

Article

Effect of Plastic Deformation-Induced Residual Stress on the Corrosion Behavior of Monoblock Dental Implants: Implications for Clinical Performance

Alejandra Partida ¹, Marco Antonio Hernández-Rodríguez ¹, Meritxell Molmeneu ², Miquel Punset ², Maria del Carmen de Lama-Odria ³, Conrado Aparicio ^{3,4} and Javier Gil ^{3,*}

¹ Faculty of Mechanical and Electrical Engineering, Universidad Autonoma de Nuevo Leon, San Nicolás de los Garza 66450, Mexico; alejandra.hernandezprtd@uanl.edu.mx (A.P.); marco.hernandezrd@uanl.edu.mx (M.A.H.-R.)

² Biomaterials, Biomechanics and Tissue Engineering Group, Department of Materials Science and Engineering and Institute for Research and Innovation in Health (IRIS), Universitat Politècnica de Catalunya—BarcelonaTech (UPC), c/Eduard Maristany 16, 08019 Barcelona, Spain; meritxell.molmeneu@upc.edu (M.M.); miquel.punset@upc.edu (M.P.)

³ Bioinspired Oral Biomaterials and Interfaces, Department Ciencia e Ingeniería de Materiales, Escola Enginyeria Barcelona Est, Universitat Politècnica de Catalunya, Av, Eduard Maristany 16, 08019 Barcelona, Spain; maria.de.lama@upc.edu (M.d.C.d.L.-O.); conrado.aparicio@upc.edu (C.A.)

⁴ ICREA—Institució Catalana de Recerca i Estudis Avançats, Generalitat de Catalunya, 08010-Barcelona, Spain

* Correspondence: javier.gil.mur@upc.edu

Highlights

What are the main findings?

- Deformation induced in Ti6Al4V monoblock dental implants generates residual stresses that promote electrochemical corrosion.
- The deformed monoblock dental implants increase the release of ions, particularly titanium and vanadium, into the physiological environment.

What are the implications of the main findings?

- The reduction in corrosion resistance is related to the magnitude of residual stress in the implant.
- The increase in ionic release is also related to the magnitude of residual stress accumulated in the implant.
- These should be considered by clinicians to ensure optimal long-term performance.

Abstract

Background/Objectives: One-piece (monoblock) dental implants are increasingly used, particularly in patients with limited bone availability. Prosthetic alignment is often achieved via plastic deformation of the titanium implant. This study aimed to evaluate the impact of such deformation-induced residual stress on the corrosion resistance of these implants. **Methods:** Two types of monoblock dental implants (spherical “S” and Mag-Conical “M”) were subjected to controlled plastic deformation. Residual stress was quantified by X-ray diffraction using the Bragg–Brentano method. Electrochemical behavior was evaluated by measuring the open-circuit potential (E_{ocp}) and performing potentiodynamic polarization tests in phosphate-buffered saline (PBS) at 37 °C. Metal ion release (Ti, V, Al) was quantified by inductively coupled plasma mass spectrometry (ICP-

Academic Editor: Massimo Del Fabbro

Received: 1 May 2026

Revised: 20 June 2026

Accepted: 7 July 2026

Published: 10 July 2026

Copyright: © 2026 by the author. Licensee MDPI, Basel, Switzerland. This article is an open access article distributed under the terms and conditions of the [Creative Commons Attribution \(CC BY\)](https://creativecommons.org/licenses/by/4.0/) license.

MS) at specific immersion time points. Surface morphology and corrosion features were examined by scanning electron microscopy (SEM). **Results:** Residual stress values increased significantly after plastic deformation. The open-circuit potential (E_{ocp}) shifted towards more electronegative values in both implant designs as deformation-induced residual stresses increased. The E_{ocp} values shifted from -0.099 V to -0.227 V in the S design and from -0.115 V to -0.141 V in the M design, comparing the as-received condition with the deformed state, respectively. Potentiodynamic tests showed an increase in corrosion rate from 0.0021 mm/year for the original implants to 0.0156 mm/year for the deformed ones. Surfaces in the stressed regions exhibited a high density of corrosion pits, indicating localized electrochemical degradation. Deformed dental implants also exhibited higher ion release, particularly of titanium and vanadium, with higher levels observed in implants with greater residual stress and lower corrosion resistance. In the deformed regions, the release of titanium and vanadium ions into the surrounding medium was nearly five-fold higher. **Conclusions:** Plastic deformation of monoblock dental implants is associated with reduced corrosion resistance. Increased residual stress correlates with enhanced electrochemical degradation and ion release, which may have relevant implications for implant selection and clinical placement.

Keywords: corrosion; monoblock dental implant; Ti6Al4V; residual stress; electrochemical degradation; bending deformation

1. Introduction

Titanium alloys, particularly Ti-6Al-4V (Grade V), have been widely used in the biomedical field, especially in dentistry and maxillofacial applications, due to their combination of desirable properties, including high mechanical strength, excellent biocompatibility, and outstanding corrosion resistance in physiological environments [1]. This corrosion resistance is primarily attributed to the spontaneous formation of a stable titanium oxide layer on the alloy surface, which acts as a protective barrier against electrochemical degradation in bodily fluids [2].

Implant-supported fixed prostheses are currently regarded as the standard treatment option for the replacement of anterior teeth. However, their success depends heavily on the quality and volume of the alveolar bone [3]. Bone density is a critical factor in achieving primary stability, which is a prerequisite for the biological process of osseointegration [4]. However, bone density may be compromised by systemic pathologies such as osteoporosis, hyperthyroidism, osteogenesis imperfecta and osteomalacia. In addition, reduced bone density may result from systemic factors, including hormonal imbalances, malnutrition and genetic disorders that affect both the density and quality of bone tissue [5]. Conventional implants are typically placed within the crestal alveolar bone and may require additional surgical procedures before implantation, particularly in cases where further preparation of the recipient bone site is required [6].

Monobloc bendable implants have been developed to facilitate implant-supported rehabilitation in patients with reduced bone density. Unlike conventional two-piece systems, they integrate the implant body and abutment into a single component. This configuration allows a minimally invasive placement technique and can reduce the need for bone augmentation procedures by obtaining primary stability through engagement of the apical cortical bone and the basal/crestal bone structures of the nasal floor, nasomaxillary buttress, and maxillary sinus walls [7]. Figure 1 presents a bendable

monobloc dental implant in its undeformed state and after deformation to a 30° inclination.

Due to anchorage in native bone, which in some cases is located far from the implant site, the interconnection (neck) between the screw and abutment allows intraoperative adjustment by bending the implant up to 20°–40°, in order to achieve parallelism and facilitate prosthetic access to the connection [8]. The monobloc dental implant allows the clinician to reshape it to achieve the appropriate angulation for proper occlusion. This adjustment must be limited to a degree that does not compromise osseointegration between the bone and the osseointegrated portion of the implant.

Through bending, the material undergoes plastic deformation, leading to an increase in dislocation density, microstructural changes, and strain hardening [9–11]. The influence of plastic deformation on the corrosion resistance of Ti-6Al-4V has been widely investigated. Zheng et al. [12] evaluated the combined effect of tensile deformation and exposure to acidic environments on the corrosion behavior of Ti-6Al-4V ELI alloy. Their results showed that tensile deformation promotes grain rotation and a reduction in texture strength, with plastic deformation exerting a stronger influence than the acidic environment on the decrease in passive film impedance.

Krawiec et al. [13] analyzed the relationship between plastic deformation, microstructural evolution, and passivity in Ti-6Al-4V and Ti-10Mo-4Zr alloys. Their findings indicated an increased presence of titanium-based compounds within the passive film under applied strain, which significantly affected the cathodic behavior of the alloys.

Similarly, Guo et al. [14] investigated the corrosion response of Ti-23Nb-0.7Ta-2Zr under compressive plastic deformation at different strain levels. Their study demonstrated that low levels of plastic deformation deteriorate corrosion resistance, whereas severe plastic deformation mitigates this effect and can eliminate the associated degradation.



Figure 1. Monobloc bendable dental implant: (a) undeformed initial state and (b) deformed state at 30° inclination.

The aim of this study is to evaluate the effect of induced plastic deformation on the corrosion behavior of Ti-6Al-4V. Plastic deformation was introduced by applying a 30° bend to the samples to assess whether the resulting mechanical strain affected the integrity and stability of the native oxide layer. This work specifically investigates whether deformation promotes the formation of a less uniform passive film, thereby increasing the corrosion rate of the material.

The electrochemical response of plastically deformed samples is compared with that of undeformed specimens to identify differences in corrosion rate, susceptibility to pitting corrosion, and localized degradation. Residual stress in the bending zone was measured and compared with that of the undeformed region. Particular attention is given to the bent

region to determine whether the plastically deformed area exhibits increased corrosion susceptibility relative to the undeformed condition.

It is hypothesized that plastic deformation induced by a 30° bending of Ti-6Al-4V increases corrosion rate and susceptibility to localized corrosion due to a loss of the homogeneity and stability of the passive oxide layer compared with undeformed controls. In contrast, the null hypothesis states that plastic deformation does not significantly influence corrosion behavior, including corrosion rate, pitting susceptibility, or passive film stability.

2. Materials and Methods

2.1. Sample Design

Commercially available Ti-6Al-4V Grade 5 bars and monobloc dental implants were supplied by MAGEOTEC (Monterrey, Nuevo León, Mexico) (Figure 1). The bars had a length of 19.5–20 mm, with base and neck diameters of 4 mm and 2 mm, respectively. Two implant geometries were investigated in this study: a spherical design (S) and a Mag-Conical design (M), as illustrated in Figure 2.

To ensure reproducibility and minimize variability in the induced plastic deformation, a standardized bending setup was designed. A custom-made wooden guide was used to define a fixed 30° angular displacement, which corresponded to the midpoint of the clinically relevant deformation range for this type of implant. This guide served as a physical stop, ensuring that all specimens were bent to the same final angular position.

Plastic deformation was applied using a lever-based single-point bending method. To reduce operator-dependent variability, all specimens were deformed by the same operator, following a consistent procedural sequence and applying a continuous manual force without interruptions until the predefined angular stop was reached. The bending process was performed at a controlled quasi-static rate, avoiding sudden loading or impact, in order to minimize strain-rate effects.

After unloading, the wooden guide also allowed for verification of the final angle, ensuring that spring-back effects were accounted for and that all samples were adjusted to the same final 30° configuration when necessary. This approach ensured consistency among the five replicates per experimental group and reduced variability in the resulting plastic strain distribution and localized dislocation density. Although the applied bending method was not fully instrumented, the use of a physical angular stop and standardized manual loading conditions provided sufficient repeatability for comparative analysis of residual stress development. Figure 2 illustrates the two sample designs used in this study in both conditions evaluated: with and without plastic deformation, respectively.

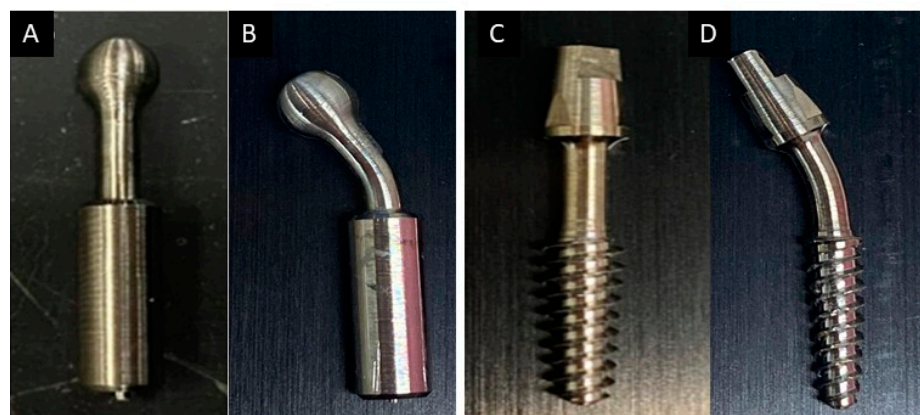


Figure 2. Monobloc dental implant geometries evaluated in this study: (A) spherical design (S) in the undeformed condition; (B) spherical design after 30° deformation (S deformed); (C) Mag-

Conical design (M) in the undeformed condition; and (D) Mag-Conical design after 30° deformation (M deformed).

2.2. Residual Stresses

Residual stresses were determined by means of a diffractometer incorporating a Bragg–Brentano configuration (D500, Siemens, Germany). The measurements were studied for the planes (213). These diffract at $2\theta = 139.5^\circ$. The elastic constants of titanium in the direction of (213) planes are $EC = (E/(1 + \nu))_{(213)} = 90.3 (1.4) \text{ GPa}$. Eleven Ψ tilt angles were used, including 0° , five positive angles, and five negative angles. The diffraction peaks were adjusted with a pseudo-Voigt function using appropriate software (WinplotR, free access on-line, Exeter, NH, USA), and then converted to interplanar distances (d_Ψ) using Bragg's equation. The d_Ψ vs $\sin^2 \Psi$ graphs and the calculation of the slope of the linear regression (A) were done with appropriate software (Origin, Microcal, Kent, WA, USA) [15]. Residual stresses were determined using the $\sin^2 \Psi$ method [15]. The slope (A) of the linear regression of the interplanar spacing (d_Ψ) versus $\sin^2 \Psi$ was calculated, and the residual stress was obtained from $\sigma = EC(1/d_0) A$, where d_0 is the interplanar spacing at $\Psi = 0^\circ$ and EC is the X-ray diffraction elastic constant associated with the crystallographic plane being analyzed (213). Five samples for each implant were used (deformed and undeformed).

2.3. Corrosion Resistance

The tests were performed using an AUTOLAB PGSTAT302N potentiostat (Metrohm, Herisau, Switzerland) controlled by Nova 2.1.6 software (Metrohm, Utrecht, The Netherlands). The electrolyte used in this study was PBS 18912-014 (Gibco, Waltham, MA, USA), whose chemical composition is presented in Table 1. This saline solution reproduces the ionic composition of the physiological human environment. The testing solution was maintained at a constant temperature of 37°C throughout the experiments.

Samples were electrically connected with colloidal silver (16031, Ted Pella, Redding, CA, USA). Mechanical fixation and insulation were performed using Loctite adhesive. Subsequently, a resin called VersoCit-2 Kit (40200089, Struers, Ballerup, Denmark) was applied to the cable–sample interface to ensure electrical insulation of the cable from the test medium (Figure 3). Electrochemical tests were conducted in PBS due to its simplified and well-controlled ionic composition, which allows reproducible evaluation of the electrochemical response without significant interference from complexing species. Additionally, the entire system was coated with an insulating varnish (Lacomit Varnish, AGG371, Agar Scientific, Rotherham, UK). An exposed area of 0.377 cm^2 was left. The experimental setup is detailed in the following figure.

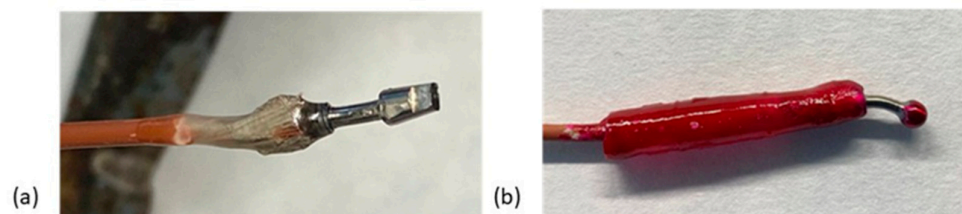


Figure 3. Sample preparation for corrosion testing. Photographic images showing the preparation of specimens for electrochemical tests: (a) internal electrical connection within the implant's internal cavity and (b) external electrical insulation of the implant, leaving only the plastically deformed region exposed.

A three-electrode cell configuration was used, with the implant as the working electrode, a saturated calomel electrode (SCE) as the reference, and a platinum wire as the counter electrode. All tests were performed at 37 °C. The corrosion tests were performed following the ISO and ASTM standards [16–19]. The experimental setup is shown in Figure 4.

Table 1. Chemical composition of PBS solution.

Component	Composition (mM)
Phosphate (as sodium phosphates)	10.00
Potassium chloride (KCl)	2.68
Sodium chloride (NaCl)	140.00

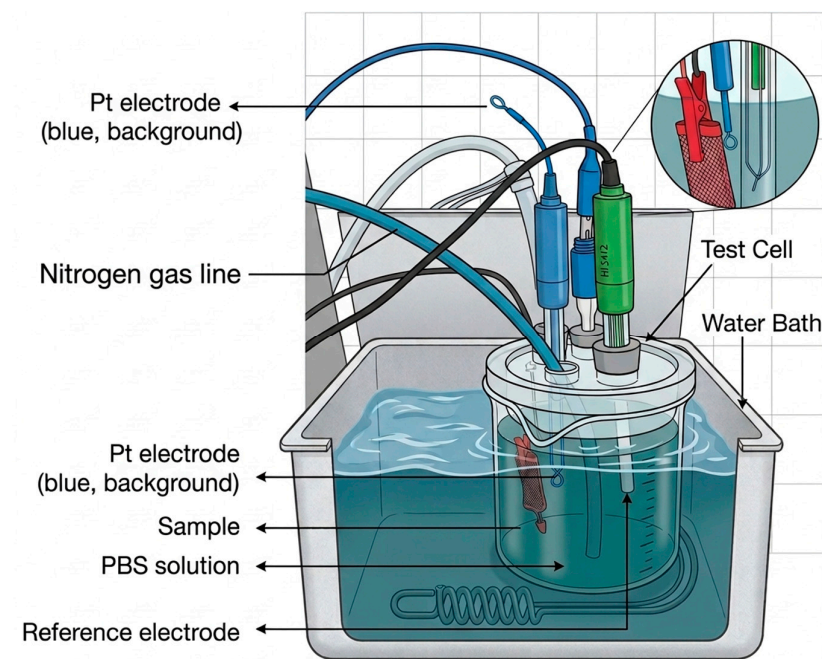


Figure 4. Schematic of the experimental setup in a water bath [20].

Potentiodynamic polarization curves were obtained for the 4 study groups taking into account the ASTM G5 standard [19]. In this test, a variable electrical potential is imposed by the potentiostat between the calomel electrode and the sample, producing a current to flow between the sample and the counter electrode. The counter electrode used was platinum [21]. Initially, the system was allowed to stabilize by means of an open-circuit test for 3 h taking results every 2 s. After stabilization, this test was launched, performing a linear polarization from −2 mV to 2 mV at a speed of 0.002 V/s with a step of 0.001 V. These parameters were registered into the Nova 2.1.6 software. The results studied are detailed below.

Key parameters included corrosion current (i_{corr}), potential (E_{corr}), and corrosion rate (mm/year), derived via Tafel extrapolation [21–23]. Twelve potentiodynamic tests were conducted in triplicate for each implant design (M and S) and condition (deformed and undeformed).

To ensure the reproducibility of the results, a constant volume of 1000 mL of solution was maintained for each assay (without recirculation). The medium was deoxygenated through continuous nitrogen (N_2) sparging, starting 30 min prior to the assay and maintained throughout its duration, while the temperature was maintained at 37 ± 1 °C.

The corrosion experiments were intentionally performed under static conditions to decouple the electrochemical response from mechanically induced surface film disruption, thereby isolating the effect of prior plastic deformation on the corrosion behavior of titanium. This approach is consistent with the standard methodology used in electrochemical characterization of passive metals, where static conditions are employed to decouple corrosion processes from mechanically assisted degradation phenomena such as tribocorrosion or corrosion-fatigue [21–24]. The literature has shown that plastic deformation alters the defect structure of the passive film and modifies the electrochemical behavior of titanium alloys even in the absence of applied mechanical loading during corrosion testing [24,25]. Therefore, static electrochemical testing is appropriate to isolate the effect of prior plastic deformation from mechanically assisted degradation mechanisms such as tribocorrosion or stress-corrosion processes.

2.4. Ion Release

Five samples from each group were used for the metal ion recovery test. After weighing the samples ($m = 0.206$ g) and following the ASTM G-102-89 [26], a weight adjustment was made at a rate of 1 mL of Hank's solution for each 0.20 g of sample, as indicated in the standard. In contrast to electrochemical tests which were conducted in PBS solution, ion release experiments were performed in Hank's balanced salt solution to better approximate physiological conditions, as the presence of calcium, phosphate, and bicarbonate ions may influence dissolution, ion complexation, and secondary phase formation during degradation. The use of different immersion media is consistent with previously reported methodologies in biomaterials research, where simplified chloride-based solutions (e.g., PBS) are employed for standardized electrochemical characterization, while complex physiological solutions such as Hank's balanced salt solution are used to evaluate degradation and ion release under biologically relevant conditions [27–29].

The 5 samples of each group were placed in the same Eppendorf with 5 mL of Hank's solution and stored at 37 °C. Five samples from each group were incubated together in a single container to ensure a standardized surface area-to-volume ratio and sufficient analyte concentration for ICP-MS detection. This setup was used to obtain a representative average ion release per group, although it did not allow assessment of inter-sample variability. Hank's solution should be extracted and stored in the refrigerator after 1, 3, 7, 14, and 21 days. After each extraction, 5 mL of fresh Hank's solution was replenished in the Eppendorf containing the samples. All the extracted samples were filtered using 0.22 μm filters before diluting in 2% ultra-pure nitric solution (Suprapur, Merck, Darmstadt, Germany).

After 21 days, the concentration of released titanium ions was measured, at the test times indicated above, by inductively coupled plasma mass spectrometry (ICP-MS) with the Agilent Technologies 7800 ICP-MS (Agilent Technologies, Santa Clara, CA, USA).

2.5. Scanning Electron Microscopy

The surfaces of the samples were observed using an SEM (JEOL JSM 5410 Microscopy Tokyo, Japan) equipped with a link LZ5 EDS (Jeol, Tokyo, Japan) operated at 10 kV, which was also used for determining the chemical composition.

2.6. Statistical Analysis

The data were statistically analyzed using Student's *t*-tests, one-way ANOVA tables and Tukey's multiple comparison tests in order to evaluate any statistically significant differences between the sample $p < 0.005$ (α -value).

3. Results

Table 2 summarizes the residual stress results for the implants investigated. The original S and M implants exhibit slight compressive residual stress likely induced by the machining of the dental implant surfaces. In contrast, the S and M implants subjected to 30° bending show tensile residual stress. The M implant exhibits higher tensile residual stress values, which may be attributed to its larger diameter compared with the S implant, resulting in greater applied force during deformation.

Table 2. Residual stress for each monoblock implant studied.

	S	M	S Deformed	M Deformed
Residual stress (MPa)	-12 ± 3	-8 ± 2	$+78 \pm 14$	$+125 \pm 13$

The open-circuit potential (E_{ocp}) results are shown in Figure 5 and listed in Table 3. The results showed a significant drop toward more active (negative) values following deformation. The E_{ocp} values shifted from -0.099 V to -0.227 V in the M design and from -0.115 V to -0.141 V in the S design, comparing the as-received condition with the deformed state, respectively. Specifically, the Mag-Conical (M) implant design shifted from -0.099 ± 0.086 V to -0.227 ± 0.007 V ($p < 0.05$), indicating a higher thermodynamic susceptibility to corrosion.

Table 3. Open-circuit potential for the different monoblock implants studied. An asterisk (*) indicates that the results show statistically significant differences ($p < 0.05$) compared to those without an asterisk. A triple asterisk (***) indicates that there are statistically significant differences ($p < 0.05$) compared to those without an asterisk, those with a single asterisk (*) and those with two asterisk (**).

Samples	E_{ocp} (V)
S	-0.115 ± 0.041
S deformed	-0.141 ± 0.041 *
M	-0.099 ± 0.086 **
M deformed	-0.227 ± 0.007 ***

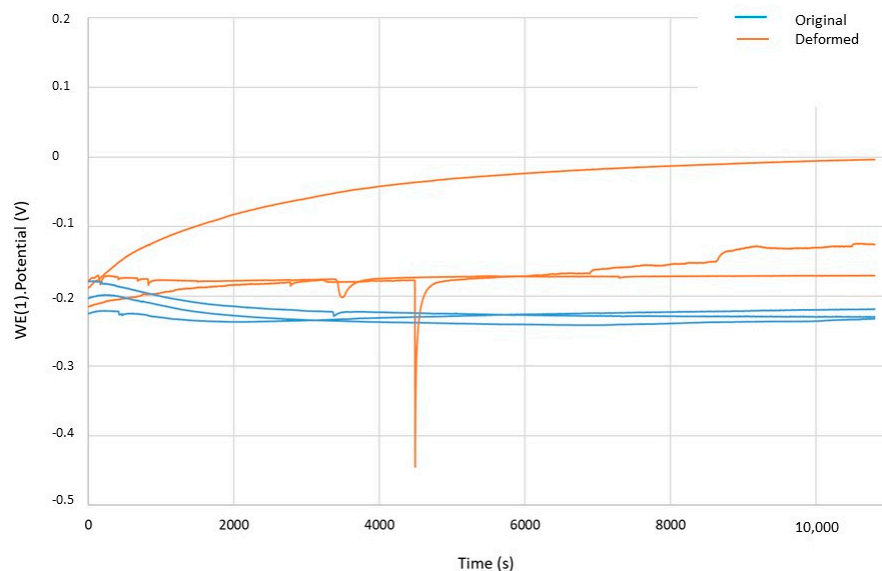


Figure 5. Curves of open-circuit potential (E_{ocp}) for the original and deformed M implants.

The corrosion potential (E_{corr}) and corrosion current (I_{corr}) were determined from the analysis of linear polarization curves. The corrosion current density (j_{corr}) was calculated by normalizing I_{corr} to the exposed geometric surface area (0.5655 cm^2). The material tested was Ti6Al4V alloy, with a density of 4.43 g/cm^3 .

Potentiodynamic test parameters are detailed in Table 4. Both deformed groups exhibited higher corrosion current densities (j_{corr}) compared to their undeformed counterparts. For instance, the M deformed group showed a j_{corr} of $1.774 \times 10^{-6} \text{ A/cm}^2$, nearly an order of magnitude higher than the original M implant ($2.337 \times 10^{-7} \text{ A/cm}^2$). Potentiodynamic tests also showed an increase in corrosion rate from 0.0021 mm/year for the original M implants to 0.0156 mm/year for the deformed ones. In the same sense, for S implants without deformation the corrosion rate was 0.0032 mm/year and 0.0099 mm/year for deformed ones. Figure 6 shows the electrochemical corrosion curves obtained from the potentiodynamic tests.

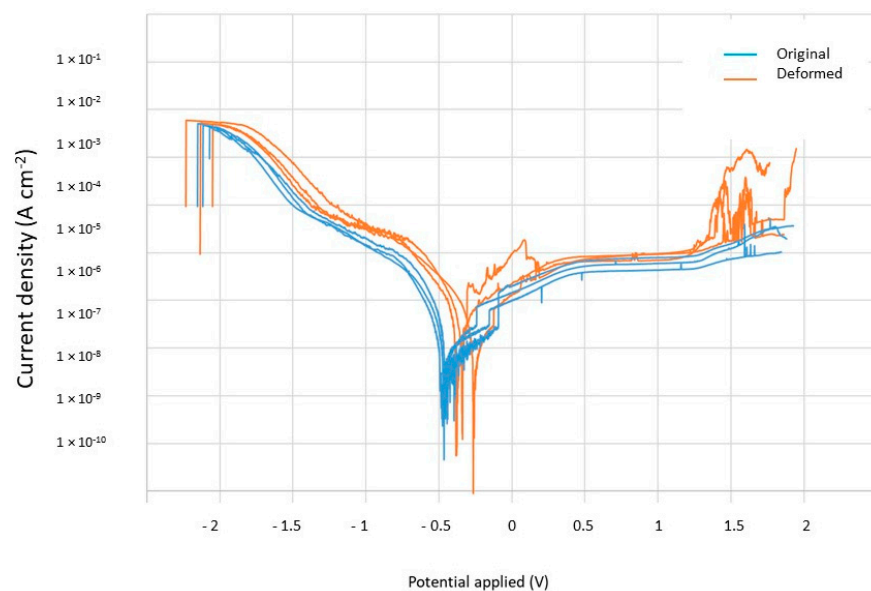


Figure 6. Potentiodynamic curves for the original and deformed monoblock dental implants.

Figures 7 and 8 show the release rates of Ti, V, and Al ions over time for undeformed and deformed implants, respectively. For both implant conditions, titanium and vanadium were the ions released in the highest amounts throughout the study period. Deformed implants exhibited substantially higher ion release rates than undeformed implants, with values reaching approximately five times those measured for the undeformed condition [26].

Table 4. Corrosion parameters obtained by potentiodynamic tests.

Samples	i_{corr} (μA) $\times 10^{-7}$	j_{corr} ($\mu\text{A/cm}^2$) $\times 10^{-7}$	E_{corr} (mV vs. SCE)	Corrosion Rate (mm/Year)
S	1.394 ± 1.122	3.698 ± 2.977	-0.468 ± 0.012	0.0032 ± 0.0026
S deformed	4.248 ± 2.378	11.27 ± 6.309	-0.331 ± 0.060	0.0099 ± 0.0055
M	8.809 ± 1.194	2.337 ± 0.3168	-0.448 ± 0.024	0.0021 ± 0.0003
M deformed	6.689 ± 2.540	17.74 ± 46.738	-0.409 ± 0.078	0.0156 ± 0.0059

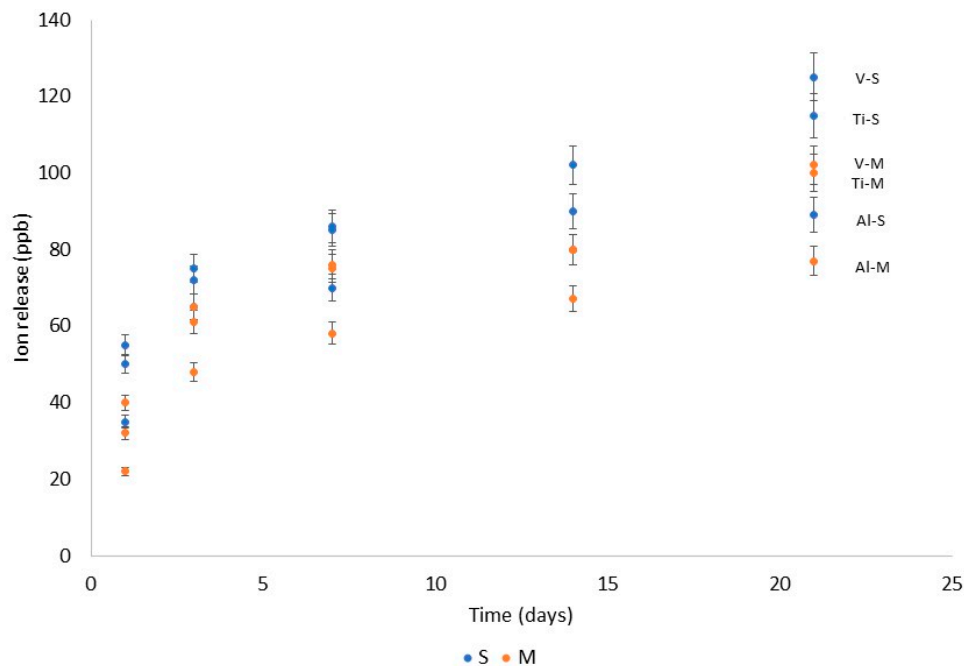


Figure 7. Ion release at different immersion times for different monoblock implants without deformation.

Scanning electron microscopy (SEM) revealed the formation of corrosion pits on the sample surfaces following corrosion testing. As shown in Figure 9, the M samples exhibited a higher pit density than the S samples, in agreement with their higher corrosion rate determined by electrochemical measurements.

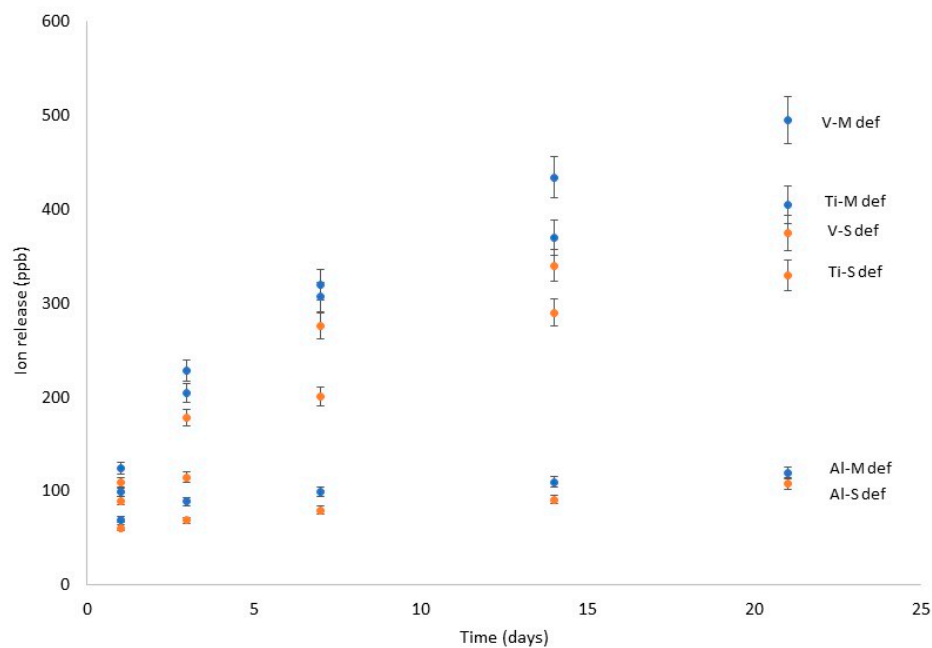


Figure 8. Ion release at different immersion times for different monoblock implants with deformation.

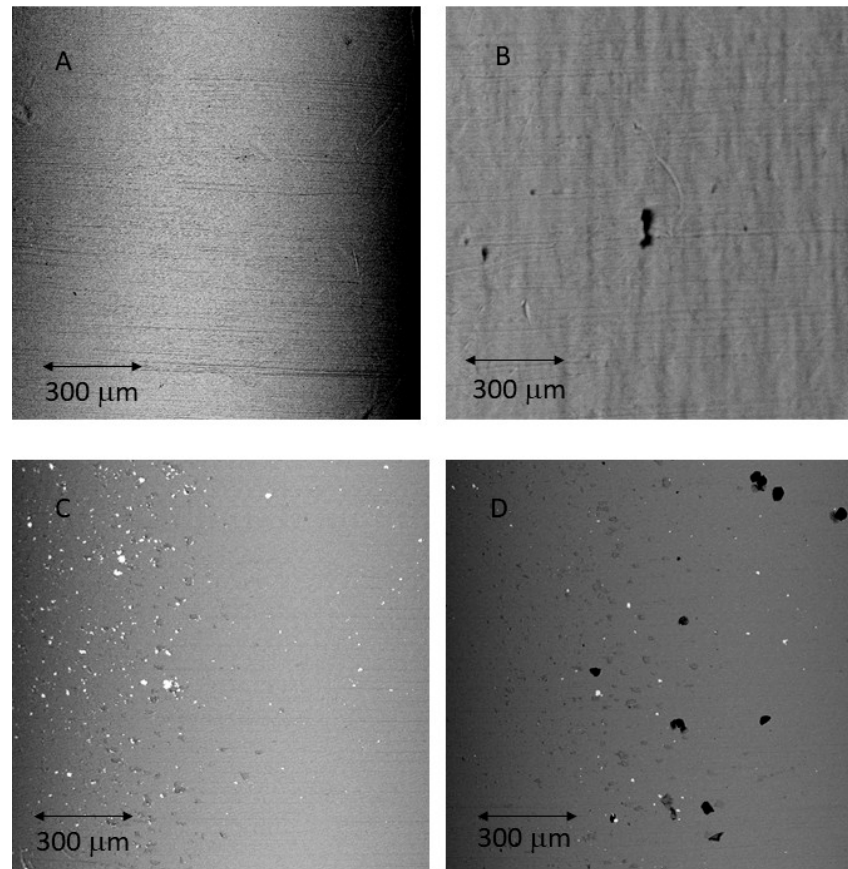


Figure 9. SEM images ($\times 500$ magnification). (A) Implant S before corrosion testing; (B) implant S after corrosion testing; (C) implant M before corrosion testing; (D) implant M after corrosion testing.

4. Discussion

The use of one-piece (monoblock) dental implants has increased, particularly in cases with limited bone availability. These implants eliminate the implant–abutment interface, thereby preventing bacterial leakage at the connection level. However, their prosthetic versatility is limited, and there is a potential risk of loss of implant–bone stability and fixation when plastic deformation is applied intraoperatively by the clinician to achieve proper prosthetic alignment and occlusion [3,5–8,30,31].

One of the key issues highlighted in this study is the reduced corrosion resistance associated with these implants. This reduction is primarily attributed to the high degree of plastic deformation induced during bending of the implant for prosthetic positioning. The applied stress generates a high density of dislocations, leading to increased residual stresses in the deformed region. This phenomenon was confirmed by grazing incidence X-ray diffraction (GIXRD) using the Bragg–Brentano method. The region with maximum stress acts as an anodic area, while the unstressed region behaves as a cathodic area, resulting in the formation of a galvanic couple within the implant itself. This electrochemical gradient promotes localized pitting corrosion, which may compromise mechanical properties and generate corrosion by-products that, as reported by several authors, may not be cytocompatible [32–36].

The results demonstrate the strong sensitivity of electrochemical corrosion behavior to residual stress. Significant differences were observed between implant S deformed with a residual stress of +78 MPa and implant M deformed with +125 MPa, with corrosion rates of 0.0099 and 0.0156 mm/year, respectively.

Unlike conventional dental implants, monoblock implants typically lack a roughened surface. This characteristic may negatively affect osseointegration, as surface roughness increases the effective bone–implant contact area and enhances interfacial biological fixation, as is widely reported in the literature [30,31]. Surface roughness is commonly achieved through abrasive particle blasting (e.g., alumina), which also induces compressive residual stresses that improve fatigue life by delaying crack initiation. In addition, these compressive stresses have been reported to increase surface hydrophilicity, thereby enhancing protein adsorption and promoting biological response [37–39].

However, this type of surface treatment is generally not applied to monoblock implants, as it may further residual stress in already plastically deformed regions, potentially promoting crack initiation and premature implant failure under cyclic masticatory loading. Importantly, shot blasting does not significantly compromise corrosion resistance as it produces a relative homogeneous stress distribution across the implant surface, thereby avoiding the formation of well-defined anodic and cathodic regions that could drive localized electrochemical activity [40].

The corrosion potential of Ti6Al4V dental implants is reported to be -0.40 V for machined surfaces and -0.15 V for roughened surfaces, both significantly more passive than the values obtained for monoblock implants. Changes in corrosion behavior due to the presence of anodic and cathodic regions have also been observed in hybrid implants (with both smooth and rough surfaces), where differences in residual stress distribution contribute to reduced corrosion resistance [41–44]. These hybrid designs are intended to minimize biofilm formation in the coronal region while maintaining a rough apical surface to promote osseointegration and ensure stable biological fixation [45,46].

It should also be considered that prosthetic components are typically manufactured from Co–Cr alloys, stainless steel, or titanium, often with compositions different from that of the monoblock implant. This mismatch may promote galvanic corrosion between dissimilar metals [47–49].

The increase in corrosion in monoblock dental implants is not merely an electrochemical phenomenon; it has clinically relevant implications that may affect both the short-term and long-term success of implant therapy. First, the presence of corrosion products may act as a cofactor in the development of peri-implant diseases, leading to an increased inflammatory response in soft tissues, disruption of the microbiological balance (with a more pathogenic biofilm), and greater susceptibility to peri-implant mucositis and its progression to peri-implantitis, with associated bone loss [24,31].

Second, corrosion may impair osseointegration, as it can alter the titanium oxide (TiO_2) layer, which is essential for biocompatibility. The degradation of the titanium dioxide layer in a low-oxygen environment (resulting from the inflammatory response) may hinder the regeneration of the TiO_2 passive layer. This can lead to the formation of non-stoichiometric oxides within the surface layer, which have been shown to exhibit a certain degree of cytotoxicity. Consequently, this may reduce osteoblast adhesion and proliferation and interfere with bone remodeling, thereby compromising long-term osseointegration [50].

Furthermore, as indicated by the results, corrosion promotes the release of titanium, aluminum, and vanadium ions [51]. These species may diffuse into peri-implant tissues, potentially inducing chronic local inflammatory responses, macrophage activation, and a nonspecific immune response, as well as exerting cytotoxic effects, particularly those associated with vanadium and aluminum at specific concentration levels [52]. It should be noted that the ion release levels observed within the experimental timeframe do not reach thresholds known to elicit such biological effects; however, progressive accumulation over time could potentially result in clinically relevant concentrations.

Corrosion is not solely a chemical process but also compromises the structural integrity of the material. Corrosion–fatigue interactions under cyclic masticatory loading may accelerate crack propagation and reduce the fatigue life of dental implants [53,54]. Corrosion is not solely a chemical degradation process but also a phenomenon that directly affects the mechanical integrity of metallic biomaterials. In the oral environment, titanium dental implants are exposed to a complex combination of electrochemical and biological factors, including fluctuating pH, chloride-rich saliva, bacterial biofilms, and proteins, all of which can influence the stability of the protective titanium oxide layer and promote localized corrosion processes. These conditions are particularly relevant in load-bearing implant systems, where mechanical and electrochemical degradation mechanisms act simultaneously.

In dental implant applications, cyclic masticatory loading introduces repeated mechanical stresses that can disrupt and reform the passive oxide layer, leading to a coupled corrosion–fatigue and fretting corrosion mechanism. At the implant–abutment interface and at the bone–implant contact region, micromotions may further intensify surface damage, enhancing ion release and accelerating crack initiation at surface defects or microstructural discontinuities. This synergistic degradation can reduce fatigue resistance and compromise long-term implant stability. Additionally, corrosion processes may contribute to surface discoloration phenomena, which can be clinically relevant in aesthetic zones due to changes in oxide layer thickness and composition.

Such environmentally assisted degradation mechanisms have been widely reported for titanium and its alloys in biomedical applications, where the interaction between mechanical loading, fretting wear, and corrosive media plays a critical role in implant longevity and clinical performance [55].

From a broader biomedical perspective, titanium alloys are extensively used in both dental and orthopedic implants due to their excellent combination of corrosion resistance, biocompatibility, and mechanical strength. However, their long-term performance remains highly dependent on surface stability and resistance to environmentally assisted fatigue and fretting damage. In particular, corrosive environments have been shown to significantly reduce fatigue resistance compared to inert conditions, emphasizing the importance of considering corrosion–fatigue interactions in implant design and clinical service conditions [56]. The theoretical framework underlying these processes is well established in classical materials science, where corrosion–fatigue is defined as the acceleration of crack initiation and propagation due to the combined action of cyclic mechanical loading and electrochemical degradation mechanisms [57]. This issue warrants further investigation, as corrosion may increase the risk of implant fracture, which is particularly critical in monoblock designs, where no modular replacement of components is feasible.

Corrosion products may also have aesthetic consequences, such as grayish discoloration of peri-implant soft tissues due to the deposition of corrosion by-products, potentially contributing to gingival recession associated with chronic inflammation processes [58–61].

Although monoblock implants lack an implant–abutment interface, thereby eliminating galvanic corrosion at that junction, they concentrate mechanical stresses within a single structure. Moreover, the inability to replace individual components means that any structural or surface degradation directly affects the entire implant, thereby increasing the risk of clinical failure.

Although the experiments were conducted in accordance with international standards for corrosion testing, certain limitations must be acknowledged. These include the use of simulated physiological environments and the absence of mechanical loading during testing. Under clinical conditions, mechanical loading may accelerate

electrochemical corrosion processes. Therefore, this study aims to increase clinicians' awareness of the potential corrosion-related implications associated with the use of monoblock dental implants.

5. Conclusions

Based on the results of this study, plastic deformation induced in Ti6Al4V monoblock dental implants generates residual stresses that influence their electrochemical behavior. Specifically, increased residual stress levels were associated with reduced corrosion resistance and enhanced release of metallic ions, particularly titanium and vanadium, into the surrounding environment. Furthermore, the magnitude of these effects was correlated with the level of residual stresses accumulated within the implant material.

These findings suggest that deformation-induced residual stresses may be an important factor affecting the electrochemical performance of Ti6Al4V dental implants. Therefore, controlling the generation of residual stresses during implant manufacturing and handling could contribute to improving their corrosion behavior. Further in vitro and in vivo studies are required to determine the potential biological and clinical relevance of these observations.

Author Contributions: J.G.: Study conception and design, analysis and interpretation of data, drafting of the manuscript, and visualization of the results. M.A.H.-R.: Study conception and design, analysis and interpretation of data, critical revision of the manuscript, administrative support, and study supervision. A.P.: Investigation, acquisition of different data and interpretation. M.P.: Investigation, analysis and interpretation of data, and visualization of the results. M.M.: Administrative support, technical methods, investigation and study supervision. M.d.C.d.L.-O.: Study conception and design, analysis and interpretation of data, and revision. C.A.: Interpretation of data, drafting of the manuscript, and visualization of the results. All authors endorsed the final version of the manuscript. All authors have read and agreed to the published version of the manuscript.

Funding: This work was funded by Minsiterio de Ciencia e Investigación. Gobierno de España. MCIN/AEI/10.13039/501100011033 and FSE+ under award PREVENTITIS-PID2022-137496OB-I00. This work is part of Maria de Maeztu Units of Excellence Programme CEX2023-001300-M/funded by MCIN/AEI/10.13039/501100011033.

Institutional Review Board Statement: Not applicable.

Informed Consent Statement: Not applicable.

Data Availability Statement: The data that support the findings of this study are available from the corresponding author upon reasonable request.

Conflicts of Interest: The authors have no conflicts of interest.

References

1. Li, S.; Yang, R.; Hao, Y.L. Biomedical titanium alloys and their additive manufacturing. *Rare Met.* **2016**, *35*, 661–671. <https://doi.org/10.1007/s12598-016-0793-5>.
2. Bocchetta, P.; Chen, L.Y.; Tardelli, J. Passive Layers and Corrosion Resistance of Biomedical Ti-6Al-4V and β -Ti Alloys. *Coatings* **2021**, *5*, 487. <https://doi.org/10.3390/coatings11050487>.
3. Chamria, C.; Alexander, J. Role of monobloc dental implants in the rehabilitation of anterior dentoalveolar. *Int. J. Oral Health Dent.* **2025**, *11*, 303–308. <https://doi.org/10.18231/j.ijohd.14497.1765857829>.
4. Raghavan, R.; Shajahan, P.A.; Rajsree, B.M. The Critical Role of Bone Density. *Int. J. Res. Med. Sci.* **2025**, *13*, 3904–3907. <https://doi.org/10.18203/2320-6012.ijrms20252818>.

5. Stromsnes, K.; Martinez, C.; Soto, S. Osteoporosis: Causes, Mechanisms, Treatment and Prevention: Role of Dietary Compounds. *Pharmaceuticals* **2024**, *17*, 17121697. <https://doi.org/10.3390/ph17121697>.
6. Vajdi Mitra, G.; Agrawal, N.; Shukla, N. An Evaluation of the Efficacy and Acceptability of Basal Implants in Traumatically Deficient Ridges of the Maxilla and the Mandible. *Cureus* **2023**, *15*, e43443. <https://doi.org/10.7759/cureus.43443>.
7. Jensen, O.T.; Adams, M.W. Secondary Stabilization of Maxillary M-4 Treatment with Unstable Implants for Immediate Function: Biomechanical Considerations and Report of 10 Cases After 1 Year in Function. *Int. J. Oral Maxillofac. Implant.* **2014**, *29*, e232–e240. <https://doi.org/10.11607/jomi.te59>.
8. Goldmann, T.; Ihde, S.; Kuzelka, J. Bendable vs. Angulated dental implants: Consideration of elastic. *Biomed. Pap. Med. Fac. Palacky Univ. Olomouc* **2008**, *152*, 309–316. <https://doi.org/10.5507/bp.2008.049>.
9. Cizek, P.; Raju, S.; Armstrong, N. Dislocation structures in a Ti–6Al–4V alloy subjected to cyclic tensile deformation. *Mater. Sci. Eng. A* **2022**, *836*, 142700. <https://doi.org/10.1016/j.msea.2022.142700>.
10. Ahmadikia, B.; Beyerlein, A.; Hestroffer, J. Designing Ti-6Al-4V microstructure for strain delocalization using neural networks. *J. Mater. Sci. Mater. Theory* **2024**, *8*, 4. <https://doi.org/10.1186/s41313-024-00055-9>.
11. Sun, H.; Liang, Y.; Li, G. Dislocation hardening and phase transformation-induced high ductility in Ti-6Al-4V with a heterogeneous martensitic microstructure under tensile load. *J. Alloy. Compd.* **2021**, *868*, 155. <https://doi.org/10.1016/j.jallcom.2021.159155>.
12. Zheng, X.; Xu, C.; Cai, Y. Effect of Plastic Deformation and Acidic Solution on the Corrosion Behavior of Ti-6Al-4V ELI Titanium Alloy. *Metals* **2023**, *13*, 1740. <https://doi.org/10.3390/met13101740>.
13. Krawiec, H.; Vignal, V.; Loch, J. Influence of plastic deformation on the microstructure and corrosion behaviour of Ti–10Mo–4Zr and Ti–6Al–4V alloys in the Ringer’s solution at 37 °C. *Corros. Sci.* **2015**, *96*, 160–170. <https://doi.org/10.1016/j.corsci.2015.04.006>.
14. Guo, W.Y.; Sun, J.; Wu, J.S. Effect of deformation on corrosion behavior of Ti–23Nb–0.7Ta–2Zr–O alloy. *Mater. Charact.* **2009**, *60*, 173–177. <https://doi.org/10.1016/j.matchar.2008.08.006>.
15. Pereira, R.; Maia, P.; Rios-Santos, J.V.; Herrero-Climent, M.; Rios-Carrasco, B.; Aparicio, C.; Gil, J. Influence of Titanium Surface Residual Stresses on Osteoblastic Response and Bacteria Colonization. *Materials* **2024**, *17*, 1626. <https://doi.org/10.3390/ma17071626>.
16. ISO/TR 20580:2022; Preparation of Metallographic Specimens. European Union: Brussels, Belgium, 2022.
17. ASTM E3-1-1(2017); Standard Guide for Preparation of Metallographic Specimens. ASTM International: West Conshohocken, PA, USA, 2017.
18. ASTM G31-21(2025); Standard Guide for Laboratory Immersion Corrosion Testing of Metals. ASTM International: West Conshohocken, PA, USA, 2025.
19. ASTM G5-14e1; Standard Reference Test Method for Making Potentiostatic and Potentiodynamic Anodic Polarization Measurements; Technical Report No.; ASTM International: West Conshohocken, PA, USA, 2014.
20. Google. (2026). Gemini, Version 1.5; Large language and image generation model. Image generated using the Gemini AI model (Google) on 28 April 2026, based on an original photograph by the author and edited to reflect the nitrogen flow and the PBS solution; Google Gemini: Singapore, 2026.
21. Ratner, B.D.; Hoffman, A.S.; Schoen, F.J.; Lemons, J.E. (Eds.) *Biomaterials Science: An Introduction to Materials in Medicine*, 3rd ed.; Elsevier: Amsterdam, The Netherlands, 2020.
22. Landolt, D.; Mischler, S. (Eds.) *Tribocorrosion of Passive Metals and Coatings*; Woodhead Publishing Series in Metals and Surface Engineering; Elsevier: Amsterdam, The Netherlands; Woodhead Publishing: Cambridge, UK, 2011.
23. Brunette, D.M.; Tengvall, P.; Textor, M.; Thomsen, P. (Eds.) *Titanium in Medicine: Material Science, Surface Science, Engineering, Biological Responses and Medical Applications*; Springer: Berlin/Heidelberg, Germany, 2001.
24. Macdonald, D.D. The point defect model for the passive state on metals and alloys. *Corros. Sci.* **2015**, *99*, 30–45. <https://doi.org/10.1016/j.corsci.2015.01.006>.
25. Wang, J.; Song, S.; Li, D. Diffusivity of a point defect in the passive film on Ti with plastic deformation in 3.5% NaCl solution. *Thin Solid Films* **2022**, *754*, 139304. <https://doi.org/10.1016/j.tsf.2022.139304>.
26. ASTM G-102-89; Standard Practice for Calculation of Corrosion Rates and Related Information from Electrochemical Measurements. ASTM International: West Conshohocken, PA, USA, 2010.
27. Guangling, S. Control of biodegradation of biocompatible magnesium alloys. *J. Corros. Sci.* **2007**, *49*, 1701. <https://doi.org/10.1016/j.corsci.2007.01.001>.

28. Hornberger, H.; Virtanen, S.; Boccaccini, A.R. Biomedical coatings on magnesium alloys—A review. *Acta Biomater.* **2012**, *8*, 2442–2455. <https://doi.org/10.1016/j.actbio.2012.04.012>.
29. Witte, F.; Hort, N.; Vogt, C.; Cohen, S.; Kainer, K.U.; Willumeit, R.; Feyerabend, F. Degradable biomaterials based on magnesium corrosion. *Acta Biomater.* **2008**, *4*, 1024–1034. <https://doi.org/10.1016/j.actbio.2008.02.011>.
30. Durrani, F.; Nahid, R.; Pandey, S.; Singh, P.; Pandey, A. One-piece implants: Careful approach for complex rehabilitation. *Natl. J. Maxillofac. Surg.* **2021**, *12*, 266–270. https://doi.org/10.4103/njms.NJMS_191_20.
31. Prithviraj, D.R.; Gupta, V.; Muley, N.; Sandhu, P. One-piece implants: Placement timing, surgical technique, loading protocol, and marginal bone loss. *J. Prosthodont.* **2013**, *22*, 237–244. <https://doi.org/10.1111/j.1532-849X.2012.00928.x>.
32. Nagay, B.E.; Cordeiro, J.M.; Barao, V.A.R. Insight Into Corrosion of Dental Implants: From Biochemical Mechanisms to Designing Corrosion-Resistant Materials. *Curr. Oral Health Rep.* **2022**, *9*, 7–21. <https://doi.org/10.1007/s40496-022-00306-z>.
33. Lozano, P.; Peña, M.; Herrero-Climent, M.; Rios-Santos, J.V.; Rios-Carrasco, B.; Brizuela, A.; Gil, J. Corrosion Behavior of Titanium Dental Implants with Implantoplasty. *Materials* **2022**, *15*, 1563. <https://doi.org/10.3390/ma15041563>.
34. García-Galvan, F.R.; Fajardo, S.; Barranco, V.; Feliu, S., Jr. Experimental Apparent Stern–Geary Coefficients for AZ31B Mg Alloy in Physiological Body Fluids for Accurate Corrosion Rate Determination. *Metals* **2021**, *11*, 391. <https://doi.org/10.3390/met11030391>.
35. Swalsky, A.; Noubissi, S.S.; Wiedemann, T.G. The systemic and local interactions related to titanium implant corrosion and hypersensitivity reactions: A narrative review of the literature. *Int. J. Implant. Dent.* **2024**, *10*, 58. <https://doi.org/10.1186/s40729-024-00578-3>.
36. Sun, Y.; Shukla, A.; Ramachandran, R.A.; Kanniyappan, H.; Yang, B.; Harlow, R.; Campbell, S.D.; Thalji, G.; Mathew, M. Fretting-corrosion at the Implant-Abutment Interface Simulating Clinically Relevant Conditions. *Dent. Mater.* **2024**, *40*, 1823–1831. <https://doi.org/10.1016/j.dental.2024.08.007>.
37. Aparicio, C.; Gil, F.J.; Fonseca, C.; Barbosa, M.; Planell, J.A. Corrosion behaviour of commercially pure titanium shot blasted with different materials and sizes of shot particles for dental implant applications. *Biomaterials* **2003**, *24*, 263–273.
38. Kunrath, M.F.; Garaicoa-Pazmino, C.; Giraldo-Osorno, P.M.; Mustafa, A.H.; Dahlin, C.; Larsson, L.; Farah Asa'ad, F. Implant surface modifications and their impact on osseointegration and peri-implant diseases through epigenetic changes: A scoping review. *J. Periodontol. Res.* **2024**, *59*, 1095–1114.
39. Rimondini, L.; Farè, S.; Brambilla, E.; Felloni, A.; Consonni, C.; Brossa, F.; Carrassi, A. The effect of surface roughness on early in vivo plaque colonization on titanium. *J. Periodontol.* **1997**, *68*, 556–562. <https://doi.org/10.1902/jop.1997.68.6.556>.
40. Toledano-Serrabona, J.; Sánchez-Garcés, M.A.; Gay-Escoda, C.; Valmaseda-Castellon, E.; Camps-Font, O.; Verdeguer, P.; Molmeneu, M.; Gil, F.J. Mechanical properties and corrosion behavior of Ti6Al4V particles obtained by Implantoplasty. An in vivo study. Part. II. *Materials* **2021**, *14*, 6519.
41. Al-Hity, R.R.; Kappert, H.F.; Viennot, S.; Dalard, F.; Grosogeat, B. Corrosion resistance measurements of dental alloys, are they correlated? *Dent. Mater.* **2007**, *23*, 679–687.
42. Rasouli, R.; Barhoum, A.; Uludag, H. A review of nanostructured surfaces and materials for dental implants: Surface coating, patterning and functionalization for improved performance. *Biomater. Sci.* **2018**, *6*, 1312–1338.
43. Fonseca, D.; de Tapia, B.; Pons, R.; Aparicio, C.; Guerra, F.; Messias, A.; Gil, J. The Effect of Implantoplasty on the Fatigue Behavior and Corrosion Resistance in Titanium Dental Implants. *Materials* **2024**, *17*, 2944. <https://doi.org/10.3390/ma17122944>.
44. Velasco-Ortega, E.; Monsalve-Guil, L.; Jiménez-Guerra, A.; Ortiz, I.; Moreno-Muñoz, J.; Nuñez-Marquez, E.; Pequerolles, M.; Perez, R.A.; Gil, F.J. Importance of the roughness and residual stresses of dental implants on fatigue and osseointegration behavior. In vivo study in rabbits. *J. Oral Impl.* **2016**, *42*, 469–476.
45. Robles, D.; Brizuela, A.; Fernández-Domínguez, M.; Gil, J. Corrosion Resistance and Titanium Ion Release of Hybrid Dental Implants. *Materials* **2023**, *16*, 3650. <https://doi.org/10.3390/ma16103650>.
46. Sanjuán Álvarez, M.; Robles, D.; Mur, J.G.; Fernández-Hernández, S.; Pérez-Pevida, E.; Brizuela-Velasco, A. Residual Stress Homogenization of Hybrid Implants. *Bioengineering* **2024**, *11*, 1149. <https://doi.org/10.3390/bioengineering11111149>.
47. Callister, W.D.; Rethwisch, D.G. *Materials Science and Engineering: An Introduction*, 10th ed.; Wiley: Hoboken, NJ, USA, 2018.
48. Gil, F.J.; Rodríguez, D.; Planell, J.A.; Cortada, M.; Giner, L.; Costa, S. Galvanic corrosion behaviour of Titanium implants coupled to dental alloys. *J. Mater. Sci. Mater. Med.* **2000**, *11*, 287–293.
49. Noronha Oliveira, M.; Schunemann, W.V.H.; Mathew, M.T.; Henriques, B.; Magini, R.S.; Teughels, W.; Souza, J.C.M. Can degradation products released from dental implants affect peri-implant tissues? *J. Periodontol. Res.* **2018**, *53*, 1–11. <https://doi.org/10.1111/jre.12479>.

50. Kotsakis, G.A.; Xie, L.; Siddiqui, D.A.; Daubert, D.; Graham, D.J.; Gil, F.J. Dynamic assessment of titanium surface oxides following mechanical damage reveals only partial passivation under inflammatory conditions. *npj Mater. Degrad.* **2024**, *8*, 98. <https://doi.org/10.1038/s41529-024-00514-1>.
51. Haugli, K.H.; Syverud, M.; Samuelsen, J.T. Ion release from three different dental alloys-effect of dynamic loading and toxicity of released elements. *Biomater. Investig. Dent.* **2020**, *7*, 71–79. <https://doi.org/10.1080/26415275.2020.1747471>.
52. Asri, R.I.M.; Harun, W.S.W.; Samykano, M.; Lah, N.A.C.; Ghani, S.A.C.; Tarlochan, F.; Raza, M.R. Corrosion and surface modification on biocompatible metals: A review. *Mater. Sci. Eng. C Mater. Biol. Appl.* **2017**, *77*, 1261–1274. <https://doi.org/10.1016/j.msec.2017.04.102>.
53. Hogg, W.W.; Jamal, M.; Zuckschwerdt, N.W.; Hess, C.M.; Bose, S.; Bandyopadhyay, A. Corrosion-fatigue of additively manufactured Ti6Al4V. *J. Mech. Behav. Biomed. Mater.* **2026**, *175*, 107289. <https://doi.org/10.1016/j.jmbbm.2025.107289>.
54. Dodig, S.; Čepelak-Dodig, D.; Gretić, D.; Čepelak, I. Tattooing: Immediate and long-term adverse reactions and complications. *Arh. Hig. Rada Toksikol.* **2024**, *75*, 219–227. <https://doi.org/10.2478/aiht-2024-75-3921>.
55. Serup, J.; Carlsen, K.H.; Sepehri, M. Tattoo complaints and complications: Diagnosis and clinical spectrum. *Curr. Probl. Dermatol.* **2015**, *48*, 48–60. <https://doi.org/10.1159/000369645>.
56. Niinomi, M. Mechanical properties of biomedical titanium alloys. *Mater. Sci. Eng. A* **1998**, *243*, 231–236. [https://doi.org/10.1016/S0921-5093\(97\)00806-X](https://doi.org/10.1016/S0921-5093(97)00806-X).
57. Velasco-Ortega, E.; Flichy-Fernández, A.; Punset, M.; Jiménez-Guerra, A.; Manero, J.M.; Gil, J. Fracture and Fatigue of Titanium Narrow Dental Implants: New Trends in Order to Improve the Mechanical Response. *Materials* **2019**, *12*, 3728. <https://doi.org/10.3390/ma12223728>.
58. Geetha, M.; Singh, A.K.; Asokamani, R.; Gogia, A.K. Ti based biomaterials, the ultimate choice for orthopaedic implants—A review. *Prog. Mater. Sci.* **2009**, *54*, 397–425. <https://doi.org/10.1016/j.pmatsci.2008.06.004>.
59. Olmedo, D.G.; Tasat, D.R.; Guglielmotti, M.B.; Cabrini, R.L. Titanium particles and osteolysis in peri-implant tissues. *J. Biomed. Mater. Res. Part A* **2003**, *65*, 43–49. <https://doi.org/10.1002/jbm.a.10432>.
60. Niinomi, M. Mechanical biocompatibilities of titanium alloys for biomedical applications. *J. Mech. Behav. Biomed. Mater.* **2008**, *1*, 30–42. <https://doi.org/10.1016/j.jmbbm.2007.07.001>.
61. Fretwurst, T.; Buzanich, G.; Nahles, S.; Woelber, J.P.; Riesemeier, H.; Nelson, K. Metal elements in tissue and their association with peri-implantitis. *Clin. Oral Implant. Res.* **2016**, *27*, 1169–1177. <https://doi.org/10.1111/clr.12721>.

Disclaimer/Publisher’s Note: The statements, opinions and data contained in all publications are solely those of the individual author(s) and contributor(s) and not of MDPI and/or the editor(s). MDPI and/or the editor(s) disclaim responsibility for any injury to people or property resulting from any ideas, methods, instructions or products referred to in the content.