



NAZARBAYEV
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**Development and characterization of Ti-15Ta alloy
fabricated by Laser Powder bed fusion for biomedical
use**

(Final Report for Capstone Project)

by

Sabina Amangeldiyeva
Gaukhar Gabit
Assar Abilkhassen

Principal Supervisor: Asma Perveen

Co-supervisor: Didier Talamona

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Declaration

We hereby declare that this report entitled “Development and characterization of Ti-15Ta alloy fabricated by Laser Powder bed fusion for biomedical use” is the result of our own project work except for quotations and citations that have been duly acknowledged. We also declare that it has not been previously or concurrently submitted for any other degree at Nazarbayev University.



Name: Sabina Amangeldiyeva

Date: December 01, 2025



Name: Gaukhar Gabit

Date: December 01, 2025



Name: Assar Abilkhassen

Date: December 01, 2025

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Abstract

Titanium alloys are often used to make implants due to their biocompatibility, strength, and corrosion resistance. However, the commonly used Ti-6Al-4V alloy contains aluminum and vanadium, which may release harmful ions during long-term implantation. To address these problems, this project investigates the development of a Ti-15Ta alloy using Laser Powder Bed Fusion (LPBF) as a potential safer alternative for biomedical applications.

In this study, Ti and Ta powders were characterized using SEM and particle size distribution analysis before being blended to achieve an 85 wt% Ti–15 wt% Ta composition. For fabrication fourteen process parameter sets were evaluated with different hatch spacing values. The optimized LPBF parameters produced parts with ~100% relative density and minimized the unmelted Ta fraction to 3.27%. Microstructural analysis revealed a fine acicular martensitic structure due to the rapid cooling rates of LPBF and incomplete Ta dissolution. Mechanical testing showed an Ultimate Tensile Strength (UTS) of 598.2 MPa, a Yield Strength of 480.6 MPa, and an elongation of 12.7%. Compression tests indicated an Ultimate Compressive Strength of 1527 MPa and a Yield Strength of 1156 MPa. Vickers microhardness results were between 294 and 324 HV. However, the Young's modulus was measured at 140.3 GPa, which is higher than the target low-modulus range. Electrochemical testing demonstrated excellent corrosion resistance, showing an average corrosion current density of $1.11 \mu\text{A}/\text{cm}^2$ and a low corrosion rate of approximately 0.0074 mm/year after stabilization. The LPBF-processed Ti-15Ta alloy successfully achieved high density, favorable strength, and excellent corrosion resistance suitable for load-bearing biomedical applications. However, the presence of non-equilibrium martensitic phases resulted in a higher-than-expected elastic modulus.

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1. Introduction

Orthopaedics and dentistry additively manufactured metallic implants are gaining more and more popularity due to their ability to combine high load-bearing capacity with complicated, patient-specific geometry and controlled porosity that enables bone ingrowth. Titanium alloys, especially Ti-6Al-4V, are now the clinical standard as they have high specific strength, corrosion resistance and tend to be very biocompatible (Niinomi, 2016). Nevertheless, Ti-6Al-4V has a Young modulus of approximately 110 GPa that is considerably greater than that of cortical bone (around 10-30 GPa). This mechanical incongruity encourages stress shielding and bone resorption and possible implant loosening, and has as a result been a key stimulus in the creation of low-modulus titanium-based biomaterials (Niinomi, 2011). Another weakness of Ti-6Al-4V is the availability of aluminium and vanadium that has already been linked to potential chronic cytotoxic and neurotoxic outcomes when the ions are emitted into neighbouring tissues. The issue has prompted a move to titanium alloys reinforced with non-toxic β -stabilising elements like niobium, zirconium, tantalum and molybdenum (Niinomi, 2016). Tantalum is one of them: it has great corrosion resistance in physiological conditions, has a stable passive oxide layer, and is generally considered to be highly biocompatible (Guo et al., 2013). As an alloying element in titanium, tantalum is also an excellent β -phase stabiliser, and can greatly decrease the elastic modulus without loss of strength, or even increase it (Zhao et al., 2021).

Ti -Ta alloys thus are a promising material of the next generation implant material. Experimental research on wrought and additively manufactured Ti-Ta alloys have demonstrated that tantalum levels of between approximately 20 and 30 wt% can be used to provide tensile strengths of over one GPa and elastic moduli at 70 and 80 GPa respectively with excellent corrosion resistance and a positive in vitro response (Polozov et al., 2025).

Meanwhile, β -type titanium alloys, such as Ti-Ta, are typically considered to be more suitable as the stiffness matchers of bone and stress shielding reducers than the conventional α or (alpha-beta) titanium alloys (Niinomi, 2016). This has been made possible through the advent of laser powder bed fusion (LPBF) or selective laser melting (SLM), a potent entry point into process to make such advanced titanium alloys in porous or lattice structures, and an in-situ alloying method starting with metallic powders. LPBF creates near-net-shape parts by selective melting of layers of powder with a scanning laser beam, where the most important parameters through which porosity, microstructure and mechanical performance can be determined are laser power, scan speed, hatch spacing and layer thickness (Sing et al., 2018). In biomedical applications, LPBF is able to fabricate patient-specific geometries and internal lattice structures to tune stiffness (Sing, 2016).

LPBF processing of Ti is easy; however, processing of Ti-Ta alloys is non-trivial due to the very different melting points and densities of titanium and tantalum. Tantalum melts at a high temperature and is very dense, and when the process is not optimised, the Ta particles have an incomplete melting process, composition inhomogeneity and defect formation (Zhao et al., 2021). It has been found that fully dense parts can be released in the case of SLM of Ti-50Ta and Ti-25Ta, but unmelted Ta particles and heterogeneities introduced during the processing are still an issue (Sing et al., 2016). These observations add to the importance of defining good process-structure-property relationships in each individual Ti-Ta composition and LPBF parameter regime. This requirement of intermediate tantalum contents has begun to be taken up recently. Polozov et al. (2025) synthesized Ti 10Ta and Ti 15Ta and Ti 20Ta when SLM was used to powdered mixed elements of Ti and Ta and demonstrated that it is possible to achieve relative densities exceeding 99 percent along with low elastic moduli and high strength with appropriate choice of parameters. The optimum range of energy density gave relative densities of over 99.7 percent, with as-built Ti-15Ta having a final tensile strength of approximately 540 Mpa, elongation of more than 20 percent and a Young modulus of approximately 90 Gpa (Polozov et al., 2025). All of

these studies prove that Ti-15Ta is one of the most promising compositions as it provides a balance between lower modulus, sufficient strength, and processability with higher Ta contents. Although this has been done, there are still a number of critical gaps. First, a thorough understanding of how LPBF process parameters affect densification, microstructure, and mechanical behavior of Ti-15Ta is still developing because many current Ti-Ta LPBF studies concentrate on higher tantalum contents, such as Ti-25Ta and Ti-50Ta, or treat Ti-15Ta only within broader composition ranges (Polozov et al., 2025). Second, a thorough characterization of the combined impact of LPBF processing, post-processing heat treatment, and subsequent corrosion behavior in physiologically relevant conditions has not yet been achieved for Ti-15Ta. Third, systematic comparisons that directly relate Ti-15Ta LPBF processing conditions to properties relevant for actual implant geometries are still scarce, despite the literature's clear indication that Ti-Ta alloys can achieve lower moduli and favorable strength-to-stiffness ratios when compared with Ti-6Al-4V.

In light of this, the current capstone project examines Ti-15Ta made via laser powder bed fusion as a potential material for orthopedic and associated biomedical uses. This work focuses on Ti-15Ta produced on an LPBF system using blended Ti and Ta powders and methodically changed process settings, building on previous findings that Ti-15Ta can exhibit a Young's modulus around 90 GPa and high ductility in SLM-processed conditions (Polozov et al., 2025). The project's main goal is to assess the microstructure, mechanical performance, and corrosion behavior of LPBF-processed Ti-15Ta as a possible substitute for Ti-6Al-4V in biomedical implants.

To achieve this aim, the specific objectives of the study are:

1. To prepare and analyze the shape and particle size distribution of Ti and Ta powders and their Ti-15Ta blend in order to guarantee appropriate flowability and compositional homogeneity for LPBF processing.

2. To achieve dense Ti-15Ta samples with few defects and a regulated fraction of unmelted Ta particles by optimizing important LPBF/SLM process parameters (laser power, scan speed, hatch spacing, and layer thickness).
3. To examine the microstructure, mechanical behavior, and electrochemical performance of the LPBF-fabricated Ti-15Ta samples.

By achieving these objectives, the project hopes to support the ongoing development of low-modulus titanium-tantalum alloys and provide processing guidelines and property data that will aid in the future design of patient-specific, laser powder bed fusion-based implants based on Ti-15Ta.

Individual contributions

Sabina Amangeldiyeva did the SEM images of mixed Ti-15Ta powders; prepared, cleaned the printer and printed the second built platform; did the calculations of the tensile test data; did corrosion tests; performed literature review. *Gaukhar Gabit* did the SEM images of pure Ti and Ta powders; prepared, cleaned the printer and printed the third built platform; performed tensile test; did literature review; did vicker's microhardness test; *Assar Abilkhassen* did the SEM images of pure Ti and Ta powders; prepared, cleaned the printer and printed the first built platform; analyzed the tensile test results; performed literature review; did the compression test and microstructure analysis.

2. Literature Review

2.1 Titanium alloys and the stress-shielding problem

The biomaterials (metallic) employed to bear loads during implantations should be of sufficient strength and with long term resistance to corrosion and biocompatibility.

Titanium and Ti-6Al-4V alloys that are commercially pure have been widely used due to their high fatigue strength as they form a passivating oxide surface and have the ability to undergo reliable osseointegration (Long and Rack, 1998). They however have very high elastic modulus (approximately 110 GPa Ti-6Al-4V) when compared to cortical bone (10-30 GPa) and cancellous bone (0.1-4 GPa) thus facilitating stress shielding and eventual bone resorption surrounding the implant. Cheong et al. (2018) clearly list the stress shielding of porous Ti-6Al-4V structures as one of the key factors of bone loss and possible implant failure.

In order to reduce stress shielding, porous titanium scaffolds and collars manufactured via selective laser melting (SLM) or electron beam melting (EBM) have been created. Such porous structures reduce the stiffness of the structure to a more bone-like value and allow the bone tissue to infiltrate the implant and enhance the stress transfer and fixation (Arabnejad et al., 2016; Wang et al., 2016). However, despite the benefits of optimized pore structures, the large intrinsic modulus of Ti-6Al-4V is still a limitation to long-term bone remodelling. In their study, Cheong et al. (2018) demonstrate that when Ti-6Al-4V porous plugs were used in an ovine condyle model, bone formation continued to be focused on the outer pores without extending to areas further. This has prompted the design of titanium alloys with lower moduli that are specifically geared towards orthopaedic use.

The most significant method of decreasing modulus without the loss of strength and biocompatibility is the creation of β -type titanium alloys with β -stabilizing elements, including Nb, Ta, Mo or Zr. These alloys may be of less than 80 GPa moduli at high strength and corrosive resistance such that they are useful in load-bearing implants. Articles that were reviewed in Soro et al. (2019) and Cheong et al. (2018) point out the promise of β -stabilised compositions to limit stress shielding in favour of $\alpha + \beta$ running alloys such as Ti-6Al-4V.

Tantalum is especially interesting as a β -stabiliser since it is extremely biocompatible, is chemically inert in the body, and is a strong stabiliser of the β -phase in titanium. According to Sing et al. (2018), tantalum reduces elastic modulus (as well as enhancing corrosion resistance) when it is present in titanium, since it stabilises the β phase, and prevents martensitic α' formation. On a larger computational and experimental scale, Cheong et al. (2018) demonstrated a direct biomechanical benefit of Ti-Ta systems with replacement of Ti-6Al-4V (110 GPa) with a low-modulus alloy Ti-70%Ta (67 GPa) in porous implants both in predicted bone ingrowth and in reduced stress shielding in finite element simulations.

Our bias toward tantalum has been established in several in vitro and in vivo studies of tantalum itself. Histological and cell-culture studies have demonstrated human osteoblasts to also attach, proliferate and retain osteoblastic phenotypes on tantalum surfaces at least as well as on Ti-6Al-4V with similar expression of osteogenic markers (Findlay et al., 2004; Welldon et al., 2008; cited in Wang et al., 2019). Moreover, the Tantalum and Ti-Ta alloys also have good prospects of becoming long-term orthopedic implants since the surface oxide layer is passive Ta₂O₅, which provides high resistance to corrosion.

The impact of tantalum level on the phase stability and mechanical properties of Ti-Ta alloys have been examined by conventional processing routes, including: casting, thermomechanical working and powder metallurgy. Ti-Ta phase formation is sensitive to tantalum content: below around 20 wt.% Ta quantities form α' or α'' martensite on cooling, but higher tantalum contents are better stabilisers of the β phase (Polozov et al., 2025). These changes at the microstructure have a direct effect on strength, ductile and elastic modulus.

The research on powder-metallurgy shows that Ti-Ta alloys are capable of producing high strength and relatively low modulus and also points out the processing challenges encountered at high Ta contents such as incomplete diffusion and Ta-rich knots. Incomplete diffusion between elemental powders has also been reported to form dual structure Ti- Ta alloys with unreacted Ta particles embedded in a Ti-Ta matrix, leading to complex local mechanical and corrosion responses. These findings underscore that tantalum content and homogenisation route strongly affect both microstructure and performance, and also that intermediate Ta contents (around 10–20 wt.%) may offer a more favourable balance between phase stability and processability.

Due to the swift development of additive manufacturing (AM), SLM or laser powder bed fusion (LPBF) has become a significant method to create patient-specific implants with predictable porosity and intricate inner structures. According to Sing et al. (2018), SLM is a laser-based powder bed manufacturing technique, in which the laser power, scan rate, hatch distance and layer thickness are some of the parameters that dictate densification, microstructure, and formation of defects.

Initial applications of TiTa through SLM utilized mainly powder alloys of commercially pure titanium and tantalum in order to have in situ alloying. Sing et al. (2016, 2018) synthesized dense Ti-50Ta parts using a 50 wt.% Ta mixture and experimentally investigated the effect of process parameters on relative density, microhardness and surface roughness. They achieved almost complete density at high laser power and medium scan rate but did see continued unmelted Ta particles because of the large melting-point gap between Ti (1668°C) and Ta (3017°C). This heterogeneity at a microscale is due to the wide two-phase (liquid + solid) field in the TiTa phase diagram and the challenge of thoroughly melting Ta among common SLM process windows. Ti -Ta parts processed by SLM usually have an equiaxed metastable β -Ti(Ta) matrix, with the β phase being retained

by the β -stabilizing action of tantalum and accelerated solidification during SLM. According to Polozov et al. (2025), Zhao et al. (2021) demonstrated that process parameters were able to be carefully controlled to enhance tensile strength and corrosion resistance, although at higher Ta content, complete homogenization was difficult. All these works point to a fundamental processing issue to TiTa alloys in SLM: enough energy needs to be provided to melt Ta completely without excessively melting titanium and resulting in keyholing, balling and residual stress conditions. The compromise is particularly vital in the case of intermediate compositions like Ti-15Ta, where β -stabilization and manufacturability have to be considered at the same time.

Besides bulk properties, TiTa alloys are available in porous and lattice structures which are more representative of cancellous bone mechanical behaviour. Periodic lattice architecture Triply periodic minimal surface (TPMS) and other periodic lattice structures enables porosity, stiffness and strength to be tuned. Soro et al. (2019) fabricated Ti25Ta scaffolds of varying porosity and demonstrated that as porosity rose (between 25% and 64%), Young's modulus decreased (36.1 ± 2.2 GPa to 14.3 ± 0.7 GPa) and yield strength, and retained large elastic admissible strains useful in implant stability.

Later research by Polozov et al. (2025) was able to apply this concept to Ti-10Ta, Ti-15Ta and Ti-20Ta lattices made in situ via SLM. They used TPMS structures (gyroid, split, Schwarz) to show elastic moduli of between approximately 10 and 19 GPa (= cancellous bone) and demonstrated that reasonably uniform mechanical properties in both lattice types in optimised SLM parameters could be achieved. Very low theoretical corrosion rates (0.001–0.007 mm/year) were also found in simulated physiological environments. Polozov et al. (2025) was based on in situ alloying of elemental powder mixtures, but compositional inhomogeneity and partially melted Ta soft spots were still a problem in high Ta loads. Their experiments and studies were more about lattice-scale mechanical and

electrochemical performance than making any fine-scale correlations between the process parameters, microstructure and local properties of a given composition like Ti-15Ta.

The biocompatibility and corrosion resistance of metallic implants are of key importance to their long-term success. Wang et al. (2019) compared porous tantalum and porous Ti-6Al-4V scaffolds that were 3D-printed and had identical pore architecture in a rabbit femoral condyle defect model. Both were comparable to cortical bone in compressive strength and cancellous bone in elastic modulus which facilitates good load transfer and reduces stress shielding. Push-out tests and histological analyses showed that both porous Ta and porous Ti achieved strong bone implant bonding within 2 weeks, and that interfacial strength and bone ingrowth were the same in the later time points.

In TiTa alloys formed by SLM, Zhao et al. (2021) and others (as summarised in Polozov et al., 2025) have demonstrated the increasing resistance in corrosion level as Ta content increases, which is manifested as low corrosion current densities, and wide passive regimes in simulated body fluids. According to Polozov et al. (2025), Ti-10Ta and Ti-15Ta were significantly influenced by low theoretical corrosion rates (~ 0.001 mm/year) and favourable pitting resistance, and Ti-20Ta demonstrated relatively high corrosion rate but was still rather resistant. The authors ascribed the differences in the subtle changes of electrochemical behaviour to the increasing fraction of β -phase and the variation of surface electrochemical activity with tantalum content.

Studies on Ti-Ta alloys made by SLM/LPBF show that a good volumetric energy density window is very important for getting high density and a good microstructure. Polozov et al. (2025) worked with in-situ Ti-10Ta, Ti-15Ta, and Ti-20Ta and found that energy densities below about 50 J/mm^3 caused high porosity ($>5\%$), while values above about 300 J/mm^3 caused too much evaporation and keyhole porosity. They found that the best combination of

relative density ($\geq 99.4\%$) and microstructural homogeneity was in the range of 56–95 J/mm³. In this range, Ti–15Ta had the best balance of density, Ta distribution, and mechanical properties compared to Ti–10Ta and Ti–20Ta (Polozov et al., 2025).

For particular Ti–Ta systems, more comprehensive parameter sets have been reported. For dense Ti–Ta produced by in-situ alloying of a 50 wt.% Ta mixture, Sing et al. (2018) showed that a combination of 360 W laser power, 400 mm/s scan speed, 0.05 mm layer thickness and 0.125 mm hatch spacing yielded “fully dense” parts (relative density $\approx 100\%$) and a melted tantalum content close to the nominal composition, corresponding to energy densities in the range 48–144 J/mm³. At lower energy densities, lack-of-fusion holes and considerable portions of unmelted Ta were found, but very high energy densities, although helpful for melting Ta, risked increased titanium vaporisation and surface instabilities (Sing et al., 2018). Similar tendencies exist in Ti–25Ta lattice work by Soro et al. (2019), who used 95 W laser power, 300 mm/s scan speed, 0.025 mm layer thickness and 0.035 mm hatch distance (energy density = 360 J/mm³) to generate dense matrix material (relative density $99.8 \pm 0.07\%$) and structurally strong scaffolds.

Across these studies, a consistent picture emerges: high-quality Ti–Ta (including Ti–15Ta) is typically obtained with relatively fine layers (0.025–0.05 mm), small hatch spacing (≈ 0.035 –0.125 mm) and energy densities on the order of 50–150 J/mm³ for in-situ alloying, or somewhat higher (≈ 300 –360 J/mm³) when using specific lattice geometries or different machines (Sing et al., 2018; Soro et al., 2019; Polozov et al., 2025). Therefore, the LPBF parameter space for Ti–15Ta in the current project was selected to fall within these literature-based energy-density ranges and layer/hatch scales. It was then experimentally refined so that the resulting samples could be meaningfully compared with previously reported Ti–Ta and Ti–15Ta alloys in terms of density and microstructure.

3. Methodology

3.1 Powder Preparation and Characterization

As a feedstock material 85 wt% commercially pure titanium (Ti) and 15 wt% tantalum (Ta) were used to obtain the samples. Before fabricating the specimens, the powders underwent a systematic preparation and characterization procedure to achieve consistent flowability, melting behavior and suitability for LPBF (Laser Powder Bed Fusion).

3.1.1 Morphology analysis using SEM

Powder morphology was examined by scanning electron microscope and particle size distribution was obtained by laser diffraction (Fig. 1-4). The parameters of D10, D50, and D90 were extracted to quantify the spread and polydispersity of powders (Fig. 3-4). D10 represents the diameter of particles below which 10% of the powder volume lies. D50 is a median particle size and represents the midpoint of the distribution. D90 indicates the particle diameter below which 90% of the powder volume lies. These values were used to calculate the span, which reflects the PSD width and defined by formula:

$$Span = \frac{D_{90} - D_{10}}{D_{50}}$$

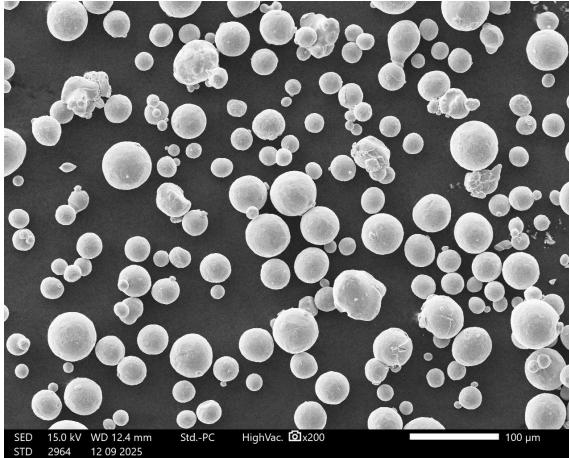


Fig. 1 x200 SEM image of Ti powder

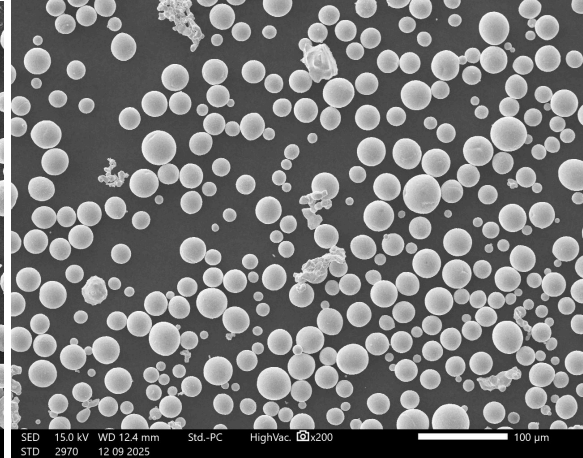


Fig. 2 x200 SEM image of Ta powder

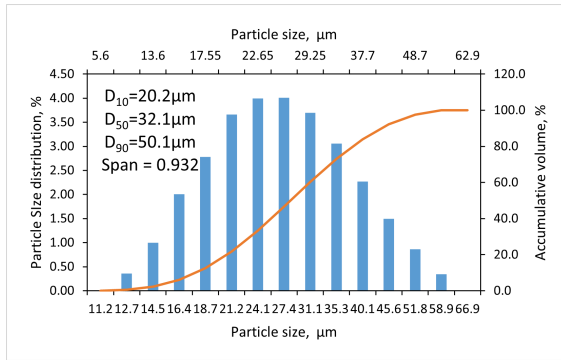


Fig. 3 PSD graph of Ta powder

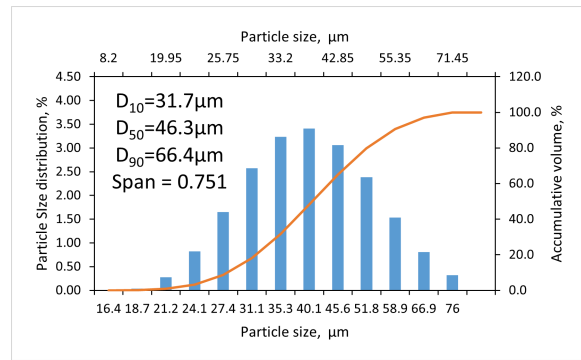


Fig. 4 PSD graph of Ti powder

3.1.2 Powder Blending

Ti and Ta powders were weighed inside a glovebox to avoid oxidation (Fig. 5). The masses of Ti and Ta were 1122g and 198g respectively. The powders were blended in an Inversina 2l tumbler mixer (Fig. 6) for 6 hours at 30 rpm, ensuring the mixer was not overfilled for effective 3D inversion mixing.



Fig. 5 Glovebox



Fig. 6 Inversina 2l tumbler mixer

After blending, each batch was stored back in the glovebox to maintain powder quality before printing. This blending procedure was performed twice, producing enough Ti-15Ta powder to prepare LPBF build platforms.

3.1.3 Microstructure testing

Microstructure studies were performed to Ti-15Ta alloy, also using scanning electron microscopy (Fig 7). SEM analysis was used to compare microstructural uniformity between powder types and between parts produced on LPBF platforms. The objectives included detection of defects, assessment of elemental distribution and evaluation of melt pool morphology.

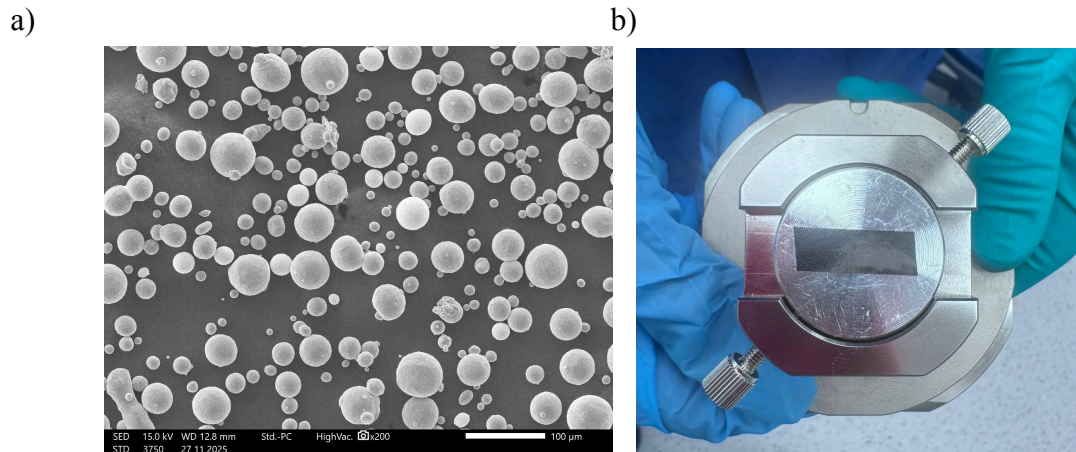


Fig. 7 a) x200 SEM image of Ti-15Ta powder, **b)** Sample for the SEM

3.2 Selective Laser Melting (SLM) Process

The fabrication of Ti-15Ta samples by selective laser melting was carried out on a Renishaw AM 400 machine (Fig. 8). Printing was performed under an argon atmosphere to prevent oxidation. The level of oxygen in the chamber maintained below 0.1% ensuring a consistent melt pool.



Fig 8. Renishaw AM 400

The key process parameters included Laser Power (P), Scanning Speed (v), Hatch Spacing (h), and Layer Thickness (t). Selection of parameters was based on the literature data and preliminary trials.

3.2.1 1st Platform Fabrication

The first platform contained CAD models of 28 cubes and 14 different process parameters were used to print them (Table 1).

Table 1 . Process Parameters

Number	Laser Power (W)	Hatch Spacing (mm)	Layer Thickness (mm)	Point Jump Speed (mm/s)
1	200	0.065	0.03	5000
2	200	0.0634	0.03	5000
3	200	0.0618	0.03	5000
4	200	0.0676	0.03	5000
5	200	0.0702	0.03	5000
6	200	0.0728	0.03	5000
7	200	0.0748	0.03	5000
8	200	0.0767	0.03	5000
9	200	0.078	0.03	5000
10	200	0.0793	0.03	5000
11	200	0.0813	0.03	5000
12	200	0.0826	0.03	5000
13	200	0.0845	0.03	5000
14	200	0.105	0.03	5000

After, among these 14 parameters the optimized one was selected based on the best combination of density and microstructural homogeneity. All samples showed ~100% relative density, therefore, the amount of unmelted Ta particles was compared, since Ta melts last.

Parameter 7 had the lowest unmelted Ta content, so it was selected as the optimal setting. The results are shown in Table 2.

Table 2. Process Parameters

Sample №	Unmelted Ta particles, %	Relative Density, %
1	3.8155	99.981

2	3.6355	99.991
3	3.864	99.991
4	3.8375	99.994
5	4.0805	99.994
6	3.602	99.968
7	3.2695	99.989
8	3.71	99.997
9	3.324	99.992
10	3.628	99.9967
11	4.18	99.9965
12	3.968	99.99658
13	3.8715	99.99885
14	3.956	99.95925

3.2.2 2nd and 3rd Platform Fabrication

During second printing cylindrical or dog-bone shaped samples were fabricated for tensile testing (Fig. 9). The diameter of the cylinder was 6mm and the gauge length was 36mm. For compression mechanical testing, 2 different cylindrical samples with diameter of 6mm and length of 4.8mm and 12 mm were printed (Fig 10). Cubic samples with 10mm sides were fabricated for microstructural investigation and hardness tests. Also cylindrical samples were printed for corrosion testing.



Fig. 9 2nd Platform Samples

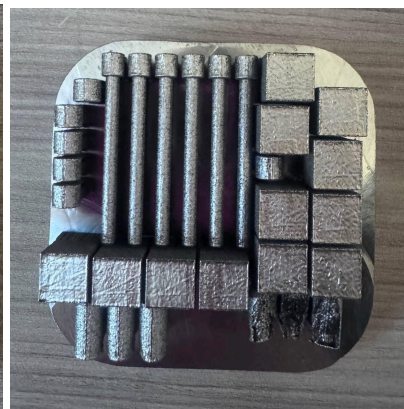


Fig. 10 3rd Platform Samples

3.3 Mechanical Testings

Tensile Test

Tensile test was carried out according to ISO 6892-1:2019 standard. The gauge length of the specimen was 36mm with a diameter of 6mm in the working section.

Testing was performed on the WDW-5T universal tensile testing machine. The extensometer was used to achieve accurate results of strain measurements and had a gauge length of 25mm. The test was performed at room temperature ($23\pm 2^{\circ}\text{C}$) and under controlled relative humidity ($50\pm 5\%$). The loading direction was perpendicular to the build direction. To ensure statistical reliability 3 specimens were tested.



Fig. 11 Sample on WDW-5T machine



Fig. 12 Sample after fracture

Compression test

Compression testing was performed for 4 cylindrical samples to evaluate bulk and structural mechanical behavior. The procedure followed ISO 7500–1:2018 requirements, similar to the methodology used in reference literatures. All cylindrical samples have a

diameter of 6mm and approximate L/D ratio of 0.83. The compression load was applied parallel to the build direction and the same equipment WDW-5T at a strain rate of 10^{-3} s^{-1} was used.

Hardness Test

Vickers microhardness test was carried out on a cross section of cubic samples with sides of 10mm using a Innova Test Falcon 500 hardness tester machine with a load of 0.5kg and a dwell time of 15 seconds. The test followed ASTM E384-17 standard. Before starting the hardness test, the top and bottom surfaces (0°) and side surfaces (90°) were polished. 9 measurements were taken on each side to ensure statistical accuracy. And to avoid interaction effects sufficient spacing was between indentations.

3.4 Electrochemical Test

To examine the electrochemical properties of the fabricated Ti-15Ta alloy specimens, electrochemical impedance spectroscopy (EIS) and potentiodynamic polarization (Tafel) testing were applied using a three-electrode electrochemical system. As a reference electrode an Ag/AgCl electrode was used and it was saturated with KCl. The counter electrode was a platinum electrode. Cylindrical Ti-15Ta specimens with the dimensions of 3mm in diameter and 37mm in lengths were used as the working electrode. Before the test, the samples were polished till a mirror finish. First, the open circle potential (OCP) was held for 2 hours to ensure stabilization. Electrochemical impedance spectroscopy was performed with the frequency range of 0.00001 Hz to 0.01 Hz and AC amplitude of 10 mV. The obtained data was fitted using a Randles circuit. Potentiodynamic polarization testing was then carried out at a scan rate of 1 mV/s with the potential range from -0.5 V to $+2.5 \text{ V}$ relative to the open circuit potential. The equivalent weight of 13.3 g/eq, Stern-Geary constant of 26 mV, and alloy density of 6.0 g/cm^3 were set for the tests.

4. Results and Discussions

4.1 *Relative Density*

The first stage of selective laser melting (SLM) was aimed at the design of a strong process window in the formation of Ti-15Ta alloy. This is complicated by the fact that the melting point of titanium (Ti, 1668°C) and tantalum (Ta, 3017°C) differ radically (Brodie et al., 2020). The efficient optimization requires adequate supply of energy to dissolve refractory Ta particles and at the same time avoid unnecessary energy supply which can lead to the formation of defects in the main Ti matrix (Brodie et al., 2020). The main success of the parameter screening was the achievement of universal success in the bulk consolidation. The set of all the tested parameter sets gave specimens with the relative density close to 100 % as the measured values and data suggest that the relative density is between 99.95925 % and 99.99885 %. This means, therefore, that the nature of the processing of the Ti-15Ta composition suggests an unexpectedly broad operating range under which high bulk density can be obtained.

4.2 *Unmelted Ta particles*

The structural integrity (high density) criteria had now been met in all sets of parameters, so the optimisation effort was switched into the problem of compositional homogeneity - an issue due to the differing melting points and densities of Ti and Ta. The optimal regime was found to be parameter Set 7 using a hatch spacing of 0.0748 mm, making the lowest fraction of unmelted Ta (3.2695%). The amount of unmelted Ta particles was determined using scanning electron microscopy (SEM). Areas with Ta content exhibit high brightness since Ta has a higher atomic number than Ti. Unmelted Ta particles were observed as isolated high-contrast structures with distinct boundaries against the matrix background. Image analysis followed to quantify the area percentage of the high-brightness regions by applying threshold techniques. This area percentage was then considered equivalent to the volume percentage of the unmelted Ta particles. The distribution of unmelted Ta throughout

the test matrix had a non-linear relationship with the energy input; thus, higher energy input (decreased hatch spacing) did not always reduce the fraction of unmelted particles further.

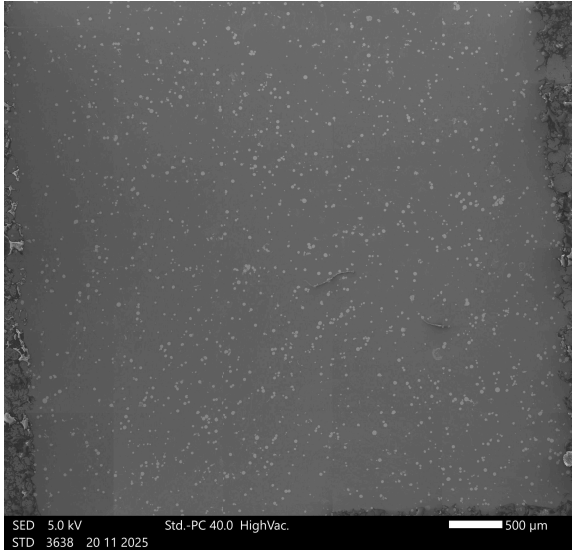


Fig. 13 SEM image of the parameter set 11

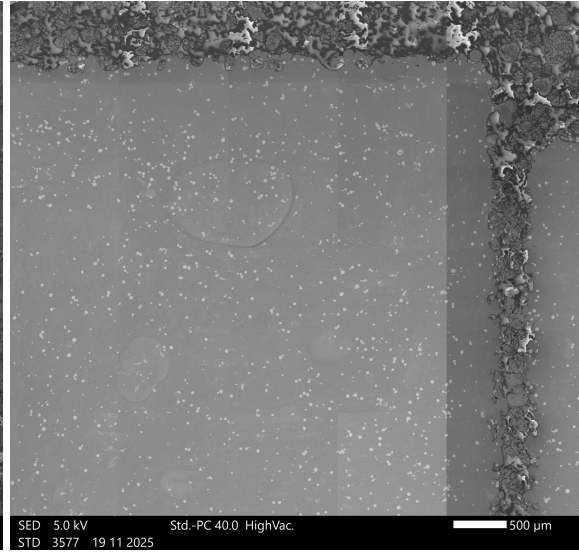


Fig. 14 SEM image of the parameter set 7

The low melting tantalum content, even at the ideal percentage of 3.27, has a strong impact on the final alloy microstructure and mechanical strength. Tantalum has two key purposes: to strengthen the solid solution and, more importantly in terms of biomedical applications, to stabilise the β -phase (Zhao et al., 2019; Polozov et al., 2025). The tantalum that remains in unmelted particles is no longer available to the rest of the Ti matrix, and therefore the active 15 wt% β -stabilizer content is actually less than that. Combined with the high cooling rates inherent to LPBF, this compositional shortcoming causes the alloy composition to be pushed out of the targeted low-modulus beta phase (Brodie et al., 2020). The material is instead highly vulnerable to developing non-equilibrium, high-stiffness martensitic phases like α as α' that solidifies quickly (Zhao et al., 2019).

4.3 Tensile Mechanical Properties

The mechanical characterisation showed a material with a significant degree of ductility, but with an Young's modulus which is much higher than the biomedical design target. The mean values suggest an average yield stress of 480.6 MPa, ultimate tensile strength of 598.2 MPa and ductility with a mean elongation of 12.7 %.

Stress Strain Curve

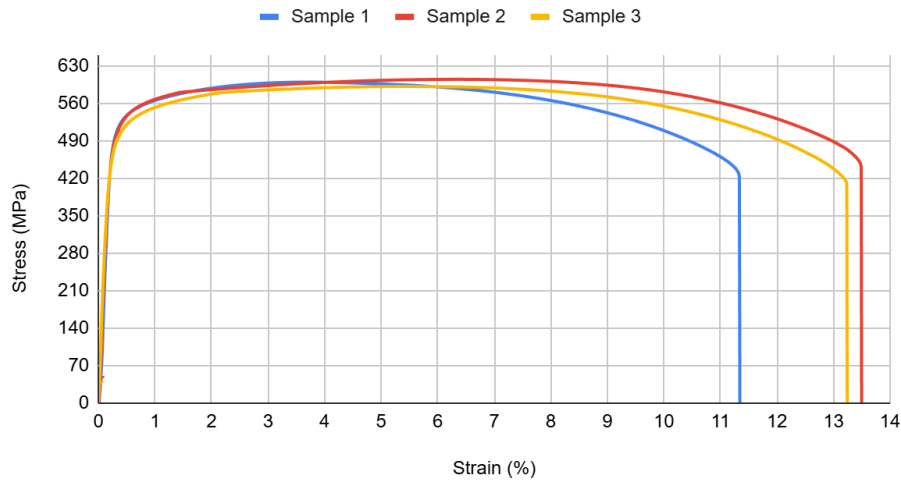


Fig. 15 Stress-Strain curve of the Ti-15Ta alloy

Table 3 . Tensile Mechanical properties of the Ti-15Ta alloy

Sample #	Young's modulus, GPa	Yield Stress, MPa	UTS, MPa	Strain to fracture, %
1	123.6	506.6	599.1	11.33
2	149.8	482.6	604.4	13.5
3	149.5	452.5	591.1	13.4
Avg.	140.3	480.6	598.2	12.7

The most noticeable observation on the as-built alloy is that it has a high average of Young's modulus of 140.3 GPa. More importantly, this stiffness is comparable to or higher than the elastic modulus of the traditional implant alloy Ti6Al4V which has a range of 110 to 148 GPa (Brodie et al., 2020; Zhao et al., 2019) . Ti -Ta alloys in biomedical implants

have been selected based on the need to make the elastic stiffness of alloying materials lower than that of conventional ones, to alleviate the impact of stress shielding leading to bone resorption, which occurs when the material implant modulus is excessively high compared to cortical bone (10 -40 Gpa)(Zhao et al., 2019). The alloy, though made of non-toxic components, does not provide the necessary mechanical compatibility to support next generation, low-modulus orthopedic implants because it attains a comparable modulus to Ti-6Al-4V.

Table 4 . Comparison table of mechanical properties of the Ti-Ta alloys in literature

Material/Condition	E (GPa)	YS (MPa)	UTS (MPa)	Elongation (%)	Reference
Ti-15Ta (As-Built)	140.3	480.6	598.2	12.7	This Work
Ti-15Ta (Pre-Alloyed SLM, Optimal)	90	460–475	540	>20	Polozov et al. (2025)
Ti-15Ta (SLM, Regime 2)	110	902	1102	5.1	Polozov et al. (2025)
Ti-25Ta (SLM, Remelt)	65	545	566	11	Brodie et al. (2020)
Ti-6Al-4V (SLM, As-Built)	110–148	1100	1200	9–12	Brodie et al., 2020; Zhao et al., 2019)

The average elongation of 12.7% is compatible with Ti-25Ta (11 -18 %) and Ti 6Al 4V (9-12 %) alloys printed with SLM (Brodie et al., 2020; Zhao et al., 2019).

The tensile strength of 598.2 MPa is up to the recommended optimum strength of 540 MPa that is reported in the papers on Ti- 15Ta. However, it is substantially less compared with the high-strength SLM Ti -Ta systems (e.g. Ti -25Ta with values above 1000 -1 MPa)(Polozov et al., 2025) . The given mechanical profile, which combines high elastic modulus, low ultimate tensile strength, and high ductility, shows that the laser-based powder-bedding (LPBF) parameters were successful to yield a dense and defect-free material, but failed to enter the microstructural phase that can achieve the combined goals of low elastic modulus and high strength.

4.4 Microstructural Characterization

Two cubical samples of Ti-15Ta were cut parallel (0°) and transversally (90°) to the build direction for microstructural analysis. These two samples first, were embedded in EPO resin using ATM OPAL, after that, ground and polished in four steps on a SAPHIR 520 to a mirror finish, and finally, etched for 30 seconds with Kroll's reagent prior to SEM analysis. The area was confirmed by EDS analysis.

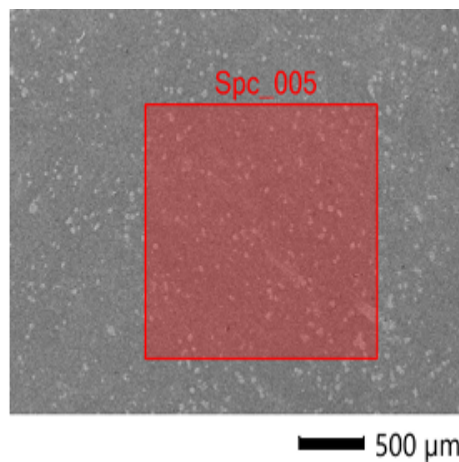


Fig. 16. SEM image (500 μm scale) of the etched Ti-15Ta sample in the 0° orientation

The low-magnification image shows the overall homogeneity of the consolidated powder and the absence of large pores or cracks. Many small bright areas are spread through the matrix as white dots. These are thought to be residual, unmelted Ta particles with a fraction of 3.27 % as measured for the optimized Parameter Set 7 in Section 4.2. These have a higher atomic number ($Z_{\text{Ta}} = 73$) than Ti ($Z_{\text{Ti}} = 22$) and therefore appear brighter due to greater backscattering contrast.

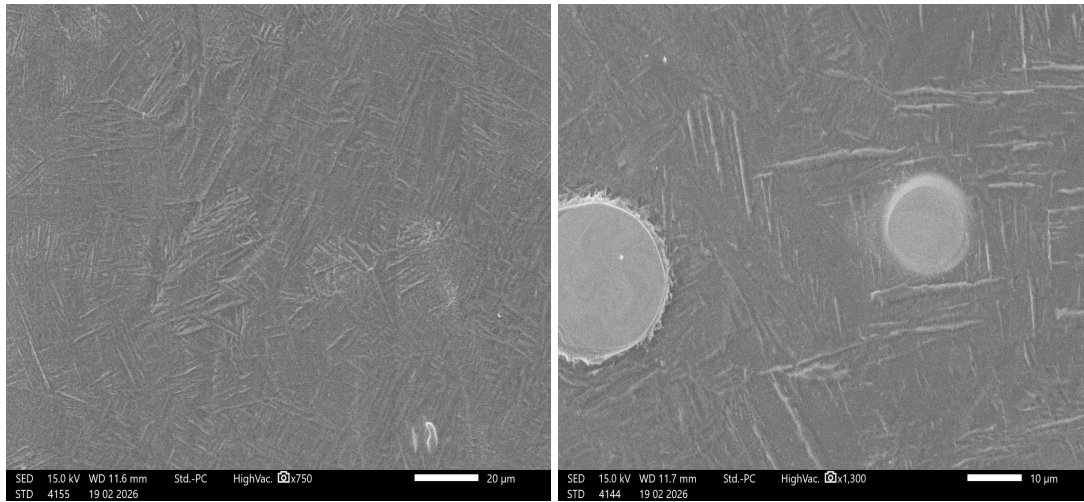


Fig. 17-18. SEM images at 10 and 20 µm scale of the etched Ti-15Ta sample in the 0° orientation

At higher magnification, the etched microstructure consists of interpenetrating acicular laths of various orientations within the grains. The lath size is in the range of hundreds of nanometres to a few micrometers, and there is no evidence of equiaxed β -grains. In Figure 18, two bright circular spots can be seen in the matrix; these are unmelted Ta particles that have been partially incorporated into the solidified Ti-rich matrix, showing a diffusion front with the surrounding microstructure.

Element	Line	Mass%	Atom%
Ti	K	87.11±0.20	96.23±0.22
Ta	M	12.89±0.13	3.77±0.04
Total		100.00	100.00
Spc_005		Fitting ratio 0.0120	

Fig. 19. EDS quantitative composition results for the area indicated in Figure 16

The EDS composition of the area marked in Figure 16 is 87.11 ± 0.20 wt % Ti and 12.89 ± 0.13 wt % Ta, or 96.23 at % Ti and 3.77 at % Ta, respectively. This is about 2 percentage points off the desired target composition of 85/15 wt %.

The lath microstructure observed in the SEM images is typical of martensite. LPBF processing involves rapid cooling (on the order of 10^5 - 10^7 K/s) which prevents the alloy from developing its equilibrium $\alpha + \beta$ microstructure, and instead "freezes" a non-equilibrium martensitic structure. At a composition of about 15 wt % Ta, this is likely to be α'' orthorhombic martensite, but could also be α' hexagonal martensite, which cannot be determined from SEM.

EDS also shows the Ta content is 12.89 wt % (not 15 wt % as expected). The melting points of titanium and tantalum are 1668 °C and 3017 °C, respectively, which means that not all the tantalum melts into the melt pool at our laser energy. The residual Ta particles remain solid, hence we found 3.27 % unmelted Ta previously (Section 4.2). Because EDS measurements probe mainly the matrix surrounding the particles, the results are lower than the bulk composition. ICP or XRF would provide a better check of the composition.

This is also why we didn't get the low modulus we wanted. To stabilise β , Ta must be dissolved into the Ti matrix - Ta trapped in unmelted particles will not help. Since the effective Ta concentration in the matrix is less than 15 wt %, it will transform into martensite, rather than the low-modulus β phase. Polozov et al. (2025) reported the same problem and demonstrated that both sputtering process optimization and subsequent homogenisation heat treatment was required to achieve a β -dominated structure.

4.5 Compression Test

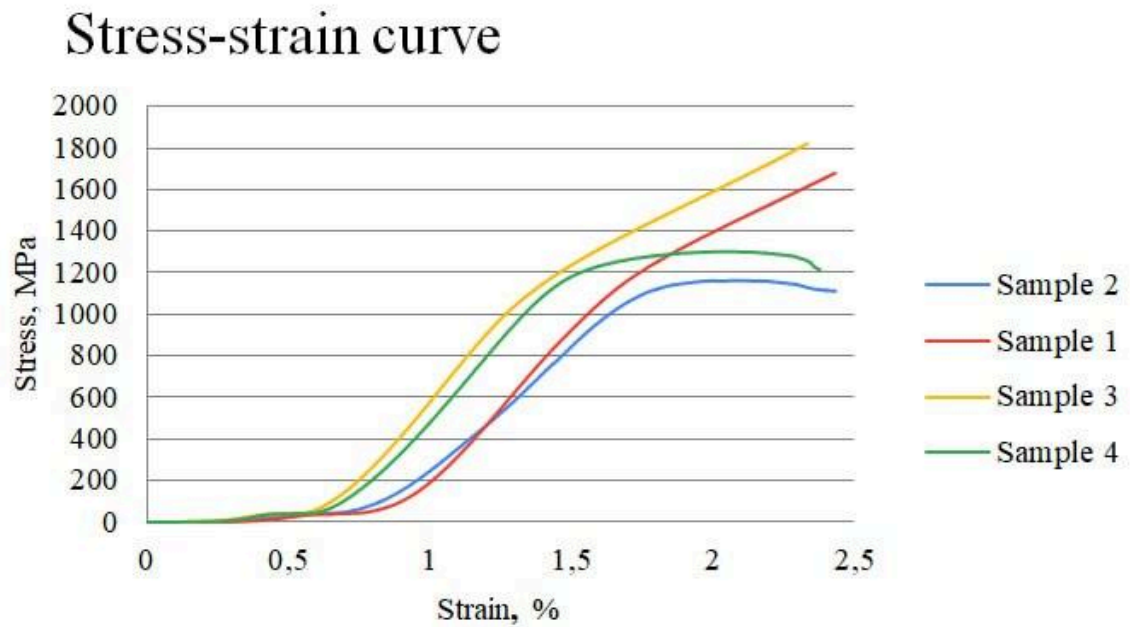


Fig 20. Compression stress–strain curves of Ti-15Ta samples with $L/D = 2$ geometry

Figure 20 shows the compression stress-strain curves, and Table 5 presents the mechanical properties of four LPBF-built Ti-15Ta samples with $L/D = 2$. Each sample displayed a distinct linear elastic region, yield and ongoing load-bearing capacity until fracture. The average compressive elastic modulus of 143.58 GPa is in agreement with the tensile modulus of 140.3 GPa reported in Section 4.3, indicating that the as-built material is elastically isotropic. These values are still well above the modulus of cortical bone (10-30 GPa), confirming the conclusion from Section 4.3 that the martensitic microstructure impairs the alloy from meeting the recommended lower modulus to reduce stress shielding (Niinomi, 2016).

Table 5. Compression test results of Ti-15Ta samples with L/D = 2 geometry

Sample #	E, GPa	Yield strength,MPa	UCS, MPa	Compression, %
1	157.01	815.86	1821.4	2.6354
2	116.42	1147.6	1163.6	2.4372
3	142.56	1640.7	1822.1	2.334
4	158.34	1019.8	1302.4	2.3791
Avg.	143.58	1155.99	1527.4	2.4464

The average compressive yield strength (1155.99 MPa) is roughly 2.4 times greater than the tensile yield strength (480.6 MPa) and the average UCS (1527.4 MPa) is around 2.6 times greater than the UTS (598.2 MPa). This difference between tensile and compressive strengths is in agreement with other reports in the literature on Ti-Ta alloys: Brodie et al. (2020) reported the tensile properties of SLM-fabricated Ti-25Ta as UTS = 566 MPa and elongation = 11%, which are of comparable magnitude to our results, but deformation mechanisms under compressive loading are different from those under tensile loading in titanium alloys. The high compressive strength reported here confirms the potential of the alloy for use in load-bearing biomedical applications, with compressive loads being more prevalent in orthopedic applications.

Of particular interest in Table 5 is the large variability in compressive yield strength, which ranges from 815.86 MPa (Sample 1) to 1640.7 MPa (Sample 3). This could be explained by

microstructural variations identified in Section 4.2, where unmelted Ta particles (3.27% in the case of the optimized process parameters) result in regions of the matrix that have lower effective Ta concentrations. Given that tantalum is a β -stabilizer only in solid solution in the Ti matrix, the regions around unmelted Ta particles will contain less effective Ta, and are therefore more likely to yield at higher stress values, forming high-stiffness, high-strength laths of martensitic phases, while more uniform regions in the matrix may have slightly more β -phase retained and yield at lower stress (Polozov et al., 2025; Zhao et al., 2021). This is evident in the stress-strain curves in Figure 20: the stress-strain curves for samples with lower yield points exhibit a smoother transition, whereas Sample 3 exhibits a single yield point with a much higher stress.

The mean compressive failure strain of 2.4464% is significantly less than the tensile elongation of 12.7% in Section 4.3. This is to be expected as fracture in dense titanium alloys under compression is generally governed by local shear, rather than diffuse plastic strain accommodating ductility in tension. The compressive behavior, with high strength and low ductility but moderately high modulus, is signatures of an as-built martensitic structure, as observed in SLM-built Ti-15Ta by Polozov et al. (2025) due to incomplete Ta dissolution and fast cooling rates that led to the formation of martensite and high modulus. As Polozov et al. (2025) also showed, process parameters and post-processing homogenization heat treatment are required to encourage β -phase retention and drive the modulus down to the 90 GPa range. The findings here are in agreement: in the as-built condition, the Ti-15Ta alloy offers high compressive strength, but lacks low modulus for bone-mimicking biomechanical performance.

4.6 Microhardness test

The microhardness test was performed to capture the microstructural resistance to plastic deformation while reducing the influence of surface roughness. The test was conducted on

both top (0°) and side (90°) surfaces with nine indentations for each. The results are shown in Table 5 and Table 6.

Table 6 . Hardness Results for 90° (side surface)

	1	2	3	4	5	6	7	8	9	<i>Average</i>	<i>Standard Deviation</i>
HV0											
.5	267.9	288.7	317.1	295.8	287.1	285.4	311.3	289.1	304.4	294.1	14.9

Table 7. Hardness Results for 0° (top surface)

	1	2	3	4	5	6	7	8	9	<i>Average</i>	<i>Standard Deviation</i>
HV0.5	344.9	355.5	323.9	270.6	313.9	296.6	298.5	344.2	368.1	324.2	31.9

The Vickers microhardness was between 294 and 324 HV. This level of hardness is significant for biomedical applications in terms of load bearing, resistance to surface deformation and maintaining structural integrity. Sing et al.(2016) reported microhardness of 242 HV for in-situ fabricated Ti-5-Ta alloy. The higher results in our study might be due to the lower concentration of tantalum, since it is a β -stabilizer, which in higher concentration provides a ductile and softer phase. In contrast, the 15wt% concentration promotes the formation of martensitic α' phase under the rapid cooling conditions of LPBF.

Our microhardness results also align with the values observed by Polozov et al. (2025), even though there is a significant difference between their UTS (1102 MPa) and our UTS (598.2 MPa). This might be due to porosity or post processing effects, but the hardness still reflects a significantly strengthened matrix.

Also, the high microhardness we observed might be due to a few factors related to the LPBF process. The rapid cooling which is typical for 3D printing, creates a fine, needle like martensitic α' structure and martensite is a metastable, supersaturated solid solution. The quick temperature drop locks the atoms in a way that prevents them from moving easily under pressure. There is also the contribution of 3.27% unmelted tantalum particles through dispersion strengthening. These unmelted hard particles tend to act as a reinforcement in the titanium matrix, pinning grain boundaries and impeding dislocation glide. In addition, this incomplete dissolution of Ta results in regions with lower effective Ta concentrations in the surrounding matrix, which makes the formation of hard martensitic phase even more susceptible. This elevates the local hardness.

4.7 Electrochemical Properties.

The corrosion behavior of the Ti-15Ta alloy was evaluated through EIS and Tafel tests in a 3.5 wt% NaCl solution. During the initial stages of the testing, the first sample was held at the open circuit potential for only one hour, however it was noticeable that during that time OCP had not fully stabilized. As a result, the next three samples were held at the open circuit potential for two hours to ensure full stabilization, where the average value was -0128 V.

Table 8. Corrosion test results

<i>Sample #</i>	<i>OCP, V</i>	<i>Ecorr, V</i>	<i>Icorr ($\mu A/cm^2$)</i>	<i>Corrosion rate, mm/year</i>
2	-0.13274	-0.25486	1.6592	0.01202
3	-0.13167	-0.19548	0.6692	0.00485
4	-0.12065	-0.18564	0.9974	0.00723

Avg.	-0.12835	-0.21199	1.1086	0.00743
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From the EIS test, the calculated solution resistance (R_s) values ranged from approximately 4.96 to 19.59 $\Omega \cdot \text{cm}^2$, while the polarization resistance (R_p) values ranged from 3550.6 to 23874 $\Omega \cdot \text{cm}^2$. Higher R_p values indicate higher resistance to charge transfer across the alloy/electrolyte interface and therefore means better corrosion resistance. The higher R_p values for the latter samples may be associated with the longer OCP stabilization period because it allowed the passive oxide layer on the Ti-15Ta surface to develop more completely before testing. It is known that titanium alloys produce a protective film of oxides mostly comprising of TiO_2 , while the addition of tantalum leads to the creation of Ta_2O_5 . The presence of the mentioned oxides reduces the rate of electron and ion transfer between the alloy and surrounding environment..

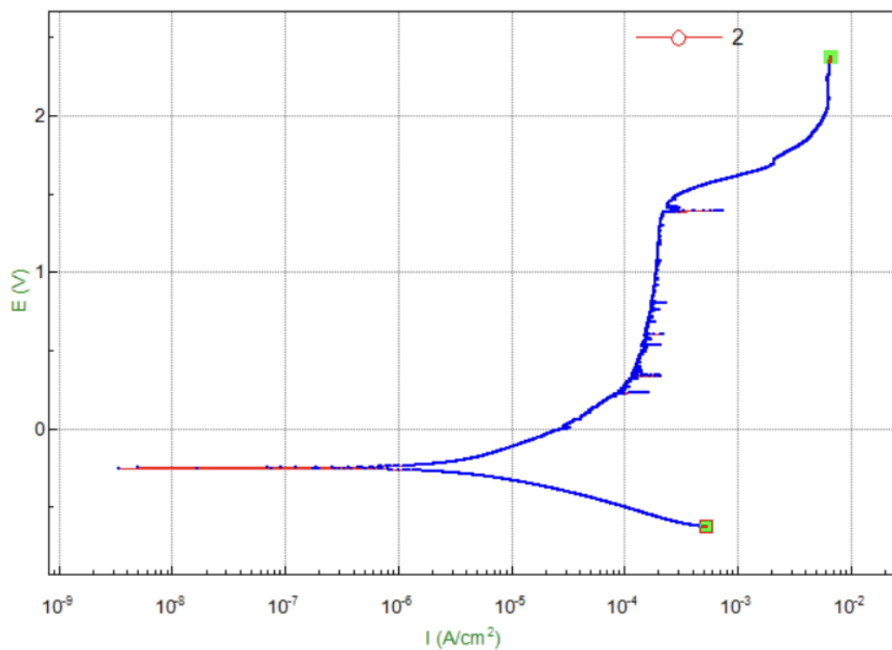


Fig 22. Potentiodynamic polarization curves for the second sample

The potentiodynamic polarization curves were characterized by typical behavior for passive titanium alloys, which is reflected in low corrosion current density and wide passive zone. The range of the corrosion current density values was 0.669-2.285 $\mu\text{A}/\text{cm}^2$. Sample 1 had the maximum value of corrosion current density (2.285 $\mu\text{A}/\text{cm}^2$) and corrosion rate (0.0166 mm/year). This result may be explained by the fact that the duration of OCP stabilization was too short, i.e., only 1 hour, and therefore the passivation layer formed poorly.

For the samples where OCP was stabilized for 2 hours, the corrosion current density was much lower and amounted to 0.669-1.659 $\mu\text{A}/\text{cm}^2$. The average value of corrosion current density for these samples was about 1.11 $\mu\text{A}/\text{cm}^2$, which corresponds to corrosion rate of 0.0074 mm/year. It can be stated that there are many perspectives for application of Ti-15Ta alloy as a material for biomedical implants because of high stability and low corrosion rate.

5. Conclusion

This research demonstrated that the development of a Ti-15Ta alloy at high density is possible through Laser Powder Bed Fusion (LPBF), which can be used as a biocompatible and non-aluminum/vanadium-containing material for orthopedic applications. The optimal processing parameters of LPBF, namely Parameter Set 7, have resulted in nearly 100% relative density with a minimum percentage of un-melted tantalum of only 3.27%.

From a mechanical standpoint, the as-built alloy proved to be a good choice for load-bearing purposes. In particular, the average yield strength of compression was around 2.4 times greater than the tensile yield strength at an average of 1155.99 MPa, which offers the necessary strength for the orthopedic field that primarily relies on compressional forces. Moreover, the alloy demonstrated sufficient ductility, with an average elongation of 12.7%.

Nevertheless, the main objective of developing an alloy with a low modulus to reduce stress shielding was not fully achieved. The calculated value for Young's modulus, 140.3 GPa, still lies within or above the level of conventional Ti-6Al-4V. The reason behind such a result is the presence of an acicular martensite microstructure that resulted from the fast cooling rate in the LPBF technique and the incomplete dissolution of tantalum in the titanium matrix. As only around 12.89 wt% of tantalum could dissolve in the matrix, the necessary stabilization of the β -phase was not attained for a modulus lower than 90 GPa.

In conclusion, although the Ti-15Ta alloy shows great promise due to its density, strength, and corrosion profile, there is a need for more research into post-build homogenization. This is key for achieving the benefits of β -phase stabilization as well as using the material as a low modulus biomaterial.

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