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Characterization of Microstructure and Electrochemical Behaviour of a New Ti-8Mo-4Sn-2Mg alloy for Biomedical Applications

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Abstract
β type Ti-Mo alloys are especially interesting for orthopedic implants due to its lightweight, excellent mechanical behavior, and corrosion resistance. However, high elastic modulus remains a major setback to its biomedical application. For this, a new βeta-Ti-8Mo-4Sn-2Mg alloy for the biomedical applicationwas designed, and its mechanical and corrosion behaviors ascertained using compressive and Potentiodynamic polarization corrosion tests, respectively. The base and new alloys' elastic moduli were determined using both ultrasonic resonance frequency and Nano-indentation techniques. The microstructures of the alloys were characterized by X-ray diffraction and scanning electron microscopy. The result shows that the elastic moduli were reduced to bone matching values without impacting mechanical performance by adding Mg and Sn. The combination of high strength, low Young's modulus, biocompatibility, and excellent corrosion resistance properties of the new βeta-Ti-8Mo-4Sn-2Mg alloy makes it a favorable candidate for orthopedic applications.

Keywords: *titanium alloys; electrochemical properties; low Young's modulus; biomedical applications;magnesium; mechanical properties*

I. INTRODUCTION

According to the landmark WHO, Global status report on road safety 2015[1], every year, the lives of approximately 1.25 million people are cut short as a result of road traffic crashes, and between 20 and 50 million more people suffer non-fatal injuries, with many incurring a disability as a result [1]. Apart from road traffic injuries, many people are prone to accidents during routine and extreme sporting activities. Most of those cases need orthopedic surgeries and implantation to cure the sufferers. The use of metallic biomaterials is popular in these hard tissue replacements [2]. However, the crucial challenge for developing suitable metallic orthopedic implants is to obtain materials that combine good biocompatibility, lightweight, excellent corrosion resistance, and a reasonable balance of high strength and low elastic modulus. This is because, despite the great numbers of metals and alloys best-known to man, remarkably few warrant even preliminary scrutiny for uses as implant materials [3]. For example, corrosion in the body environment poses an extreme concern in the design of most orthopedictools and implants because of the relatively corrosive environment, incompatible mechanical properties, and poor tolerances of the body minute concentrations of most metallic corrosion products [4]. The difficulties stated above suggest the development of low-cost materials with appropriate mechanical and electrochemical compatibility is highly essential.

Ti and some of its alloys come as the first choice because of its lightness, its combination of favorable characteristics and immunity to corrosion [5], excellent biocompatibility [6], and reasonable balance of high specific strength, comparatively low Young's modulus, and the capacity for osseointegration [7,8]. However, the commonly used Ti-alloys, Ti–6Al–4V alloy, presents elastic modulus (~110 GPa), which is much higher than natural bone (~20-40GPa) [9]. Recent studies on Ti–6Al–4V alloy has indicated the release of small amounts of Vanadium (V) and Aluminium (Al) ions in the human body. Vanadium and Aluminium ions are held responsible for long-term health problems, including Alzheimer's disease and neuropathy [9, 10]. For this, researchers on biomedical Ti alloys are focused on the development and control of implant toxicity via alloying additions. Previous studies have shown that the most promising compromises are alloys present Nb, Zr, Mo, Sn, and Ta (β-stabilizing elements) as alloying elements [14, 15, 16].

However, the elastic moduli have not yet matched that of human bone.

Our design strategy aims to reduce the elastic modulus, total bio-compatibility and an improvement in the electrochemical properties by modifying the Ti-based alloy's chemical composition and microstructural characteristics. A systematic evaluation of possible alloying elements to meet the above criterion resulted in the selection and inclusion of magnesium (Mg) and Tin (Sn) in binary Ti-Mo alloy. Even thoughthe Ti–Mo-alloy system is very intriguing, there are only a few or scarce studies dealing with multi-component Ti-Mo alloys and, more importantly, Mg and their potential use hard tissue biomaterials. Although there have been some biomedical Ti alloy formulations consisting of Ti-Mo-Nb [18], Ti-Mo-Zr [19], Ti-Nb-Sn [20], Ti-Mo-Sn-Nb[21] and Ti-Mo-Nb-Zr [22], etc., to the best of our knowledge, it is still scarce or uncommon in the literatures. It rarely has any group considered using Mg to develop or fabricate permanent or long term orthopedic implant alloy. Even as Mg is an essential element for the human body and is largely found in bone tissue, its presence is beneficial to bone strength and growth [23].

Additionally, Mg has a density similar to that of natural bone (1.8–2 g/cm^3) and has been reported to support bone cells [24]. Equally, Mg is also a co-factor for several metabolic enzymes and stabilizes the structure of DNA and RNA [25, 26]. The only possible limitation can be the corrosion rate, in which the presence of Sn, Mo, and Ti can ameliorate [23]. Therefore, in the biomaterials' research field, magnesium can be of great practical significance and expected to rival the commonly used Ti-6Al-4V alloy for orthopedic applications.

This study combined compositional design and heat treatment (annealing) to fabricate a metastable β-type Ti-8Mo-4Sn-2Mg alloy with ultralow modulus akin to that of human bone. Furthermore, the Effect of elemental additions on microstructure and electrochemical behavior of Ti-8Mo alloy were evaluated.

II. EXPERIMENTAL WORK

A. Materials preparation

Previous experimental investigations on the Ti-Mo alloy systems revealed that α′′ and β phases coexist within 5 -10 at—% Mo threshold [19]. Hence, we chose Ti-8at. % Mo alloy to evaluate the influence of alloying additions.Hence, the new low-elastic modulus Ti-8Mo-4Sn-2Mg is designed using theapproach discussed elsewhere [19, 27, 28]. The starting binary Ti-8Mo and the new multi-component Ti-8Mo-4Sn-2Mgalloys were prepared from commercially pure Ti, Mo, Sn, and Mg metals of high quality (~99.97% purity) by vacuum arc-melting method. The actual compositions (See Table 1 and 2) of the resulting ingots were analyzed by inductively coupled plasma atomic emission spectrometry (ICP-AES) at the lab of Metallurgical and Materials Engineering, University of Nigeria, Nsukka, Enugu State. (Atomic percent are used here and throughout the text unless otherwise specified). To ensure good standard, the ingots (30g) were inverted and remelted at least six times to ensure chemical homogeneity before casting into cylindrical rods. The cast rods were heat-treated at 960°C under a high vacuum for 24 h inside a tubular furnace, followed by quenching in the water at room temperature.

B. Structural and mechanical properties characterization

The phase's characterization was performed using X-Ray diffraction (XRD, Siemens D500 diffractometer) operated at room temperature at 40 kV, 30 mA, and CuKα radiation (λ =1.5418 Å). The normal coupled (θ-2θ) scans were performed at a 2θ range of 20° to 90° with 0.02 step size and 1*s* dwell time. The phases were distinguished through comparison with simulated diffractograms using powder Cell software [30]. The microstructural characterization was investigated after grinding and polishing through standard metallographic techniques and then etched in Krol's reagents (3ml HF, 6ml HNO_3, and 100ml H_2O) by scanning electron microscopy (using Hitachi's Secondary electron microscopy).Thermogravimetric analyses (TGA) studies were carried out on a Setaram LABSYS Evo STA Instrument. Two entire heating-cooling cycles from 0 to 1200°C at a heating and cooling rate of 10 °C/min was used to investigate the existence of martensitic transformation and other transformation.

The change in hardness due to the alloying addition was measured using a Vickers micro-hardness tester under the load of 5N as defined by standards ASTME92. Each sample was measured at least ten times; then, the average was taken. Besides, compression tests were performed using an Olsen 8500 series testing machine in the air at room temperature with a strain rate of 4 x 10^{-3} s^{-1}. The dimension of test samples was 12 mm x 8mm for length and diameter, respectively (according to ASTM D882-11). The true stress-strain data are obtained from this equipment following the process detailed in [31]. The load and extension were measured using a load cell fixed on the machine. The ultimate tensile strength of the specimens was obtained from the true stress-strain curve. The samples' elongation was achieved by calculating the specimens' gage length before and after the tensile tests.

In the case of biomaterials for load-bearing applications, the elastic modulus is an important criterion because the stiffness mismatch between implant material and surrounding bone leads to 'stress shielding' and the bone [32]. The present study's elastic modulus was determined by two methods,namely, ultrasonic and nano-indentation techniques. For the elastic modulus (EM) determined by free resonance (using Olympus 38DL PLUS model) techniques, the obtained moduli values were an average of eight (8) measurements. For comparison, Young's modulus of the staring materials, Ti-8Mo and Ti-6Al-4V alloy as the reference, were also determined under the same condition. The equation's details for the dynamic Young's modulus (E) have been explained elsewhere [22, 32].

To compare with the ultrasound technique, the nano-indentation technique (NHT) was also used to determine the materials' elastic properties at a nanoscale level in a single experiment, after polishing the surface down to 2500 grit with emery papers under running water. The indentation process is continuously monitored concerning force, displacement, and time. To ensure good standards, the experiment was repeated five times. The equation for reduced elastic modulus (E_r) is detailed elsewhere [32].

C. Electrochemical and cytotoxicity evaluation

Due to the nature of the human body, a metallic material in living tissue is usually susceptible to biochemical corrosion attack, and since the corrosion resistance and biocompatibility of the Ti-(4, 6, 10, 12, 20, 25) Mo alloys have already been investigated in previous studies by [22]. To study Sn and Mg micro-additions' Effect on the initial alloy's electrochemical behavior, corrosion analysis was performed in Hank's simulated body fluid (37 ± 0.5) °C. Small disc samples and controls cut from Grade 2 CP Ti were ground to a 2500 grit finish with emery papers under running water and then ultrasonicated in ethanol. Electrochemical experiments were conducted in a three-electrode (CH instrument 660D) glass cell, the working electrode area of ~8 mm^2, a platinum counter electrode, and the Ag/AgCl as a reference electrode. The working electrolyte was ~50 ml of unstirred naturally aerated Hank's balanced physiological solution, with a subsequent $NaHCO_3$ addition (details of Hank's composition has been discussed in [33], maintained at 37±1°C. Before the potentiodynamicstudies, cathodic treatment was carried out at -0.9V for 600s to remove any oxide films present on each specimen's surface. The measurements were performed using the potentiostat (model VersaSTAT 4). Potentiodynamic polarization scan was performed on the samples at the rate of 0.335mV/s in the potential range of -1 to 2V vs. SCE, and a step size of 1 mV, after reaching a steady state.

To ensure good standard, the experiment was repeated three times with different samples after polishing using waterproof emery papers before each test and undertaken in stirred solution maintained at 37±1◦C (i.e., normal body temperature) pH of 5.74.

In terms of cytotoxicity test, the Neutral Red Uptake (NRU) assay, one of the most common and fastest methods to assessing cell membrane integrity, was used to evaluate the cytotoxicity of the new Ti-8Mo-4Sn-2Mg alloy. The red blood cell cultures are incubated for one week, two weeks, and three weeks at approximately 37°C±1. This technique is based on viable (living) cells' ability to incorporate NRU dye in the Lysosomes. In that instance, the dye is added to cells after being cultured in the material's extract. The dye is then desorbed in acidified ethanol solution. However, the quantity of dye absorbed depends on cell condition. Uniquely, the viable cells exhibit the maximum absorption, while dead cells have no absorption. Since the absorbance intensity of material is closely related to the number of living cells, the cell viability can be evaluated by measuring red-light absorbance at 540nm with a spectrophotometer following the work by [13]. The cell survival rate is then calculated relative absorbance to a negative control sample (the NRU growth medium is used as the negative control). The absorption was measured within 1 hr of adding NRU desorb solution at 540 nm with VersaMax ELISA Microplate Reader. For the purpose, the cell viability was calculated as a ratio of living cells to the number of viable cells in the appropriate negative indicator. To ensure good standard, the experiment was repeated three times with different samples.

Table 1: A list of proposed structures with the theoretical and experimental determined lattice parameters, a (*Å*) *revealing the values of the critical formation and cohesive*

S/N	Structures	*Lattice parameters Å*		*Formation energy [EFm]* (ev)	Cohesive energy (ev)
		Theory	*Exper.*		
1	Ti-8Mo	4.64	3.15	40.0	2.15
3	Ti-8Mo-4Mn-2Sn	4.82	-	-55.3	5.38
4	Ti-8Mo-4Fe-2Mg	4.72		84.1	7.6
5	Ti-8Mo-4Sn-2Mg	4.80	3.29	-92.7	8.72
6	Ti-8Mo-4Fe-2Ca	4.60	-	-69.1	8.61
7	Ti-8Mo-4Mn-2Mg	4.81		73.9	8.1
8	Ti-8Mo-4Sn-2Ca	4.90	-	-53.8	8.15
9	Ti-8Mo-4Fe-2Mn	3.98		64.7	5.30
10	Ti-8Mo-3Fe-4Sn	392		76.2	6.6

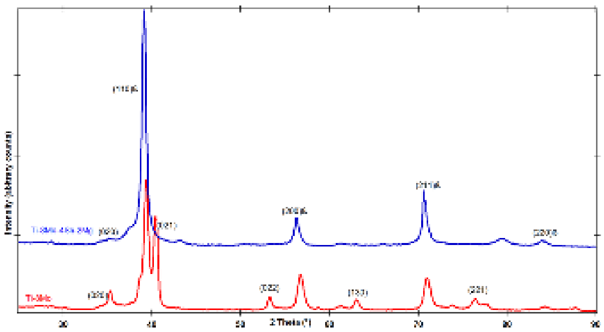

Fig. 1: XRD patterns of the starting binary Ti-8Mo and the newTi-8Mo-4Sn-2Mgs multicomponent alloys studied

III. RESULTS AND DISCUSSIONS

A. MicrostructuralAnalysis

It is known that alloy properties are governed by phase content and depend additively on the ratio and properties of the constituent phases. Therefore, phase composition is an important factor in the design of a biomedical material. As demonstrated by the results, several representative XRD spectra consisting of the starting Ti-8Mo alloy and newly developed Ti–8Mo–4Sn-2Mg multi-component alloy treated at 1070°C for 1 h are presented in Fig. 1 revealing the relevant peaks of the phases and lattice parameters, as determined by pattern refinement via Powder Cell software [30] as listed in Table 1. The result indicates the presence of β-phase associated with the strong (002), (200), (220), and (211) diffraction peaks in both alloys. Two additional low diffraction peaks associated with (020), (021), and (022], indexed as α"-phase is evident. The distinguishable lattice parameters of 3.15 and 3.29 Å for the starting Ti-8Mo and the new Ti8Mo-4Sn-2Mg represent a 5.45% reduction in the lattice constant suggests that the alloying additions (Sn and Mg) had an effect or improved the β-phase stability.

Additionally, the observed weakening or less noticed intensity of the α" peaks in the newly designed alloy further reflectsthe sensitivity of the Ti-8Mo alloy to the micro-additions. Their absence, then, would suggest that sufficient β-phase stabilization was achieved by the Sn and Mg micro additions. This is attributable to the increasing degrees of supersaturation or solid solution strengthening gained by adding the microalloying elements. It is expected that such changes might result in a great reduction of grain size and, eventually, improvement in mechanical property.

The microstructure was also investigated via SEM analysis, micrographs presented in Fig. 2(a)–(b). Under this analysis, the initial Ti-8Mo alloy exhibited a dual β plus α" phase microstructure. It can be observed that the binary Ti-8Mo alloy contains large equiaxed grains and display a mean diameter on the order of 500 µm (Fig. 2(a). Aside from the absence of the acicular martensitic structures around the grain boundaries evident in the Ti-8Mo microstructure, as expected, a significant change inthe Ti–8Mo-4Sn-2Mg microstructure [Fig. 2(b)] is evident. By contrast, the grain sizes of the multi-component Ti-8Mo-4Sn-2Mg alloy (Fig. 2(b) is more than ten times smaller than that of the starting Ti-8Mo alloy and displays a mean diameter on the order of 10– 50µm. Thus, the Sn/Mg micro-additions can modify the microstructure and decrease the initial grain size in the Ti-8Mo alloy system.

Additionally, the new alloy's grain sizes became more homogeneous than the starting microstructure, further revealing the Ti-Mo system's sensitivity to the micro-additions. Invariably, a significant change in property can be expected, as microstructure strongly affects almost all the properties of materials [34]. Based on the result, we can attest that the microstructural characterization (Fig. 2) complements the X-ray diffraction data (Fig.1) as both reveal improved stability of the new alloy would ensure their physical and mechanical compatibility when the alloy is employed in an orthopedic implant for repairing or as a replacement tissue in the human body.

B. Mechanical property evaluation

In order to evaluate the Effect of Sn and Mg alloying additions on the mechanical behavior, elastic modulus, hardness, and tensile test experiments were carried out on the heat-treated samples. Micro-hardness measurement is a very simple approach to evaluate the Effect of alloying additions on the mechanical properties, such as wear resistance. The Vickers micro-hardness values for the two alloys compared with as-received Ti–6Al–4V alloy and other conventional biomedical Ti alloys are presented in Table 2. The mean hardness of Ti–8Mo-4Sn-2Mg alloy (373–388 HV) is higher compared to that of Ti–6Al–4V alloy (288 HV) and the overall hardness of Ti–8Mo alloy (349–353 HV). For the case of Ti-8Mo alloy, the result can be ascribed to the coexistence of extensive α"-precipitation in the microstructure, which presents a hardness about half of the β phase [16]; variation within a given alloy was minimal. The values are consistent with the observed microstructure.

A summary of the results obtained from compressive stress tests of the starting Ti-8Mo alloy and Ti-8Mo-4Sn-2Mg multi-component alloy, achieved by dividing the load with the immediate area recorded during a tensile test of the alloys in the air at room temperature are presented in Table 2. The true stress-strain curves are shown in Fig. 3. From the demonstrated mechanical test results, it is obvious that the Ti-8Mo has lower mechanical properties due to the large grain size. By contrast, the result shows a strong relationship between mechanical behaviors and alloying content. As the Sn and Mg are added to binary Ti-8Mo alloy, there was a clear increase in the elongation, yield, and Ultimate tensile strength. It can be seen that the tensile strength of the starting Ti-8Mo alloy of 570 MPa has increased by

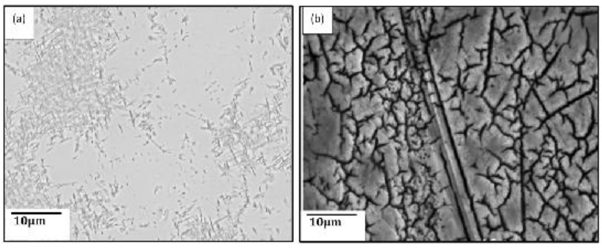

Fig 2: SEM microstructure of the alloys studied in as homogenized condition (a) binary Ti-8Mo, (b)Ti-8Mo-4Sn-3Nb-2Mg multi-component alloy, revealing the internal structure

More than 72%, due to the micro-alloying (Sn and Mg) additions. The new alloy's mean values were 982.9 MPa, 867.62 MPa and about 37.8% for tensile compressive strength, yield strength, and elongation (%), respectively. The mechanical properties elucidated above could be interpreted in the light of the constitution of the alloy. In the case of the new Ti-8Mo-4Sn-2Mg, the coexistence of mostly β-phase alongside α″ phase and suppression of ω phase precipitation are major pointers for the improved mechanical behavior above Ti-8Mo and Ti-6Al-4V alloys. Moreover, the increased solid solution strengthening offered by the alloying content and a decrease in grain size are also of interest, in agreement with the Hall-Petch relation.

Because of the importance of mechanical compatibility, the elastic modulus was carefully measured using free resonance wave and (nano-indentation) techniques. The values of Young's modulus obtained from both techniques are 36.8 GPa (38.9 GPa), respectively, for Ti-8Mo-4Sn-2Mg and 78GPa (80GPa), and 110 GPa (112GPa), for the starting binary Ti-8Mo materials and the reference Ti–6Al–4V alloy, respectively. This indicates a more than ~39% reduction in Young's modulus of the initial alloy, and generally lower than most commonly used alloys for some biomedical application [35]. As can be seen in Table 2, Ti–8Mo-4Sn-2Mg alloy presents a lower elastic modulus value much closer and comparable to cortex bone, the high ductility, better strength-to-modulus ratio than the Co–Cr–Mo alloys and shows better yield strength values compared to Ti–6Al-4V (solubilized). It is crucial to mention here that the presence of Mg and Sn in the alloy is effective in reducing Young's modulus and increasing the ductility, evident in Fig 4. In contrast, the presence of Mo is responsible for increasing strength [37]. Therefore, it is envisaged that if an extra mismatch of these elements is done, a further reduction in Young modulus and other properties can be expected.

Furthermore, a good agreement between the two techniques is evident. The observed small discrepancy can be due to variation in experimental equipment and environmental related error. For example, while ultrasound is done in an open laboratory, the nano-indenter is tested in an air-conditioned maintained laboratory room. Young's modulus' present value, E for the new alloy, is low compared to those obtained in previous studies [7, 26]. Therefore, we can conclude that Sn and Mg composition's synergic addition has a marked influence on the elastic properties of Ti-8Mo alloy. Thus, the low Young's modulus of Ti-8Mo-4Sn-2Mg alloy is attributed to phase constitution (coexistence of mostly β-phase alongside little fraction the α″ phase), compared to Ti-8Mo alloy where α″ is the predominant structure. This is in good agreement with what was reported by Davis et al. [38]. They suggested that α″ martensitic phase has less strain value (softer) than the β phase and hardness of about half of the β phase structure.

C. Electrochemical analysis

Electrochemical characterization of the Ti-8Mo-4Sn-2Mg and Ti-8Mo alloys was performed through the samples' potentiodynamic polarization study over the anodic and cathodic range. The representative polarization traces for the newly developed alloy with the reference Cp-Ti are presented in Fig. 4, while key parameters determined from this are presented in Table 3. Even though the test was repeated three times using the same procedure, surprisingly, the linear polarization measurements reveal minimal variation concerning the Cp-Ti; and implicitly, there is no statistically significant difference in their E_{corr} values between Ti-8Mo-4Sn-2Mg and Cp-Ti both in the classical cathodic and anodic branch in the Tafel region (Fig. 5). The corrosion potentials (E_{corr}), corrosion current densities (I_{corr}) and passive current densities (I_{pass}) obtained from the polarization curves of the Ti-8Mo-4Sn-2Mg and Cp-Ti using the Tafel extrapolation

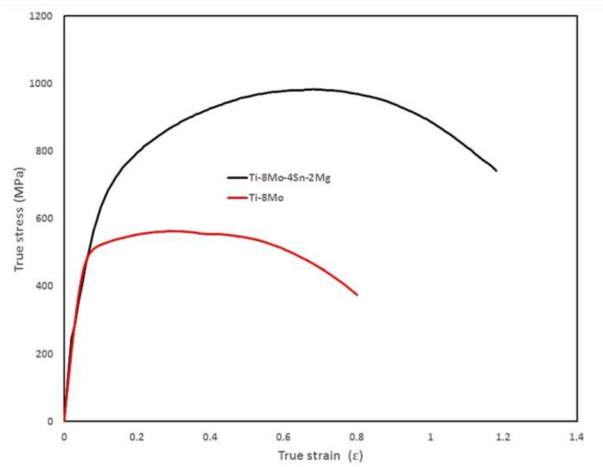

Fig. 3: True stress-strain curves of the alloy studied compared with initial alloy

method are − 0.24 ± 0.05V/SCE, 0.2 ± 0.1 μA/cm² and 3.02 ± 02 μA/cm² respectively , compared with the values of − 0.25 ± 0.04 V/SCE , 0.2 ± 0.1 μA/cm², and 3.85 ± 0.2 μA/cm² for the CP Ti, already in use as biomedical alloy. The uniformity in these values speaks well toward the potential application of the Ti-8Mo-4Sn-2Mg alloys in a clinical role. There is this clear uniformity, but between 1.8 to 2 E_{corr}, the cathodic branch of the Tafel plot for Ti-8Mo-4Sn-2Mg alloy was drifting towards the less noble arm. This is considered a lesser concern, given the expected environment within the body falls into the anodic regime [23]; however, further work would be needed to confirm this.

It is well known that corrosion resistance is influenced by various factors such as chemical composition, surface roughness, grain size, and texture. From the passive current densities data, it is reasonable to infer that the chemical composition or the addition of Sn and Mg to binary Ti-8Mo does not appear to modify so much the protection characteristics of its spontaneous oxides in the physiological fluid (Hank's solution). [23, 39].The smaller-grain structure of alloy likely aided the corrosion resistance coupled with the fact that the alloy derives its corrosion resistance to the formation of surface oxide films of TiO_2.

D. Cytotoxicity Evaluation

To assess the biocompatibility of the Ti-8Mo-4M0-2Sn alloy for biomedical application, a cytotoxic evaluation study was carried out using the Neutral Red Uptake assay. The results obtained from the cytotoxicity analysis of the Ti-8Mo-4Sn-2Mg alloy were compared to that of Ti–6Al–4V alloy, and the cytotoxicity control are presented in Fig 6. As shown in Fig. 6, the percentage of viable cells of the Ti-8Mo-4Sn-2Mg alloy is near to that of Ti–6Al–4V alloy for the 7, 14, and 21 days periods. This is attributed mostly to the presence of the high biocompatible alloying elements, i.e., Mo, Sn, and Mg elements, in the alloy, as well as the stability of oxide layer formed on the surface of the alloy, which inhibits the metal–ions release, in good agreement with what was reported by Hartwig [40]. They reported that Mg played a key role in the oxide layer's stability formed on the Ti-alloys' surface. Thus, based on the present results, one can infer that the new Ti-8Mo-4Sn-2Mg alloy is biocompatible and has better potential surpassing or rivaling the commonly used biomaterials, Ti–6Al–4V alloy.

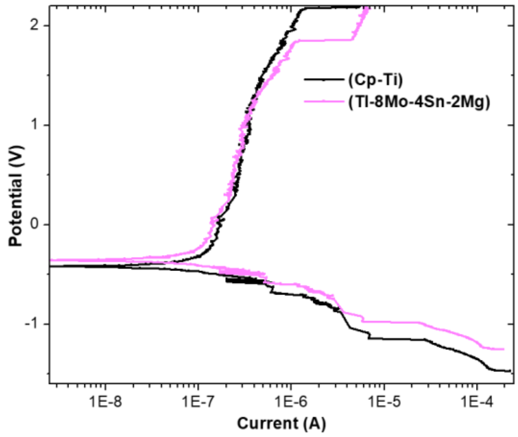

Fig. 4: Potentiodynamic polarization curves for the Ti-8Mo-3Sn-2Mg and reference Ti-8Mo after 2hrs immersion in Hank's solution at 37 ± 1° C, Scanning rate: 0.5Mv/s.

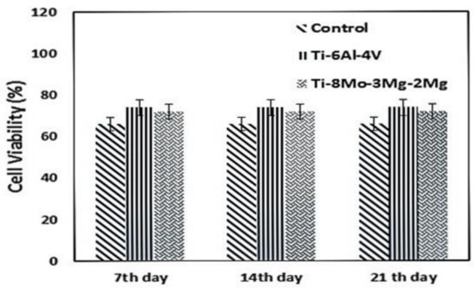

Fig.5. Mean ± SD percent of cell viability in extracts for one week, two weeks, and three weeks of-quenched-8Mo-4Sn-2Mg alloy and Ti–6Al–4V evaluated by NRU assay.

Table 2: Mechanical properties of the new alloy studied with typical biomaterials and shape memory alloys compared.

S/N	Reported alloy	*Hardness (HV)*	*E(GPa)*	UTS (MPa)	Strength/modulus ratio [x10⁻³]	ε(%)	Ref.
1	Ti-7.89Mo-3.97Sn-2.1Mg	373-388	36.8/38.9	982.9	12.3	37.8	This work
2	Ti-7.98Mo	349-353	78	597	4.87	13.01	This work
3	Ti-6Al-4V(annealed)	288	110	825–900	1.7-4.7	10–15	[13]
4	Ti-6Al-7Nb (swaged)	-	105	900	-	12	[23]
5	CP Ti(Grade 1-4)	-	102.7	170-485	1.7-4.7	15-24	[7]
6	Ti-6Al-7Nb(soln.)	-	110	910-990	-	21	[32]
7	Gum Metal	-	42	-	-	-	[31]
8	Ti-10Mo	390	95	-	-	-	[31,33]
9	ASTMF75(Co-Cr-Mo)	-	220	825-869	7.2-7.9	6-10	[31]
10	Cortical Bone	-	28.8-40	-	-	-	[5,35]

Table 3: Corrosion parameters (and standard deviation values) determined from the potentiodynamic polarization curve measured for the Ti-8Mo-4Sn-2Mg and Cp-Ti in physiological fluid solution at 37◊±1 C. (Hank's)

Alloy	The corrosion parameters (and standard deviation values) determined for the new Ti-8Mo-4Sn-2Mg alloy			
	$E_{corr.}$ (V_{SCE})	$j_{corr.}$(μA cm⁻²)	$j_{pass.}$(μA cm⁻²)	Reference
CP –Ti	-0.33(0.021)	0.33(0.07)	2.17(0.04)	This work
Ti-6Al-4V	-0.327(-)	0.326(-)	0.323(-)	[13]
Ti-8Mo-6Nb-4Zr	-0.301(0.021)	0.214(0.07)	1.9(0.05)	This work
Ti-8Mo	-0.316(0.022)	0.217(0.06)	1.92(0.04)	[22]

IV. CONCLUSIONS

A new low elastic modulus metastable β-type Ti-8Mo-4Sn-2Mg alloy is presented in this study. The microstructure, mechanical properties, biocompatibility, and corrosion resistance of this newly developed Ti alloy are characterized, and the main results are summarized as follows:

1. The present study shows that Sn and Mg micro-addition to binary Ti-8Mo alloy led to microstructural refinement and improved elastic modulus to 36.8 GPa(38.9 GPa) based on free resonance (nano-indentation) measurements, respectively. The agreement level in the two methods is very remarkable concerning metallic biomaterial for orthopedic implant applications.

2. Although further studies may be considered, the outcome of the mechanical, biocompatibility and electrochemical tests show that this new Ti-8M0-4Sn-2Mg can be a favorable candidate for some biomedical applications than the commonly used biomaterials alloys
3. The cytocompatibility result from one to three weeks of investigations shows that the new Ti-8Mo-4Sn-2Mg alloy is similar to Ti–6Al–4Valloy.

REFERENCES

[1] https://www.who.int/violence_injury_prevention/road_safety_status/2015/en/
[2] Myron Spector, a promising decade for biomedical materials, Biomed. Mater. 11 Editorial, (2016) 1, doi: 10.1088/010201 1748-6041/11/1/010201.
[3] R.Sangeeth kumar, Dr.S.V.Ramana, Inter.J. Engn.Trends and Technology (IJETT) 28; 2 (2015)91-97.
[4] http://www.corrosion-doctors.org/Implants/Metals.htm
[5] M. Geetha, A.K. Singh, R. Asokamani and A.K. Gogia, Ti-based biomaterials, the ultimate choice for orthopedic implants – A review, Progress in Materials Science 54 (2009) 397–425.
[6] DM Brunette, P.T.M.Texlon, Thompson, Titanium in medicine, Springer Berlin, Heidelberg, New York, 2001.
[7] Eisenbarth, E., Velten, D., Muller, M., Thull, R. &Breme, J. Biocompatibility of beta-stabilizing elements of titanium alloys. Biomaterials 25, 5705–5713 (2004).
[8] Wilson, Anne P. (2000). "Chapter 7: Cytotoxicity and viability" in Masters, John R.W. Animal cell culture: A practical Approach Vol.1 (3rd Ed.). Oxford: Oxford University Press. ISBN 978-0-19-963796-6.
[9] Carlos Oldani and Alejandro Dominguez (2012). Titanium as a Biomaterial for Implants, Recent Advances in Arthroplasty, Dr.SamoFokter (Ed.), InTech, DOI: 10.5772/27413.
[10] Nag, S., Banerjee, R. & Fraser, H. L. Microstructural evolution and strengthening mechanisms in Ti-Nb-Zr-Ta, Ti-Mo-Zr-Fe and Ti-15Mo biocompatible alloys. Mater. Sci. Eng. C 25, 357–362 (2005).
[11] MitsuoNiinomi, Recent research, and development in titanium alloys for biomedical applications and healthcare goods Science and Technology of Adv. Mater. 4 (2003) 445–454.
[12] Y. Abd-elrhman, M. Gepreel, A. A. Abdel-Moniema, Sengo Kobayashi, Compatibility assessment of new V-free low cost Ti-4.7Mo-4.5Fe alloy, Materials and Design 97 (2016) 445–453.
[13] Y. X. Tong B. Guo, Y. F. Zheng, C. Y. Chung, and L. W. Ma, J. Mater. Eng. & Perf.20: 4(2011)762-766
[14] Nnamchi, P.S., Todd, I., Rainforth, MW, Department of Materials Science, and Eng. The University of Sheffield,UK(Ph.D. Thesis 2014.).
[15] Hao, Y.L., Li, SJ, Sun, S.Y., Zheng, C.Y., Yang, R., Elastic deformation behavior of Ti-24Nb-4Zr-7.9Sn for biomedical applications, Acta Biomater.3, (2007)277–286.
[16] L.J. Xu,Y.Y. Chen, Zh.G. Liu, and FT Kong, The microstructure and properties of Ti-Mo–Nb alloys for biomedical application, J. Alloy &Compds, 453: 1–2,(2008), 320–324.
[17] Diego Rafael Nespeque Correa et al., Effect of the substitutional elements on the microstructure of the Ti-15Mo-Zr and Ti-15Zr-Mo systems alloys, j. Mater. Res. Tech 4; 2 (2015) 180-185.
[18] Hiroaki Matsumoto, Sadao Watanabe, ShujiHanada, Microstructures and mechanical properties of metastable β TiNbSn alloys cold-rolled and heat-treated, J. Alloy &Compds. 439, 1–2, (2007), 146–155.
[19] Diego Rafael Nespeque Correa et al., Effect of the substitutional elements on the microstructure of the Ti-15Mo-Zr and Ti-15Zr-Mo systems alloys, j. Mater. Res. Tech 4; 2 (2015) 180-185.
[20] Z. Yu, M. Zhang, Y. Tian, J. Cheng, X. Ma, H. Liu, et al., Designation and development of biomedical Ti alloys with finer biomechanical compatibility in long-term surgical implants, Front. Mater. Sci. 8 (2014) 219–229.
[21] P.S.Nnamchi and R.E.Njoku: Alloy design and property Evaluation of Ti-Mo-Nb-Sn alloy for biomedical application, Nigeria J. Tech., 32. No. 3. (2013), 410 – 416
[22] Paul S. Nnamchi, C.S. Obayi, Iain Todd, M.W. Rainforth. Mechanical and Electrochemical Characterization of New Ti-Mo–Nb–Zr Alloys for Biomedical Applications, Journal of the Mechanical Behaviour of Biomedical Materials, 60, 2016, pp 68-77.
[23] Witte F, Kaese V, Haferkamp H, Switzer E, Linderberg AM, Wirth CJ, Windhagen H. Invivo corrosion of foue magnesium alloys and their associated bone response, Biomaterials,26 (2005),3557-3563.
[24] Emsley J. The elements. Oxford: Clarendon Press; 1998.
[25] Hartwig A. Role of magnesium in genomic stability. Mutat Res 475, (2001)113–21.
[26] AgungPurnama, HendraHermawan, Jacques Couet, Diego Mantovani, Assessing the biocompatibility of degradable metallic materials: state-of-the-art and focus on genetic regulation, ActaBiomaterialia 6 (2010) 1800-1807.
[27] Ghosh, S. Delsante, G. Borzone, M. Asta and R. Ferro, Phase stability and cohesive properties of Ti-Zn intermetallic: first principle calculations and experimental results, ActaMaterialia 54 (2006) 4977–4997.
[28] Raabe D., Sander B., Fri'ak M., Ma D, Neugebauer. J. Theory guided the bottom-up design of β-Titanium alloy as biomaterial based on first principle calculation: Theory and experiments, ActaMaterialia 55, (2007), 4475–4487.
[29] S Kirklin, JE Saal, B Meredig, A Thompson, JW Doak, M Aykol, S Rühl, The Open Quantum Materials Database (OQMD): assessing the accuracy of DFT formation energies, npj Computational Materials 1, 15010
[30] Kraus W and Nolze G. J. Powder Cell: a program for representing and manipulating crystal structures and calculating the resulting X-ray powder patterns. Journal of Applied Crystallography. 1996; 29(3):301-303.
[31] P. Mujumdar, S.B. Singh and M. Chakraborty, Elastic modulus of biomedical Ti alloy by nanoindentation and ultrasonic techniques: A comparative study, Mater. Sci. & Eng. A 489 (2008) 419–425.

[32] Julie Lévesque, HendraHermawan, Dominique Dube, Diego Mantovani. Design of a pseudo-physiological test bench specific to the development of biodegradable metallic biomaterials.ActaBiomaterialia, 2008,4(2), p. 284-295.

[33] Zhao X., Niinomi M., Nakai M., and Hieda J.; "Beta type Ti-Mo alloys with changeable Young's modulus for spinal fixation applications,"ActaBiomater. 8 (2012) 1990–1997.

[34] S.E.Haghighi, H.B.Lu, G.Y.Jian, G.H.Cao, D.Habibi, L.C.Zhang, Effectofα″martensite on the microstructure and mechanical properties of beta-type Ti–Fe–Ta alloys,Mater. Des. 76 (2015) 47–54.

[35] Friedel J., in: J. M. Ziman (Ed.); the Physics of Metals, Cambridge University Press, Cambridge, 1969, p. 340.

[36] MK Gouda, K. Nakamura, M.A.H. Gepreel, First-principles study on the Effect of alloying elements on the elastic deformation response β-titanium alloys, J. Appl. Phys. 117 (2015) 214905, http://dx.doi.org/10.1063/1.4921972.

[37] R. Davis, H.M. Flower, DRF West, Martensitic transformations in Ti-Mo alloys, J. Mater. Sci. 14 (1979) 712–722.

[38] Li L, Gao J, Wang Y. Evaluation of cytotoxicity and corrosion behavior of alkali heat-treated magnesium in simulated body fluid. Surf Coat Technol. 185, (2004):92–8.

[39] Diego Rafael Nespeque Correa et al., Effect of the substitutional elements on the microstructure of the Ti-15Mo-Zr and Ti-15Zr-Mo systems alloys, j. Mater. Res. Tech 4; 2 (2015) 180-185.

[40] Hartwig A. Role of magnesium in genomic stability. Mutat Res., 475 (2001); 113–21.