



Tribo-electrochemical characterization of metallic biomaterials for total joint replacement

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ABSTRACT

Knee and hip joint replacement implants involve a sliding contact between the femoral component and the tibial or acetabular component immersed in body fluids, thus making the metallic parts susceptible to tribocorrosion. Micro-motions occur at points of fixation leading to debris and ion release by fretting corrosion. β -Titanium alloys are potential biomaterials for joint prostheses due to their biocompatibility and compatibility with the mechanical properties of bone. The biotribocorrosion behavior of Ti–29Nb–13Ta–4.6Zr was studied in Hank's balanced salt solution at open circuit potential and at an applied potential in the passive region. Reciprocating sliding tribocorrosion tests were carried out against technical grade ultra high molecular weight polyethylene, while fretting corrosion tests were carried out against alumina. The wear of the alloy is insignificant when sliding against polyethylene. However, depassivation does take place, but the tested alloy showed an ability to recover its passive state during sliding. The abrasivity of the alloy depends on the electrochemical conditions of the contact, while the wear of polyethylene proceeds through third body formation and material transfer. Under fretting corrosion conditions recovery of the passive state was also achieved. In a fretting contact wear of the alloy proceeds through plastic deformation of the bulk material and wear resistance depends on the electrochemical conditions.

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1. Introduction

Human joints operate by low-friction articular cartilage bearing surfaces, which are conforming and self-regenerating [1–3]. When natural joints are severely damaged, e.g. due to osteoarthritis, they are often replaced by artificial implants. In total joint replacement the implant components are generally made of metal–metal, metal–polymer, ceramic–ceramic or ceramic–polymer couples. Metal on polyethylene is a very common material coupling in total joint replacement [4].

Knee and hip replacement joints involve a sliding contact at the articulation between the femoral component and the tibial or acetabular component during motion of the human body [5–7]. As a result the metallic components of the artificial joint are susceptible to sliding tribocorrosion (see Fig. 1). Tribocorrosion is the irreversible transformation of a material due to the simultaneous action of corrosion and wear taking place in a sliding tribological contact. It involves numerous synergy effects between mechanical and electrochemical phenomena, usually leading to an acceleration of material loss [8–10].

Fretting corrosion is a particular form of tribocorrosion involving a small amplitude relative displacement or vibration, usually between surfaces that are meant to be fixed to each other [11]. In the particular case of orthopedic implants micro-motions are known to occur at points of fixation [12], while corrosion is caused by the body fluids, which contain various inorganic and organic ions and molecules [13] (see Fig. 1). Fretting corrosion has been identified at the stem/neck and neck/head contacts of modular implants, at the stem/bone and stem cement interfaces of cemented and uncemented implants, and at the screw/plate junction of fixation plates [14]. In contrast to sliding, during fretting a considerable part of the displacement may be accommodated in the contact by elastic deformation and thus the elastic properties of biomaterials can affect the implants behavior and functionality. Furthermore, due to the closed geometry of the contact electrolyte replenishment is difficult and the electrochemical conditions might differ from those of the bulk. Furthermore, debris is easily trapped and thus the behavior of the third body is critical [15].

The commonly used metallic biomaterials for orthopedic applications (stainless steel, titanium and CoCrMo alloys) owe their high corrosion resistance to the spontaneous formation of a passive surface oxide layer which forms the interface between the alloy and the environment [16,17]. The properties of surface films dictate the results of chemical and mechanical interactions at the

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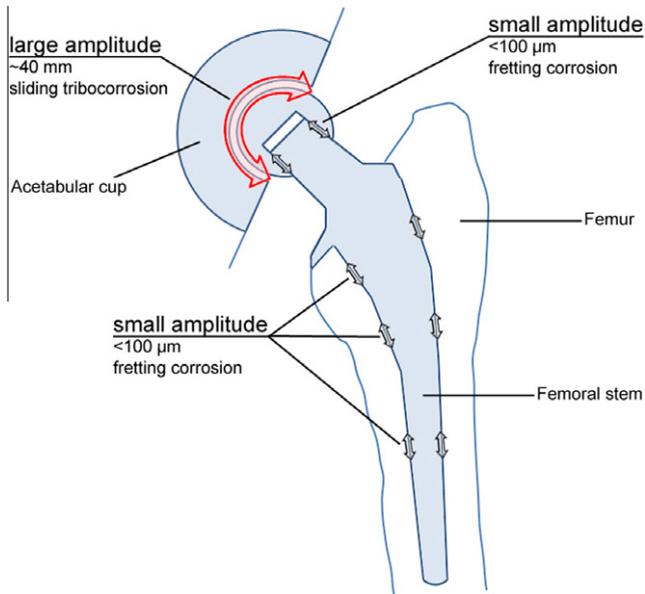


Fig. 1. A schematical representation of a total hip joint replacement prosthesis. The types of motion and surface degradation mechanisms of the implant metallic components are shown.

interface. Both sliding tribocorrosion and fretting corrosion of passive metals lead to local damage or removal of the passive film, as well as detachment of metal particles, leading to mechanical wear. As the counterbody moves on the depassivated surface area re-oxidizes, a process involving a charge transfer reaction at the interface which yields dissolved metal ions and a solid oxide. As a result in both tribocorrosion situations passive film degradation induces wear-accelerated corrosion. The repeated removal of oxide films produces particles and ions, which can result in adverse biological reactions and can lead to mechanical failure of the device [18–21]. Therefore, an approach combining both electrochemistry and tribology is a necessary means to study these complex phenomena and to assess the biocompatibility of candidate metallic materials.

In recent years different electrochemical techniques have been combined with tribology [22] and methodologies have been developed [23] that allow the study of tribo-electrochemical systems. Two types of electrochemical tests are commonly combined with tribology: chronopotentiometry and potentiostatic polarization. During chronopotentiometric measurements the evolution of the open circuit potential (E_{oc}) of the sample is monitored before, during and after a mechanical perturbation. The open circuit

potential is the difference in electrical potential spontaneously established between the metal and the solution. Upon disruption of the passive surface film the underlying metal is uncovered and the E_{oc} drops to more cathodic values, indicating the increased oxidation tendency of the exposed metal. In such a case the measured E_{oc} is a mixed potential resulting from galvanic coupling between passive and depassivated areas in the sample surface [24]. On the other hand, during potentiostatic polarization tests one can control the potential and thus study the response of the material under different oxidizing conditions, such as in the case of local tissue inflammation when oxidizing agents will be present in the environment. The potential of the sample is externally imposed and the evolution of the current is monitored during sliding or fretting. In this case disruption of the passive surface films leads to an increase in the measured anodic current originating from the accelerated oxidation reactions taking place at the uncovered metal surface [25].

The scope of this work is to evaluate the combination of electrochemistry and tribology in the study of the wear behavior of biomedical alloys. For this, laboratory set-ups are used to test a newly developed β -titanium alloy in different tribological contacts in physiological solution. The selected counterparts were UHMWPE and alumina in order to simulate the ball/acetabular cup sliding contact and the femoral stem/ball fretting contact, respectively.

2. Materials and methods

2.1. Characteristics of the alloy

In this work a titanium alloy with β microstructure (see Fig. 2), Ti–29Nb–13Ta–4.6Zr, was studied [26,27]. β -Alloys exhibit a modulus of elasticity considerably lower than commonly used Ti–6Al–4V and Co–28Cr–6Mo alloys and thus decrease the risk of stress shielding of the femur [28]. The added benefit of using Nb, Zr and Ta as alloying elements is that they tend to form dense surface oxides, increasing the stability of the passive layer [29].

The alloy was prepared by melting in a vacuum arc electric furnace (non-consumable) and casting in the form of pancakes, followed by forging. A vacuum arc electric furnace in which the charge can be melted in a water cooled copper crucible under vacuum (10^{-3} mbar) was employed. Prior to melting the furnace chamber was purged with argon gas once and then again filled with the gas up to 532 mbar. The charge was melted by arcing emitted from a tungsten bit brazed to a copper stringer rod suspended above the charge. A d.c. potential of up to 30–32 V and current of 1000 A was applied between the tungsten cathode and charge material, which formed the anode. A stirring coil around

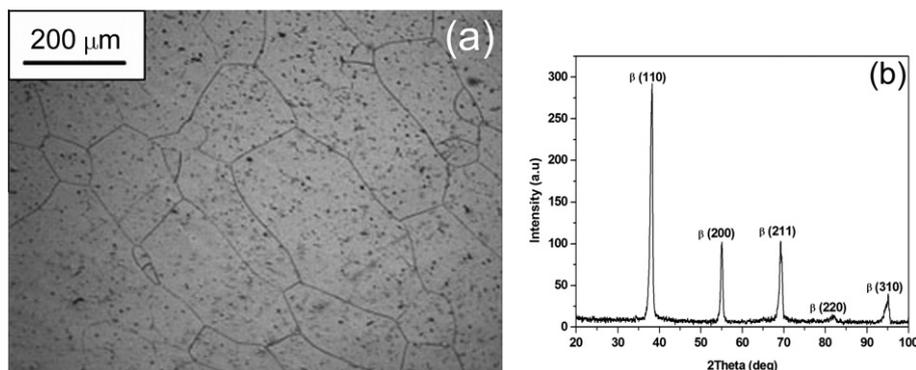


Fig. 2. (a) Optical micrograph and (b) X-ray diffraction spectrum of the Ti–29Nb–13Ta–4.6Zr alloy.

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