Medical Engineering and Physics 000 (2018) 1–9

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Factors influencing taper failure of modular revision hip stems

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ARTICLE INFO

Article history:
Received 29 September 2017
Revised 19 January 2018
Accepted 11 February 2018
Available online xxx

Keywords:
Hip revision
Taper junction
Fracture
Corrosion
Fatigue
Strength
Titanium
Finite element model
Surface damage

ABSTRACT

Stem modularity of revision hip implant systems offers the advantage of the restoration of individual patient geometry but introduces additional interfaces, which are subjected to repetitive bending loading and have a propensity for fretting corrosion. The male stem taper is the weakest part of the modular junction due to its reduced cross section compared to the outside diameter of the stem. Taper fractures can be the consequence of overloading in combination with corrosion. The purpose of this study was to assess the influence of implant design factors, patient factors, and surgical factors on the risk of taper failure of the modular junction of revision stems.

An analytical bending model was used to estimate the strength of the taper connection for pristine, fatigued and corroded conditions. Additionally, a finite element contact model of the taper connection was developed to assess the relative motion and potential for surface damage at the taper interface under physiological loading for varyied assembly and design parameters.

Increasing the male taper diameter was shown to be the most effective means for increasing taper strength but would require a concurrent increase in the outer implant diameter to limit a greater risk of total surface damage for a thinner female taper wall. Increasing the assembly force decreases the total surface damage but not local magnitudes, which are probably responsible for crack initiation. It is suggested that in unfavourable loading conditions a monobloc implant system will reduce the risk of failure.

1. Introduction

The numbers of revisions of both primary and revision hip implantations are steadily increasing [1]. Revision rates for first revision at 10 years in national arthroplasty registries of between 3% and 20% are reported [2–10]. Revision rates of primary implantations in the same registries are much lower (between 2% and 4% if metal-on-metal bearing articulations are omitted) [6,64]. Revision implantation is more challenging than a primary procedure as bone stock has often been lost due to stress shielding and the operative removal of the primary implant and the cement mantle, mostly in the proximal region. In anticipation of further revision procedures revision stems are typically uncemented, and are therefore made of titanium alloys which are compatible with bone ingrowth. They are anchored by press-fit in the distal femur. This necessitates a femoral stem that is longer for a revision implant than for a primary stem [11]. Modularity has found widespread application in revision femoral stems as it allows for intra-operative adaptation to the conditions of the femoral bone. Modular systems also require less inventory than monobloc (non-modular) systems. Modular systems incorporate a taper junction between prosthesis neck component and stem, generally located in the proximal third of the implant, allowing the distal stem to be implanted and the proximal neck component to be fitted to the anatomy and assembled subsequently.

Failure of modular connections has been reported to occur by dislocation, corrosion and fracture [12–18]. Failure by fracture of a revision stem is reported at rates of 0.9% to 3.6% [4]. Fracture is induced mechanically by fretting, involving the repetitive mechanical disruption of the protective oxide surface layer of the bulk metal due to oscillating relative motion between two surfaces in contact. By definition the amplitude is smaller than the width of nominal contact area. The resulting damage can occur due to wear, fatigue or corrosion. The process of passive oxide film removal due to fretting, followed by corrosion, is called “fretting corrosion”. Modular junctions in revision stems are subject to high bending moments, due to their offset (lever arm) from the joint force vector, especially in situations without proximal bone support. Patient anatomy and body weight also influence bending loading of the taper [14,16,18–20]. The diameter of the intramedullary canal limits the maximum diameter of the stem that can be implanted. Increased loading cannot always be compensated by implant dimensions, leading to limited flexural strength [14,17–19,23,24]. This also applies to the taper dimensions, which are limited by the out-

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https://doi.org/10.1016/j.medengphy.2018.02.001
1350-4533/

side diameter of the implant. Failure has also been documented for the stem-head taper junction, particularly in association with the re-introduction of large diameter metal-on-metal joints, which can generate increased joint friction moments [13–18].

Failure of revision stems is generally due to fatigue fracture [24,25]. Fracture of an implant occurs when stresses exceed the material strength. Pristine components have the greatest strength, which decreases with fatigue loading and corrosion. Higher assembly forces have the potential to reduce fretting by increasing the press-fit [26–28]. A parametric analysis of the major factors responsible for prosthesis failure is not available.

The purpose of this study was therefore to investigate the influence of joint loading, stem taper geometry, material characteristics and assembly force on potential taper failure of modular revision hip stems.

2. Materials and methods

Stress magnitudes were assessed analytically for different realistic implant geometries and patient loading scenarios, and related to the strength of the material in various states of degradation by fatigue and corrosion. The influence of the modular taper design and assembly force on degradation of the material by surface damage were also analysed, using finite element (FE) analysis. For validation of the analytical and numerical models, experimental determination of the fracture load, seating depth and gap opening at the taper connection under different assembly loads were performed.

A clinically successful modular THA revision prosthesis system (MRP, Peter Brehm GmbH, Germany) was used as basis for the analysis [22]. The femoral implant of the MRP consists of a distal stem and a proximal neck piece (Fig. 1A). For this study a neck piece with a CCD angle of 130° was used (Fig. 1C). Both components are made from Ti6Al4V alloy (Ti) and have an outside diameter of 20 mm. The components are joined by a taper connection with a diameter of ~12 mm for the male taper, a taper angle of 1.4° (from the mid-axis) and a contact length between male and female components of 19.5 mm when assembled. The lateralised proximal neck piece in combination with an L4 prosthesis head (+16 mm) produces a horizontal offset of 53 mm. This was investigated as a worst-case loading scenario of the taper connection ("high offset"; Fig. 1C), since this results in the highest bending moment possible at the modular junction. This combination is not approved by the manufacturer but is used clinically. For comparison a "short" (−5 mm) head with a standard neck piece producing a horizontal offset of 30 mm was also investigated ("low offset"; Fig. 1C).

2.1. Analytical

A simple beam bending model was used to calculate the maximum bending stresses σ_{max} for the outside surface of the male taper of the distal stem, according to joint loading and implant geometry (Fig. 1B). The joint load was assumed to act through the centre of the prosthesis head, in the plane of the implant and at an angle θ to the vertical implant axis. The bending moment M acting around the centre of the male taper of the distal stem, at the level of the open end of the proximal neck piece, was calculated from the joint force vector and its distance Δ from this point (Fig. 1D). The radius R of the male taper of the distal stem at this point was varied. Peak joint forces F measured in vivo during walking and stumbling for a light (60 kg) and a heavy patient (120 kg) [29] were applied. Calculated stresses were compared with the strength of pristine Ti6Al4V ($\hat{\sigma}_{pristine}$ = 1000 MPa) [53], a fatigued material at 10^7cycles ($\hat{\sigma}_{fatigue}$ = 750 MPa) [30] and a severely corroded material ($\hat{\sigma}_{corrosion}$ = 200 MPa) [30].

The analysis assumes linear bending theory. Stresses due to the axial and shear force components are neglected as they are relatively small [63]. It was assumed that there was no bony support of the proximal neck piece.

2.2. Numerical

Fretting damage can occur when small relative motions between two interface surfaces in contact occur [21,26], which abrade the protective oxide layers. In a fluid environment, metal ions can leave the bare metal surface, which is known as fretting corrosion [31]. In titanium alloys this leads to a roughened surface that causes stress concentrations as sites of crack nucleation [32].

A finite element model of the stem taper junction was generated (Abaqus 6.14, Dassault Systèmes, France) based on CAD data of the MRP-System (Fig. 2A). Contact analysis of the taper interface was implemented to investigate mechanical and design factors potentially influencing fretting damage. Linearly elastic, homogeneous, isotropic material properties for the titanium alloy Ti6Al4V were used (E = 113.8 GPa; ν = 0.34). The friction coefficient for contact of the alloy surfaces was set to μ = 0.35 [33,34].

A mesh convergence analysis resulted in a suitable element size of 0.8 mm on the contact surface of the taper junction (proximal neck piece and distal stem), resulting in 105,837 elements for the distal stem and changes of less than 2% in gap opening and seating depth between mesh refinements. Relative shear interface motion and contact pressure were sampled at each node of the surface of the male taper. Assembly forces were varied between 0.5 kN and 40.0 kN, representing very low intraoperative values and very high laboratory values, respectively. Physiological joint loading for a walking cycle of a 75 kg patient was applied, according to in-vivo measurements in joint replacement patients [29].

Relative shear interface motion and pressure at each node were multiplied to obtain a factor representing relative surface damage for ten equally spaced time intervals of the loading cycle (based on Archard's law [35,36]). Total surface damage was determined by summing over all nodes and all time intervals.

The influence of prosthesis design was investigated by the variation of the male taper diameter between 8.0 mm and 16.0 mm, with the outside diameter of the implant maintained at 20.0 mm, the length of contact at 19.5 mm and the taper angle at 1.4° (Fig. 2B). For this parametric analysis an assembly force of 11.0 kN was applied, followed by the joint force in a gait cycle [29].

2.3. Validation

For experimental validation of the analytical bending stress model, the fracture load of the modular implant was determined and compared with the analytical prediction. The proximal neck piece was assembled under 9.0 kN on a replica of the distal stem piece (produced by the implant manufacturer) using a materials testing machine (Zwick Z010, Zwick GmbH & Co.KG, Germany; in accordance with ISO 7206–10; 0.04 mm/s). The prosthesis head was then loaded vertically until fracture of the taper (n = 3). The peak force was recorded.

For experimental validation of the FE model, the opening of the gap at the taper interface was measured in the radial direction at the lateral side using a chromatic confocal sensor (DT IFS 2403–1.5, Micro-Epsilon Messtechnik GmbH, Germany; measurement range: 1.5 mm, resolution 60 nm) mounted on the outside surface on an aluminium clamp (Fig. 3A). The change in distance to the male taper surface was measured through a 1.5 mm diameter hole drilled through the proximal neck piece at the lower edge of the lateral side. Measurements were repeated three times.

The two prosthesis components were assembled quasi-statically at 0.04 mm/s (ISO 7206–10) with forces of 0.5, 5.0 and 9.0 kN, along the taper axis using a materials testing machine (Zwick Z010, Zwick GmbH & Co.KG, Germany). After assembly, a joint