



Trans tibial Bone-Anchored Prosthesis: Experimental and Theoretical Study

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Abstract

In this study, an experimental and numerical approach is used to evaluate a trans tibial bone-anchored prosthesis system that includes an osseointegration implant composed of Ti-6Al-7Nb. The work addresses three factors that are critical to clinical and biomechanical performance: (1) the mechanical properties and corrosion behavior of the implant materials, (2) enhancement of bioactivity via hydroxylapatite (HAP) surface coatings, and (3) structural response of the tibia-implant construct under physiologically representative gait loads. A compressive test was also realized, and the good mechanical behavior of Ti-6Al-7Nb ($\sigma_u = 1271$ MPa; $\sigma_y = 755$ MPa) revealed its ability to be used for load-bearing applications. The HAP coatings had a homogeneous nanostructure, greatly enhanced corrosion resistance (89% protection efficiency), and induced apatite-like formation in simulated body fluid. Peak ground reaction force was 1060 N, recorded using gait analysis, and used for implant biomechanics analysis in FE modelling. The equivalent stress of the model was low (11.95 MPa), its deformation small (0.114 mm), and the safety factor high (11.7), indicating that the implant-bone interface had good structural stability. Taken together, these data demonstrate that Ti-6Al-7Nb osseointegrated prostheses for trans tibial amputees exhibit good mechanical reliability and bioactivity, offering a stable load-transfer profile suitable for long-term clinical application.

Keywords: Bone-Anchored, Ti-6Al-7Nb Alloy, Hydroxyapatite Coating, Finite Element Analysis, Gait Analysis.

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1. Introduction

Traditional prosthesis fitting for individuals with trans tibial amputations comprises a socket designed for the residual limb and a supportive girdle to ensure a secure fit and complete range of motion. Sockets can be challenging for some individuals due to limited mobility, contact dermatitis, blisters, and discomfort during sitting or engaging in weight-bearing activities [1]. Osseointegration, however, a method of leeching the skeleton to allow attachment for prosthetic purposes, will provide greater potential and stability for amputees. Their review considers recent advances, advantages, and remaining challenges of this strategy in the context of current prostheses [2]. This is where prosthesis osseointegration aims to aid, or at least lessen, some of these issues. Research found that 90% of implants in the mandible had survived after 15 years [3]. Osseointegration involves screwing the fixture into close contact with bone to make a lasting connection for an external prosthesis. During osseointegration, the fixture is placed very close to bone tissue to create a strong, long-lasting attachment for the external prosthesis [4]. This method improves limb control, eases prosthetic handling, and eliminates range-of-motion limitations caused by sockets and socket-related skin problems [5–8]. A transcutaneous osseointegration implant might be a solution for prostheses attached to bone [9]. Osseointegration occurs when the

titanium oxide surface of an implant and bone tissue come into close contact without fibrous tissue interposed [10]. A titanium cylinder with threads is surgically implanted into the medullary cavity of the remaining bone. This treatment is simple. It is recommended to wait six months for the implant to fully integrate before inserting a skin-penetrating extension (abutment) to secure the external prosthesis. The following step is a well-planned rehabilitation program [11]. The abutment's detachable design makes it easy to swap it out in the event of damage or twisting. This method helps certain amputees (often those with amputations from trauma or tumors) cope with their prostheses better and have more control over their limbs. It also enhances their quality of life [12–14]. Prosthetic joint infections may occur in people with diabetes, RA, renal failure, malnutrition, immunosuppression, or wound infections, or those with *S. aureus* nasal carriage [15]. Obesity is a distinct risk factor for periprosthetic infection [16]. Having diabetes appears to make osseointegration more difficult, and smoking prevents bones from healing properly [17–20]. A bone-anchored prosthesis (BAP) that utilizes an osseointegration implant (OI) is an excellent alternative to socket-suspended prostheses (SSP) for amputee patients who suffer from pain, pressure sores, and restricted mobility [21]. Among the many advantages of OI is the direct skeletal link it provides to prosthetic limbs [22]. Better sitting and walking conditions, enhanced osseoperception, and more stable, physiological prosthetic control

are all outcomes of this approach, which also eliminates the socket-residuum interface and its associated problems. [23–26]. There are currently two OI systems available for purchase. Integrum AB, Sweden markets the oldest and most extensively studied titanium screw fixation system, OPRA, as the Osseointegrated Prostheses for the Rehabilitation of Amputees Implant system [27]. The Swedish Brånemark group developed the system. There is another, slightly more contemporary OI method that the osseointegration organizations of Germany, the Netherlands, and Australia developed and use: the press-fit attachment technique. It may be purchased as an osseointegrated femoral or tibial prosthesis or as an integral leg prosthesis (ILP) [28–33]. A press-fit implant can achieve full weight-bearing much more quickly than screw-type implants because it requires less overall treatment time. [34]. A recent systematic review on safety found that press-fit implants had a somewhat higher femoral OI survival rate than screw implants [35]. The objectives of this work were to evaluate the mechanical properties of a Ti6Al7Nb alloy implant, examine the gait of a patient who had a below-knee amputation and an osseointegration prosthesis, and use Ansys Workbench 17.2 to conduct numerical simulations for a transtibial bone model with an osseointegration implant.

2. Experimental Procedures

2.1 Material Selection for the Implant

The experimental materials used were cylindrical sections of the Ti-6Al-7Nb-ASTM F1295 alloy [36], with a diameter of 20 mm and a length of 20 mm, as shown in Figure 1. In this case, Baoji Jinsheng Metal Material Co., LTD. supplies the samples. Mechanical testing, particularly compressive testing, relies on standardized specimens, and this sample served that purpose. The required geometries were achieved using wire electrical discharge machining. Haas Automation Inc. of Oxnard, California, USA, uses a CNC lathe, ST-10, and a CNC milling machine, VF-1, to produce prefabricated rods of reduced size, which are then adjusted to meet the necessary standards. They were then used to create and alter smaller prefabricated rods according to the specified criteria. X-Ray Fluorescence (XRF) testing was conducted in the material laboratory at Iraq's Ministry of Science and Technology in Baghdad to perform the chemical analysis. The chemical composition of the Ti-6Al-7Nb alloy, as determined by XRF testing, is described in detail in Table 1.



Figure 1. The Ti-6Al-7Nb specimen

2.2 Mechanical Parameters

The mechanical properties and tensile strength of the Ti-6Al-7Nb used in osseointegration implants may be evaluated using compression testing equipment. The mechanical characteristics were determined by assessing the following parameters: Essential for

understanding a material's durability and stretching pressure tolerance, as well as its mechanical capabilities, robustness, and suitability for various uses like implant material in prostheses, are its tensile strength (σ_u , MPa), yield strength (σ_y , MPa), and elastic modulus (E , GPa). A Ti-6Al-7Nb specimen was subjected to a constant 2 mm/min crosshead speed until it broke or shattered in a static compression test. To guarantee consistency and collect essential data, this approach was applied to three samples in total. Cyclic stress application is the standard procedure for fatigue testing spinning specimens made of the Ti-6Al-7Nb alloy. This type of test often assesses the fatigue strength and lifetime of materials, in particular under rotating loads. This knowledge of a material's fatigue strength under long-term cyclic loading with rotational motion can be extremely useful for engineering and design. The S-N curve for human bone was based on the extensive data from Swanson and his co-workers' research [37]. The specimens are subjected to a cycle of compressive and tensile loads with zero mean stress during the test procedure. To investigate the prosthetic function of the osseointegrated TT-prosthesis, results from the experiments described above are essential.

Table 1. Ti-6Al-7Nb Base Alloy's chemical composition (wt%)

Elements	wt. %
Al	6.000
Nb	7.100
Fe	0.190
C	0.018
N	0.024
H	0.005
O	0.080
Ti	86.2
Residual element	0.383
Sum	99.617

2.3 In-vitro Test

The coating solution was prepared by dissolving 2 g PVP K30 (CDH, India) in 8 mL of water and stirring vigorously with a magnetic stirrer (VS-130SH, Korea) until complete dissolution. The PVP adheres well to the coating layer and provides an anchor. Polyvinylpyrrolidone (PVP) can be a suitable candidate for this purpose. PVP has excellent water-absorbing properties and good biocompatibility [38, 39]. The solution was then brought to a volume of 70 mL by adding 60 mL of absolute ethanol (99%, Sigma-Aldrich, USA) with vigorous stirring. Then, 10 g of hydroxyapatite (<40 nm, Sky Spring Nanomaterials) were added to the mixture and stirred vigorously with a magnetic stirrer. The coating application process consisted of heating the specimen made of Ti-6Al-7Nb alloy to 80°C for 2 minutes, then coating it using a spray gun connected to a 5-bar compressor with a gun cup loaded with the solution ready for spraying, as shown in Figure 2. After the coating layer was applied, the Ti-6Al-7Nb alloy disc was allowed to dry for 2 minutes. The materials were characterised using XRD, FESEM, and EDX, as well as an adhesion test. Corrosion test. The corrosion behavior was investigated via OCP and polarization curve (Tafel) measurements in SBF solution using a potentiostat (CHI 604E; China). The composition of SBF is shown in Table 2, and the preparation details are provided in the Methods section. The formation of a bony tissue-like layer on the surface of osseointegrated implants represents one of the essential conditions for good bonding with live bone and tissue. [40].



Figure 2. Coating system with pyrolysis spray.

Table 2. Chemical composition of simulated body fluid (SBF) solution [41].

ITEM	Concentration (g/L)
NaCl	8.035
NaHCO ₃	0.355
KCl	0.255
K ₂ HPO ₄	0.231
MgCl ₂ ·6H ₂ O	0.311
CaCl ₂	0.292
Na ₂ SO ₄	0.072

2.4 Experimental Evaluation of Loads

The patient shown in Figure 3 was examined on a P-walk plate in the prosthetics and orthotics engineering department at Al-Nahrain University, with ethical approval dated 09/2025. The patient's details were a height of 1.75 meters, male gender, a right-side below-knee amputation, and a weight of 85 kilograms. The objective was to determine the curve load time and the maximum ground reaction force in different walking situations for the prosthetic osseointegration limb. The maximal force magnitude will be used in Finite Element Modeling (FEM) using ANSYS 17.2 to analyze the total deformation and stress distribution of the implant and bone.



Figure 3. Patient with a transtemporal bone anchored.

3. Finite Element Method Model Analysis

The model prosthetic implant seen in Fig. 5 may be classified into two components:

Fixed Intramedullary Stem: The residual limb, or the portion of the leg remaining after amputation, is engineered to accommodate this

prosthetic device inside its medullary canal. The prosthesis, often made of a biocompatible material, is reinforced and anchored by the intramedullary stem. **Transcutaneous Component (Abutment):** This element functions as a link between the artificial and host systems' legs, attaching to the intramedullary stem and penetrating through the skin. The prosthetic elements are securely connected to the intramedullary shaft by abutments, ensuring a stable connection. The SolidWorks platform model of the implant and bone allows the study, optimization, and visualization of the prosthetic model. This enables researchers and engineers to titrate the exact dimensions of the implant to optimise fit, biomechanical compatibility, and hence function. It is particularly useful for people who have a BK amputation. The technology enables accurate alterations and improvements to the prosthetic design, improving performance and fit for each user.

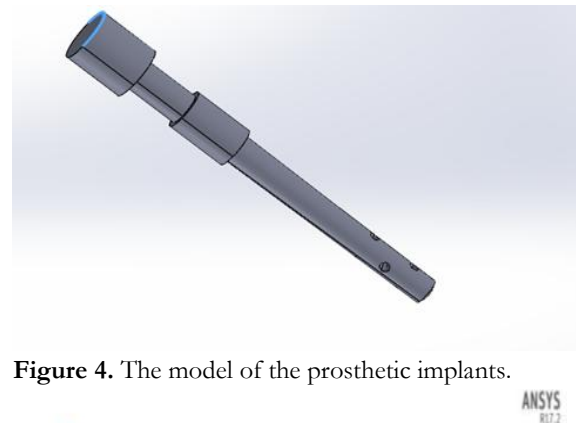


Figure 4. The model of the prosthetic implants.

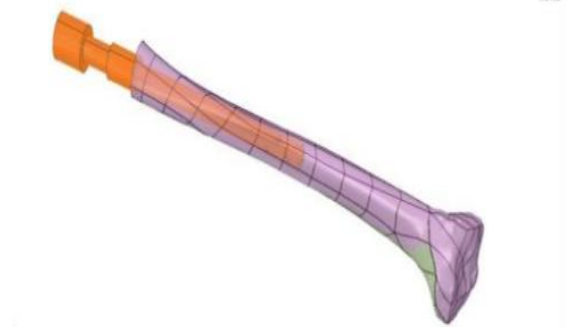


Figure 5. The model of the implant/bone assembly.

The study employed ANSYS Workbench (version 17.2) for FEM-based numerical simulations to investigate the relationship between total deformation and stress distributions. Ten-noded tetrahedral elements (SOLID187), as depicted in Figure 7, were applied to mesh the implant and tibial bone. Such polyhedral complexes are frequently employed to discretize and approximate the geometry of solid bodies in FEM simulations. Structures [42].

Modeling of interactions among diverse model parts, facilitating the investigation of forces and contact behavior in the mechanical connection between the prosthesis and the bone, as well as in load transmission. The model's analysis incorporates boundary conditions, including rigid support at the implant and an applied load simulating the patient's weight at the proximal end of the tibia, as seen in Fig. 8. These fundamental boundary conditions are necessary to model and assess the performance and efficiency of the prosthetic system under specific loading and support situations. Table 3 delineates the mechanical characteristics of cortical bone.

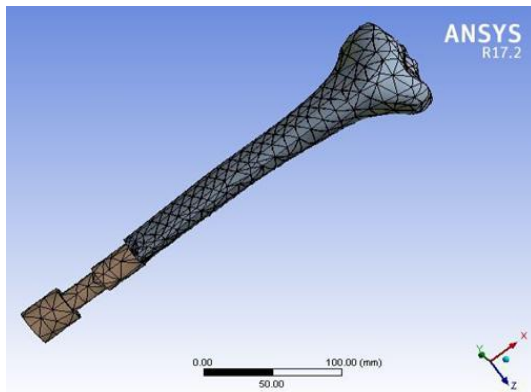


Figure 6. Meshed model with ten-node tetrahedral elements.

Table 3. Mechanical properties of the cortical bone [43].

Material	σ_y (MPa)	σ_u (MPa)	E (GPa)
Cortical bone	175	205	20

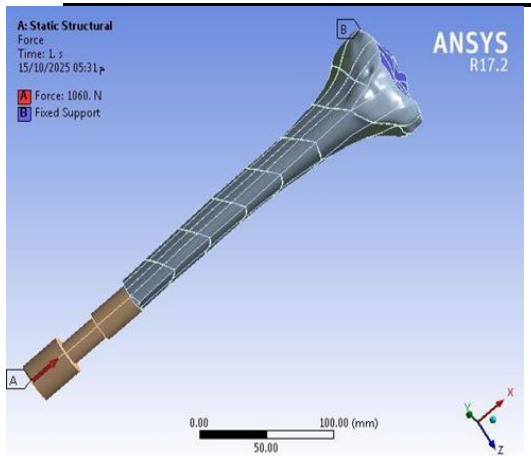


Figure 7. Boundary condition of the model.

4. Results and Discussion

4.1 Compression properties results

The statistics of average parameters, average compressive strength (CS), and average compression yield strength (YSc), obtained through static compression test, are as follows: Mean value for compressive strength is 1271.483 MPa. On the contrary, the YSc average exceeded the reference values with 755.064 MPa (10). Fig. 2. The typical compression stress-strain curves of the corresponding groups were presented in Fig. 9. A remarkable combination of high strength and increased ductility for the Ti-6Al-7Nb alloy. Materials with these same properties are needed for biomedical applications, especially for prosthetic devices that must withstand high-impact loads without failure. The Ti-6Al-7Nb compression test results provide insight into how the material responds to compressive loads experienced during weight-bearing activities such as standing or walking. Implant-related materials, in particular osseointegrated prosthetic materials, should have high strength that will resist loads without damaging or crushing. The experimental compressive test data for the Ti-6Al-7Nb material confirm the material's strength under mechanical loading in biomedical applications. This protects the function and performance of prosthetic implants by enhancing their longevity, safety, and bone-tissue compatibility.

4.2. XRD test

Figure 10 shows the XRD pattern of the uncoated sample, with titanium-labeled Miller indices at each peak location. As shown in

Figure 11, the XRD pattern of the coated sample displays peaks for hydroxyapatite and titanium, with Miller indices indicated. The hydroxyapatite coating on the samples' surfaces diminished the titanium peaks.

Table 4. Mechanical properties of Ti-6Al-7Nb alloy in compression test.

Sample	YSc(MPa)	σ_u (MPa)
1	785,189	1,269.75
2	699,971	1,271.20
3	780,032	1,273.50

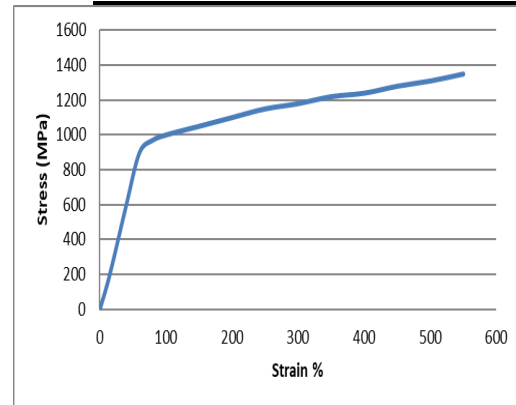


Figure 8. stress -strain curve for compression test.

To demonstrate this, the augmented peaks for titanium and hydroxyapatite are shown with their Miller indices in Figure 11. On the other hand, it is titanium according to the same index as the sample without coating. As illustrated in the subsequent images, the titanium peaks decrease as the sample coating layers are deposited, and the hydroxyapatite peaks increase. Characteristic peaks of hydroxyapatite and the Ti-6Al-7Nb alloy are present in the titanium spectrum, as indicated by the Miller indices. The titanium peaks in the uncoated sample's XRD pattern have reduced intensity compared to those in the coated sample, indicating a greater hydroxyapatite coating. In addition, the number of titanium peaks has decreased to two, indicating that the hydroxyapatite coating coverage has improved. Hydroxyapatite coats the sample surface so well that titanium is strongly suppressed. The coating formed on the surface is necessary to enhance the material's biocompatibility and osseointegration. By comparing the XRD patterns of the samples, it can be observed that there is an increase in intensity (hydroxyapatite) and a decrease in intensity (titanium).

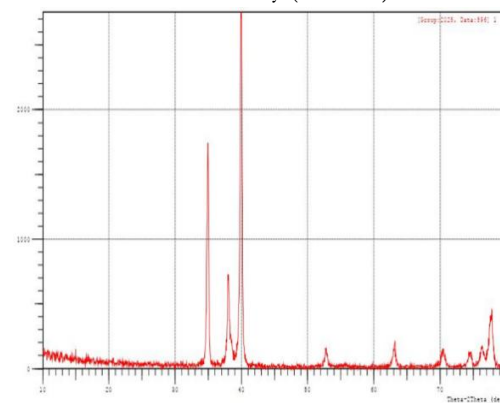


Figure 9. The XRD pattern of uncoated Ti-6Al-7Nb alloy.

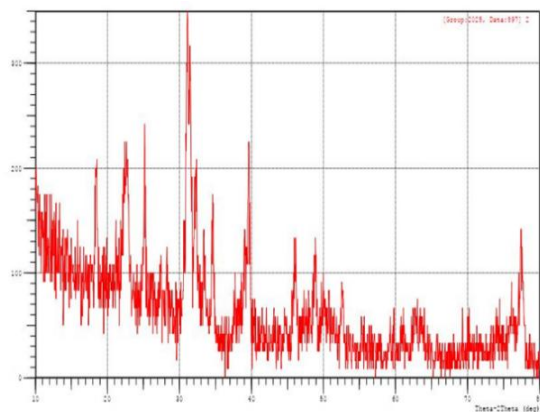


Figure 10. The XRD pattern for the coated sample.

4.3. FESEM test

Figure 12 presents an FESEM image of the basic sample, whereas Figure 13 shows photos of the coated sample at two magnifications, revealing a substantial coating across all samples. The sample with the coating layer exhibits an extremely porous structure. This occurs because the coating's sufficient thickness does not change as it is heated, so no cracks form. Moreover, the particle size of coated samples is in the range of 42.41-81.16 nm, and this would be considered optimal for higher osseointegration with implant and bone, indicating increased agglomeration with an increase in the number of layers coated. The FESEM images of the HaP coatings are presented after coating [3]. The test results show that the sprayed coatings are crack-free and void-free, with a smooth surface. In addition, the HaP particles were evenly distributed on the film. Grain shape and the regularity of grains in the HaP material were also affected by the doping concentration.

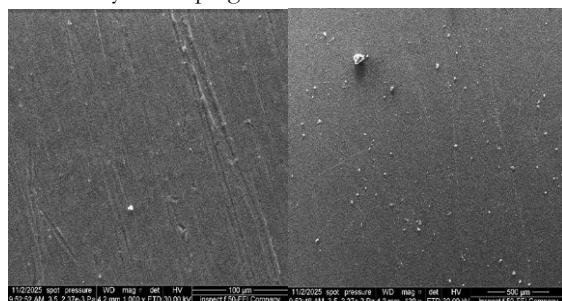


Figure 11. base sample of Ti-6Al-7Nb alloy.

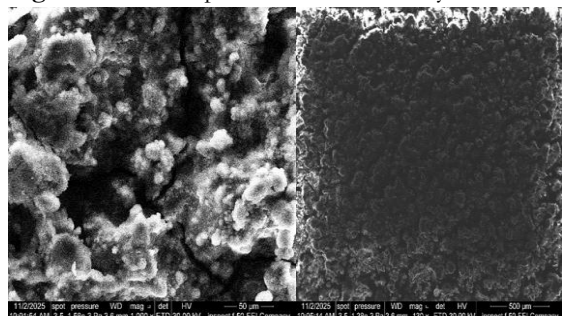


Figure 12. coated sample of Ti-6Al-7Nb alloy.

4.4. EDX Test

EDX analysis results of both (coated and uncoated) samples are depicted in Figures 14 and 15, with the major elements of the uncoated sample being titanium, niobium, and aluminum. The coated layer was predominantly composed of calcium, phosphorus, and oxygen, while titanium, niobium, or zirconium could not be

found. This occurred as the film was spread evenly across the metal components.

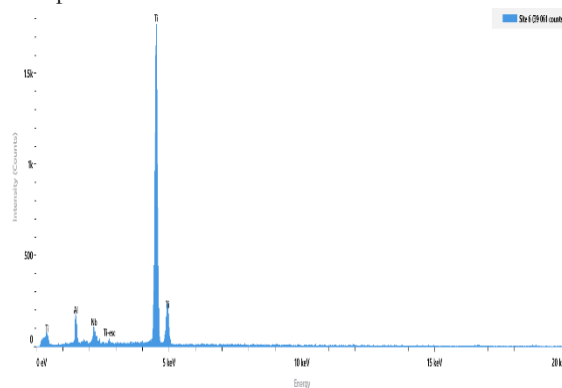


Figure 13. The EDX test for an uncoated sample of Ti-6Al-7Nb alloy.

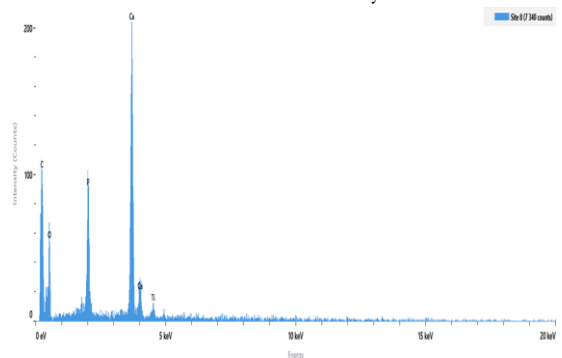


Figure 14. The EDX test for the coated sample of Ti-6Al-7Nb alloy.

4.5. Corrosion test

Electrochemical methods are extensively used to assess corrosion resistance and forecast the protective properties of certain biocompatible metals [44]. The OCP test results for the samples shown in Figure 16 and Table 5 demonstrate elevated OCP values, indicating that the coating enhances the samples' passivity. The OCP curves for the samples showed a marginal rise over 1200 seconds. The analysis of the electrochemical parameters indicated that with the HAp layer, the Eocp moved to more positive values, from -0.580 to -0.410 V.

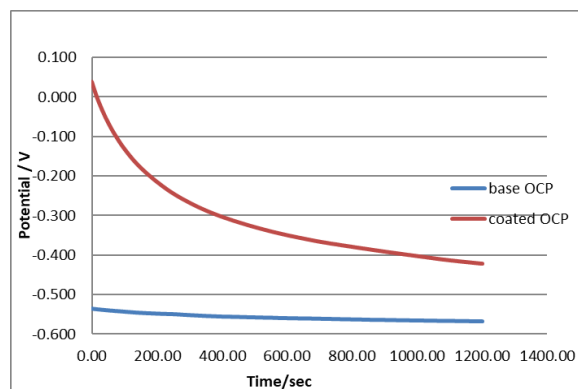


Figure 15. OCP testing for all samples.

Table 5. OCP test results for all samples.

ITEM	OCP (Volt)
Base OCP	-0.580
Coated OCP	-0.410

The polarization curve test for all samples shows enhanced corrosion resistance with increasing coated-layer thickness, as shown in Figure 17 and Table 6. The corrosion rate decreased from 2.427×10^{-2} mmpy for the untreated sample to 2.664×10^{-3} mmpy for the coated sample, representing a reduction of almost 21 times and resulting in a protection effectiveness of nearly 89%. The efficacy of protection may be determined using the following equation (Eq. 1) [45]:

$$PE \% = [(icorr.uncoated - icorr.coated) / icorr.uncoated] \times 100 \dots \dots \dots (1)$$

In this context, PE % signifies protection effectiveness, i_{corr} , uncoated represents the corrosion current of the uncoated sample, and i_{corr} , coated denotes the corrosion current of the coated sample. The Polarization Resistance (R_p) may be calculated using the Stern-Geary equation shown below (Eq. 2) [46]:

$$R_p = (b_a \times b_c) / [2.303(b_a + b_c) \sigma] \dots \dots \dots (2)$$

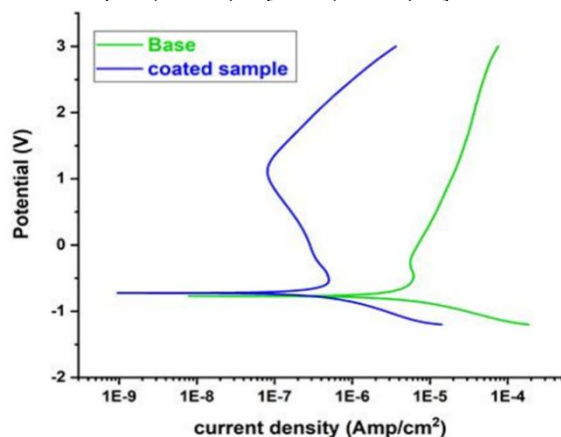


Figure 16. Polarization curve (Tafel) for all samples.

Table 6. Corrosion characteristics for all samples.

ITEM	E corr. (volt)	I corrected. (Amp.)	Corr. Rate mmpy
Base	-0.784	2.738×10^{-6}	2.427×10^{-2}
Coated sample	-0.725	3.005×10^{-7}	2.664×10^{-3}

4.6 Adhesion test

This study used the conventional adhesion cross-cut tape test, ASTM D3359, to assess the adherence of coated materials. Figure 18 and Table 7 provide the outcomes of the adhesion-coated sample, with a minimum removed area of 0% and a classification of 5B.

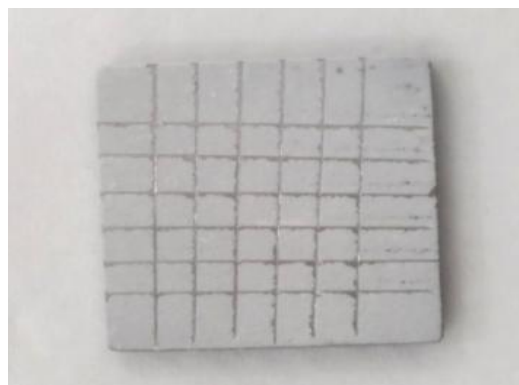


Figure 17. The adhesion test images for the coated sample.

Table 7. The adhesion test results according to ASTM D3359.

Item	Classification	Percentages are removed
HA-coated sample	5B	0%

4.7 After immersing in the SBF solution

4.7.1 XRD test

The XRD pattern shown in Figure 19 reveals the hydroxyapatite and titanium peaks for the uncoated sample after one month of immersion in SBF solution. An indication of bone osseointegration is the formation of hydroxyapatite peaks on the surface of the Ti-6Al-7Nb base. Figure 20 shows that hydroxyapatite appears in the XRD pattern of the coated sample after immersion in SBF solution for 1 month. The reason is that after one month, the surface was adequately covered by the new hydroxyapatite layer that developed from the SBF solution. This layer also served as a good signal for bone osseointegration.

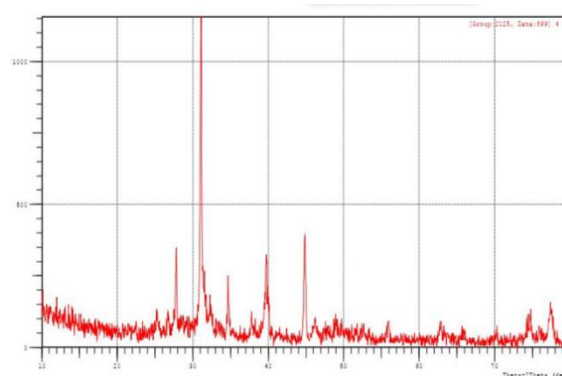


Figure 18. XRD pattern for the uncoated sample after immersing in SBF solution for one month.

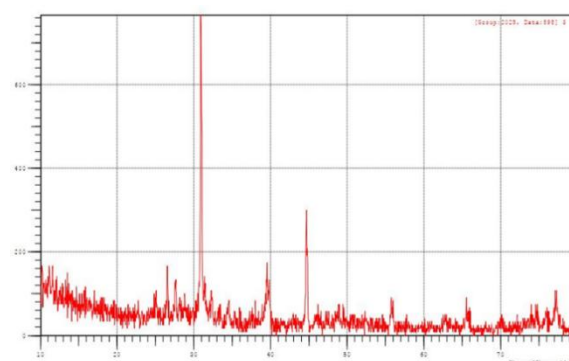


Figure 19. XRD pattern for the 5-layer sample after immersing in SBF solution for one month.

4.7.2 FESEM test

The FESEM images of the uncoated and coated samples after 1 month SBF immersion are represented in Fig 1.* Figure 21. The average particle size observed in the FESEM image for the new hydroxyapatite layer was between 43.44 nm and 80.17 nm. For the coated sample, Figure 22 shows that the particle sizes observed in the FESEM images of the new-layer hydroxyapatite were between 67.17 nm and 93.07 nm, with a significantly different appearance from what we observed in the previous immersion. A thick new hydroxyapatite layer originating from the SBF solution with a large number of fissures is convincingly formed on the base sample. The cracks account for the presence of titanium peaks in the earlier EDX analysis. The cracks are caused by the fact that the metallic substrate expanded faster than the ceramic layer during sintering. The coated sample with a homogeneously developed hydroxyapatite layer, devoid of fissures and covering the whole surface, accounts for the absence of titanium peaks in the preceding EDX analysis.

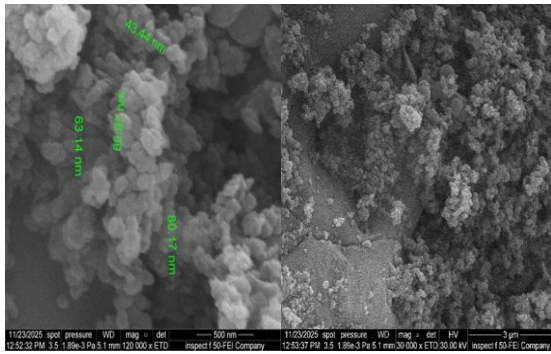


Figure 20. FESEM Images for uncoated base samples after immersing in SBF solution for one month.

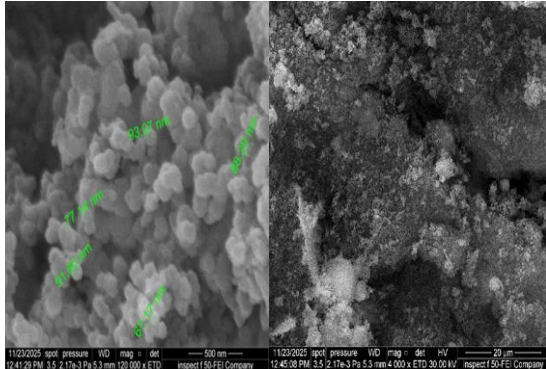


Figure 21. FESEM Images for coated base samples after immersing in SBF solution for one month.

4.7.3 EDX test

3.2 EDX Analysis The results of the EDX analysis of the uncoated sample after immersion in SBF solution for 4 weeks are shown in Figure 23. Calcium, phosphorus, and oxygen are present, which confirms the formation of hydroxyapatite. Besides, trace elements such as Si, Cl, K, Mg, Na, and S are present in hydroxyapatite precipitated from SBF solution. The titanium, zirconium, and niobium are due to the base alloy, as highlighted by fissuring in the hydroxyapatite layer. The five-layer-coated sample (Figure 24) has mostly the same elements as the uncoated one, except for halftime, which is explained by the final coating with layer five that covers the surface and blocks Ti, Al, and Nb.

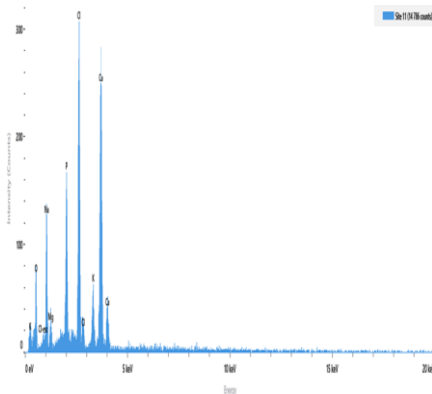


Figure 22. EDX test after immersing in SBF solution for one month for the uncoated sample.

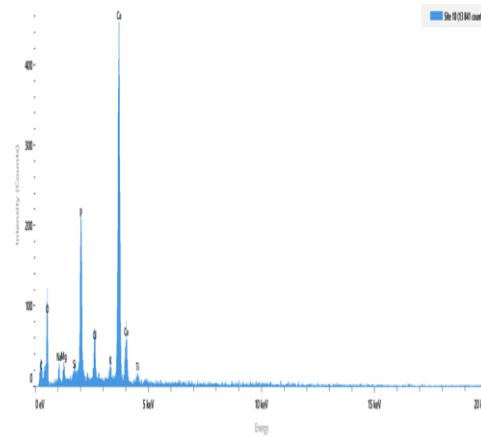


Figure 23. EDX test after immersing in SBF solution for one month for the coated sample.

4.8 P-WALK results

The P-WALK: We measured the forces and moments applied to an abutment of an implant during one gait cycle using a five-cycle plantar pressure analysis. This information might be critical to the loading of the implant. It is important to understand the peak stress that occurs during heel strike and toe-off. Any important variations between the right and left legs must be noted. An evaluation of loads on the implant's abutment and of its function and stability during walking can be conducted by considering loading conditions at heel strike and toe-off (peak forces and moments). Temporal variations comprised of force and pressure curves are illustrated in Fig. 25. Red and blue lines indicate the left and right leg, respectively. The maximum force is 1060 N; data are from the protocol described above, with measurements on a test bench once for the right-side appropriate prosthetic foot and once for the opposite side.

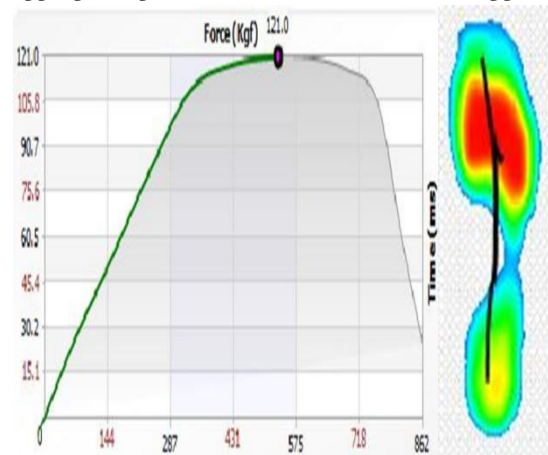


Figure 24. Force vs. Time

4.9 FEM Analysis Results

This safety factor, also referred to as bone strength, is distributed throughout the whole structural model (for example, see figure 26). The computed results demonstrate that the model is very safe and can withstand mechanical loads, with a maximum safety factor of 11.697 times the minimum allowable value, typically in the range of 1.5 to 2 for biomedical applications [47-50]. The blue colour in the figure indicates a trusted shape for use in clinical environments, since most of the structure and equipment have achieved a suitable level of safety. The findings also suggest that well-designed prosthetic limbs are important for achieving safety and functional performance

during prosthesis use. These findings could help make future prosthetic devices more effective.

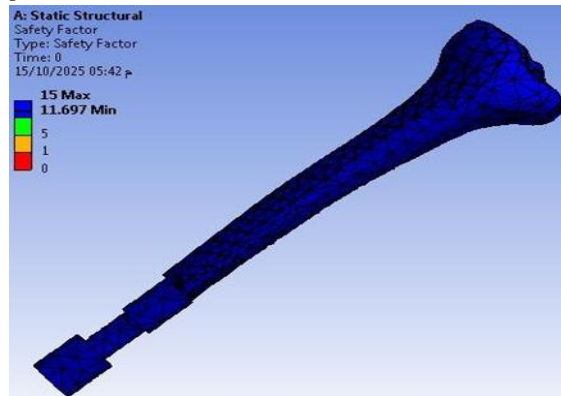


Figure 25. Safety factor.

An examination of the artificial bone's structural stress was conducted using ANSYS software. The equivalent (von Mises) stress was evaluated under certain loading circumstances. Mechanical pressures impact various portions of the artificial bone, and the resultant picture illustrates the distribution of stresses throughout the structure. Dark blue represents the lowest stress values, whereas brighter hues indicate regions of greater stress in the picture. The results demonstrate that the maximum equivalent stress may reach 11.946 MPa, which is within the allowed range for materials used to simulate bone (fig. 27). The findings show that the structure is well-designed and can withstand mechanical loads without deforming or failing, as most components are not under significant stress. The data also show that the pivot points are near the most stressed locations, so there needs to be further design refinement to ensure the stress is distributed more evenly.

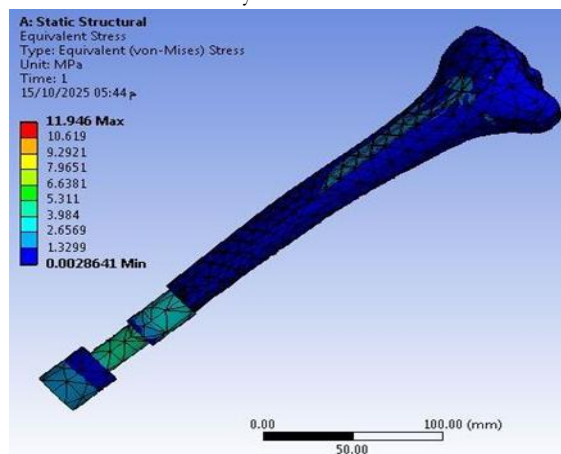


Figure 26. Von-Mises stress.

The total deformation of the artificial bone was analyzed under loading conditions. The picture reflects how the deformations are distributed throughout the structure, and thus provides important information about how bone responds to mechanical forces. Absolute deformation measures the extent to which a shape changes when it deforms. Colors in the image correspond to different levels of distortion: dark blue indicates the least, and brighter shades indicate areas of higher deformation. It is found that the maximum deformation is 0.11448 mm, as shown in Fig. 28, which is within the predetermined level considered satisfactory in the technical specifications of prosthetic legs. The results show that most of the

structure undergoes elastic deformation, demonstrating the design's ability to maintain its shape and function under mechanical loading. Among the most deformed areas are those under load, so the force distribution needs to be engineered to make these areas as nondeformable as possible. These findings are critical for future investigations, as they refine our understanding of how stressors affect the performance of synthetic bones. They also enable more efficient and user-friendly designs, improving prosthetic limb performance.

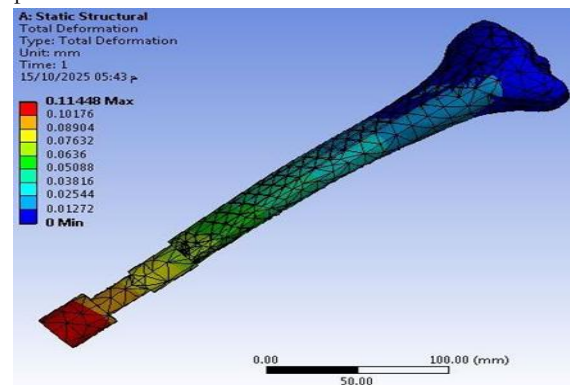


Figure 27. Total deformation.

5. Conclusion

1. The compressive strength of the Ti-6Al-7Nb alloy was found to be 1271.483 MPa, while its compression yield strength was 755.064 MPa, confirming its ability to support weight.
2. Analysis of Gait: Implant loading patterns were evaluated using force plate data collected throughout the gait cycle, which revealed ground response pressures and moments acting on the implant abutment. The highest recorded force was 1060 N.
3. The polarization curve test for all samples reveals enhanced corrosion properties with coated layers. The corrosion rate decreased from 2.427×10^{-2} mmpy for the untreated sample to 2.664×10^{-3} mmpy for the coated sample, representing a reduction of almost 21 times and resulting in a protection effectiveness of nearly 89%. The analysis of the electrochemical parameters indicated that with the HAP layer, the Eocp moved to more positive values, from -0.580 to -0.410 V.
4. The XRD analysis reveals the presence of intense HAP peaks, whereas the titanium peaks vanish entirely. This suggests that a fresh layer of hydroxyapatite has formed in the SBF solution. This new layer is visible in the FESEM images, providing strong evidence of osseointegration. Tests using potentiodynamic polarization and EIS reveal that HAP coatings are much more resistant to corrosion than the untreated material. The use of a simple, inexpensive spray pyrolysis approach to produce HAP makes this inquiry stand out.
5. The model is safe and can withstand mechanical stresses; its safety factor is 11.697 times the minimum criterion. The findings show that bone-simulating materials can withstand an equivalent stress of 11.946 MPa, the maximum. Since most components are unloaded, the structure is appropriately designed to resist distortion or damage. The reverse value of the maximum deformation is 0.11448mm, which satisfies the technical requirements of prosthetic limbs. The majority of the structure undergoes minor deformation, indicating that, to a great extent, it retains its form and utility under mechanical forces. Such observations offer insights into how stress influences

the functioning of artificial bone, which is important for future studies. They also enable user-friendly, efficient designs that improve prosthetic limb performance.

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