Effects of pH and Elevated Glucose Levels on the Electrochemical Behavior of Dental Implants

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Implant failure is more likely to occur in persons with medically compromising systemic conditions, such as diabetes related to high blood glucose levels and inflammatory diseases related to pH levels lower than those in healthy people. The aim of this study was to investigate the effects of lower pH level and simulated-hyperglycemia on implant corrosion as these effects are critical to biocompatibility and osseointegration. The electrochemical corrosion properties of titanium implants were studied in four different solutions: Ringer’s physiological solution at pH = 7.0 and pH = 5.5 and Ringer’s physiological solution containing 15 mM dextrose at pH = 7 and pH = 5.5. Corrosion behaviors of dental implants were determined by cyclic polarization test and electrochemical impedance spectroscopy. Surface alterations were studied using a scanning electron microscope. All test electrolytes led to apparent differences in corrosion behavior of the implants. The implants under conditions of test exhibited statistically significant increases in $I_{corr}$ from 0.2372 to 1.007 $\mu$Acm$^{-2}$, corrosion rates from 1.904 to 8.085 mpy, and a decrease in polarization resistances from 304 to 74 $\Omega$. Implants in dextrose-containing solutions were more prone to corrosion than those in Ringer’s solutions alone. Increasing the acidity also yielded greater corrosion rates for the dextrose-containing solutions and the solutions without dextrose.

Key Words: corrosion, implants, diabetes, pH, EIS, SEM

INTRODUCTION

Titanium is considered one of the most ideal metals for in vivo applications because of its excellent biocompatibility.1,2 Therefore, titanium and its alloys have been used extensively in the past several decades as materials for medical devices, orthopedic implants, and dental implants.3 In implantology, it is necessary to know the interaction between biomaterials and the oral or physiological environment and surrounding tissues, especially the longevity of these biomedical implants.4 Because corrosion is responsible for gradual degradation of materials through the release of metal ions into the body environments, the biocompatibility of a material is correlated with its corrosion tendency.5 In contrast to these properties, passivating alloys can generate metal ions that react with human fluid, especially with chloride ions, forming complexes and precipitates. In addition, they can form hydroxides or oxides with water and produce a change in the pH.6 These pH changes can produce potential and current gradients, so there is a possibility of local acceleration of corrosion on some areas of the implant. A pH value can vary from 5.4 until 7.8 during inflammatory process.7 Although implant treatment is very successful in healthy persons, implant failure is more likely to occur in persons with medically compromising systemic conditions, such as diabetes,8 and other inflammatory diseases associated with lower pH values.9 The risk to patients with diabetes is particularly disconcerting because of the growing incidence of this disease.10

It has been predicted that the number of patients with diabetes mellitus worldwide will...
increase from an estimated 135 million in 1995 to 300 million in 2025.11

There are two types of diabetes mellitus. Type 1 diabetes mellitus is caused by the autoimmune destruction of \( \beta \)-cells, which results in partial or complete insulin deficiency. Type 2 diabetes mellitus is characterized by variable degrees of insulin resistance in peripheral tissue, impaired insulin secretion, and increased glucose production. Although the common symptoms of diabetes that contribute to the failure of endosseous dental implants are not exactly known, poorly controlled diabetes mellitus is associated with adverse systemic sequelae such as increased susceptibility to infection, delayed wound healing, and microvascular complications.12,13

Because corrosion of endosseous implants is a primary variable in the response of bone and other periodontal tissues to implant placement, several studies have used a variety of electrochemical techniques to measure the corrosion properties of implant materials and have documented the corrosion of many alloys in biological environments.14,15 However, these studies have generally been restricted to the use of flat or cylindrical “bulk” samples in commercial electrolytes (ie, saline, serum, or artificial saliva).16,17

The aim of this study was to investigate the effect of pH and dextrose on the electrochemical corrosion behavior of actual dental implants. Ringer’s solution with 2 different pH values (5.5 and 7.0) and 2 different dextrose levels (0 and 15 mM dextrose) was used to simulate clinical conditions and diseases that are relatively more common in elderly patients.

**MATERIALS AND METHODS**

Twenty screw-type titanium implants (Neoss AB, Mölnlycke, Sweden) 4 mm in diameter and 11 mm in length were used in this study. This implant has an altered surface, which has been subjected to a multistage preparation and cleaning procedure during production. The result of this process is a biomodal surface presenting a coarse level of surface roughness. The implant has thread-cutting and thread-forming features to facilitate firm stability in all qualities of bone. The implant is double threaded and designed with a positive tolerance to achieve compression and increase stability in poor bone quality. All implants were received in the original manufacturer’s packaging and opened under sterile conditions.

The electrochemical properties of titanium dental implants were studied in the following four different solutions (\( n = 5 \)):

- **Group R7**: Ringer’s physiological solution at pH = 7.0;
- **Group R5**: Ringer’s physiological solution at pH = 5.5;
- **Group RD7**: Ringer’s physiological solution containing 15 mM dextrose at pH = 7;
- **Group RD5**: Ringer’s physiological solution containing 15 mM dextrose at pH = 5.5.

Dextrose (15 mM) was chosen because it is equivalent to a blood sugar level of 270 mg percent, which is common in patient with uncontrolled diabetes or undiagnosed type II diabetes.12,13

Electrochemical measurements were carried out with a computerized CHI Instrument 660 B System (CH Instruments Inc, Austin, Tex) in a conventional three-electrode cell. The cell components were washed several times with 70% ethanol and kept in alcohol to ensure sterility. The cells were then rinsed with sterile double-distilled water, filled with sterile medium, and sealed to avoid contamination. A platinum wire was used as the counter electrode. The reference electrodes were a saturated calomel electrode. The working electrode was a titanium dental implant specimen with an exposed testing area of 0.3 cm\(^2\). The volume of the electrolyte in the corrosion cell was 50 mL. The water-jacketed cell was connected to a constant temperature circulator operating at 37 ± 1°C to simulate mouth conditions. The working electrode was placed in the holder/electrode assembly when the electrolyte reached the test temperature. During each experiment, solutions were mixed with a magnetic stirrer.

Before testing, all test samples were allowed to reach a steady-state potential (open circuit potential [OCP]) for a period of two hours. After the resting potential was established, electrochemical impedance spectroscopy (EIS) was used to evaluate the corrosion mechanism of the implants and the characteristics of any coating that may have formed. The EIS measurements were made at the OCP using a 5 mV amplitude sinusoidal signal over frequencies ranging from 100.000 to 0.02 Hz.

Tafel and cyclic polarization curves were deter-
mined in a similar manner with respect to reference electrode. Tafel curves were recorded by scanning the potential −400 and +200 from the OCP at a sweep rate of 5 mV/s. Cyclic polarization curves were determined using a rate of 5 mV/s from 100 mV below OCP to a vertex potential of 1700 mV over OCP and a final potential of 0 mV. From these curves, the corrosion potentials (E_{corr}) and the current densities (I_{corr}) were determined by extrapolating the anodic Tafel regions. The corrosion rate in mils per year (mpy) was calculated from the corrosion current using Faraday’s law equation. Hysteresis in the polarization curves was used to characterize alloy susceptibility to pitting and crevice corrosion (Table 1).

Each measurement was repeated three times per sample and mean values for corrosion parameters were calculated. All data were statistically compared using analysis of variance and Tukey’s test (α = 0.05). These tests were chosen as the multiple comparison technique when necessary.

To evaluate the changes to the surface topography of the implants before and after corrosion analyses, the implants were examined at 1000 magnification using a scanning electron microscope (SEM; QUANTA 400, FEI Company, Eindhoven, Netherlands) at an accelerating voltage of 10 kV.

## RESULTS

The representative cyclic polarization curves for groups are provided in Figure 1 to show the range of corrosion behavior of the dental implant exposed to different electrolytes. All groups had an active-to-passive transition behavior. In this study, all the groups showed few hysteresis loops between the forward and reverse scans (Figure 1), indicating that the titanium dental implants for all test conditions were prone to pitting corrosion.

The Tafel plots of the polarization curves from the titanium dental implant electrodes immersed into different solutions are shown in Figure 2. Values of the electrochemical corrosion parameters regarding corrosion potentials, current densities, and corrosion rates are shown in Table 2.

The material with higher current density at a given potential is more prone to corrode. The ranking of the I_{corr} values (μA/cm²) among test groups was as follows: RD5 > RD7 > R5 > R7. All the differences between the groups were statistically significant (P < .05). The materials with higher I_{corr} at a given potential were more prone to corrode. Group R7 had the most passive density value, whereas Group RD5 had the most active critical density value (Table 2).

The corrosion rates of the implants in solutions with dextrose were significantly higher than those with Ringer’s solutions alone (Table 2). Differences in corrosion rates between the same solutions with different acidities were also statistically significant (P < .05). The corrosion rates of the groups confirmed that group RD5 at 8.085 mpy exhibited the most mass loss.

Figure 3a and b shows the EIS and equivalent
The fitting parameters of the EIS are summarized in Table 2. The results of EIS measurements showed that $R_p$ significantly decreased with the presence of dextrose in the medium ($P < .05$). Also, increased acidity yielded dramatically decreased polarization resistance values for both Ringer’s solutions and Ringer’s solutions with dextrose. All the differences between $R_p$ values were statistically significant ($P < .05$), except groups R5 and R7 (Table 2). The ranking of the $R_p$ values among the test groups were as follows: R7 > R5 > RD7 > RD5.

Figures 4 and 5 show the SEM images of implants before and after testing. In Figure 4, which served as a control for SEM analysis, implant surfaces were entirely regular and had no pits, grooves, and so on. Figure 5a shows the slight deterioration on the surface after it was immersed in R7 electrolyte. The formation of small pits and grooves on the surface of the implant in group R5 can be easily observed (Figure 5b). The SEM images of the implants in groups RD7 and RD5, shown in Figure 5c and d, reveal the degree of surface degradation. For both groups, larger pits were visible on the surface but at different rates corresponding to results of corrosion measurements.

In this study, the corrosion rates of the dental implants were significantly higher under conditions with media containing dextrose than in media only.
The EIS data were used to characterize and enzymatic properties,19 it is necessary to oxidation because of its thermal, ionic, microbiological, bacterial plaque and saliva, causes metal biodegradation. Because the oral environment, which contains biomaterials because of their superior mechanical, physical, and chemical properties, such as low density, high mechanical resistance, low elasticity coefficient, and high corrosion resistance.1,2,18

Titanium and its alloys are used successfully as commercial titanium alloys, Ti-6Al-4V and Ti-13Nb-13Zr, in Ringer’s solution at 2 pH values (5.5 and 7.0). The corrosion properties were examined by using potentiodynamic anodic polarization, cyclic polarization, and EIS. The following findings were observed: (1) The effect of pH on corrosion resistance in Ringer’s solution was more accentuated for the Ti-13Nb-13Zr alloy than for the Ti-6Al-4V alloy; (2) the value of \( I_{corr} \) was larger for the Ti-13Nb-13Zr alloy when at an acid pH; (3) after anodic polarization, both alloys presented an anodic film with two layers; (4) the alloys did not show a tendency to localized corrosion in the Ringer’s solution.

Messer et al15 investigated the corrosion properties of machined titanium implants in blood, cultures of monocytic cells, and solutions containing elevated dextrose concentrations. Implant corrosion was estimated by OCPs, linear polarization resistance, and EIS. Implants under conditions of inflammatory stress exhibited more negative \( E_{corr} \) values, suggesting an increased potential for corrosion. Linear polarization measurements detected increased corrosion rates in the presence of elevated dextrose conditions over phosphate-buffered saline conditions. The EIS measurements suggested that implants underwent surface passivation reactions that may have limited corrosion over

### Discussion

Titanium and its alloys are used successfully as biomaterials because of their superior mechanical, physical, and chemical properties, such as low density, high mechanical resistance, low elasticity coefficient, and high corrosion resistance.1,2,18 Because the oral environment, which contains bacterial plaque and saliva, causes metal biodegradation because of its thermal, ionic, microbiological, and enzymatic properties,19 it is necessary to develop and evaluate materials resistant to wear and corrosion. The high corrosion resistance of titanium, which is responsible for chemical stability in the human body, results from the formation of a dense and stable layer of titanium oxide on its surface.20 This layer is formed rapidly as the reactivity of the titanium with oxygen forms several oxides, including TiO\(_2\), which is the most common.20 The EIS data were used to characterize implant surface and to provide a better understanding of the effects of solution on the interface/ alloy in this study. The EIS findings showed that a protective oxide layer formed on the dental implant surface for all test electrolytes; but the stability of these films was not equal among the groups. Because we used only one type of alloy in this study, it can be postulated that differences resulted purely from different solutions representing sound and diabetic conditions.

It is difficult to make direct comparisons between the current study and earlier ones depending on different types of implants and/or media and methods including pH alterations or high glucose levels only.4,7,15,21,22 This research, according to our knowledge, is the first study that used 2 different pH and dextrose levels simultaneously to observe implant corrosion. The corrosion rate of the implants, which were clearly visible in the SEM images, were significantly higher under conditions with lower pH values and higher dextrose concentrations than in Ringer’s solution in our study; this finding fully or partially concurs with the results of earlier studies.7,15,21,22 Souza et al7 studied the electrochemical behavior of 2 commercial titanium alloys, Ti-6Al-4V and Ti-13Nb-13Zr, in Ringer’s solution at 2 pH values (5.5 and 7.0). The corrosion properties were examined by using potentiodynamic anodic polarization, cyclic polarization, and EIS. The following findings were observed: (1) The effect of pH on corrosion resistance in Ringer’s solution was more accentuated for the Ti-13Nb-13Zr alloy than for the Ti-6Al-4V alloy; (2) the value of \( I_{corr} \) was larger for the Ti-13Nb-13Zr alloy when at an acid pH; (3) after anodic polarization, both alloys presented an anodic film with two layers; (4) the alloys did not show a tendency to localized corrosion in the Ringer’s solution.

### Table 2

<table>
<thead>
<tr>
<th>Groups†</th>
<th>( E_{corr}^* \text{ (mV)} )</th>
<th>( I_{corr}^* \text{ (µAcm}^{-2})</th>
<th>Corrosion rate* \text{ (mpy)}</th>
<th>( R_p^* \text{ (Ω)} )</th>
<th>( R_t \text{ (Ω)} )</th>
<th>( Q \text{ (1/Ω)} )</th>
<th>( W \text{ (Ωs}^{1/2})</th>
</tr>
</thead>
<tbody>
<tr>
<td>R5</td>
<td>-246 (4.472)†</td>
<td>0.4036 (0.016)†</td>
<td>3.240 (0.142)†</td>
<td>268 (19.235)†</td>
<td>11</td>
<td>9.496 × 10(^{-5})</td>
<td>0.95 1.009 × 10(^{-5})</td>
</tr>
<tr>
<td>R7</td>
<td>-313 (4.123)†</td>
<td>0.2372 (0.015)†</td>
<td>1.904 (0.153)†</td>
<td>304 (11.662)†</td>
<td>17</td>
<td>7.254 × 10(^{-5})</td>
<td>0.95 6.66 × 10(^{-6})</td>
</tr>
<tr>
<td>RDS</td>
<td>-270 (3.808)†</td>
<td>1.007 (0.056)†</td>
<td>8.085 (0.056)†</td>
<td>74 (3.162)†</td>
<td>29</td>
<td>6.322 × 10(^{-5})</td>
<td>0.95 1.28 × 10(^{-5})</td>
</tr>
<tr>
<td>RD7</td>
<td>-285 (3.606)†</td>
<td>0.5031 (0.037)†</td>
<td>4.038 (0.377)†</td>
<td>207 (34.928)†</td>
<td>31</td>
<td>7.552 × 10(^{-5})</td>
<td>0.95 1.155 × 10(^{-5})</td>
</tr>
</tbody>
</table>

*Different letters and symbols within the same column indicate statistically significant differences for means between groups (\( P < 0.05 \)).
†Group R7 indicates Ringer’s physiological solution at pH 7; group R5, Ringer’s physiological solution at pH 5.5; group RD7, Ringer’s physiological solution containing 15 mM dextrose at pH 7; group RDS, Ringer’s physiological solution containing 15 mM dextrose at pH 5.5.
FIGURES 4 AND 5. **Figure 4.** Scanning electron microscope (SEM) micrograph of dental implant before corrosion testing. **Figure 5.** Postcorrosion SEM micrographs. (a) Group R7 (Ringer’s physiological solution at pH = 7.0) (b) Group R5 (Ringer’s physiological solution at pH = 5.5) (c) Group RD7 (Ringer’s physiological solution containing 15 mM dextrose at pH = 7) (d) Group RD5 (Ringer’s physiological solution containing 15 mM dextrose at pH = 5.5).
the short term of this test. The measurements suggested that inflammatory stress and hyperglycemia may increase the corrosion of dental titanium implants.

Sartori et al.\textsuperscript{21} assessed the influence of a fluoride medium with different pHs on the corrosion resistance of titanium dental implants using an SEM and energy dispersive X-ray spectroscopy. Forty-two dental implants from 3 commercial brands were used. We simulated 5 years of regular mouth rinsing with sodium fluoride content of 1500 ppm and 2 different pHs by immersing the specimens into the medium for 184 hours. The SEM and energy dispersive X-ray spectroscopy analyses showed no evidence of corrosion on the specimens’ surfaces after being submitted to fluoride ions or after fluoride ions were incorporated in the set surface. Therefore, it has been concluded that the fluoride concentration and the pH of the solutions did not alter implant corrosion resistance.

**Conclusions**

This study suggests that the corrosion risk of titanium dental implants is greater under conditions of low pH values related to biological inflammation and high glucose concentrations and that this might be associated with higher implant failures in patients with diabetes. Further clinical studies with larger sample sizes are needed to better understand this matter. The selected solutions (Ringer’s physiological solution with or without dextrose) were not exact imitations of how the in vivo environment would be in contact with implant surface because of the lack of blood and blood-related products, including proteins. Their effects on the corrosion behavior of titanium implants could also change the results and needs to be investigated.

**Abbreviations**

EDS: energy dispersive X-ray spectroscopy

OCP: open circuit potential

SEM: scanning electron microscope

**References**


