Published in final edited form as:

J Mech Behav Biomed Mater. 2010 November ; 3(8): 549–558. doi:10.1016/j.jmbbm.2010.06.006.

Tribological Properties of Ti-Based Alloys in a Simulated Bone-Implant Interface with Ringer's Solution at Fretting Contacts

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Abstract

The wear properties of oxidized and non-oxidized gamma-TiAl (a potential biomaterial) as well as Ti-6Al-4V and CP-Ti disks were studied and characterized by means of standard wear tests using a custom made bone pin arrangement. The Ti-base disks were oxidized in air at 500°C and 800°C for one hour. The tribological properties of the oxides formed over the disks were studied using a linear reciprocating wear testing machine under both dry and simulated biological conditions using Ringer's solution. Loss of metal oxide and coefficient of friction values were determined from the wear testing. From the results, abrasion and adhesion were the primary wear mechanisms in each of the three alloy-bone pairs. Specifically, the oxide formed on gamma-TiAl possessed the highest COF and wear resistance of the three materials which were studied. Also, as expected, bone wears down faster than the Ti-base metal oxide.

Keywords

Titanium alloy; Titanium Oxide; Tribology; Fretting; Bone-Implant

Introduction

It has been recognized that early osseointegration is required in order to catalog an implantation surgery a medical success (Liner et al., 1988; Sieber et al., 1999). The degree and rapidity of the fixation of bone cells on the implant surface is dependent on the surface composition and topography (Böhler et al., 2002; Kim and Kim, 1993; Northmore-Ball et al, 1993; Wimhurst et al, 2001). Consequently, a variety of methods have been developed to generate irregular surfaces to enhance the clinical performance of the implants within the human body (Scott et al., 1993). One of the techniques used to roughen titanium-base implant surfaces is usually by forming an oxide layer over the titanium device. This procedure is estimated to increase the surface area of the implant by approximately 25 to 35 percent, providing additional adherence area, thus improving the osseointegration between the tissue being formed by new bone-cells and the surface of the implant (Taljanovic et al, 2003). The combined mechanical and physical properties of titanium alloys provide implants which are highly damage tolerant. The problem, however, comes with the existence of continuous micro-motion between implants and bone tissue. The fact that the bone and the metal implant cannot be joined together perfectly, combined with the continuous movement between both, results in implant loosening and

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infection caused by the presence of particulate wear products. For this reason the friction and wear resistance of metal implants, especially titanium-base devices with enhanced surface topography, play an important role in determining the performance of biomaterials. From this perspective, different materials and surface finishes have been explored, although with each new possibility new problems arise. Based on a new approach by Sieber et al. 1993, from experimenting with various material-material pairs, many studies have evaluated the effect of wear particles on the rejection reactions in the implanted area. In general all of them agree that the presence of wear debris supplements the abrasive wear of the implant, making this phenomenon a continuous cycle, since when the oxide layer is breached by mechanical abrasion, the implant surface recovers automatically by generating a passive oxide layer, which will wear down again and again. These metallic particles (ions) become enzyme inhibitors which may cause infections or necrosis, which ultimately may require removal surgeries or in severe cases amputation and even death (Böhler et al., 2002; Chu and Liu, 2008; Harman et al., 1997; Kim and Kim, 1993; Lalor et al., 1991; Northmore-Ball et al, 1993; Shimmin et al., 2008; Wilmhurst et al, 2001; Witt and Swann, 1991).

A recent study (Choubey et al., 2004) exploring the fretting wear resistance of common Ti-alloy biomaterials (CP Titanium, Ti-6Al-4V, Ti-5Al-2.5Fe and Ti-13Nb-13Zr) in Hank's solution observed the formation of slip bands, cracking and extensive plastic deformation as predominant wear mechanisms in these materials. Also, the coefficient of friction (COF) of the alloys was lower when measured in Hank's solution. There is little other information on the fretting wear of Ti-alloys in a simulated bone-implant interface and in a simulated body fluid environment, which has major significance and a more valid outlook as far as the application of Ti-implant materials is concerned. From this perspective, the current research addresses the friction and wear performance of some relevant Ti-base alloys such as Ti-6Al-4V, CP-Titanium and gamma-TiAl (a potential implant material) in a simulated bone-implant interface, submerged in Ringer's solution. A comparison is made between the behaviors of these Ti-base materials in dry as well as lubricated conditions after oxidation at 500°C and 800°C.

Experimental Procedure

Specimen Preparation

Disks of 25.4 mm diameter (1 mm thick) were machined using electric-discharge machining (EDM) from the as-received titanium materials: Ti-48Al-2Cr-2Nb (at.%), Ti-6Al-4V (wt.%) and commercially pure (CP) titanium (95 grade). Because the titanium disks were the moving component in this fretting study, the surfaces were prepared to be as flat as possible. The samples were ground to a 600 grit finish using standard emery paper. Prior to the oxidation process and the fretting tests, both the samples and the wear pins (fabricated from bone) were cleaned using 70% isopropyl alcohol pads. A conventional oven was used to oxidize these titanium alloy disks. Unoxidized disks were tested for comparison. Two sets of disks were treated (oxidized), one at 500°C and another at 800°C for one hour. After oxidation the samples were allowed to reach ambient temperature prior to conducting the wear tests. A linear reciprocating tribometer was used to measure the wear resistance and COF of the oxide in each of the alloys in a dry and lubricated environment.

Bone Pin Preparation

A custom made bone pin was prepared from the femur of consumable livestock. The complete femur was immersed in water at 100°C for 6 hours to remove soft tissue attached to the bone. In order to prevent changes in the mechanical properties of the bone, the femur was never allowed to touch the heat source while being slowly heated. After this process, the remnant soft tissue attached to the bone was mechanically removed by hand. The bone was then

submerged in a bleach-water mixture for 120 hours. This mixture, which consisted of 1 liter of bleach per 15 liters of water, was replaced every 24 hours. The femur was finally cleaned and dried in ambient air and stored in a refrigerator. The denser extremities of the bone (epiphysis) were cut and grinded until the final geometry of the bone pin was obtained as shown in Figure 1. The tip of the bone pin had a radius of curvature similar to that of a standard 3.175 mm ball. The inside of the bone was modified using a manually operated threaded perforator. An aluminum rod was tightly screwed into the inside of the bone. This rod transmits the force applied by the weights in the wear testing machine arm across the bone pin and on to the titanium samples. The bone pin was refurbished after each fretting test in order to maintain the radius of curvature of the primary contact tip.

Wet and dry tests were run for ten minutes with an applied force of 3N. The sliding velocity was set to 900mm/min with 5 Hz reciprocatory tangential sliding amplitude of 5 mm. All lubricated tests were performed using Lactated Ringer's Injection USP Solution. Each experiment was run using fresh solution obtained from newly opened bags at ambient temperature ($25^{\circ}\text{C} \pm 5$). The composition of the Ringer's solution was 600 mg Sodium Chloride USP, 310 mg Sodium Lactate USP, 30 mg Potassium Chloride USP, 20 mg Calcium Chloride USP per 100mL. The solution's pH was approximately 7.4. During the lubricated tests, the disks were completely submerged in Ringer's solution. All non-lubricated tests were carried out at ambient temperature ($25^{\circ}\text{C} \pm 5$) and relative humidity of $67 \pm 2\%$. Continuous monitoring of the humidity levels was done using a Davis Vantage Pro-2 weather station. Mass loss (volume loss) of every sample and pin were measured after every test using a high resolution five digit analytical balance with a resolution of 0.01mg. A detailed micro-structural characterization of the oxidized and worn surfaces was performed using a Scanning Electron Microscope (Phillips XL30s SEM-EDS), and a High Resolution Optical Microscope (Nikon Eclipse 80i). Images of surface topography and wear tracks were taken using the SEM and the optical microscope. By analyzing the images, the surface roughness was determined and the predominant wear mechanism was observed and evaluated. Prior to using the SEM, EDX and optical microscope, all the samples were carefully cleaned with water at ambient temperature and dried. Since some of the Ti-base alloys generated brittle oxide films after the thermal treatments, the cleaning process was performed with a low pressure water source. Data from all the wear tests was obtained using the commercial software provided for the tribometer. Using the tribometer software and Matlab® code, 50 points per cycle were eliminated and only the positive peaks were retained. Since the linear reciprocating movement is comparable to the actual physical movement of a joint, the resulting plots behave cyclically, thus everything from zero to the most negative value was discarded. The moving average slope was assigned and plotted for each experimental setup.

In total, forty eight titanium alloy disks were prepared, oxidized and used to perform the wear tests on the oxide layer generated in lubricated and non-lubricated mediums. All the tests were repeated using a custom made bone pin. Four disks of each of the alloys, gamma-TiAl, Ti-6Al-4V and CP titanium were oxidized (two disks at 500°C and two at 800°C). Two tribology test runs were performed, one on either side of each disk. Weight measurements were taken before and after polishing, oxidizing and performing each of the tribology tests on every titanium alloy disk using the five digit precision analytical balance.

Results and Discussion

Oxide morphology

The results for the weight (mass) gained after oxidation for one hour at two different temperatures (500°C and 800°C) are shown in Table 1. The difference in the roughness, porosity and density of the oxide layer formed on each of the Ti-alloys treated at 500°C is noticeable in Figure 2. The images illustrate a consistently more porous and rough pattern in

the oxide generated on the surface of the gamma-TiAl disks compared to the other two materials. This may explain the difference in the mass gained after oxidation of the samples. The existence of oxide peaks on the surface of gamma-TiAl is visible while the oxide formed on Ti-6Al-4V alloy has fewer oxide peaks and more porosity (holes). The oxide from the CP-Ti looks compact and uniform with the consistency of a painted coating rather than a layer. Interestingly on visual inspection, the gamma-TiAl samples emitted a blue color, while Ti-6Al-4V was golden and blue patched pattern was observed on the CP-Ti samples. For the disks treated at 800°C, the Ti-6Al-4V disks showed the highest increment in mass after oxidation. However, the oxides formed on the gamma-TiAl and CP-Ti disks were strongly adhered to the substrate while the oxide on Ti-6Al-4V was brittle and spalled off from the substrate metal. The weak adherence of the oxide layer formed on Ti-6Al-4V at 800°C was not seen with the samples treated at 500°C. Figure 2 illustrates a SEM image of the oxide layers formed on the three Ti-base alloys at 500°C and 800°C. From the image, the porosity and the roughness peaks of the Ti-oxide surface are observable. The visible porosity of the oxide formed on gamma-TiAl is a desirable characteristic, which may actually permit better mechanical bonding with bone in an implant material. From Table 2, it appears that the oxide layer is rich in alumina. Although aluminum ions have been linked to different enzyme restraining reactions, the aluminum oxide is an amphoteric substance making this oxide resistant to strong acids and alkali environments (Choubey et al, 2004). Also, it possesses high strength and is highly wear resistant. These qualities are ideal for preventing implant damage due to the harsh biological environment of the human body (Jacobs et al., 1998). Based on the mixed oxide formed at this temperature and the absence of chromium or niobium it can be cautiously assumed that no atomic diffusion of chromium or niobium occurred into the surface of the disks. However, niobium is well known for forming a porous layer similar to the one shown in Figure 2 after diffusion into TiO₂, especially at a relatively high temperature (Diebold, 2003). This possibly means that although the surface concentration of Nb was not high enough to be detected with the EDX, diffusion of Nb into the oxide layer may have occurred, resulting in a porous outside layer. Different surface morphology and colors were observed in the gamma-TiAl samples after being oxidized at 800°C, as shown in Figure 2. Table 2 highlights the components of the oxide. Although the continuous growth of the aluminum oxide film by the removal of oxygen anions from the TiO₂ was expected, contrary to the disks treated at 500°C, at this temperature the oxide layer is rich in titanium. Also, no evidence for an intermetallic TiAl alloy at the interface was found. The presence of niobium and chromium in the oxide layer is clearly evidenced in Table 2 and hence the diffusion of both elements from the substrate and into the oxide is substantiated for oxidation at 800°C.

Tribology tests

Dry tests—Figure 3 shows the coefficient of friction (COF) values versus number of cycles of the implant materials tested against bone pin in a dry medium. All three Ti-base alloys tested showed the same trends with a high COF at the beginning of the test which decreased as the number of cycles increased. The COF values for the three Ti alloys surfaces appear to depend on the roughness of the oxide layer with gamma-TiAl exhibiting the highest value and CP-Ti having the least. The COF values remained constant in the case of gamma-TiAl for about 300 cycles after which a drop occurred probably caused by the continuous wearing of the contact point of the bone pin which changed the initial force parameter from a normal applied load to a distributed force. Figure 4A shows surface images of the wear track formed on gamma-TiAl for both treatment temperatures. For the disk treated at 500°C a more marked and deeper wear track is visible indicating that the oxide film is not very wear resistant. On the contrary, a very superficial trail shows that the oxide formed at 800°C possesses greater resistance to wear than the one formed at 500°C confirmed by the lower mean mass removed indicated in Table 3. It was clear during experimental testing that the custom made contact bone pin wears down more than the gamma-TiAl samples as a result of which the contact point of the bone pin becomes

larger and the protective oxide film wears down with time, reflected in the decreasing COF values as the test progresses as seen in Figure 3. Debris from both the sample and the bone pin were also visible. As already mentioned, while it is normally assumed that the lower the COF, the smaller the amount of material removed by wear, this reasoning does not apply to our results. Although the COF values of the gamma-TiAl disks are larger than those for Ti-6Al-4V and CP-Ti, the amount of mass removed after the wear tests in the gamma-TiAl samples was considerably smaller than for the other two alloys. As shown in the Figure 4B, the wear scars for the Ti-6Al-4V are wider and deeper, which accounts for the larger amount of oxide lost after the wear tests. Unlike the gamma-TiAl samples, for Ti-6Al-4V samples, the wear scratches were mainly located on the sides. Also, a more uniform abrasion was observed on the Ti-6Al-4V treated at 800°C as evidenced in Figure 4B. This indicates that the resistance to wear of this oxide layer was lower and instead of wearing down the bone pin, as the gamma-TiAl samples did, the oxide on the surface of Ti-6Al-4V suffered greater damage. The wear paths generated on the CP-Ti samples (Figure 4C) are less profound and more homogeneous than on the other Ti-samples. There were areas within the surface of the track that remained intact and were not affected or damaged by the wear tests. In general, in the dry tests using a bone-pin, no plastic deformation or grain pullout was visible on any of the Ti-samples. On the contrary, the presence of abrasion on the surface of all the samples was evident, especially for the Ti-6Al-4V samples oxidized at 800°C in which some parts of the oxide layer were spalled off due to the constant fretting. The clearly visible white color of the wear paths in the optical images is a result of the adhesion of the bone pin debris on the oxidized surfaces during the tests. While it appears the oxides on Ti-6Al-4V do not possess the ability to withstand fretting, the low COF values and low mass loss after fretting in CP-Ti are attributed to the lower roughness on the surface of the oxide which minimizes the resistance to movement, thus aiding free relative motion and consequentially a low amount of wear. Even though the high COF and roughness of the oxide layer in gamma-TiAl is desirable in terms of osseointegration and attachment to the surrounding tissue, it also makes this oxide more vulnerable to wear. Nevertheless, in terms of mass lost, gamma-TiAl appears to possess greater wear resistance than the Ti-6Al-4V and CP-Ti oxides under the experimental conditions used in this study.

Figure 5 shows smoother trends of the COF for the control samples of the three Ti-base alloys which were not oxidized. In the case of the unoxidized samples, the starting COF values were smaller but increased and reached values similar to those of the oxidized samples towards the end of testing. This is probably due to sample work hardening initially followed by possible formation of localized oxides in the latter stages of testing. The mean mass loss values, due to wear, for the three materials are presented in Table 3. The gamma-TiAl samples lost the smallest amount of mass after the tribology tests, when compared to Ti-6Al-4V and CP-Ti. Furthermore as seen in Figure 6, the wear scratches formed on the surface of the gamma-TiAl surfaces are considerably shallower and less marked than the wear tracks seen on the other two materials. It is obvious from Tables 3 and 4 that the oxidized samples lose more mass than the unoxidized samples in the fretting tests under the same conditions. In general there were almost no wear particulates from the bone pin after the tests and consequently no weight changes could be quantified. Debris from the Ti-base alloy samples after the tests was noticed, but relatively small when compared to the tests with the oxidized samples. This again is an indication of the brittle nature of the oxide layer obtained from thermal treatment. Also, other than surface scratches, no fractures or cracks, pits, oxide, grain pull out or plastic deformation were visible on the disk surfaces after the wear tests.

Lubricated tests—COF versus number of cycles for the oxidized Ti-base implant materials which were completely submerged in Ringer's solution and fretted against a bone pin are plotted in Figure 7. Generally, as expected, there is a considerable reduction in the COF values obtained in the lubricated tests when compared to the dry tests. Consequently, there is a decrease in the mass lost for the lubricated tests when compared to dry tests. The optical images shown in

Figure 8 confirm this behavior. Contrary to the dry tests, in the presence of Ringer's solution, the bone pin did not wear down as quickly. In fact, the bone pin absorbed a lot of lubricant during the tests, and its surface changed from a porous texture to being dense and filled. This liquid absorption appears to strengthen the bone pin, thus making it more resistant to wear. In general, for gamma-TiAl only parts of the oxide film wore down, mainly in a specific path across the full length of exposure which also explains the small amount of material loss and low COF values. As observed in Figure 7, for Ti-6Al-4V and CP-Ti, there is a considerable reduction in the COF values obtained in the lubricated tests when compared to the dry tests. In general, there is a large difference in the amount of material removed between the lubricated and dry tests. Also, as expected, the usage of lubrication decreased the values for the coefficient of friction in all the materials for all the treatments. Also when comparing the images of the sample surfaces after both dry and lubricated tests, for all the alloys tested, an increased concentration of scratches and generally greater damage on the surface of the oxides is clearly visible on the sample surfaces from the lubricated tests. However, from the weight measurements it can be concluded that although a greater localization of wear is revealed in the lubricated tests, under dry conditions, the scars were deeper reflecting greater mass removal. Debris from both the Ti-samples and the bone pin were visible in all the tests. From the tribology tests, all the samples treated at 500°C, with the exception of gamma-TiAl, suffered similar mass loss and COF values. In the case of oxidation at 800°C the weight losses were not similar. Ti-6Al-4V samples attained the largest mass loss after wear testing for both oxidation temperatures. Finally, it appears that the presence of a lubricant does affect the behavior and mechanisms that dominate the wearing down of the bone/implant pair.

Figure 9 shows the values of the COF for the control (unoxidized) samples of the three alloys after tribology tests against a bone pin in a lubricated medium where similar behavior is observed for all three materials. It appears that the Ringer's solution serves as an excellent lubricant in aiding the rapid stabilization of the values of COF at lower values in these fretting tests. As shown in Table 4, the gamma-TiAl samples suffered the smallest amount of mass removed and the least amount of surface damage after the wear tests (see Figure 10). In contrast, as also seen in Figure 10, large scratches were observed all around the wear path for the Ti-6Al-4V and CP-Ti samples with the wear scratches being sufficiently deep to erase the transverse grinding scratches formed during preparation of the samples with the 600 grit papers prior to the wear tests. In general, the amount of mass removed after the wear tests in the lubricated medium was smaller when compared to the tests performed without lubrication for the same parameters. There were almost no wear particulates from the bone pin after the tests. Debris from the metal samples after the tests was noticed, although the amount of particulates was relatively small when compared to the tests with the oxidized samples. Also, other than surface scratches, no fractures or cracks, pits, oxide, grain pull out or plastic deformation were visible on the disk surfaces after the wear tests.

Conclusions

The tribomechanical properties of thermally induced oxide layers on gamma-TiAl (Ti-48Al-2Nb-2Cr at.%) at 500 °C and 800 °C were studied and compared to two popular biomaterials (Ti-6Al-4V and CP-Ti) under the experimental conditions described above. Wear tests simulating the bone-implant interface by lubricating with Ringer's solution were performed and examined. Similar studies were carried out in a dry environment. The following conclusions are drawn from the results of this study:

1. The highest COF values were observed for the oxidized gamma-TiAl surfaces, whereas the COF values were lower for the Ti-alloy surfaces which were not oxidized.

2. Abrasion was the predominant wear mechanism on all the three Ti-base alloys in both dry and lubricated media, although in the non-lubricated cases adhesion was also observed.
3. The bone pins wear down relatively faster compared to the metal oxide during testing and adhesion wear resulted from the bone debris embedded in the oxide surface.
4. Material removal was greater in the lubricated medium although COF values were lower.
5. Generally, the gamma-TiAl samples exhibited the least amount of mass removed after the wear tests in both dry and lubricated mediums and it appears that the oxides formed on gamma-TiAl possess high wear resistance in a simulated bone-implant interface environment and could potentially serve as an orthopedic biomaterial.

Acknowledgments

This project was supported by the grant SO6GM-08103 from the National Institutes of Health (NIH), National Institute of General Medical Sciences (NIGMS)/ MBRS-SCORE, University of Puerto Rico, Mayagüez Campus. Most of the sample characterization was carried out at HP Puerto Rico in Aguadilla, Puerto Rico.

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Figure 1.
Left- Denser extremities of femoral bone of consumable livestock. Right- Finished custom-made bone pin attached to aluminum arm connector.

Figure 2.

Images of the oxide layer formed on each of the three alloys after oxidation in air for 1 hour at 500°C and 800°C. The images were taken using a SEM at 1000×, 15kV and 350×, 10kV.

Figure 3.

Comparative plot for coefficient of friction (COF) versus number of cycles for the oxidized materials tested against bone pin at 3N load in dry medium, 5Hz frequency, 10mm displacement.

Figure 4.

(A) *Left:* Micrograph (10×) of the wear track in gamma-TiAl (500°C). *Right:* Micrograph (5×) of the wear track in gamma-TiAl (800°C). (B) *Left:* Micrograph (5×) of the wear track in Ti-6Al-4V (500°C). *Right:* Micrograph (5×) of the wear track in Ti-6Al-4V (800°C). (C) *Left:* Micrograph (5×) of the wear track in CP-Ti (500°C). *Right:* Micrograph (5×) of the wear track in CP-Ti (800°C). All samples were tested against a bone pin at 3N load in a dry medium for 500 cycles, 5Hz frequency, 10 mm displacement.

Figure 5.

Comparative plot for coefficient of friction (COF) versus number of cycles of Ti-base materials studied against a bone pin at 3N load in dry medium, 5Hz frequency, 10mm displacement.

Figure 6.

(A) *Left:* Micrograph (5×) of the wear track in non-oxidized gamma-TiAl. *Right:* Micrograph (20×) of the wear track in gamma-TiAl tested against bone pin. (B): *Left:* Micrograph (5×) of the wear track in non-oxidized Ti-6Al-4V. *Right:* Micrograph (20×) of the wear track in Ti-6Al-4V tested against bone pin. (C) *Left:* Micrograph (5×) of the wear track in non-oxidized CP-Ti. *Right:* Micrograph (20×) of the wear track in CP-Ti tested against bone pin. All samples were tested against a bone pin at 3N load in a dry medium for 500 cycles, 5Hz frequency, 10 mm displacement.

Figure 7.

Comparative plot for coefficient of friction (COF) versus number of cycles of implant materials tested against a bone pin at 3N load in lubricated medium, 5Hz frequency, 10mm displacement.

Figure 8.

(A) *Left:* Micrograph (20×) of the wear track in gamma-TiAl (500°C). *Right:* Micrograph (10×) of the wear track in gamma-TiAl (800°C). (B) *Left:* Micrograph (5×) of the wear track in Ti-6Al-4V (500°C). *Right:* Micrograph (5×) of the wear track in Ti-6Al-4V (800°C). (C) *Left:* Micrograph (5×) of the wear track in CP-Ti (500°C). *Right:* Micrograph (10×) of the wear track in CP-Ti (800°C). All samples were tested against a bone pin at 3N load in a lubricated medium for 500 cycles, 5Hz frequency, 10 mm displacement.

Figure 9.

Comparative plot for coefficient of friction (COF) versus number of cycles of Ti-base materials studied against bone pin at 3N load in lubricated medium, 5Hz frequency, 10mm displacement.

Figure 10.

(A) *Left:* Micrograph (5×) of the wear track in non-oxidized gamma-TiAl. *Right:* Micrograph (20×) of the wear track in gamma-TiAl. (B) *Left:* Micrograph (5×) of the wear track in non-oxidized Ti-6Al-4V. *Right:* Micrograph (20×) of the wear track in Ti-6Al-4V tested against bone pin at 3N load in a lubricated medium for 500 cycles, 5Hz frequency, and 10mm displacement. (C) *Left:* Micrograph (5×) of the wear track in non-oxidized CP-Ti. *Right:* Micrograph (20×) of the wear track in CP-Ti. All samples were tested against bone pin at 3N load in a lubricated medium for 500 cycles, 5Hz frequency, and 10mm displacement.

Table 1

Mean values of mass gained of all three Ti-base alloys after oxidation.

Material	Mean Mass Gained (g)	
	Oxidized at 500°C	Oxidized at 800°C
Gamma-TiAl	0.04239	0.0101
Ti-6Al-4V	0.01324	0.02116
CP-Ti	0.00193	0.00983

Table 2

Composition of the gamma-TiAl oxide layer, for oxidation in air at 500°C and 800°C obtained using EDAX analysis. Note that Cu appears as impurity from brass wires used in EDM process.

Element	500°C		800°C	
	Wt. %	At. %	Wt. %	At. %
O	52.66	71.79	45.57	67.84
Al	18.86	15.25	19.04	16.44
Nb	----	----	1.95	0.49
Ti	28.48	12.97	27.11	13.19
Cr	----	----	0.92	0.41
Cu	----	----	4.42	1.62

Table 3

Mean values of mass loss after wear tests for all the oxidized Ti-base alloys under dry and lubricated fretting against bone pin.

Material	Mean Mass Loss (g)			
	500°C		800°C	
	Dry	Lubricated	Dry	Lubricated
Gamma-TiAl	0.01255	0.00744	0.01222	0.0071
Ti-6Al-4V	0.0131	0.00901	0.02669	0.00979
CP-Ti	0.01391	0.00869	0.01432	0.00924

Table 4

Mean values of mass loss after wear tests for all the non-oxidized Ti-base alloys under dry and lubricated fretting against bone pin.

Materials	Mean Mass Loss (g)	
	Dry	Lubricated
Gamma-TiAl	0.00763	0.00700
Ti-6Al-4V	0.00824	0.00768
CP-Ti	0.00861	0.00880