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Study of the Effect of Plastic Deformation on the Corrosion Resistance of 316L and NiTi in Simulated Saliva

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***Abstract* - Biomedical materials such as Stainless Steel SS316L and Nickel-Titanium (NiTi) are widely used in orthodontic practices due to their mechanical properties and corrosion resistance. However, clinical manipulation often induces plastic deformation, which can disrupt the passive oxide layers protecting the metal surfaces. This study aims to evaluate and compare the effect of various levels of plastic deformation via bending on the corrosion rate of SS316L and NiTi wires wiyh wire diameter of 0.012" in an artificial saliva medium. Electrochemical polarization testing was performed using a Potentiostat/Galvanostat CorrTest CS300. Multiple bending diameters ranging from 2 mm to straight wire (0% deformation). The results showed that plastic deformation increases the corrosion rate of 0.012-inch wires for both materials, with NiTi showing extreme sensitivity, experiencing a corrosion rate jump up to 463 times under extreme bending due to TiO_2 film breakdown. Overall, SS316L demonstrated better corrosion resistance than NiTi after undergoing plastic deformation, particularly in smaller dimensions, though both materials strictly maintained an excellent corrosion resistance classification according to Fontana standards (<0.02 mm/year).**

Keywords: Plastic Deformation, Corrosion Rate, SS316L, NiTi, Artificial Saliva, Orthodontic Wire.

I. INTRODUCTION

Nickel–titanium (NiTi) and stainless steel 316L (SS316L) are the most widely used metallic materials for orthodontic archwires because of their excellent mechanical properties and relatively good corrosion resistance in the oral environment [1,2]. NiTi wires are primarily employed during the initial stages of orthodontic treatment due to their superelasticity and shape-memory effect, which enable the application of nearly constant corrective forces. In contrast, SS316L wires are commonly used during the finishing stages because of their higher stiffness and superior dimensional stability.

Despite their favorable properties, orthodontic wires are continuously exposed to saliva, which acts as an electrolyte containing chloride and other ions that can initiate electrochemical corrosion processes. Corrosion of orthodontic materials is a major concern because it may deteriorate the mechanical performance of the wires and promote the release of metallic ions, particularly nickel and chromium, into the oral cavity. The release of these ions has been associated with adverse biological effects, including allergic reactions, local inflammation, and reduced biocompatibility [3].

During clinical application, orthodontic wires are frequently subjected to bending and other mechanical manipulations to conform to the dental arch geometry and treatment requirements. These procedures introduce plastic deformation and residual stresses within the material. Plastic deformation increases dislocation density and may damage the protective passive oxide film formed on the wire surface. In NiTi, the passive layer mainly consists of titanium dioxide (TiO_2), whereas SS316L is protected by a chromium oxide (Cr_2O_3) film. Disruption of these passive layers can create localized defects and micro-cracks that facilitate electrolyte penetration and accelerate corrosion reactions [4].

Previous studies have demonstrated that mechanical deformation can alter the electrochemical behavior of metallic biomaterials [5]. However, the influence of different levels of plastic deformation on the corrosion behavior of orthodontic wires with different diameters has not been fully understood. Since orthodontic wires routinely experience various degrees of bending during clinical procedures, understanding the relationship between plastic deformation and corrosion resistance is essential for predicting their long-term performance and ensuring patient safety.

Therefore, this study aims to investigate the effect of plastic deformation on the corrosion behavior of SS316L and NiTi orthodontic wires with different diameters by means of electrochemical polarization testing in artificial saliva. The findings are expected to provide a better understanding of the degradation mechanisms of orthodontic wires and contribute to the optimization of material selection and clinical manipulation procedures in orthodontic treatment.

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II. METHODOLOGY

2.1 Specimen Preparation

Commercial orthodontic archwires made of stainless steel 316L (SS316L) and nickel–titanium (NiTi) alloys were used as test specimens. The diameter of wires is 0.012". Each wire was cut into specimens with a uniform length of 50 mm using a precision wire cutter to minimize surface damage at the cut ends.

Prior to testing, all specimens were ultrasonically cleaned in ethanol for 10 minutes to remove contaminants such as grease, dust, and manufacturing residues. The specimens were subsequently rinsed with distilled water and dried at room temperature. To ensure consistency in the electrochemical measurements, the exposed surface area of each specimen was carefully controlled, while the non-exposed sections were insulated using chemically resistant epoxy resin.

Plastic deformation was introduced by bending the wires into circular loop configurations using cylindrical mandrels of different diameters. Figure 1 shows the geometry of the specimens. The loop diameters (D) are 2, 4, 8, 15, 23 and 30 mm producing different levels of plastic strain in the wire material. Smaller loop diameters generated higher deformation levels, while larger diameters produced lower deformation levels. Straight, undeformed wires were used as reference specimens representing 0% deformation.

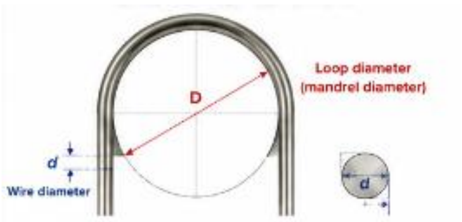

Figure 1: Geometry of bent wire

The bending process was carried out gradually using a custom bending fixture to avoid sudden loading and to ensure uniform deformation along the wire length. After bending, the geometry of each specimen was verified using a digital caliper with an accuracy of ±0.01 mm. Each test condition was prepared in triplicate to ensure the reproducibility and reliability of the experimental results.

2.2 Electrochemical Testing

Electrochemical corrosion testing was performed in artificial saliva solution to simulate the oral environment encountered by orthodontic wires during clinical service. The artificial saliva was prepared according to the formulation reported in the literature and contained potassium chloride (KCl), sodium chloride (NaCl), potassium dihydrogen phosphate (KH_2PO_4), potassium thiocyanate (KSCN), disodium hydrogen phosphate (Na_2HPO_4), sodium bicarbonate ($NaHCO_3$), and urea dissolved in deionized water. The solution pH was adjusted to approximately 6.8–7.0 to represent normal human saliva conditions.

Prior to each electrochemical measurement, the test solution was prepared fresh and maintained at a temperature of 37 ± 1 °C using a thermostatically controlled water bath to simulate the physiological temperature of the oral cavity. The solution was continuously stirred at a low speed to ensure homogeneous distribution of ions throughout the electrolyte.

Electrochemical measurements were conducted using a Potentiostat/Galvanostat CorrTest Type CS300 connected to a computer equipped with CorrTest software for data acquisition and analysis. A conventional three-electrode electrochemical cell was employed, consisting of:

- Working electrode (WE): SS316L or NiTi wire specimen.
- Reference electrode (RE): Ag/AgCl electrode.
- Counter electrode (CE): Platinum electrode.

Before polarization testing, each specimen was immersed in the artificial saliva solution and allowed to stabilize for 30 minutes to reach a steady-state open circuit potential (OCP). The OCP was continuously monitored and recorded during this stabilization period.

For each wire material, diameter, and deformation level, at least three independent measurements were conducted. The average values of Ecorr, Icorr, and corrosion rate were calculated and reported along with the corresponding standard deviations. Statistical analysis was performed to evaluate the significance of the effects of wire diameter and plastic deformation on the corrosion behavior of SS316L and NiTi orthodontic wires.

III. RESULTS AND DISCUSSION

3.1 Corrosion Test of 316L Wire

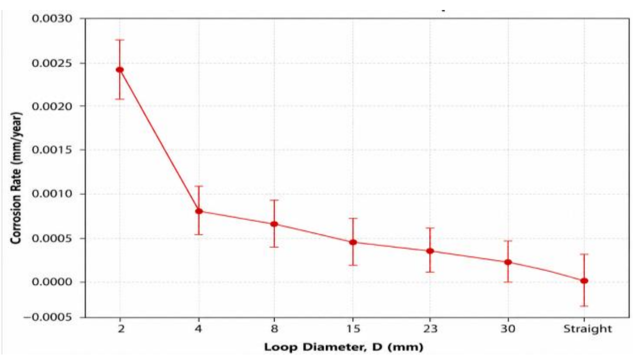

Figure 2: Corrosion rate of SS316L

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Figure 2 show the results of corrosion rate of SS316L. The straight condition has the lowest corrosion rate at 0.0000162 mm/year. As the degree of deformation increased (at a 2 mm bending diameter), the corrosion rate escalated to 0.002439 mm/year.

Based on the graph, the corrosion rate of SS316L wire in artificial saliva increased significantly as the bending diameter (loop diameter, D) decreased. The straight wire (0% deformation) exhibited the lowest corrosion rate, approaching zero. As the wire was bent to progressively smaller diameters, the corrosion rate increased gradually and reached its highest value at D = 2 mm.

The increase in corrosion rate at smaller bending diameters is associated with the greater plastic deformation experienced by the wire. Plastic deformation leads to an increase in dislocation density, residual stresses, and crystal defects in the SS316L material. These conditions may cause non-uniformity in the passive chromium oxide (Cr_2O_3) layer that naturally protects the SS316L surface against corrosion. In regions subjected to high deformation, the passive layer becomes more susceptible to microcracking and localized damage, allowing chloride ions present in artificial saliva to penetrate more easily and initiate corrosion processes.

The relatively sharp increase in corrosion rate observed at D = 4 mm and particularly at D = 2 mm suggests the existence of a critical deformation level at which passive film degradation and residual stress accumulation become sufficiently severe to accelerate metal dissolution. In contrast, at larger bending diameters (D ≥ 15 mm), the induced plastic deformation remains relatively low, enabling the passive film of SS316L to maintain its protective capability in the artificial saliva environment.

Overall, these results indicate that the corrosion resistance of SS316L orthodontic wires is strongly influenced by the degree of plastic deformation induced by bending. The smaller the bending diameter, the higher the level of deformation and, consequently, the greater the corrosion rate. These findings suggest that orthodontic wire bending with excessively small radii should be avoided, as it can reduce the corrosion resistance of the material and potentially increase metal ion release during service in the oral cavity.

3.2 Corrosion Test of NiTi wires

Figure 3 show the corrosion rate of NiTi wires. At the maximum bending diameter of 2 mm, the corrosion rate reached 0.011205 mm/year, which is approximately 463 times higher than its straight counterpart (0.0000242 mm/ year). This sharp escalation highlights that the TiO2 passive layer in thin NiTi wires is highly susceptible to mechanical disruption and micro-cracking [5].

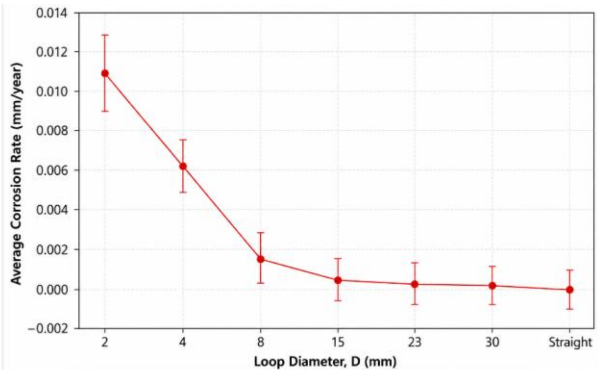

Figure 3: Corrosion rate of NiTi

Based on the Figure 3, the corrosion rate of NiTi wire in artificial saliva exhibited a trend similar to that of SS316L, increasing as the bending diameter (loop diameter, D) decreased. However, the increase in corrosion rate in NiTi was considerably more pronounced. The straight wire (0% deformation) and wires bent at larger diameters (D ≥ 15 mm) showed very low corrosion rates, approaching zero. As the bending diameter decreased to D = 8 mm, the corrosion rate began to increase noticeably and then rose sharply at D = 4 mm, reaching its maximum value at D = 2 mm.

This increase in corrosion rate is closely associated with the magnitude of plastic deformation experienced by the NiTi wire. Bending to smaller diameters produces higher bending strains on the wire surface. Plastic deformation leads to an increase in dislocation density, the formation of stress-induced martensite, and the development of residual stresses within the NiTi material. These conditions can disrupt the stability of the passive titanium dioxide (TiO_2) layer that naturally protects the NiTi surface against corrosion.

At high levels of deformation, the TiO_2 passive layer may develop microcracks, thickness non-uniformities, or localized damage due to stress concentration in specific regions. The degradation of the passive layer facilitates the penetration of aggressive ions, particularly chloride ions present in artificial saliva, thereby accelerating electrochemical reactions. As a result, the metal dissolution rate increases and the corrosion resistance of NiTi decreases significantly.

The sharp increase in corrosion rate observed at D = 4 mm and especially at D = 2 mm suggests the existence of a critical deformation level, at which residual stress accumulation and passive film degradation become sufficiently severe to markedly accelerate corrosion. In contrast, at larger bending diameters (D ≥ 15 mm), the induced plastic deformation remains relatively low, allowing

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the TiO_2 passive film to maintain its stability and protective properties in the artificial saliva environment.

Overall, these results indicate that the corrosion resistance of NiTi orthodontic wires is highly sensitive to plastic deformation induced by bending. The smaller the bending diameter, the greater the deformation and, consequently, the higher the corrosion rate. Compared with SS316L, NiTi exhibits a much steeper increase in corrosion rate at very small bending diameters, suggesting that deformation-induced degradation of the TiO_2 passive layer plays a significant role in governing the corrosion behavior of NiTi in artificial saliva.

A comparison between the corrosion behavior of NiTi and SS316L wires in artificial saliva indicates that both materials exhibit an increase in corrosion rate as the loop diameter (D) decreases, indicating that plastic deformation adversely affects the corrosion resistance of both alloys. However, the extent of the increase differs significantly between the two materials.

At large loop diameters ($D \geq 15$ mm) and in the straight condition, both materials show relatively low corrosion rates because the induced deformation is insufficient to significantly damage their passive films. Nevertheless, when the loop diameter decreases to 4 mm and particularly to 2 mm, the corrosion rate of NiTi rises much more sharply than that of SS316L. This finding suggests that NiTi is more sensitive to severe plastic deformation in terms of corrosion degradation [8].

The difference in behavior can be attributed to the distinct characteristics of their passive films and deformation mechanisms. SS316L is protected by a chromium oxide (Cr_2O_3) passive layer, which generally exhibits good stability and self-repair capability [9]. Although plastic deformation increases dislocation density and residual stresses in SS316L, the Cr_2O_3 film remains relatively effective in maintaining passivity under moderate deformation conditions.

In contrast, NiTi is protected by a titanium dioxide (TiO_2) passive film and undergoes more complex microstructural changes during deformation, including the formation of stress-induced martensite and the generation of substantial residual stresses. Severe bending can produce microcracks and localized defects in the TiO_2 layer, making it more susceptible to chloride attack in artificial saliva. Consequently, the passive film breakdown occurs more readily, resulting in a significantly higher increase in corrosion rate.

Furthermore, the higher sensitivity of NiTi to plastic deformation may have important clinical implications. Excessive bending of NiTi orthodontic wires could accelerate the release of metallic ions, particularly nickel ions, which may raise concerns regarding biocompatibility and allergic reactions in susceptible patients. In contrast, SS316L demonstrates comparatively better corrosion stability under equivalent deformation conditions.

Overall, the results demonstrate that although both materials experience reduced corrosion resistance with increasing plastic deformation, SS316L exhibits superior corrosion stability, whereas NiTi is considerably more susceptible to deformation-induced corrosion degradation, especially at very small loop diameters (D = 2–4 mm). From the results of corrosion test, both SS316L and NiTi wire show the same effect of plastic deformation. However the corrosion rate remained within the "Excellent" corrosion resistance category (<0.02 mm/year) [7].

IV. CONCLUSIONS

The corrosion behavior of SS316L and NiTi orthodontic wires in artificial saliva was strongly influenced by the degree of plastic deformation induced by bending. For both materials, the corrosion rate increased as the loop diameter (D) decreased, indicating that smaller bending diameters generated higher levels of plastic strain and greater degradation of the protective passive films. Wires in the straight condition and those bent at relatively large diameters ($D \geq 15$ mm) exhibited low corrosion rates, whereas a significant increase in corrosion was observed at D = 4 mm and especially at D = 2 mm, suggesting the existence of a critical deformation level that accelerates corrosion processes.

A comparison between the two materials revealed that NiTi was considerably more sensitive to deformation-induced corrosion than SS316L. The sharp increase in the corrosion rate of NiTi at small loop diameters was attributed to the instability and localized breakdown of the TiO_2 passive film, combined with the formation of stress-induced martensite and high residual stresses. In contrast, the Cr_2O_3 passive layer on SS316L remained relatively more stable and retained its protective capability under comparable deformation conditions. These findings indicate that excessive bending of orthodontic wires, particularly NiTi wires, should be minimized to preserve corrosion resistance and reduce the potential risk of metal ion release in the oral environment.

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