Published in final edited form as:

J Arthroplasty. 2018 September ; 33(9): 2707–2711. doi:10.1016/j.arth.2018.03.055.

What Factors Drive Taper Corrosion?

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Abstract

Adverse local tissue reactions to corrosion products can lead to total hip replacement (THR) failure. Although this problem has been well known for more than 25 years, it has seemingly increased in frequency over recent years. The occurrence of corrosion is multifactorial—depending on implant, patient and surgeon factors. As of now, there is no “one-size-fits-all” solution to prevent corrosion in THR devices. Thus, it is imperative to fully understand the exact mechanisms of modular junction corrosion to prevent premature implant failure. This review highlights a few key concepts that need to be explored to minimize the impact of corrosion. Key concepts include 1) the prevention of micromotion, 2) the role of implant alloy metallurgy in the corrosion process, 3) the *in vivo* generation of a corrosive environment, and 4) potential unanticipated problems.

Introduction

The generation of tribocorrosion products within modular junctions is an established cause of failure in total hip arthroplasty (THA). Tribocorrosion debris from cobalt alloy components has been shown to cause symptomatic adverse local tissue reactions (ALTR) in up to 4.9% of patients with primary total hip replacements^{1–6}. This prevalence figure has evolved as more surgeons have become cognizant of the problem of corrosion and fretting processes and the resulting tissue response. It is possible that the occurrence of ALTRs is often misdiagnosed as it could mask as infection, instability or inflammation⁷.

The recent orthopedic literature is rich with articles relating implant failure to mechanically assisted crevice corrosion (MACC)^{8–10}. MACC is the overarching term describing corrosion processes within the confined environment between two corresponding taper surfaces¹¹. MACC also includes the most frequently described damage process which is fretting corrosion. There is an overall consensus that the problem of MACC, fretting corrosion and the potential subsequent tissue response is multifactorial^{7,12}. Roughly, the main factors can be separated into implant-, surgical- and patient-related factors. Current concepts have been recently summarized by us⁷ and by others^{12,13}. A pervasive clinical trend involves

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abandoning cobalt alloy heads in favor of ceramic heads and abandoning cobalt alloy stems in favor of Ti-alloy stems. Evidence has suggested that this combination promises a reduction of the risk of ALTR^{14,15}. However, there may be other damage modes with this combination of materials that could affect the implant longevity. Considering the large number of implant designs, patient variability, and surgical considerations, it is unlikely that there is a “one-size-fits-all” solution to prevent corrosion, wear, or other damage modes that could result in premature implant failure. It is the purpose of this review to highlight a few key concepts that need to be explored to minimize the impact of MACC including 1) the prevention of micromotion, 2) the role of implant alloy metallurgy in the corrosion process, 3) the *in vivo* generation of a corrosive environment, and 4) potential unanticipated problems.

Preventing micromotion

Fretting corrosion is characterized by oscillating micromotion between two surfaces, leading to passive film disruption, thus leaving the surface temporarily unprotected and enabling corrosion. Although fretting corrosion appears to be the most frequently reported damage process within modular junctions, there are numerous simultaneously or situationally acting damage modes. A recent study of 364 head/stem pairs from metal on polyethylene (MoP) bearings has described the large variety of underlying damage modes that can be identified from damage features on the retrieval surfaces¹⁶. Some damage modes, such as etching, pitting corrosion, and intergranular corrosion, appear to be purely electrochemically driven, whereas others are entirely mechanically dominated such as material transfer, creep, and fretting wear. Thus, taper damage is not necessarily purely corrosion driven. Yet, in the majority of cases it seems most likely that fretting corrosion, triggered by micromotion, is at least the initiator of other damage modes. For example, micromotion could lead to a widening of the crevice and subsequent fluid infiltration. The crevice environment, along with patient specific biological factors, can lead to chemical alterations that enable purely chemically driven damage modes¹⁶. Another scenario is that micromotion can lead to surface fatigue, also leading to a widening of the crevice and the transformation over time to a toggling-type motion leading to severe plastic deformation and wear¹⁷. Such cases can lead to a complete destruction of the taper as has been reported for tapers made from titanium alloys^{18,19}. Either way, it becomes clear that preventing micromotion is the best way to avoid any type of damage process within modular junctions.

Laboratory tests have taught us that high assembly loads and clean taper surfaces enable optimum stability^{20–23}. However, the surgeon cannot always ensure a consistent impaction force. The surgical approach and patient factors such as bone quality play an important role, and can therefore limit the surgeon’s ability to ensure the best seating conditions⁷. Therefore, it is important that the implant design and machining tolerances promote seating conditions that minimize the risk of micromotion. Factors such as material couple, angular mismatch, and surface topography need to be considered^{12,24–26}. For example, we recently found that the contact mechanics at the taper surfaces are highly variable depending on the geometry of the machining lines on the stem taper²⁴. High contact pressure over a larger contact area may help ensure locking and minimize micromotion but are often at odds since larger contact area tends to decrease contact pressure. Besides initial stability, the design has

to also support long term stability under cyclic loading where additional factors such as contact area, flexural rigidity and head size have been shown to be important^{12,27}. The ideal combination of all design factors needs to be further investigated, but it may be different depending on the material couple (Ti-alloy vs Co-alloy, Co-alloy vs Co-alloy, metal vs ceramic), and taper type (head/neck, neck/stem). Material couples need to be further distinguished between different types of Ti-alloy (Ti6Al4V, Ti6Al7Nb, TMZF®) and manufacturing processes (wrought, cast, powder metallurgical, additive manufacturing).

Understanding the role of implant alloy metallurgy

Despite efforts to limit micromotion, its prevention cannot be guaranteed. A recent study in our laboratories has shown that among surgically retrieved THAs, 86% of stem tapers and 80% of head tapers had only minimal to mild damage, whereas only 4% and 9% were severely damaged, respectively. Therefore, in the majority of cases only minimal micromotion has occurred. Cases that were moderately to severely worn exhibited various damage modes. It is evident that in moderate to severely worn retrievals, metallurgical factors are increasingly important in driving the damage process. Most prominently, 18% of moderate to severely damaged heads exhibited a distinct damage feature of several tens of micrometers deep troughs running parallel to the taper axis. Due to its column-like appearance, this damage feature has been called *column damage* (Figure 1)²⁸. The occurrence and appearance of column damage is dictated by the banded microstructure of the wrought bar stock material that is frequently used for femoral heads—especially 28 or 32 mm head sizes. Column damage occurs along the banded microstructure running longitudinally through the bar stock material^{16,28}. The resulting corrosion behavior in alloys with the banded microstructure is markedly inferior in comparison to a homogeneous alloy²⁹. Energy dispersive x-ray spectroscopy suggests that the banded structure is related to a local depletion of molybdenum within the alloy, which is a crucial element for the corrosion resistance of the alloy²⁸.

Other microstructural features such as carbides and intermetallic phases were also associated with specific damage patterns such as intergranular corrosion, phase boundary corrosion and pitting²⁹. The volume fraction, type, and size of such hard phases can vary widely^{30–33} and are not specified in the material standards^{34,35}. The role of cobalt alloy microstructure has been discussed at length with respect to its wear behavior—especially for metal-on-metal bearings^{33,36}. Unfortunately, it appears that there is still an insufficient understanding of its role in terms of corrosion. Current studies suggest that a more homogeneous implant alloy microstructure may be a key strategy to minimize material loss due to corrosion^{29,37}. Improvement of the alloy microstructure may be crucial to decrease the metal ion release in cases where micromotion has occurred and corrosive processes have been activated.

Another frequently occurring damage feature—so called “imprinting”—has been reported in several studies^{16,17,38,39}. It is characterized by the imprinting of the usually rougher stem taper topography into the smoother head taper (Figure 2). Interestingly, this damage mode occurs independent of the hardness of either head or stem, but appears to be mainly driven by the difference in topography²⁸. It is also not limited to femoral heads, but can also occur in sleeve adapters as well. Often, the entire female taper surface of heads or the sleeve

adapters is affected. Recent studies suggest that imprinting is fretting driven and affects 83% of moderate to severely damaged MoP THAs^{16,28}. It can be expected that imprinting is a major contributor to the reportedly larger material loss on the female taper compared to the stem.

Understanding the chemical environment

Both cobalt and titanium alloys are considered “corrosion resistant” alloys due to their ability to form a protective passive film. The generally favorable corrosion behavior is one of the qualities that lead to the widespread use of these alloys as implant materials. Even if micromotion occurs, and even if the alloy can be more or less well equipped to withstand electrochemical corrosion, it remains surprising that severe damage such as pitting and intergranular corrosion can occur in vivo. Why does the material corrode severely in some cases and not in others? One possibility is inherent to the process of crevice corrosion: oxygen depletion causes a reduced pH and thus a more corrosive environment. Another possibility is that fretting currents cause periodic local passive film disruption⁴⁰. Yet, to date, no fretting test has successfully reproduced all damage features as observed on retrieved components, especially with respect to electrochemically caused damage features. Munoz et al performed corrosion tests on cobalt alloy in synovial fluid retrieved from TKR patients during revision surgery⁴¹. The resulting corrosion behavior was highly variable. This could indicate that patient specific composition changes in the joint fluid may have a significant impact on the corrosion behavior within the modular junction. Gilbert et al pointed out the capability of inflammatory cell types such as macrophages to promote corrosion due to their release of reactive oxygen species and termed the phenomenon “cell induced corrosion”⁴². Although some of the damage features initially attributed to cell induced corrosion were later shown to be caused by electrocautery during revision surgery^{43,44}, a more recent study has shown evidence of numerous cells seated within the contact area of the femoral head taper¹⁶. It has yet to be shown, however, whether specific cell types can contribute in a meaningful way to the generation of a corrosive environment within the modular junction.

What lies ahead?

In order to minimize the risk of ALTRs due to MACC it has become a common practice to abandon cobalt alloy heads in favor of alumina ceramic heads. Initial clinical studies of patients undergoing revision surgery for ALTR due to MACC have shown promising results, where ceramic head/titanium sleeve exchange led to fast decline of blood metal levels and resolution of symptoms¹⁵. Indeed it appears that overall less metallic material loss can be detected in systems with a ceramic head¹⁴. The use of ceramic heads increased from 37% to 53% between 2012 and 2016 among the medical centers participating in The American Joint Replacement Registry. During the same time, the use of metal heads decreased from 60% to 43%⁴⁵. The inert nature of the ceramic material is mainly responsible for the reduction in metal ion release, and—when coupled with a titanium alloy stem—the elution of cobalt and chromium ions can be completely avoided. Although this trend may be promising, the long-term clinical performance of ceramic heads with modern titanium alloy stems must be followed closely in the coming years. THAs composed of titanium alloys and ceramic heads may face other unforeseen challenges. Titanium alloy has a known higher fracture risk⁴⁶,

and ceramic head fracture, although it has a low probability of occurrence, is catastrophic when it does occur. In addition, several new damage modes have been observed in recent years, for example oxide induced stress corrosion cracking which is limited to titanium alloy⁴⁷. Although so far stress corrosion cracking has only been reported for neck/stem tapers and not in head/neck tapers, the mechanical and chemical conditions enabling this damage mode are not completely understood. There are recent reports of catastrophic failures of titanium alloy stems^{17,18,47}. It appears that this gross taper failure is not limited to, but occurs most frequently in stems made from beta-phase titanium alloy. So far it is not clear if the severe damage—affecting often the entire trunnion as well as the neck—is mainly related to the materials, mechanical factors, head size, the smooth surface topography²⁴, or a combination of all these factors¹². Another case of a titanium neck/stem trunnion exhibited complete dissolution of the beta phase after only one month in vivo⁴⁸. These examples demonstrate that avoiding cobalt alloy may not necessarily eliminate adverse clinical outcomes related to MACC at modular junctions. The fact that additive manufacturing (e. 3-D printing) has also made a leap forward into the portfolio of implant designs, further underlines the importance of a better understanding of implant alloy metallurgy—not only for cobalt alloys, but also for titanium alloys. Finally, as no bench-top test to date has been able to perfectly simulate in vivo failure mechanisms, innovative preclinical testing methods are required to better predict failure mechanisms whether they are related to corrosion, wear, fatigue, or a combination of these processes.

Summary

Reducing the risk of THA failure due to corrosion or other damage processes in modular junctions remains an important task. Continuous research on clinical outcomes, retrieval analysis, and mechanical and electrochemical testing is needed to minimize modular junction-related implant failures. The multifactorial nature of corrosion and fretting processes needs to be considered in order to improve clinical performance, and care must be taken to ensure that new corrosion preventive strategies do not create new problems. As new concepts and innovations are explored, for example cell-induced corrosion and additive manufacturing, it is important to have a full understanding of the ideal implant alloy metallurgy and contact mechanics to avoid the introduction of new failure mechanisms.

Acknowledgments

Parts of this work have been funded by NIH/NIAMS grants R01AR070181 and R03AR066829.

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Figure 1.

Scanning electron micrographs (SEM) of column damage. A) Column-like troughs are running parallel to the taper axis, and perpendicular to the initial machining lines. B) At a higher magnification, it can be seen that the troughs are characterized by etching marks along with fine pitting at some locations.

Figure 2.

Typical SEM image of an imprinted femoral head taper surface. In the lower half of the image the original head taper topography can still be seen. In the upper half of the image an imprint of the typically wider stem taper topography is visible. The imprinted surface appears fairly smooth, but some residual fretting marks remain (arrows). Thus, it appears likely that imprinting is a fretting driven process.