■ SPECIALTY UPDATE: HIP

Corrosion at the head-neck interface of current designs of modular femoral components

ESSENTIAL QUESTIONS AND ANSWERS RELATING TO CORROSION IN MODULAR HEAD–NECK JUNCTIONS

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There is increasing global awareness of adverse reactions to metal debris and elevated serum metal ion concentrations following the use of second generation metal-on-metal total hip arthroplasties. The high incidence of these complications can be largely attributed to corrosion at the head-neck interface. Severe corrosion of the taper is identified most commonly in association with larger diameter femoral heads. However, there is emerging evidence of varying levels of corrosion observed in retrieved components with smaller diameter femoral heads. This same mechanism of galvanic and mechanically-assisted crevice corrosion has been observed in metal-on-polyethylene and ceramic components, suggesting an inherent biomechanical problem with current designs of the head-neck interface.

We provide a review of the fundamental questions and answers clinicians and researchers must understand regarding corrosion of the taper, and its relevance to current orthopaedic practice.

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The concept of modularity in primary and revision total hip arthroplasties (THAs) is well established. Modern modular femoral components offer surgeons the flexibility to tailor the size, offset and biomaterials of the femoral head, neck and stem to the anatomy of the patient and the biomechanics of the hip, whilst reducing the size of the inventory required at operation. However, this provides the femoral component with an additional interface and a potential source of wear and corrosion.

Corrosion of the taper at the head-neck interface was first identified in THA retrievals by Collier et al[1] in 1991, and was attributed to galvanically-accelerated crevice corrosion, which was previously undetected by laboratory testing. However, the biological impact was poorly understood and seemed to be largely subclinical until the introduction of second generation metal-on-metal (MoM) THAs. Adverse reactions to metal debris (ARMD) and elevated serum metal ion concentrations have been observed in patients with large diameter femoral head MoM THAs when compared with resurfacings with similar bearing surfaces.[2] This has led to renewed interest and re-evaluation of modular junctions. Periprosthetic osteolysis and localised soft-tissue reactions such as aseptic lymphocytic vasculitis-associated lesions (ALVAL), pseudotumours and metallosis comprise a spectrum of ARMDs.[3,4]

Corrosion and fretting at modular taper junctions of MoM components contribute to the generation of metal wear debris and raised serum cobalt (Co) and chromium (Cr) ion concentrations.[5-7] However, this is not limited to MoM components or those with larger diameter femoral heads. There is increasing evidence of taper corrosion with ARMD and elevated serum metal ions in metal-on-polyethylene (MoP) and ceramic-on-polyethylene (CoP) THAs.[5,8-10] Globally, the adoption of larger diameter femoral heads for modular primary and revision THAs, has renewed scientific as well as media interest on associated ARMD and the potential risks of elevated serum metal ion concentrations. We provide a guide to the fundamental questions and answers clinicians must understand regarding taper corrosion and its relevance to current orthopaedic practice.

What processes occur at modular junctions?

Passive metals commonly used in THAs, which include titanium (Ti) alloys, cobalt chromium (CoCr) and stainless steel (SS), produce a protective oxide film on their surfaces (passivation layer) rendering them relatively resistant to corrosion *in vivo*. However, galvanic corrosion, pitting and fretting have been observed at the modular head-neck interfaces of retrieved components.[11]

Galvanic corrosion, resulting from differences in anodic potential between two dissimilar metals, has long been known to occur when Ti is coupled with SS. However retrieved components with CoCr on Ti interfaces have also shown evidence of galvanic corrosion.[12] Crevice corrosion occurs in an electrochemical microenvironment formed when fluid enters the crevice between the male and female tapers and stagnates. There are many theories concerning the electrochemical processes involved in the initiation of crevice corrosion and how they may initiate hydrolysis of metal ions producing a positively charged, low pH, high chloride and oxygen depleted solution in the crevice. This solution increases the solubility of the passivation layer and therefore its breakdown.[13] Metal ions are released into the joint fluid as oxides, hydroxides and chlorides. In CoCr alloys, Co is preferentially ionised leaving a Cr-oxide rich passivation layer.[14]

More recent studies have focused on the key role of fretting and 'mechanically assisted crevice corrosion' (MACC). Loading of THAs during walking causes stress and micromotion at the taper interface, leading to repetitive fracture of the protective oxide film. The exposed metal spontaneously re-oxidises *in vivo,* causing further oxygen depletion in the crevice, thereby accelerating the process of crevice corrosion.[14,15] Fretting involves more severe mechanical damage to the metal surface with deep surface asperities beyond the passive layer providing further isolated electrochemical environments for pitting corrosion to occur. Once fretting has been initiated, the corrosion processes may continue in the absence of loading.[16] Retrieved corroded CoCr head tapers have a typical macroscopic appearance in which the topography of the male taper is imprinted onto the female taper.[17]

Modular junctions are designed to be free of movement *in vivo,* and therefore not subject to traditional modes of material wear. Retrieved tapers show evidence of etching and delamination at these interfaces, suggesting some macroscopic movement at the interface.[11] Furthermore, corrosion at the interface may lead to instability and macromotion, making them further susceptible to abrasive wear.[18] Various cases of spontaneous dissociation of the head–neck junction have been recently noted in the literature in association with corrosion at the interface.[19] In contrast, other joint retrieval studies have demonstrated evidence of increased bonding, and even cold welding at the femoral head-neck interface,[20] particularly in association with Ti-to-Ti taper sleeves.

What are the local and systemic consequences of taper corrosion?

The products of corrosion generated at the head-neck interface vary in size from < 1 to 500 micrometres.[6] Most particulate debris is < 5 micrometres in size, plate-shaped and similar, regardless of the material combination used.[6,21] The most abundant particles produced are chromium orthophosphate and mixed oxides, as well as chlorides of chromium, molybdenum and Ti.[6] The biological response to these corrosion products is similar to that produced by bearing surface metal debris,[9] but there is evidence suggesting that they may be more biologically active, possibly as a result of higher concentrations of ionised particles.[7] They trigger an immune response comprising macrophages, giant cells and lymphocytes, leading to localised soft-tissue ARMD and peri-articular osteolysis.[22]

There is concern regarding the systemic distribution of submicrometer-sized particulate debris and metal ions, which have been discovered in organs such as the liver, spleen, abdominal lymph nodes and placenta of patients with THAs.[23,24] The potential for carcinogenicity has been demonstrated.[25] Nevertheless, population-based studies are yet to establish a causal association with any malignancy.[26,27] Cobalt toxicity is an extremely rare phenomenon associated with THA.[28]

Can using different material combinations and surface treatments control corrosion?

Femoral head components are typically manufactured from either CoCr or ceramic due to their favourable wear properties at the bearing surface. In contrast, stems may be manufactured from CoCr, SS or Ti. The stiffness of Ti more closely resembles that of cortical bone, and therefore reduces the effects of stress shielding at the proximal femur, making them more biomechanically suited for use in femoral stems.

Retrieval and *in vitro* studies have consistently demonstrated that CoCr alloy couples are less susceptible to fretting corrosion than CoCr coupled with either Ti or SS.[8,29] This is likely to be due to the effects of galvanic corrosion. However, CoCr couples also have significantly lower pull-off forces and turn-off moments, indicating a weaker interface.[29,30]

In vitro and retrieval studies indicate that interfaces involving ceramic heads have the highest resistance to fretting and corrosion.[10,31,32] Carli et al[33] conducted a systematic review of published cases of revision THAs for symptomatic taper corrosion, finding that only five of 776 cases did not involve a metal head. Whilst most cases (419) involved MoM interfaces, 352 involved a MoP interface. The components which were revised and those used for the revision were documented in 12 articles comprising 24 cases. In 20 (83%), the male taper was found to be macroscopically intact, and in 15 out of the 20 cases the existing head was revised to a new ceramic head without revising the stem. Khatod et al[34] reported that use of ceramic heads in revision surgery significantly decreases the likelihood of further revision.

Surface coatings and treatments may be used to improve the corrosion properties of a material, whilst retaining favourable mechanical properties, especially in Ti stems.

Few studies have attempted to investigate this. However, Goldberg and Gilbert[16] used *in vitro* electrochemical studies to illustrate that certain coatings can improve fretting and resistance to corrosion. Ceramic coatings, such as diamond like carbon, have the potential for failure due to fatigue and delamination resulting from their brittle nature, although modern techniques of surface engineering have improved this situation. An alternative surface may be produced by thermal transformation of metal surfaces, such as oxidised zirconium, making them more resistant to fractures of the surface and delamination compared with applied coatings.[35] The durability of coatings or surface treatments and their effect on corrosion when exposed to the high stresses at modular interfaces is yet to be established, however, remains a promising solution.

Is surface topography important?

There is wide variation in the surface finish that manufacturers use for tapers in their attempts to improve fixation at the head-neck junction.[36] The design of male tapers was changed to incorporate increased roughness often with threaded surfaces to accommodate ceramic heads, which are less deformable than CoCr. The significance of surface topography in relation to corrosion at modular junctions is poorly understood. In an *in vitro* study, Panagiotidou et al[37] demonstrated that rough-surfaced tapers produce more fretting and corrosion than smooth tapers under normal loading and perform poorly under high loads. Further studies would be required to verify this finding, particularly in relation to the use of ceramic heads.

What is the importance of the diameter of the femoral head?

In recent decades, surgeons have favoured the use of larger femoral heads to increase stability and impingement-free range of movement, whilst promoting fluid film lubrication. The relationship between the diameter of the femoral head and corrosion has been scrutinised after the well-publicised failures of large head MoM THAs.[2,38] Various clinical and laboratory studies have shown that increasing the size of the head leads to increasing corrosion at the head-neck interface with greater ARMDs and elevated metal ion levels in the blood.[31,39-41] In their systematic review of all cases revised for symptomatic taper corrosion, Carli et al[33] found that the median head size was 46 mm for MoM and 36 mm for MoP components.

Elkins, Callaghan and Brown[42] used finite element analysis (FEA) to illustrate increasing stresses at the modular interface with increasing femoral head diameter, but there was no concomitant increase in stability beyond 40 mm diameter. In support of these findings, a clinical study showed questionable improvement in function or range of movement with femoral heads whose diameter was > 36 mm.[43]

Increasing the diameter of the femoral head in itself causes greater frictional torque forces, which are transmitted to the head-neck modular interface as rotational stresses and cause a secondary detrimental effect on corrosion.[31,44] This force is amplified by increased friction at the bearing surface caused by abnormal wear and poor lubrication. With regards to component positioning, high inclination angles of the acetabular component are known to result in edge loading and high rates of wear after the run-in phase,[45] which is also known to be the case with hip resurfacing.[46,47] The relationship with acetabular version is less clear, and it can be extrapolated that large femoral heads are more sensitive to malpositioning, particularly of the acetabular component, as a result of the increased frictional torque produced.

How does head-neck offset affect corrosion?

The patterns of wear on retrieved trunnions are suggestive of mediolateral toggling, the magnitude of which is directly related to the head-neck offset. Macroscopic fretting has been demonstrated in increasing magnitude as the femoral head-neck offset is increased,[15,48] a finding supported by FEA data.[49] Where Gilbert, Mehta and Pinder[29] were only able to show a statistically non-significant increase in corrosion with higher head-neck offset for CoCr on Ti, Panagiotidou et al[31] performed electrochemical studies showing a significant increase in corrosion with increasing head-neck offset across various material combinations.

The effect of head-neck offset on corrosion will be of importance to surgeons when selecting offset femoral heads in order to balance soft-tissue tension or leg length, suggesting that ‘plus heads’ are detrimental. Where possible, increased offset and leg length should be conferred from the stem rather than the femoral head.

Does taper geometry affect corrosion?

The traditional 12/14 taper was originally designed for smaller diameter femoral heads and may need to be modified to accommodate higher stresses with large diameter heads. In a systematic review of components which were revised for symptomatic taper corrosion, the size of the taper was documented in 431 cases, of which all were 12/14, or smaller, in diameter.[33] Despite this, the literature is relatively deficient in identifying the importance of various parameters of taper geometry. Figure 1 depicts a standard design of taper for the head-neck junction.

Increasing the diameter of the male taper should theoretically reduce the effect of torque produced by large femoral heads at the interface. The flexibility of the neck, which is directly related to its thickness, has been shown in several studies to be an important determinant of corrosion at the interface.[8,10,50] Clinically, a balance needs to be achieved between the thickness of the femoral neck and impingement-free movement of the hip.

The contact surface area of the taper is also an important determinant of corrosion, possibly by altering the concentration of stress. However, studies are conflicting regarding the nature of this relationship,[37,50] most likely due to confounding factors in determining the contact area, including

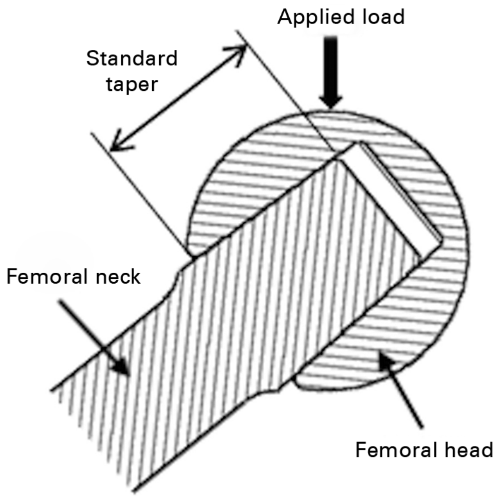

Fig. 1

Typical head-neck taper design.

the size, geometry and surface topography of the taper and of the head-neck offset. Modern tapers are shorter with increased flexibility, making them more susceptible to corrosion than their predecessors.[51] Closer geometric matching of the male and female taper in order to minimise micromotion may reduce fretting, but at the expense of difficult intra-operative assembly.

It should be noted that 12/14 tapers from different manufacturers might have differing angles and lengths, which can alter the contact biomechanics. Therefore, mixing and matching is not advisable, particularly for less deformable ceramic heads.

Can the surgeon influence long-term corrosion through the conditions of assembly of the components?

Various intra-operative techniques have been evaluated to determine the optimal method in assembling the femoral head to the stem. Orthopaedic surgeons vary in the amount of force applied, ranging from simple hand application to mallet strikes of varying intensity. A study showed that axial impact forces applied by surgeons varied between 273 N and 7848 N.[52] Impact assembly leads to less corrosion than application by hand.[53] Higher impaction loads lead to reduced corrosion and increased strength of the interface,[30,54] although sufficient strength is achieved with impaction forces of 4 kN. The required impact load may vary with different combinations of material and this may be especially relevant in ceramic heads where there is evidence of loosening of the modular connection during normal activities.[52] Contrary to popular belief, repeated impactions do not seem to increase the force required to dissociate the femoral head from its neck.[30]

Shrink-fit stems, for which the head and neck are coupled in the factory, offer an alternative to intra-operatively assembled components. These modular couples have shown higher pull-off strengths and less visible evidence of corrosion in a retrieval study.[12] This is at the expense of losing the ability to manipulate offset and head-size intra-operatively.

There is strong evidence from *in vitro* studies to suggest that dry non-contaminated assembly of the head and neck is protective against, does not prevent corrosion, possibly by decreasing the onset load of fretting.[29,55] Furthermore, Weisse et al[56] showed that contamination of the taper with blood or bone reduces the resistance of ceramic heads to fracture.

In conclusion, corrosion at modular interfaces is an important cause of failure after primary and revision THA. MACC is the primary mechanism of corrosion leading to the generation of metal particles and ions that may be more biologically active than those produced at the bearing surfaces. Severe head-neck taper corrosion has been identified most commonly in association with MoM THAs secondary to increased frictional torque and bending produced by large diameter femoral heads. Emerging evidence suggests that corrosion of the taper occurs with all the combinations of bearing surface which are in clinical use. With the increasing popularity of larger diameter femoral heads, the design of the modular taper interface must be adapted to cope with the additional loading stresses in order to reduce both the risk of corrosion and the rates of failure.

Although ceramic heads are less susceptible to corrosion of the taper than CoCr heads, the interface is weaker and subject to increased movement within normal loading conditions. The role of ceramic surface coatings and surface treatment is an exciting prospect that could offer increased resistance against corrosion, whilst allowing the material of the component to retain its biomechanical properties. Further research will be required to establish the efficacy and durability of coatings and surface treatments.

Surgeons wishing to minimise head-neck corrosion should select components with smooth tapers, thicker necks and a large contact surface area. Manufacturers vary significantly in most aspects of taper design and this should be considered during the selection of components. The use of high-offset femoral heads and head diameters of > 40 mm should be avoided. Where possible, increased femoral offset should be conferred by appropriate selection of the stem, rather than the femoral head. Intra-operatively, components should be assembled dry using a single hammer strike with a high-impact load of > 4 kN.

Supplementary material

ë Figures showing taper corrosion and the imprinting effect are available alongside the online version of this article at www.bjj.boneandjoint.org.uk.

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M. Khan: Contributed to writing the paper.

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