface free energy
VHN: Vickers Hardness score
VO: Van Oss theoretical model
XPS: X-ray photoelectron spectroscopy

INTRODUCTION

The implant-soft tissue interface has been shown to have similarities with the supporting apparatus of natural teeth, having an oral epithelium, sulcular epithelium, junctional epithelium and underlying connective tissue.[1] Maintenance of soft tissue integrity around an implant is crucial. The so called 'biologic width" around implants in usually 3-4mm apico-coronally and comprises of two zones, the coronal 2mm of junctional epithelium, attached directly via a basal lamina and hemidesmosomes, and the remaining 2mm of connective tissue.[1,2,3] Together they form a barrier capable of biologically protecting the peri-implant tissues and preventing bacterial penetration that could jeopardise either the initial healing, or longterm behaviour of the implants.[3] However, the peri-implant mucosa is less effective in limiting the extent of the inflammatory process than is gingiva around natural teeth and may be jeopardised by peri-implantitis.[4] This is a site specific, plaque induced inflammatory process, affecting the peri-implant soft tissues.[4-7] The first stage involves bacterial adhesion to the abutment surfaces leading to peri-implant mucositis.[5] This is followed by a host immune response, characterised by an inflammatory cell lesion rich in leukocytes and vascular structures.[4] It is usually associated with pocket formation, suppuration, swelling, colour changes, bleeding on gentle probing and radiographic evidence of bone destruction, peri-implantitis and late implant failure.[5,8] The associated microflora are complex, but similar to those found in the gingival tissues of natural teeth

1. **D Herbst:** *BSc, BChD, MChD, FCD (SA).* Senior Specialist, Department of Prosthodontics, University of Pretoria.
2. **H Dullabh:** *BChD, MSc, MDent.* Professor and Head of Department, Department of Prosthodontics, University of Pretoria.
3. **L Sykes:** *BSc, BDS, MDent.* Principal Specialist and Associate Professor, Department of Prosthodontics, University of Pretoria.
4. **C Vorster:** *BChD, MChD.* Registrar, Department of Prosthodontics, University of Pretoria.

Corresponding author

LM Sykes:
Principal Specialist and Associate Professor, Department of Prosthodontics, University of Pretoria. Fax: +27 012 323 0561.
E-mail: Leanne.sykes@up.ac.za

affected by periodontitis, with a high proportion of anaerobic gram-negative rods, motile organisms and spirochetes.[7] These may all contribute to implant failure.[6] Healthy implant sulci have few of these organisms and a higher proportion of coccoid cells.[7]

The transmucosal components of implants play a significant role in the maintenance of integration of dental implants.[9] The characteristics of the material of these components are important features in determining implant and abutment success, yet very little is known about the mechanisms of bacterial interactions with many of these materials.[6] An adequate gingival attachment to the transmucosal components is needed to protect the underlying tissues and preserve implant integration.[10] An ideal transmucosal implant component should allow for sufficient epithelial adhesion, but should inhibit bacterial adsorption.[10] Thus, material type and surface characteristics play crucial roles in influencing plaque accumulation and the subsequent bone and soft tissue reactions.[1, 2]

The properties of a material and how it behaves when in contact with body fluids determine the reactions of the surrounding tissues and cells. The surface characteristics of the material are influenced by differences in preparation and sterilization methods and may be very different from those of the bulk of the material.[3] Surface modifications may influence the adhesion of salivary macromolecules as well as of bacteria.[10] Micro-organism adhesion and subsequent plaque formation is affected by surface free energy (SFE), surface roughness[11,10,12] hydrophilicity, surface chemistry, surface charge and the presence of proteins.[13] Titanium exhibits a propensity for increased bacteria and plaque adhesion due to its high SFE.[12,14] Chemical characterisation of surfaces can be evaluated by X-ray photoelectron spectroscopy (XPS), which allows for evaluation of specific elements and their chemical states and can assess the thickness of surface oxide layers as well as contaminates.[3] This information can help in determining the optimal material composition needed to enhance soft tissue attachment and to decrease the amount of plaque accumulation on the abutment materials.[3]

Dental plaque is formed by the adhesion and subsequent stagnation of micro-organisms to form a pellicle on intra-oral surfaces.[15] Adhesion involves a complex mechanism of actions, whereas stagnation may be associated with soft diets, poor oral hygiene, diminished salivary flow and poorly finished restorations.[12] Micro-organism adhesion is divided into four phases: transport of the micro-organism to the surface; initial adhesion, attachment, colonization and biofilm formation.[12] After initial adhesion a protein-rich pellicle film rapidly absorbs onto all surfaces exposed to the oral environment.[10,12] This early, thin biofilm consists of salivary proteins and bacteria such as *Streptococcus mitis* and Actinomyces species.[16] Later colonisers include *Streptococcus mutans* and anaerobic bacteria, contributing to a thicker biofilm, known as dental plaque.[10,16]

Initial adhesion is both a bio-chemical and a physio-chemical process. Bio-chemically, specific ligand-receptor interactions occur between complementary surface components[17], while the physio-chemical mechanism involves the bacterium interacting with the surface, via short-range forces.[12,17] These interactions occur at distances of less than 2nm from the surface.[12] The water layer between the bacterium and the surface, which inhibits attachment, is removed by the dehydrating capacity of the bacteria.[12] Long range forces become active at distances above 50nm and include Van der Waal's and electrostatic forces, which are weak, hence this bacterial adhesion is considered reversible. Subsequent and more specific reactions including covalent, ionic and hydrogen bonding render the attachment irreversible.[18] Each of these force interfaces is associated with energy exchange, the interfacial free-energy being a function of the free-energy of the individual interacting surfaces, i.e. bacterium / liquid interface, surface / liquid interface and solid / bacterium interface. The free energy balance of all interfaces involved in bacterial adhesion can be used to determine whether adhesion is energetically favourable.[19] It can be shown by the following formula:[12,15,20]

$$\Delta G_{adh} = \gamma_{sb} - \gamma_{si} - \gamma_{bi}$$

The interfacial free energy of the adhesion of the bacteria (ΔG_{adh}) is thus related to the solid-bacterium interfacial free energy (γ_{sb}), the solid-liquid interfacial free energy (γ_{si}) and the bacterium-liquid interfacial free energy (γ_{bi}). The SFE of the different species of Streptococci interact with each other and the adhesive potential of a bacterium, in a suspended medium onto a solid substratum, can be determined according to the following theoretical calculation (Figure One):

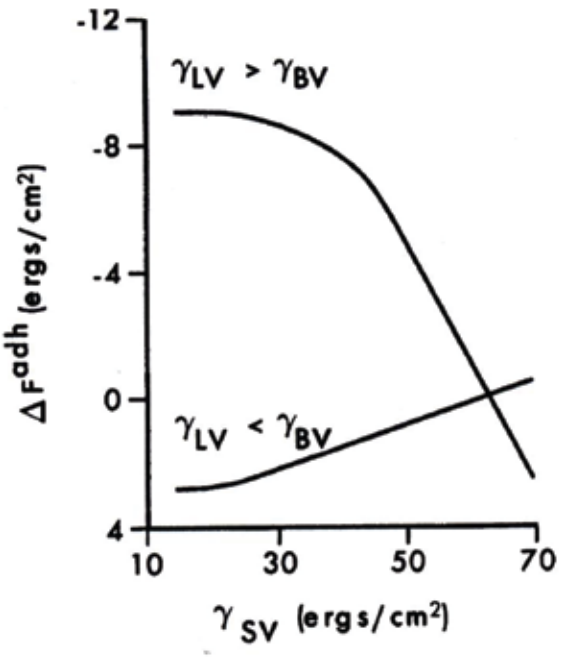

Figure 1: Theoretical equation of the free energy of adhesion (Δ^{Fadh})

SFE of bacterium (γ_{bv})

SFE of substratum (γ_{sv})

Surface tension of the suspending medium (γ_{lv})

When the surface tension of the suspending medium (γ_{lv}) is greater than the SFE of the bacterium (γ_{bv}), the free

energy of adhesion (ΔF^{adh}) becomes progressively negative with increasing SFE of the substratum (γ_{sv}). This results in enhanced bacterial adhesion on low energy (hydrophobic) substrata.[12] The opposite pattern can also be predicted when $\gamma_{lv} < \gamma_{bv}$ which results in enhanced adhesion on high energy (hydrophilic) substrata.[12,20]

SFE of a substratum affects the adherence and spreading of micro-organisms.[15] The substratum SFE plays a role when the water film between interacting surfaces has to be removed before the short–range forces can come into play. Bacteria with low SFE (γ_{bv}), adhere more definitively to low SFE (hydrophobic) substrata, whilst bacteria with a high SFE (γ_{bv}), show greater adherence to high SFE (hydrophilic) substrata.[12,20,21] Eighty percent of early bacterial plaque micro-organisms have a relatively high SFE (γ_{bv}).[21,22] Saliva has a comparatively low SFE (γ_{lv}), and this brings about the $\gamma_{lv} < \gamma_{bv}$ situation frequently encountered intra-orally. It may be concluded that the higher the substratum SFE (γ_{sv}), the more easily bacterial adhesion will occur.

The protein- rich pellicle also has an effect on the final substratum SFE, increasing the values of low SFE substrata and decreasing those of high SFE substrata, thus converging the levels.[12,21] The pellicle coating also reduces the number of adhering bacteria, irrespective of the substrata SFE. However, the original SFE levels retain an influence on bacterial adhesion,[21] with the least retention on low or very high SFE substrata.[23]

Hydrophobic surfaces (low SFE), harbour significantly less plaque than do hydrophilic surfaces (high SFE).[12] The SFE of solids is also directly related to the binding force of bacteria, which implies that bacteria are more easily removed from low SFE solids, than from those with higher SFE.[23] Lowering the SFE of intra-oral hard surfaces and materials, results in retardation of plaque formation and maturation, through a reduction in the initial adhesion and a decrease in the retention capacity of micro-organisms.[12,22]

Surface roughness is even more important in plaque growth than SFE in terms of initial bacterial adhesion and stagnation.[21,23,24,25] Micro-organisms are sheltered in niches of surface irregularities, protected from mechanical shear, whilst they are easily removed from smooth surfaces.[21] Surface roughness also increases available attachment surface area.[21] Crowns with rough surfaces are frequently surrounded by an inflamed periodontium, with a high bleeding index, increased crevicular fluid and increased inflammatory cell infiltrate.[26] This is also the case in implant abutment surfaces. A study comparing rough and polished titanium abutments, found the former to harbour 25 times more bacteria in the subgingival area than did the latter.[12] However, a lowered SFE was associated with a corresponding drop in plaque quantity and pathogenicity.[24] The Ra value is a value depicting surface roughness.[18,24] A threshold Ra value of <0.2 µm has been suggested to prevent bacterial colonisation.[25]

Titanium is a bio-compatible medical and dental material,[27,28] which has been used for crowns, fixed and removable partial denture frameworks and implant components, including transmucosal abutments. The material is available in four different grades, based on the incorporation of small amounts of oxygen, nitrogen, hydrogen, iron, and carbon during purification.[27,28] Its bio-compatibility is due to the surface oxide film offering good resistance to corrosion. However, procedures such as application of acidulated fluorides, the use of metal instruments and tooth brushing have all been shown to increase its surface roughness.[25] Other materials such as gold and ceramics[2] have been investigated for use in abutments, but none were ideal in terms of anti-microbial requirements or soft tissue attachment.[1]

Studies have found significantly lower bacterial adhesion, particularly that of *Streptococcus mutans*, to cobalt chromium alloys (CCM) and ceramic surfaces than to titanium, with the CCM also exhibiting low levels of cytotoxicity.[16] It has been postulated that if CCM has a lower SFE than commercially pure titanium (cpTi), bacteria with a high SFE will be less adherent to the former. Similarly, if CCM is harder than cpTi, it should be more resistant to surface roughening. These factors may make it a suitable alternative material for transmucosal implant abutments.[29]

OBJECTIVES

The aim of this study was to assess and compare the surface characteristics (SFE and hardness) of surgical grade cobalt chromium alloy to commercially pure titanium, Grade 4.

MATERIALS AND METHOD

Nine cylinders, measuring 8mm in diameter and 15mm in height, were machined from commercially pure titanium Grade 4 to ASTM F67-95 and nine from surgical grade cobalt chromium alloy by Southern Implants (Irene, South Africa). Each cylinder was then mounted in an acrylic resin block (Clarofast, Struers) using a mounting press (Imptech HA) and following a cycle of 8.5 minutes at 125°C, after which samples were removed and cooled.

All samples were metallographically polished to achieve a mirror-like appearance and the desired Ra value of <0.1µm, thus ensuring surface roughness would not affect the contact angle measurements. Polishing methods differed for each material and were performed following an industrially accepted protocol, using the Rotopol-11 and Rotopol-1 (Struers).

The cpTi samples were plane ground at 10N for five minutes with water at 300rpm, until a homogenously roughened surface was obtained. A 200mm MD-Piano 220 diamond disc (Struers) was used as the plane grinding surface. This was followed by a fine grinding cycle at 15N for 5 minutes without water at 150rpm. A 9µm monocrystalline diamond suspension (Aka-Mono) was used as an abrasive on a 200mm MD-Largo fine grinding disc (Struers). Final polishing was done at 15N for 10 minutes at 150rpm. The abrasive liquid used was a mixture of 10ml 30% H_2O_2 (Alpha), added to 90ml colloidal silica solution (Akasel). During the final polishing, 5ml of the abrasive liquid was applied to the polishing surface every five minutes. An MD-Nap polishing cloth (Struers) was used as a polishing surface. CCM samples were polished, using the same procedure, except that the initial plane grounding procedure was carried out at 40N for 30 minutes. Samples were removed from the acrylic resin mountings and ultrasonically cleaned in 90% (v/v) alcohol for five minutes and

rinsed with 90% (v/v) alcohol. They were then ultrasonically cleaned in sterile distilled water for five minutes, followed by a final rinse with sterile distilled water. They were dried in an oven for 15 minutes at 30°C, to ensure evaporation of water and were stored in closed Petri dishes in a desiccator at room temperature.

X-ray photoelectron spectroscopy

After polishing and cleaning, the surfaces of each material were characterised to check for surface contaminants, using an x-ray photoelectron spectrometer (Perkin Elmer, Massachusetts, USA).

Atomic force microscopy

An atomic force microscope, the TMX 2000 Discoverer Scanning Probe Microscope (Topometrix, Santa Clara, USA), was used to characterise the surface roughness (Ra value) and to verify that it was lower than 5nm, which would then allow for accurate comparison between the surface free energies of the two material types. Four samples from each material group were randomly selected to test surface roughness. Four measurements were taken on each sample, at random positions, to represent the overall surface roughness and the average values recorded. The data were pooled and the mean Ra for the sample determined.

Wettability

Contact angles were obtained, using the sessile-drop method with a Standard Contact Angle Tension meter (Ramé-Hart, New Jersey, USA). One µl droplets were applied to each sample. Images were then captured with a video camera and the shape of the drop was automatically converted to the contact angle with an image analysing system. Contact angles were measured at room temperature (22OC), during the first second after application of the droplet onto the surface. Three probe liquids of different polarities were used to characterise the nine samples from each material group by determining the respective contact angles. These liquids were: di-iodomethane (Fluka, Switzerland), formamide (Calbiochem, USA) and distilled water.

Surface free energy determination.

The final contact angle used for calculation of SFE was the average of the left and right contact angles. The SFE of the two different materials was calculated using the Van Oss (VO) theoretical model. The VO model yields the dispersive ($\square^{LW}$) and the polar acid-based ($\square^{AB}$) components which are further divided into an acidic ($\square^{+}$) and a basic ($\square^{-}$) part according to the following equation:

$$0.5(1 + \cos)\square_L = (\square_S^{LW}\ \square_L^{LW})^{½} + (\square_S^{-}\ \square_L^{+})^{½} + (\square_S^{+}\ \square_L^{-})^{½}$$

where $\square$ refers to the SFE, the subscripts L and S to the liquid and solid, the superscript LW refer to the dispersive component, and the + and – to the acid and base components. The contact angle measurements of three different liquids (of which two were polar liquids) with known $\square_L^{LW}$, $\square_L^{+}$, and $\square_L^{-}$ values, were recorded on an Excel spreadsheet and the above mentioned equation was calculated three times to determine the $\square_S^{LW}$, $\square_S^{-}$, and $\square_S^{+}$ for cpTi and CCM. The total SFE of each specimen was then calculated as the sum of its dispersive and acid-based components in the following equation:

$$\square_S = \square_S^{LW} + \square_S^{AB}, \text{ where } \square_S^{AB} = 2\ (\square_S^{+}\ \square_S^{-})^{½}.$$

The possible effect of spreading pressure was considered, being the contribution to surface free energy of the adsorption of an external layer from the atmosphere. If the SFE is higher than 60ml/m^2, the spreading pressure must be taken into account in applying a correction factor for final SFE calculation. In the present study, SFE values are lower than this limit and the effect can be neglected.

Surface hardness

Surface hardness is the resistance to deformity, evaluated by means of a Vickers hardness micro-indentation test, using a Vickers Hardness Tester (Future-Tech, Tokyo, Japan). A Vickers diamond pyramid was indented into the surface of each sample at a 100N applied load, for five seconds. The resulting indentation was imaged microscopically and the final Vickers Hardness score (VHN) was automatically calculated by the instrument.

STATISTICAL ANALYSIS OF DATA

Data was captured on an Excel spreadsheet, designed to calculate SFE (mJ/m^2) by employing the Van Oss theoretical model. The observed data for the SFE components, including $\square_S^{LW}$, $\square_S^{AB}$, $\square_S^{+}$, and $\square_S^{-}$, were summarised by metal surface, using the descriptive statistics of mean and standard deviation. Similarly, Vickers Hardness scores and atomic force microscopy measurements were summarised. The two different metal surfaces were compared with respect to their SFE and Vickers Hardness scores, using Student's two-sample t-test. The latter was also done for contact angles, recorded for each of the three liquids. Testing was done at the 0.05 level of significance (p-value).

RESULTS

The Ra values for surface roughness are shown in Table 1. These ranged from 4.69nm to 3.32nm, were similar for both metals, and were all less than 0.1µm.

Table 1: Atomic force microscopy results (Ra values in nm).

Metal	Mean	SD	Minimum	Maximum
Commercially pure Titanium (grade IV)	4.35	0.35	3.89	4.69
Surgical grade Cobalt Chromium alloy	3.53	0.18	3.32	3.76

The X-ray photoelectron spectroscopy (XPS) analyses revealed clean surfaces, free of contaminants that could affect the SFE of all samples. The process characterised the elemental composition of the superficial 1-10nm of the surfaces. The cpTi samples had titanium (Ti) and oxygen (O_2) as their main elements, with limited traces of carbon (C) also being detected. The CCM alloy samples were characterised with cobalt (Co), chromium (Cr), molybdenum (Mo), oxygen (O_2) and carbon (C).

Table 2: Mean contact angle, by liquid and metal surface (mJ/m^2)

Metal	Water	Di-iodomethane	Formamide
Commercially pure titanium (grade IV)	40.59	41.89	34.3
Surgical grade cobalt chromium alloy	49.23	46.16	51.24
P-value	0.0019	<0.0001	<0.0001

Results of the sessile drop measurements utilising water, di-iodomethane and formamide for each metal type are depicted in Table 2.

Surface free energy values were calculated with the Van Oss theoretical model using the contact angle measurement results. Mean SFE values were 47.91mJ/m^2 for cpTi and 38.11mJ/m^2 for CCM samples (Table 3).

Table 3: Mean and standard deviations of contact angle measurements, Lifshitz-van der Waals (γ_S^{LW}), Lewis acid-base (γ_S^{AB}), Lewis-acid (γ_S^+), Lewis-base (γ_S^-) surface energy components and total SFE (γ_S), by metal surface.

SFE (mJ/m^2)					
Metal	γ_S	γ_S^{LW}	γ_S^{AB}	γ_S^+	γ_S^-
Commercially pure titanium (Grade IV)	47.91	38.65	9.26	0.63	38.08
SD	1.69	0.84	0.56	0.11	2.47
Surgical grade cobalt chromium alloy	38.11	36.39	1.72	003	40.47
SD	1.44	0.18	0.45	0.01	2.20
P-value	<0.0001	<0.0001	<0.0001	<0.0001	0.4810

A significant difference (p<0.0001) was obtained between the total SFE values (γ_S) of the cpTi and the CCM alloys. The mean polar component (γ_S^{AB}) ranged between 1.72mJ/m^2 for cpTi and 9.26mJ/m^2 for CCM samples. Both metal surfaces were strongly basic (γ_S^-) but weakly acidic (γ_S^+).

The Vickers Surface Hardness scores for the nine areas measured in each of the metals is depicted in Table 4. This score was significantly higher for the CCM samples than for the cpTi (p<0.0001; 443.85 vs. 247.98).

Table 4: Mean and standard deviation of VHN.

Metal	Mean	SD	Minimum	Maximum
Commercially pure Titanium (grade IV)	247.98	0.82	246.25	249
Surgical grade Cobalt Chromium alloy	443.85	1.22	442.3	445.6

DISCUSSION

The surface properties of a material, including surface roughness and SFE, are important determinants for bacterial adhesion and also for the formation of biofilm. Little information is available for SFE interactions of cpTi and CCM abutment materials. Solid surfaces with an Ra value below 0.1µm have no effect on the contact angle measurement for determination of a substratum SFE.[12] Quarrymen *et al* reported that a threshold Ra value of below 0.2µm will not further influence the degree of bacterial adhesion to a surface. In this study, all samples were polished until a surface roughness of <0.1µm was obtained. The cpTi samples yielded Ra values with a mean of 4.35nm (0.35), whilst the mean value for the CCM samples was 3.53 (0.18), thereby allowing for comparison of the total SFE and of the SFE for the components.

Polishing to a final finish was more difficult for the cpTi samples. This may be explained by the fact that the mean Vickers Hardness score of the cpTi was 247.98 (0.82), compared with 443.85 (1.22) for the CCM. This difference was statistically significant (p<0.0001) and may contribute to the poorer wear resistance of titanium.[27]

Material type, surface modifications and surface roughening have all been shown to play an important role in bacterial adhesion to implant abutments.[5] Researchers have found that smoother abutment surfaces are related to decreased plaque accumulation and reduced incidence of peri-implant infections.[18] In this regard, CCM may be more resistant to surface modifications and surface roughening during daily oral hygiene procedures than cpTi. The relative softness of cpTi makes it more susceptible to progressive roughening during professional or personal hygiene procedures.[18] This study confirmed these findings.

Chemical characterisation of surfaces involves analysis of surface features by means of their XPS scores, enabling the assessment of the chemical state of specific elements and the detection of organic and other contaminants.[6] Preparation and polishing methods have been shown to introduce contaminants to the surface, which tend to mask the properties of the underlying material.[3] In this study, XPS analysis was used to confirm the sample surfaces were adequately cleaned after polishing. CpTi surfaces were found to be repeatedly contaminated, with high readings of carbon and traces of sodium, especially when Teepol soap was used before a final rinsing with sterile water. CCM samples did not show significant contamination of the surfaces, irrespective of the cleaning procedure that was used. This confirms results from a study by Rompen *et al*, which reported a strong binding of proteins and amino-acids to titanium, which created a surface difficult to clean with normal decontamination procedures, following polishing and laboratory handling.[3] CCM may provide a surface more easily cleaned than a cpTi surface and one that is less hazardous to the oral environment.

Although several methods are used to measure contact angles on solid substrata, the sessile drop still remains standard.[13] Contact angle measurements were evaluated with three different liquids, including water, formamide (polar liquid) and di-iodomethane (apolar liquid) and were used to determine the total SFE of both metals. The cpTi samples exhibited a mean total SFE of 47.91 (1.69)mJ/m^2 which demonstrated a moderate hydrophobic character. CCM alloy samples yielded a statistically significantly different (p<0.0001) mean total SFE of 38.11 (1.44)mJ/m^2 and were more hydrophobic, suggesting oral bacteria might be less adherent than to cpTi.

The Lifshitz-van-der Waals (non-polar component) and the Lewis acid-base (polar components) of SFE were also determined through the mathematical equation in Excel from the VO theoretical model.[30] The polar component was further divided into an acidic or electron-acceptor component (γ^+) and a basic or electron-donor component (γ^-).[11] This study found a pronounced basic character (γ_S^-) of the two metals investigated, with a mean of 38.08 (2.47)mJ/m^2 and 40.47 (2.20) mJ/m^2 for the cpTi and CCM samples respectively. Sardin *et al*.

reported that this component of SFE has little or no effect on streptococci adherence to intra-oral substrata. However, others have found that the determination of the Lifshitz-van-der Waal ($\square_S^{LW}$) and the Lewis acid-base ($\square_S^{AB}$) components are important for anticipating bacterial adherence on different abutment materials.[11] Bacterial adhesion to a surface has been shown to be more dependent on the balance between the Lifshitz-van-der Waal ($\square_S^{LW}$) and the Lewis acid-base ($\square_S^{AB}$) interactions.[11]

Smooth titanium implant abutment surfaces can be colonised by pioneer oral bacteria,[11] owing to their inherently high total SFE.[12] This colonisation may lead to the initiation of peri-implant mucositis and possible subsequent peri-implantitis, especially in patients with poor oral hygiene. In this study, CCM exhibited a lower total SFE than that of cpTi, which may yield a surface that is less prone to colonisation with oral bacteria.

CpTi is susceptible to surface modifications and roughening brought about by daily oral hygiene procedures.[27] The CCM is harder and thus is more resistant to wear and to surface modifications that may be introduced during routine oral hygiene procedures.

CONCLUSION

Surface characteristics, including total SFE and surface roughness, play a significant role in the initial bacterial adhesion to, and colonisation of, intra-oral hard surfaces. This study demonstrated that CCM might be a suitable alternative implant abutment material. CCM not only yielded a lower total SFE than cpTi, making it more wear resistant, but also presents a surface to which intra-oral bacteria with a high SFE would be less adherent .

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