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TRIBOLOGICAL IMPLICATIONS OF THP MODULARIZATION

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Abstract: *The aim of the paper is to highlight the advantages of the modularization of the femoral component of the Total Hip Prosthesis (THP), especially the tribological implications it presumes (fretting wear of the modular junctions, fatigue and corrosion by fretting, which can contribute to the increase of metal ions concentrations in the blood and the possibility of the femoral stem fracture at the femoral head junction). Experimental laboratory studies focused on the analysis of the tribological phenomena occurring at the conical junctions between the femoral head and neck, as well as the neck and femoral stem, highlighting the relevant qualitative tribological aspects. Femoral stems with two junctions allow even greater capacity for the independent fit of the proximal femoral, offset and limb length with the metaphysical dimension/ stem body. For this purpose, experimental junctions, tested on a universal MTS testing machine, were built. It emphasizes the importance of the materials used for the conical or trapezoidal joints components and the measures to reduce the micro-motion that lead to fretting phenomena and fretting corrosion. The influence of conicity on the tribological phenomena is analysed.*

Keywords: *THP, modular junctions, micro-motion, fretting wear, fretting corrosion, taper damage.*

1. INTRODUCTION

As H.J. Cooper et al. stated, [1], modularization of total hip prostheses (THP) started from the need to simplify the implants subsequent revision by changing only the femoral head. Also, the head modularity allows legs length adjustment, and also allows the use of other materials (ceramics) as bearing option. If the stem must be kept under revision, exposure can be improved by removing the head, which also offers the opportunity to apply a new head.

A. Srinivasan et al., [2], specified that this could be beneficial in terms of subsequent bearing wear. But, an increase in the use of modular interfaces can have negative effects. This because it can lead to an increase of the fretting corrosion and corrosion cracking at the conical junction. Corrosion products of taper junctions may contribute to the joint wear with the third body. Although the corrosion of taper is relatively rare in hips with metal on polyethylene (MoP) joints, corrosion products can lead to adverse local tissues reactions (ALTR). The authors defined the taper between the femoral head hole and the femoral stem trunnion as the "head-neck junction". This term will also be used by the authors of the present paper in their research.

A complete review of the femoral modularization, theoretical reason for modularity, and explored clinical results was published by H. Krishnan et al. [3]. Clinically relevant issues reported using modular neck femoral stems were also examined in this research. Beside this, the failure mechanism is important to be known to determine whether modular neck femoral stems will be used in the future and how patients who already have implants should be monitored.

S. Hussenbocus et al., [4], specified that "THP corrosion reduction seems to be related to geometric parameters, materials combinations and femoral head size". They described the pathogenesis, risk factors, clinical evaluation and corrosion management of taper from the head-neck junction. Under this aim, failed neck adapters were implanted during almost 2 years in a total of about 5000 devices. After this period, titanium neck adapters were replaced by cobalt-chromium adapters.

T.M. Grupp et al., [5], showed that the primary micro-motions initiated fretting within the modular connection of the femoral taper neck. A continuous abrasion and repassing process was carried out with a subsequent cold welding at the titanium alloy modular interface. Titanium oxide layers of 10-30 µm were observed on the surface. Surface cracks caused by fretting or fretting corrosion finally lead to fatigue fracture of titanium alloy modular neck adapters. Using a cobalt-chromium neck, micro-motions can be reduced three times, especially in the case of contaminated taper connection. The incidence of fretting corrosion was also substantially lower in the case of cobalt-chromium neck.

J.R. Goldberg et al., [6], investigated the effects of various factors (materials combination, metallurgical condition, bending stiffness, head and neck moment arm, neck length and implantation time) on corrosion and fretting of modular taper surfaces. The obtained results suggest that the in vivo corrosion of hip modular taper interfaces is attributed to a mechanically assisted corrosion process. The corrosion process that appears as a result of fretting can be reduced by using higher diameter necks that increase their rigidity. However, the increase of the neck diameter must be balanced, taking into account the decrease in the range of motion and the resulting stability of the joint.

S.Y. Jauch et al., [7], studied modular prosthesis neck adapter failures with the aim to investigate the influence of materials combinations and of assembly conditions on the micro-motions size at the stem-neck interface during cyclic loading. The largest observed micro-motions were located at the lateral edge of the taper neck connection, which is consistent with the cracks' location of the clinically failed prostheses. Higher micro-motions were observed in the case of titanium neck adapters and contaminated interfaces compared to cobalt-chromium neck adapters. Based on the fact that the excessive micro-motions from the stem-neck interface could be involved in the implant failure process, the main conclusion of their studies was that particular attention should be paid to the cleaning of the interface before assembling. The same attention should be paid, when adapters with titanium neck and titanium stems are used.

J.L. Gilbert et al., [8], tested hip femoral stems made of stainless steel (ASTM F-1568) combined with CoCr alloy heads (SS/CoCr), in an in vitro corrosion test, to evaluate the tendency to mechanically assisted corrosion. Three different aspects of the modular design were evaluated: (1) comparison of CoCr/CoCr materials combinations, (2) wet/ dry assembly for SS/CoCr couplings and (3) the 0, and 6 mm offset of the head for the SS/CoCr couplings. Fretting corrosion tests were performed by a range of cyclic loads up to 3300 N and continuous cyclic loading at 3300 N for 1 M cycles. The results of these studies showed that SS/CoCr couplings were more prone to the fretting corrosion than the CoCr/CoCr couplings. Dry assembly increased the debut loading, but did not prevent the fretting corrosion. 6 mm offset heads had greater visual evidences of fretting damage. Also, "micro-motion measurements indicated fretting motions in the range of 10 – 25 µm, where the 0 mm offset heads tended to piston on the stem trunnion, while the 6 mm offset heads tend to break".

S.I. Jauch et al., [9], had mechanically tested and investigated micro-motions at the stem-neck interface of two different prostheses models: Metha (Aesculap AG) and H-Max M (Limacorporate). Metha prostheses demonstrated a substantial number of in vivo fractures for Ti-Ti couplings, but there are no fractures documented for Ti-CoCr couplings. In contrast, for H-Max M prostheses with a Ti-Ti coupling, only a clinical failure was reported. The main results showed that "for Ti-Ti couplings, Metha prosthesis showed a tendency towards higher micro-motions compared to H-Max M (6.5 ± 1.6 µm vs. 3.6 ± 1.5 µm). Independent of design, the prostheses with Ti neck adapter have caused significantly higher micro-motions at the interface, than those with CoCr adapter (5.1 ± 2.1 µm vs. 0.8 ± 1.6 µm). No differences were observed in micro-motions between the Metha prosthesis with CoCr neck and H-Max M with Ti neck (2.6 ± 2.0 µm)".

Results of the experimental tests realized by P. Wodecki et al., [10], have shown that THP with modular femoral components (stem-neck interface) makes it possible to adapt to extramedullary femoral parameters (anteversion, offset and length). In this way, muscular function and stability are improved. However, "the addition of a new interface has some disadvantages, like: reduced mechanical strength, fretting corrosion and fatigue fracture of the material".

R.T. Mikkelsen et al., [11], conducted an analysis of modular neck femoral stems, compared to the non-modular femoral stems in total hip arthroplasty (THA), with regard to the clinical outcome of metallic ions levels and radiological findings. They showed that the modular neck femoral stem was introduced to optimize the outcome of THA, but this created concerns about pain, high levels of metallic ions in blood, and adverse reactions to metal debris, such as pseudotumor, linked to the corrosion between the femoral neck and stem.

A.M. Kop and E. Swarts, [12], analysed sixteen cases of double taper cones of recovered Margron hip prostheses. These exhibited a significant fretting and a corrosive cracking of the neck-stem taper with an average duration of 39 months after implantation. The remaining recoveries showed no corrosion after an average time of 2.7 months in situ. The final results demonstrated that the increased modularity can lead to fretting and corrosive cracking, generation of metallic ions and particles debris even in the case of a modern conical design and corrosion-resistant materials. All these, fretting and corrosive cracking, generation of metallic ions and particles debris, can contribute to periprosthetic osteolysis and loss of fixation.

M.B. Ellman et al., [13], have shown that a possible complication of modularity increase is the components fracture. This has been demonstrated in the case of fracture of the modular femoral neck. "The combined effects of cracking and fretting corrosion of the large diameter femoral head, long metal-to-metal modular neck, patient size and activity level, have all played integral roles in creating a susceptible environment for this classic fatigue fracture model".

H.H. Ding et al., [14], investigated the influences of diamond-like carbon (DLC) coatings and roughness on the fretting behaviour of Ti6Al4V. It has been shown that "without DLC coating, the friction coefficient was high, and under high motion conditions, the wear volume was high. Smoother surfaces have extended the sliding and rough regime to lower motion conditions under normal force conditions. For DLC coating tests, the coating response wear maps were divided into three areas: the coating working area (low motion conditions and low normal force), coating failure area (high conditions of motion and normal force) and the transition area. In the coating working area, DLC coatings could protect the substrate with low friction, low wear volume and slight damage of the coating. The state of operation has occurred under the gross sliding regime". The increase in normal force and motion accelerated the failure of the coating.

N.J. Hallab et al., [15], studied the fretting corrosion differences of the metal-metal and ceramic-metal modular junctions of total hip replacements (THR). Ceramics femoral head (zirconia, ZrO_2) and metal (Co alloy) on Co alloy stem components were investigated using an in vitro comparison of fretting. In vitro fretting corrosion testing consisted in the monitoring and potentiodynamic analysis of the metal loss in 28 mm zirconia and Co alloy femoral heads with similar surface roughness (R_a = 0.46 µm), on identical Co alloy stems, at 2.2 kN for 1 x 10^6 cycles, at 2 Hz. A metal release about 11 times larger at Co and a 3-times increase at Cr and the potentiodynamic fretting of metal-metal modular junctions, compared to ceramics-metal junctions was observed.

T. McTighe et. al., [16], published a review of the risk factors and benefits of modular taper junctions in THA. Fretting corrosion is one of the factors that produce the decline in clinical acceptance of hip modular implants. A main mechanism behind the fretting corrosion is the stress, whose increasing at the modular junction will increase proportionally the fretting corrosion. Stryker Ortopedics, Mahwah, NJ withdrawaled products (e.g. Rejuvanate™ and ABGII™) that had reduced tapered support (13 mm vs. 15 mm and 17 mm) with high bending and torsional moments. These produce much greater stresses at the modular junction and potentially lead to a faster corrosion speed, compared to the style of stems that keep the neck. Conical adapters for neck can have design limitations in that they have sockets that can interfere with the range of motion or can cause pressure, generating debris and/ or dislocations.

In another work [17], T. McTighe et al., reviewed the developments of the short femoral stems, which offer many advantages. First, with a few short stems' patterns, most of the femoral neck is kept. Surgically, this facilitates a minimally invasive surgical approach and attenuates soft and bone tissues damage. Femoral neck preservation, which provides a more natural barrier to the particles' debris migration, is associated with lower blood losses and less time and energy for hip rehabilitation, reducing the stress shielding of proximal femur (load's redistribution and subsequent loss of proximal femoral bone mass) and reducing the pain at the end of the coast. Considering all these advantages, the use of a short stem can make quicker and less painful the patient's rehabilitation. The new design feature inherent to short stem implants – namely, the preservation of bone and native proximal tissue – provides theoretically an easier revision if or when it becomes necessary. For these reasons, the short stem procedures also have wider indications as compared to recovery of hip surface. Finally, many models with short stems do not require many stem sizes.

R. Grunert et al., [18], have shown that modularity in THA allows the reconstruction of the hip biomechanical parameters. Starting from the observation that models of femoral stems structured using taper junctions contribute to the implant fracture, the authors assumed that a new modular neck-stem interface may result in lower implant breakage compared to conventional femoral stems. Realizing a new modular stem for THA, the authors developed three different variants of the interface mechanisms. These provide a simple connection between the stem and the modular neck, and allow an intraoperative adjustment. The authors have shown that with the new design of the three manufactured prototypes, it should be possible to detach intraoperative the modular stem and neck, to adapt to the anatomical situation. It has been also shown that modular implants have to be used with caution because of the high risk of breaking, fretting of taper, corrosion and disconnection.

In a previous work, L. Capitanu et al., [19], presented a study in which they analysed the loss of cemented stem stability of the THP under physiological stress conditions. The experimental study was conducted on a universal dynamic testing machine, MTS® Bionix. The authors reported the failure by the fretting fatigue of the cemented fixation of the stem at 2200000 variable loading cycles.

M.G. Bryant et al., [20], evaluated a bi-modular prosthesis (Rejuvenate™, Stryker, USA) with a Ti-Mo-Zn-Fe alloy femoral stem (TMZF™, Stryker, USA) and femoral modular neck made of CoCrMo (Vitallium™ Stryker, USA). Large amounts of material loss at the medial proximal edge of the CoCrMo neck trunnion were noticed. On the modular taper interface, a conserved stain was observed, surrounded by high wear areas. By CMM analyse, three areas of interest were highlighted, which were subjected to further examination. They indicated a reference area (i.e., unworn material outside the contact area), a fixed spot (i.e. the region located within the extremely worn area, where the original topography was maintained) and very worn areas (areas with the largest loss of material).

S.Y. Jauch et al., [21], investigated the magnitude of micro-motions at the stem-neck interface and behaviour during daily activities, the neck connection of a design made of different alloys. Modular hip prostheses (Metha®, Aesculap AG, Germany) with neck adapters ($CoCr_{29}Mo_6$ or Ti6Al4V) were incorporated into PMMA and subjected to cyclic loading with peak loads ranging from walking (F_{max} = 2.3 kN) to stumbling (F_{max} = 5.3 kN). The micro-translation and rotation motions from the taper interface and the layout characteristics during assembling and loading were determined using four eddy current sensors. Studies have shown that placement during loading after assembly of the implant was dependent on the size of the load, but not on the coupled material.

Characterization of fretting corrosion behaviour of the surface and debris from the head-taper interface, of two different hip models has been realized by C.T. dos Santos et. al., [22]. The first one was a SS/Ti cementless model with the stem made of ASTM F136 Ti6Al4V alloy and metallic head made of ASTM F138 austenitic stainless steel. The second one was a SS/SS cemented model with both components made of ASTMF 138 stainless steel. The results obtained after fretting corrosion tests according to ASTM F1875 standard criteria, showed that micro-motions caused mechanical wear and loss of material in the head-taper interface, resulting in fretting corrosion. After 10 million cycles, it has been demonstrated that the SS/Ti model was more resistant to fretting corrosion than the SS/SS model. But, in both cases there were various morphologies of residues. Small and crowded particles were observed in the SS/Ti model, and in the SS/SS model, irregular particles. All these released particles can cause local tissue reactions in the human body and the loss of THP stability.

Fretting of CoCrMo and Ti6Al4V alloys in modular prostheses was analysed by A.O. Oladokun et al., [23]. Fretting behaviour of CoCr-CoCr and CoCr-Ti couplings and their damage mechanisms were investigated. A tribometer with ball on plate contact with in situ electrochemistry was used to characterize the damage caused by tribocorrosion on the contact of the two couplings. The amplitudes of fretting motions of 10, 25 and 50 μm were evaluated at an initial contact pressure of 1 GPa. The results reveal a greater loss of metal volume in CoCr-CoCr alloys couplings compared to CoCr-Ti alloys, and the open-loop potential indicates a depassivation of the protective oxide layer at displacement amplitudes > 25 μm. The damage mechanisms of the CoCr-CoCr and CoCr-Ti contacts have been identified as the prevalent mechanisms of wear and fatigue.

2. EXPERIMENTAL STUDY

2.1 Theoretical approach

Currently, in the speciality literature, there is no theoretical approach of the fretting wear speed. The wear speeds are quantified applying the Archard's classic approach. It reports the wear volume as the product between the sliding distance and the normal load. A wear coefficient is then extrapolated and it is assumed that it determines the wear resistance of the studied material. This approach does not work when the friction coefficient is not constant. Consequently, it seems more relevant to consider the mechanical work of interfacial shearing as significant parameter of wear.

By identifying the wear energy coefficients, quantification of wear can be rationalized and the wear resistance of studied tribosystems can be quantified.

This seems to be a convenient approach to interpret different wear mechanisms. The energy balance confirms that a small part of the dissipated energy is consumed through plasticity, while most of it participates in the heat flow and debris through the interface. When introducing a load energy approach, an accumulated density of dissipated variable energy is considered to quantify the formation of the transformed tribological structure (TTS).

The described methodology allows the implementation of the "Archard" or "dissipated energy" law to predict the fretting wear. However, the wear energy approach is presented here as a unified prediction of a single wear energy coefficient in a wider range of strokes (from 50 mm to 1.3 mm) than Archard's law and as such has no a wide range of applications (S Fouvry et al., 2003 [24], R. Magaziner et al., 2008 [25], T. Liskiewicz and S. Fouvry, 2005 [26]).

Calculation of the volumetric wear is based on the wear energy law in Eq. (1), where the mechanical interfacial shearing work is the predominant parameter for determining the wear. Total volumetric wear W_v is obtained from product of the total accumulated local energy E that is dissipated and a worn energy coefficient α

$$W_v = \alpha \cdot E \tag{1}$$

where

$$E = Q \cdot s \tag{2}$$

and Q is the shear traction, and s is the relative displacement of the contact surfaces, giving

$$W_v = \alpha \cdot Q \cdot s \tag{3}$$

Dividing both members of Eq. (3) to the contact area, the depth of linear wear W_d can be calculated using Eq. (4), where τ is the shear stress of the contact surface

$$W_d = \alpha \cdot \tau \cdot s \tag{4}$$

For the numerical implementation of this wear law, the wear depth of in contact surfaces generated in a single loading cycle of the components (such as loading applied on hip in vivo, for a single walking step) is first determined. Subsequently, if the prostheses components will be as typically, the object of millions of loading cycles during their lifetime, this wear depth in a single cycle is multiplied by a wear coefficient, β, so that an analysis can be realized in an acceptable time period. The "wear scaling" factor represents a certain number of loading cycles, and its value depends on how accurate the wear evolution is calculated and the way it evolves over time. The in-contact surfaces geometry of the components changes after measuring the wear depth, to reflect the wear that would have occurred after a certain number of β cycles. Calculated wear can be applied to only one component or both, in equal or unequal quantities, depending on combinations of materials in contact. The process is then repeated using the updated geometry up to the number of loading cycles that have been applied, or until a predetermined wear depth has been reached.

Digitization of a loading cycle in several time intervals n is necessary to accurately model the effect on the wear of the variable load distribution over time, during the loading cycle (as it appears during walking). As such, the wear depth for a single loading cycle (the depth of cyclic wear W_c) can be calculated using Eq. (5),

$$W_c = \sum_{i=1}^{n} \alpha \cdot \tau_i \cdot s_i \tag{5}$$

where τ_i and s_i are the shearing stress of the surface and the relative displacement, calculated at the end of a certain time interval, i.

The total wear depth W_d generated in a specified total number of loading cycles N can be determined from Eq. (6) where j is the specific "step of analysis" which reflects the evolution of wear.

$$W_c = \sum_{j=1}^{(N/\beta)} \beta \sum_{i=1}^{n} \alpha \cdot \tau_i \cdot s_i \tag{6}$$

The accuracy and effectiveness of this approach depend on many factors and on the magnitude of the wear energy coefficient that has been used. The number of time intervals used for loading cycle discretization and the "wear coefficient" factor β requires a careful analysis of their influence on the accuracy and duration of the analysis.

2.2 Experimental approach

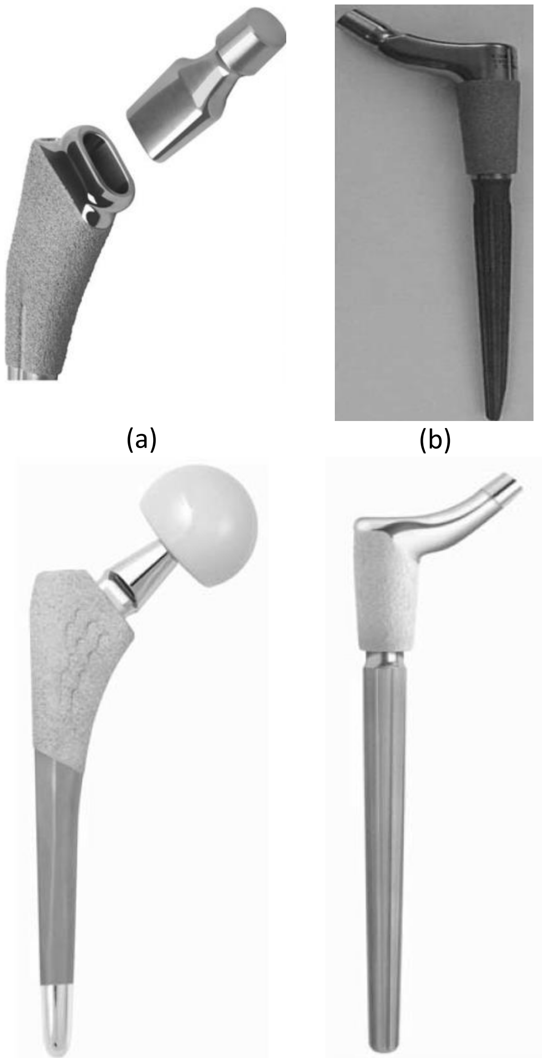
The fretting tests were performed on an Instron® device (Fig. 1) which simulates fatigue loading of a hip cemented stem during a walking cycle. It is installed on a MTS® Bionix multi-axial dynamic servo-hydraulic testing machine [19], equipped with a hip implants testing system.

Figure 1. Instron® fatigue testing device of total hip prosthesis

The device is mounted in a sealed chamber containing saline solution with a high concentration of sodium chloride (~ 0.2 M) at a temperature of 38 ^{0}C. The assembly has a temperature regulator and a circulation pump for in vivo testing. Flexible holder for the prosthetic stem allows testing of a large variety of hip prostheses stem geometries, offset angles, and materials. The device applies compressive, bending and torsional stresses to meet ISO 7206-4 requirements.

Tests were performed on THP made by assembling femoral stems with two junctions, with femoral head-neck junction combinations.

Femoral stems with two-junctions (Fig. 2) allow a greater independent fitting capacity of the proximal femoral version, of the offset, and limbs length with the metaphyseal dimension/ stem body.



(a) (b)

(c) (d)

Figure 2. Dual modular femoral stems. (a) Kinectiv and (b) ZMR - Zimmer, Warsaw, IN, USA (c) Rejuvenate and (d) Restoration - Stryker, Mahwah, NJ, USA

In order to study the fretting of the THP's modular junctions, taper and trapezoidal junctions of a femoral neck adapter, made of CoCr and Ti6Al4V (Fig. 3), were realized to match the stems with which current modular prostheses were equipped. They were subjected to dynamic fatigue tests, up to 2500000 cycles on the Instron® device.

The neck adapters were used in combination with femoral heads made of CoCr, Ti6Al4V and ceramics (Fig. 4).

To analyse how fretting corrosion is influenced by combinations of coupled materials, the authors have developed and analysed different couplings of the Kinectiv femoral stems with different neck junctions with the femoral heads. Combining the neck adapters with the three types of femoral heads, resulted six femoral head-femoral neck combinations: Ti6Al4V/Ti6Al4V, Ti6Al4V/CoCr, Ti6Al4V/ceramics, CoCr/Ti6Al4V, CoCr/CoCr, and CoCr/ceramics. Three of the six resulting combinations are shown in Fig. 5.

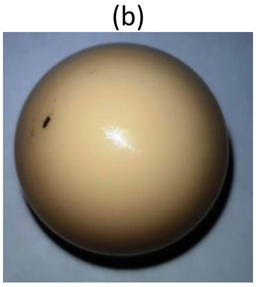
(a)

(b)

Figure 3. Taper and trapezoidal junctions of a femoral neck adapter made of CoCr (a) and Ti6Al4V (b)

(a)

(b)



(c)

Figure 4. Femoral heads made of CoCr (a), Ti6Al4V (b) and ceramics (c), used

Figure 5. Femoral head – femoral neck combinations that have been tested

As shown in a previous work [19], the relatively large difference between the maximum and the minimum friction torque value, and the visual observation of the prosthesis's functioning through the transparent fastening holder, appeared to be a sign of the fretting wear manifestation. Testing was stopped when this difference occurred, at 2450000 cycles, and the prosthesis was removed from the Instron® device and qualitatively analysed. Tests were performed during this number of 2450000 cycles on the Instron® device with variable loading application of the normal load and of the flexion-extension (FE), abduction-adduction (AA) and external-internal rotation (IOR) motions.

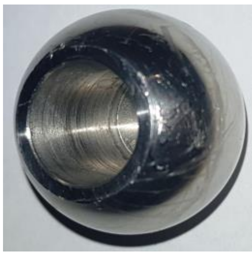
3. RESULTS AND FINDINGS

After the tests, the resulting surfaces of the femoral head – femoral neck and femoral stem – femoral neck taper junctions were inspected visually and microscopically. All of these surfaces have shown, to a greater or lesser degree, evidences of fretting wear and fretting corrosion.

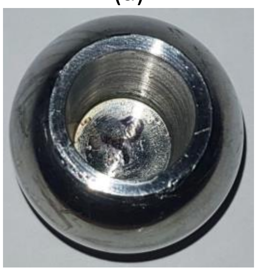
Fretting traces were revealed both on the taper and trapezoidal junctions of the femoral neck adapters, as well as inside the conical holes of the femoral heads.

Figs. 6 and 7 show the fretting and corrosion wear images on the surfaces of these junctions.

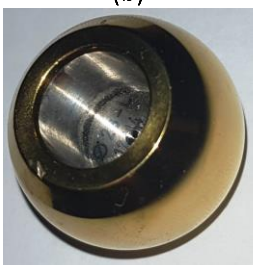
Figure 6. Taper and trapezoidal junctions of femoral neck adapters in CoCr (a) and Ti6Al4V (b) after 2450000 fretting cycles



(a)



(b)



(c)

Figure 7. The fretting and corrosion wear of the inner taper of the femoral heads made of CoCr (a), Ti6Al4V (b) and ceramics (c) after 2450000 fretting cycles

4. DISCUSSION

Mechanistic model of tribocorrosion is one that takes into account all possible mechanical and chemical interactions, as follows:

$$W_T = W_m + \Delta W_{cw} + W_c + \Delta W_{wc} \qquad (7)$$

where W_T is total tribocorrosive wear, W_m and W_c are the material losses due to pure mechanical wear and pure corrosive wear. The terms ΔW_{cw} and $\Delta W_{wc,}$ respectively, represent the wear enhanced by corrosion and corrosion enhanced by wear, respectively.

4.1 Mechanical wear model

The current mechanical wear model uses a local form of Archard's equation to calculate the wear depth. In a ball on plane configuration, the local wear depth of each point on the surface is given by:

$$\Delta h(x, y) = W\frac{K}{R} \cdot P(x, y)\Delta t \cdot v \qquad (8)$$

where H, K, P, v and Δt are material hardness, Archard's adimensional wear coefficient, local contact pressure, sliding speed and time step, respectively [19].

To use Archard's wear formula, the wear volume was divided to the wear track surface. Although not uniform, it provides a rough estimate of the wear depth.

Contact conditions vary from plate to ball. The balls are always in contact with the plate and move the contact’s positions on the plate. Therefore, the numerical implementation of wear on the two surfaces must be different. Wear calculated from Eq. 5, taking into account the ball hardness is used directly to modify the geometry of the ball at each stage. However, the wear on the plate is calculated at each stage of the time, using a change factor.

Since the wear of the plate does not occur all the time, the wear calculated in Eq. 8 taking into account the plate hardness is divided by the ratio of the length of the wear track to the nominal contact surface, to find the balance. These wear values will then be deducted from the surface profiles, and the surfaces will be changed at each step of the time.

In simulation of the contact’s mechanics, all parameters in Eq. 8 are calculated, except K. K is determined by calibration of the model. Depth of the mechanical wear calculated from

Eq. 5, is used to locally change the geometry of the surface, based on the asperity pressure's distribution. Total mechanical wear can be calculated by summing the mechanical wear from the loading cycles.

4.2 Corrosive wear model

Corrosive wear model is based on Faraday's law and the calculation of the volume of metal ions transferred to the surface and used to form the oxide. The formulation used for corrosive wear is as follows:

$$V_c = \frac{QM}{nF\rho} \tag{9}$$

where V_c is the volume of metal removed by the anodic reactions. Q is the total electrical charge passed and is calculated by integrating the current over time ($Q = \int_0^t idt$) when an excessive potential is being applied. M is the atomic mass of the metal, n is the charge for the oxidation reaction, ρ is the passive metal density, and F is Faraday's constant.

In an actual tribocorrosive wear environment, the anodic and cathodic currents are always equal to the free corrosion potential (E_{corr}), and the anodic current measured under this condition, will be true I_{corr} and should be used as a representation of the corrosion contribution. Applying the over-potentials on a sliding system will significantly alter the steady state so that the cathodic reaction becomes negligible, and the current measured will be only the anodic current from the working electrode. Working under such conditions is different from a real natural tribocorrosion state, which occurs around the E_{corr} of the metal alloy. Applying excessive potential can change the pH of the surface and may have an impact on corrosion. Therefore, strictly, I_{corr} used in any tribocorrosion study of metal alloys to measure the contribution of all mechanical and chemical components of tribocorrosion wear should be determined at E_{corr}. Despite this, most studies in the tribocorrosion area are applied over-potential and change surfaces from their natural free corrosion state.

An important parameter to be considered in the electrochemical wear model in Eq. 9, is the total electrical charge (Q). Calculation of Q will be possible when the current passage through the passive film is successfully captured. The electrochemical model is presented at the asperity scale, and the corresponding current density is calculated in a deterministic manner, considering the inhomogeneous nature of the surface asperities. The summation of the local current densities results in the calculation of the macro-current density, being comparable to the experimental results.

5. CONCLUSION

Failures of neck adapter at different models of modular prostheses extended research in this domain, and most scientists considered that the micro-motions at the stem-neck interface were responsible for these implant failures.

In the present paper the results of a study, in which the influence of materials combinations and assembly conditions on the micro-motions size at the stem-neck interface during cyclic loading were investigated, are presented.

The largest micro-motions were observed at the lateral edge of the taper neck connection. Larger micro-motions were shown in the case of titanium neck adapters, compared with CoCr ones and in the case of interfaces contaminated with fats or debris.

Fretting and fretting corrosion have occurred on all modular neck-stem recoveries, regardless of model. However, mixed metal couplings exhibited more corrosion than homogeneous couplings. This is due to the lower elasticity modulus of the titanium alloy used for the stem, which allows a greater metal transfer and surface damage when is loaded on a modular cobalt alloy neck.

The quantification of wear can be rationalized, and the wear resistance of the studied tribosystems can be quantified by identifying the wear energetical coefficients. In this way it is possible to interpret the various wear mechanisms.

The main results obtained after the studies realized in this project confirmed that different materials can be used to optimize the mechanical and tribological properties of hip modular prostheses. Even if this type of prostheses is flexible to fit to the anatomical variations, the micro-motions associated with the modular components can lead to fretting corrosion. Finally, the release of debris is produced and can cause negative local tissue reactions from the human body and the loss of THP.

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