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Do Stem Taper Microgrooves Influence Taper Corrosion in Total Hip Arthroplasty? A Matched Cohort Retrieval Study

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Abstract

Background—Previous studies identified imprinting of the stem morphology onto the interior head bore, leading researchers to hypothesize an influence of taper topography on mechanically assisted crevice corrosion (MACC). The purpose of this study was to analyze whether micro-grooved stem tapers result in greater fretting corrosion damage than smooth stem tapers.

Methods—A matched cohort of 120 retrieved head-stem pairs from metal-on-polyethylene bearings was created controlling for implantation time, flexural rigidity, apparent length of

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engagement, and head size. There were two groups of 60 heads each, mated with either smooth or micro-grooved stem tapers. A high precision roundness machine was used to measure and categorize the surface morphology. Fretting corrosion damage at the head/neck junction was characterized using the Higgs-Goldberg scoring method. Fourteen of the most damaged heads, were analyzed for the maximum depth of material loss and focused ion beam (FIB) cross-sectioned to view oxide and base metal.

Results—Fretting corrosion damage was not different between the two cohorts at the femoral head ($p = 0.14$, Mann Whitney) or stem tapers ($p = 0.35$). There was no difference in the maximum depths of material loss between the cohorts ($p = 0.71$). Cross sectioning revealed contact damage, signs of micro-motion, and chromium rich oxide layers in both cohorts. Micro-groove imprinting did not appear to have a different effect on the fretting corrosion behavior.

Conclusion—The results of this matched cohort retrieval study do not support the hypothesis that taper surfaces with micro-grooved stems exhibit increased *in vivo* fretting corrosion damage or material release.

Keywords

mechanically assisted crevice corrosion; fretting; corrosion; microgrooves

Introduction

Modularity is highly valued in modern lower extremity total joint arthroplasty, because it allows for intraoperative flexibility, the ability to fine-tune a hip arthroplasty based on patient-specific anatomy, and the ability to combine different component materials. However, starting in the 1980s, fretting and corrosion were identified at the modular connection between femoral heads and stem tapers, and were associated with the release of metal ions and corrosion products into periprosthetic tissues [1–3]. Metal ion release and taper corrosion have since re-emerged as a clinical concern in certain designs of metal-on-metal (MOM) hips [4], certain modular femoral stems [5], as well as in contemporary metal-on-polyethylene bearings [6]. The mechanism of fretting and corrosion damage in modular tapered interfaces, known as mechanically assisted crevice corrosion (MACC), is ubiquitous and multifactorial, with surgeon, patient, and implant factors potentially playing a role in its clinical presentation [7]. Among the numerous clinical and implant variables associated with visual evidence of MACC damage in previous retrieval studies, researchers have identified the following contributing factors: implantation time; flexural rigidity and length of the femoral stem taper; mixing of alloys between the head and stem; and incorporating multiple modularity into the stem, for example a modular neck [7–9]. Regardless of the etiology and ubiquity of MACC, there appears to have been only a small number of patients developing clinically significant adverse local tissue reactions from the release of metal ions and corrosion products from taper interfaces [10]. The reasons underlying a patient's specific response to metal ions and corrosion products remain poorly understood.

The surface finish of the stem taper has recently been proposed as yet another design factor potentially contributing to MACC [11–13]. Historically, and up to the present day, manufacturers produce stem tapers with either a smooth or micro-grooved surface finish

[13]. A micro-grooved stem taper, which has the appearance of microscopic screw threads or gramophone ridges, may influence the strength of the head-stem modular connection [13]. Previous studies of retrieved cobalt-chromium (CoCr) alloy femoral heads in metal on metal (MOM) hip arthroplasty have identified imprinting of the stem taper micro-grooves onto the interior head bore [14], leading researchers to hypothesize that stem taper surface finish may influence taper corrosion. Kop and colleagues [11] studied a cohort of retrieved stems with modular necks and reported that CoCr alloy necks with a micro-grooved finish were more corroded when comparing with smooth finished CoCr stems, but noted the opposite trend with Ti-6Al-4V alloy necks. Panagiotidou et al. [12] performed an *in vitro* test of CoCr heads on Ti-6Al-4V alloy stems and found breaching of the passive film on the micro-grooved, but not the smooth neck tapers. Overall, the available published literature regarding the influence of micro-grooves on MACC is limited and contradictory.

The purpose of this matched cohort retrieval study was to evaluate the effect of stem taper surface morphology on *in vivo* fretting and corrosion damage. We tested the hypothesis that the micro-grooved surface morphology of the stem taper increases *in vivo* fretting and corrosion damage at the head-stem modular junction. Accordingly, we asked: (1) whether micro-grooved stem tapers result in greater *in vivo* fretting and corrosion damage to CoCr heads than smooth stem tapers; (2) what device and patient factors explained the variation of head and stem taper fretting corrosion damage in the two matched cohorts; and (3) whether the mechanism of MACC differs between smooth and micro-grooved stem tapers.

Methods and Materials

Study Design, Cohort Selection, and Clinical Information

The matched cohort design was based on the combined retrieval collections of two academic engineering-based programs working in collaboration with 12 clinical revision centers from the Northeast, Midwest, South, and Western regions of the United States. Retrievals were collected as part of a 14-year ongoing institutional review board-approved revision and retrieval program. A power analysis was conducted and revealed that a total sample size of 120 was sufficient to detect a difference in corrosion score of 0.5 between the smooth and micro-grooved taper cohorts with 82% power and 95% confidence.

From our combined inter-institutional database of over 3,000 total hip arthroplasties (THAs), we first identified 398 sets of matched CoCr head/femoral stem taper pairs [15]. A high precision roundness machine (Talyrond 585, Taylor Hobson, UK) was used to measure surface morphology and to categorize the stem tapers into smooth vs. micro-grooved categories [15]. Ten millimeter long axial profiles were measured with a 5µm diamond tip stylus [15]. Commercial software (Ultra, Taylor Hobson, UK) was used to analyze a 1mm as-manufactured representative region [15]. Three parameters were calculated from the profiles: average surface roughness, amplitude, and wavelength of micro-grooves (if present) [15]. We classified stem tapers as micro-grooved if they exhibited a periodic surface profile with a wavelength > 100 µm and an amplitude of > 4µm (Figure 1).

The metallic head and stem material compositions of all samples were confirmed using an X-ray fluorescence (XRF) detector (Thermo Scientific, Niton XL3t GOLDD+, Waltham,

MA, USA). Stems were fabricated from CoCr or Ti-6Al-4V alloys and were used in a metal-polyethylene (M-PE) bearing THA device (table 1 and 2). We excluded MOM bearings, as well as any CoCr head with a taper adapter sleeve. We also excluded prostheses with a modular femoral stem from the study because of previous research suggesting that these tapers were associated with increased femoral head corrosion in MOM bearings [8]. Thus, this study focused exclusively on M-PE hip bearings incorporating a CoCr femoral head and a monolithic femoral stem of CoCr or Ti-6Al-4V alloy, with no other sources of modularity in the head or stem.

Devices in the smooth and micro-grooved taper cohorts were matched based on the following four criteria (in order of importance) based on important variables published in previous retrieval studies of taper corrosion [7, 8]: (1) implantation time (most important); (2) stem-neck flexural rigidity; (3) apparent length of taper engagement; and (4) head size (least important). In this study, the CoCr heads had the same manufacturer as the stems they were implanted with, eliminating manufacturer mixing as a confounding factor. Flexural rigidity of the stem tapers was calculated using the equation used by Goldberg and colleagues [7]. The flexural rigidity of the stem tapers were calculated using the Young's modulus (E) of the alloy multiplied by the second moment of area ($I = \pi (d)^4 / 64$, where d =diameter of stem at the distal contact point of the head taper). The diameters of the necks were measured by two independent observers and were assumed to be circular. The apparent length of taper engagement was measured from the proximal edge of the femoral stem taper to the point of femoral head engagement.

In addition to the retrieved components, clinical data (implantation time, age, sex, body mass index (BMI), University of California Los Angeles (UCLA) activity score, and reason for implant revision) were collected for all patients in the smooth and micro-grooved taper cohorts (Tables 1 and 2, respectively). For the smooth taper cohort, the mean $\pm$ standard deviation implantation time was 6.4 ± 5 years (range, 0.25 to 19 years), the mean patient age at implantation was 61 ± 13 years, 31 of 60 (52%) were female, the mean BMI was 29 ± 6 kg/m², and the mean UCLA activity score was 5 ± 2 (Table 1). For the micro-grooved taper cohort, the mean $\pm$ standard deviation implantation time was 6.2 ± 5 years (range, 0.25 to 18.75 years), the mean patient age at implantation was 61 ± 12 years, 29 of 60 (48%) were female, the mean BMI was 29 ± 7 kg/m², and the mean UCLA activity score was 6 ± 2 (Table 2). There was no difference in the implantation time ($p=0.68$), gender ($p = 0.48$), BMI ($p = 0.98$), or UCLA activity levels ($p=0.45$) between the smooth or micro-grooved taper cohorts. The most frequently reported reasons for revision in both cohorts were infection and loosening (Tables 1, 2). According to the medical records, none of the heads or stems in either cohort were revised as a result of an adverse local tissue reaction.

Modular Interface Damage Evaluation and Scoring System

Devices were cleaned in accordance with institutional procedures. The CoCr head and neck tapers were inspected visually and under a digital optical microscope (Keyence; Itasca, IL, USA) for evidence of fretting and corrosion. Composite fretting and corrosion damage at the modular CoCr head and metal stem interfaces were characterized using a previously published 4-point semi-quantitative scoring technique [8, 16]: with a score of: 1, indicating

minimal fretting or corrosion (fretting on < 10% surface and no corrosion damage); 2, indicating mild damage (fretting on > 10% surface and/or corrosion attack confined to one or more small areas); 3, indicating moderate damage (fretting > 30% and/or aggressive local corrosion attack with corrosion debris); and 4, indicating severe damage (fretting over majority (> 50%) of mating surface with severe corrosion attack and abundant corrosion debris).

Components were scored independently by the same three investigators. In the event of disagreement between the scores, the three investigators convened to adjudicate the discrepancy and arrive at a consensus score for the taper. The investigators were blinded to the cohort status of the stems during scoring.

Maximum Depth of Material Loss

Each of the 120 femoral head bores were screened by measuring with four axial profiles, 90° spacing between each trace using a high precision roundness machine (Talyrond 585, Taylor Hobson, UK). These profiles were analyzed by fitting a least squares line through the as-manufactured surface and measuring the maximum deviation from the fitted line, indicating the magnitude of material loss. From the results of this analysis, the most severely damaged femoral heads (n=7 from each cohort, femoral head damage score = 4) were chosen for more detailed observation. The femoral heads were centered and leveled on the roundness machine followed by the measurement of four axial profiles from the interior of the bore, with 90° spacing between each trace. Without moving the femoral head, circumferential measurements were taken every 0.25 mm. The axial profiles were screened for the location of the maximum depth of material loss, and the corresponding circumferential profiles were then analyzed. Regions with material loss were excluded, and the remaining portion of the profile, which represented the as-manufactured surface, were used to fit a least squares circle representing the as-manufactured surface. The maximum depth of material loss was estimated as the maximum deviation from the fitted circle. The selection of regions to be excluded and measurement of maximum depth of material loss was performed independently by two trained investigators.

Metallurgic Characterization at the Taper surfaces

Six retrieved femoral head bores (3 CoCr smooth stem tapers and 3 Ti-6Al-4V micro-grooved stem tapers) were analyzed to determine the impact of stem taper topography on the corrosion mechanisms. Optical microscopy (Zeiss Axio Zoom.Z16 stereo microscope, Oberkochen, Germany) was performed to determine areas of interest on the taper surface, and describe the specimens at a macro level. Focused ion beam (FIB) cross-sectioning (FEI Versa 3D dual beam microscope, Hillsboro, OR, USA) was performed by depositing a layer of platinum on the region of interest (outer exposed surface). Then gallium, ions were directed to the surface to cut a wedge-shaped trench allowing for the cross-sectional viewing of the oxide and base metal layers (Figure 2). Samples were also cross-sectioned for microstructural analysis; standard metallographic procedures, in general accordance with ASTM E3 [17]. Scanning electron microscopy (SEM, FEI Versa 3D dual beam microscope, Hillsboro, OR, USA), energy dispersive spectroscopy (EDS, FEI Versa 3D dual beam microscope, Hillsboro, OR, USA) were also performed on the specimens to describe the

corrosion damage mechanisms and the material composition. This SEM was equipped with a field emission electron gun allowing for high-resolution imaging of the features identified.

Statistical Analysis

Preliminary evaluation of the visual corrosion damage scoring data demonstrated a non-normal distribution. Hence, nonparametric statistical analyses were performed using statistical software (JMP 10.0; SAS Institute, Cary, NC, USA). Mann-Whitney U, Kruskal-Wallis (with post-hoc Dunn tests when necessary), and Wilcoxon tests were used to assess differences in taper damage grouped by categorical parameters. The level of significance chosen for these hypothesis tests was $p < 0.05$.

Results

The two cohorts were statistically matched for implantation time ($p=0.68$), stem neck flexural rigidity ($p=0.73$, Wilcoxon), and apparent length of taper engagement ($p=0.18$) (Table 3). Within this study, there was a difference in distribution between femoral stem neck diameters ($P < .001$). The microgrooved cohort had a larger neck diameter with a mean $\pm$ SD of 13.8 ± 0.08 mm (range, 13.8-14 mm), and the smooth cohort had a mean $\pm$ SD neck diameter of 13 ± 0.75 mm (range, 14 to 12.1 mm). The smooth cohort was composed of 10% Ti-6Al-4V alloy stems ($n=6$) and 90% CoCr stems ($n=54$). The micro-grooved cohort was composed of 55% Ti-6Al-4V alloy stems ($n=33$) and 45% CoCr ($n=27$). Due to differences in taper trunnion dimensions, mainly neck diameter, a notable discrepancy of alloy distribution between the smooth and micro-grooved cohorts emerged through our efforts to maintain statistically similar flexural rigidity values. However, within the entire cohort, neither the stem alloy ($p=0.61$) nor the femoral stem diameter ($p=0.11$) was correlated with an increase in femoral head damage score.

Mild to severe damage (Fretting Corrosion Score ≥ 2) was observed in 74% of the 120 CoCr femoral heads, 77% of the heads mated with micro-grooved stems (46/60, Figure 3A) and 72% of the heads mated with smooth stems (43/60, Figure 3B). Fretting and corrosion damage scores for the CoCr femoral head female tapers were not different between the two cohorts ($p = 0.14$, Figure 3C). The median fretting and corrosion score was 2 for the head tapers in both cohorts. Likewise, the fretting and corrosion damage scores for the male stem tapers was not different between the two cohorts ($p = 0.35$, Figure 4). The median fretting and corrosion score was 1 for the micro-grooved stems tapers and 2 for the smooth stems tapers.

The maximum depth material loss of the most severely damaged femoral heads ($n=7$ femoral heads paired with smooth stem, $n=7$ femoral heads paired with micro-grooved stems) was not different between the two cohorts ($p = 0.71$, Mann-Whitney U test, Figure 5). The implantation time ($p = 0.26$), stem neck flexural rigidity ($p = 0.71$), apparent length of taper engagement ($p = 0.26$) and head size ($p = 0.13$) were all similar between the two groups. The smooth cohort was exclusively composed of CoCr femoral stems and the micro-grooved cohort was exclusively composed of Ti-6Al-4V femoral stems.

Substantial metal loss was not observed optically or visually for either the micro-grooved or smooth head-bore surfaces (Figures 6 and 7, respectively). Microscopic inspection revealed contact damage in the form of insertion and removal deformation. Signs of micro-motion were observed with fretting damage apparent on both types of head-bore surfaces; however, damage was more readily apparent on the micro-grooved imprinted head-bore surface (Figure 8A). In contrast, the more intimate metal-to-metal contact on the smooth stem taper may have smeared the fretting damage on the head-bore surface making them less readily apparent (Figure 8B).

A chromium rich oxide layer was apparent in some locations on the surface of both the smooth and micro-grooved femoral head bores (Figure 9). The discontinuities in the oxide layer did not appear to vary substantively for smooth vs. micro-grooved surfaces. There was a noted difference in microstructure near the outer surface of the head-bore. Micro-groove imprinting did not appear to have a different effect on the fretting corrosion behavior of the head bore surface of head-stem pairs. This finding was consistent with the depth of material loss analysis suggesting no difference in material loss between these stem pairs. Additionally, there was no obvious effect of galvanic coupling between the CoCr and Ti-6Al-4V on the morphology or amount of material loss observed.

Discussion

The results of this matched cohort retrieval study do not support the hypothesis that taper surfaces with micro-grooved stems exhibit increased *in vivo* fretting corrosion. Recognizing that *in vivo* fretting and corrosion damage is multifactorial, we accounted for implant, patient, and clinical factors that may influence *in vivo* taper corrosion with the matched cohort design and by *post hoc* statistical analyses. Based on our statistical power calculations, we believe that our sample size was sufficiently large to detect a difference in corrosion scores between the smooth and micro-grooved cohorts, if indeed such a relationship existed. It is clear, however, based on our observations of severe fretting and corrosion damage among heads and stems with both smooth and micro-grooved tapers (e.g., Figures 2 and 3), that independent modification of taper surface morphology alone is unlikely to present a promising solution for fretting corrosion damage. Furthermore, based on the results of this study, it likewise appears implausible that historical changes in taper surface morphology are responsible for today's renewed clinical concern with taper fretting corrosion [10].

This study has limitations. By design, all of our retrievals included a metal-on-polyethylene bearing, and the only source of modularity was between the head and the stem. Thus, the results of our study may not be applicable to other bearings, such as MOM, ceramic-on-polyethylene (COP), or to stems incorporating a modular neck. Also, the devices we examined were obtained at revision and as such represent clinical failures. There was a population limitation resulting in moderately low BMI, as a result of the retrieval collection and the restrictions used to match devices within this cohort. Although none of these revisions were performed due to an adverse local tissue reaction due to metal release, they nevertheless may not be representative of the same distribution of fretting and corrosion damage that might be observed on well-functioning implants, which could be assessed via

autopsy retrievals. Additionally, although we have made every attempt to account for the multifactorial nature of the problem we are studying, it was not possible to exactly match all of the confounding factors between the smooth and micro-grooved cohorts, such as head size. We have attempted to mitigate this concern, to the extent possible, by our *post hoc* statistical analysis, in which we sought to identify factors that would explain the variation in fretting and corrosion scores in our two cohorts.

Surface morphology has been speculated to increase fretting corrosion damage due to observations of imprinting within the femoral head bore of micro-grooved tapers [11–13, 16, 18], and the speculation that increased crevices through the grooved texture could increase corrosion. However, previous research of smooth and micro-grooved tapers showed that other confounding factors, including implantation time, stem neck flexural rigidity, apparent length of taper engagement and femoral head size were associated with increased damage and not surface topography [15]. These findings are supported in the literature, in which an increase in implantation time [19], a decrease in stem neck flexural rigidity [10, 19], an increase in apparent length of taper engagement [10, 18, 20], and an increase in femoral head size [10, 18, 21] correlated with increased MACC taper damage. Specifically, flexural rigidity of the stem appears to be an important factor due to the increased flexibility of the trunnion potentially causing more fretting within the interior of the femoral head bore.

The theory of mechanically assisted crevice corrosion relies on the species within the femoral head bore allowing for an acidic corrosive environment; however, it also relies on increased fretting and breakdown of metal oxide layers to expose virgin metal for further corrosion [2]. This theory does not include elements of galvanic corrosion through dissimilar metals. Instead, the theory reports that mechanical breakdown of metal oxide layers through stresses within the oxide through normal gait and fretting at the head-neck junction allows for more corrosion [2]. The present study did not include femoral stem alloy as a criteria for matching the cohort, because it was not statistically significant in the initial investigation of surface finish [15], or within this study cohort. This result is due to the involvement of the elastic modulus of the stem material in the calculation of flexural rigidity [19] accounting for the difference in stem-taper flexibility.

Previous research has shown a strong correlation between lower fretting corrosion scores and material loss [22]. However, a weaker correlation has been observed with the more severely damaged femoral head bores, specifically those with a score of 3 or 4 [23, 24]. To compare the material loss between the smaller cohort of the 14 most highly damaged implants, we chose to describe the maximum depth of material loss of these components by calculating the best fit of the femoral head bore at different z heights along the taper. For this method to be successful, it was required to have as-manufactured surfaces around the circumference of the femoral head bore. This methodology allowed for a more accurate description of clinically relevant tapers, with increased corrosion at the head-neck junction. This method was validated by two independent trained investigators. Accuracy, when comparing measurements from both investigators, was within 0.1 μm . This investigation found that there was no difference between these two groups independent of femoral stem material, supporting the previous interpretations involving the fretting corrosion score.

Observation of chromium rich oxides on the surface of CoCr femoral head bores has been described as part of the protective mechanism for the metals used in total joint arthroplasty [25–27]. However, this protective oxide can be disrupted through fretting or gouging of the surface allowing for exposed virgin metal [26]. In the literature, there have been observations of imprinting, pitting, and discoloration on the surface within the femoral head bore [11, 12, 18, 20, 24]. Bryant et al. [27] observed oxides rich in chromium with cracks forming throughout the oxide within femoral head bores. They also observed clear fretting lines within the femoral head bores. Material loss can naturally occur through the re-passivation of the metal, after fretting breaks the protective oxide layer [2]. This allows for ion release into the joint space [25]. Wang et al. [28] observed the transition between bulk CoCr material and oxide formation through material bands consisting of an oxide layer, transition layer and bulk material. It was thought that the introduction of micro-grooves could allow for crevice formation, altered contact stresses and localized corrosion [18]. Through our investigation, a more apparent fretting mechanism was observed in the femoral head bores paired with micro-grooved stems, compared with the smearing mechanism observed in smooth stems. However, there was no difference in the amount of pitting or fretting within the femoral head bores.

This study describes the similarities of femoral head bore and femoral stem damage between both smooth and micro-grooved femoral stem surface morphologies when controlling for other confounding factors (implantation time, stem neck flexural rigidity, apparent length of taper engagement, and femoral head size). To strengthen the assertion that the damage at the head-neck taper junction was statistically similar between these two cohorts, an investigation of maximum material loss was also performed involving the 14 most severely damaged components. This investigation confirmed the previous findings indicating that there is no difference between femoral head fretting corrosion damage and surface morphology [15]. In conclusion, this study does not support the hypothesis that stem taper surface morphologies are associated with fretting corrosion damage or material release in M-PE total hip arthroplasty.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Figure 1.

A: Macro image of micro-grooved stem taper topography with B, surface profile of micro-grooved trunnion topography. C: Macro image of smooth stem taper topography with D, surface profile of smooth trunnion topography.

Figure 2. Schematic illustration showing the basic orientation of the ion and electron beams used for gathering cross-sectional images of the surfaces examined. SEM, scanning electron microscopy. FIB, focused ion beam.

Figure 3. Exemplar highly worn femoral head bores paired with microgrooved (A) and smooth (B) topography femoral stems. (C) Plot describing the statistically similar fretting corrosion scores across microgrooved and smooth surface topographies. Outliers represented by a circle.

Figure 4. Exemplar highly worn femoral stems with microgrooved (A) and smooth (B) topography stem tapers. (C) Plot describing the statistically similar fretting corrosion scores across microgrooved and smooth surface topographies. Outliers represented by a circle.

Figure 5.
The maximum depth of material loss for microgrooved and smooth femoral stem topography was not statistically different, $P=0.71$.

Figure 6. SEM image sequence (stitch) from the bore opening to the base from a head bore surface from a micro-grooved stem. Red boxes in the low magnification image shown at the bottom of the figure correspond to the areas displayed in the high magnification images connected above. The dark mud-cracked surface layer was found to be rich in chromium oxide.

Smooth

Figure 7. SEM image sequence (stitch) from the bore opening to the base from a head-bore surface from a smooth stem. Red boxes in the low magnification image shown at the bottom of the figure correspond to the areas displayed in the high magnification images connected above. The dark mud-cracked surface layer was found to be rich in chromium oxide.

Figure 8.

A: representative SEM image of the micro-fretting damage (blue arrow) observed in the valleys of the micro-grooved imprints on the head bore surface that was in contact with a micro-grooved stem. B: Representative SEM image of the smeared micro-fretting damage (purple arrows) observed on the head bore surface that was in contact with a smooth stem.

Figure 9. FIB cross-section showing oxide at the top surface of the micro-groove imprinted head-bore surface. The layer on top of the oxide is a platinum layer deposited for ion milling.

Table 1

Clinical and Device Information Corresponding to the Micro-grooved Stem Taper Cohort

Case number	Implantation time (years)	Age at insertion (years)	Sex	BMI (kg/m ²)	Maximum UCLA score	Revision reason	Head size (mm)	Manufacturer	Stem type	Stem Alloy	Flexural rigidity (10 ⁹ N/m ²)
1	0.25	56	M	34.39	6	Infection	44	Zimmer	M/L Taper	Ti Alloy	200.99
2	1.75	67	F	31.89	NA	Infection	32	Depuy	Tri-Lock	Ti Alloy	182.56
3	1.00	75	F	21.95	3	Loosening	36	Depuy	Tri-Lock	Ti Alloy	186.36
4	1.25	67	F	24.89	2	Loosening	32	Depuy	Tri-Lock	Ti Alloy	186.90
5	1.25	NA	F	34.54	NA	Infection	32	Zimmer	Versys FMT	Ti Alloy	195.83
6	1.75	71	F	35.78	7	Infection	32	Zimmer	Versys Fiber Metal Mesh	Ti Alloy	199.26
7	1.75	59	M	27.46	6	Infection	32	Zimmer	VerSys Fiber Metal Taper	Ti Alloy	200.99
8	1.00	80	M	29.68	NA	Infection	36	Zimmer	Versys Fiber Metal	Ti Alloy	200.99
9	1.75	54	F	29.44	4	Loosening	32	Zimmer	M/L Taper	Ti Alloy	202.15
10	1.50	83	M	22.55	8	Infection	32	Zimmer	M/L Taper	Ti Alloy	202.15
11	1.00	62	M	NA	6	Instability	28	Smith and Nephew	Echelon	CoCr	334.87
12	1.75	65	F	22.71	6	Loosening	32	Zimmer	Versys Beaded	CoCr	356.05
13	2.25	49	F	35.08	6	Infection	40	Depuy	Corail	Ti Alloy	180.95
14	2.25	61	F	22.83	7	NA	28	Depuy	Tri-Lock	Ti Alloy	185.27
15	2.00	55	M	33.12	6	Other	36	Depuy	Tri-Lock	Ti Alloy	185.27
16	2.00	61	M	39.60	2	Loosening	32	Zimmer	M/L Taper	Ti Alloy	196.97
17	2.25	63	F	51.38	4	Infection	32	Zimmer	Versys FMT	Ti Alloy	198.11
18	2.00	71	F	34.13	2	Infection	28	Zimmer	Versys Fiber Metal	Ti Alloy	200.99
19	2.00	57	M	10.83	6	Infection	32	Zimmer	M/L Taper	Ti Alloy	201.57
20	2.59	59	F	NA	3	Loosening	36	Zimmer	APR II-T	Ti Alloy	207.43
21	2.00	NA	F	NA	NA	Other	32	Zimmer	APR II-T	Ti Alloy	208.62
22	2.25	68	M	15.07	3	Infection	32	Zimmer	Versys Beaded Fullcoat	CoCr	347.87

Case number	Implantation time (years)	Age at insertion (years)	Sex	BMI (kg/m ²)	Maximum UCLA score	Revision reason	Head size (mm)	Manufacturer	Stem type	Stem Alloy	Flexural rigidity (10 ⁹ N/m ²)
23	3.50	39	M	20.57	NA	Infection	36	Zimmer	Trabecular Metal Taper	Ti Alloy	188.55
24	3.75	57	M	32.28	7	Loosening	32	Zimmer	M/L Taper	Ti Alloy	200.99
25	3.75	57	F	21.43	NA	Loosening	32	Zimmer	Versys Fiber Metal Taper w/ Cactct.	Ti Alloy	201.57
26	3.75	39	M	33.23	3	Loosening	32	Zimmer	M/L Taper	Ti Alloy	201.57
27	3.50	52	M	29.15	8	Infection	32	Zimmer	Versys Beaded	CoCr	364.38
28	4.25	70	F	26.37	4	Loosening	32	Smith and Nephew	Echelon	CoCr	334.87
29	4.25	61	M	24.95	6	Loosening	32	Zimmer	Versys Cemented LD/FX	CoCr	353.99
30	4.50	37	F	27.46	5	Fracture	28	Zimmer	Versys Beaded	CoCr	367.54
31	4.23	78	M	NA	3	Loosening	36	Zimmer	Versys Beaded	CoCr	369.31
32	5.25	46	F	21.48	7	Fracture	28	Depuy	Corail	Ti Alloy	203.90
33	5.25	51	F	NA	2	Loosening	28	Smith and Nephew	Spectron EF	CoCr	333.89
34	5.75	68	M	30.71	NA	Instability	28	Zimmer	Versys Heritage	CoCr	351.94
35	5.00	NA	M	31.47	9	Infection	32	Zimmer	Versys Beaded Fullcoat	CoCr	359.16
36	6.50	43	M	37.25	5	Loosening	28	Zimmer	Versys Fiber Metal	Ti Alloy	197.54
37	6.58	43	M	25.00	5	Infection	28	Zimmer	Versys FMT	Ti Alloy	200.41
38	6.25	NA	M	25.24	6	Infection	28	Zimmer	Versys Cemented	CoCr	355.02
39	7.00	77	F	18.88	NA	Infection	32	Zimmer	Versys Fiber metal Midcoat	Ti Alloy	198.68
40	7.01	56	F	NA	2	Other	36	Zimmer	M/L Taper	Ti Alloy	201.76
41	7.25	83	M	31.42	10	Loosening	32	Zimmer	Versys Cemented	CoCr	352.97
42	8.50	76	M	28.06	9	Fracture	32	Zimmer	M/L Taper	Ti Alloy	201.57
43	8.50	45	M	32.98	8	Loosening	28	Zimmer	Revision Stem	CoCr	352.97
44	8.25	75	F	22.76	6	Infection	28	Zimmer	Advocate Cemented Stem	CoCr	353.99

Case number	Implantation time (years)	Age at insertion (years)	Sex	BMI (kg/m ²)	Maximum UCLA score	Revision reason	Head size (mm)	Manufacturer	Stem type	Stem Alloy	Flexural rigidity (10 ⁵ g/Nm ²)
45	9.00	75	F	30.18	5	Loosening	28	Zimmer	Versys Cemented Collarless Taper	CoCr	349.90
46	9.01	63	F	38.67	6	Pain	28	Zimmer	Versys Cemented	CoCr	353.99
47	9.75	74	M	25.40	6	Infection	32	Zimmer	Versys Beaded	CoCr	362.29
48	10.25	50	F	34.46	2	Infection	36	Zimmer	M/L Taper	Ti Alloy	191.33
49	10.25	64	F	NA	NA	Instability	28	Zimmer	Versys Cemented Plus	CoCr	349.90
50	11.00	63	F	24.00	6	Pain	28	Zimmer	Versys Cemented	CoCr	352.97
51	11.25	70	M	29.53	6	Other	28	Depuy	Solution	CoCr	363.33
52	11.95	73	F	26.29	7	Loosening	28	Depuy	AML	CoCr	495.69
53	12.25	42	M	32.55	9	Infection	32	Zimmer	Versys Fiber Metal Taper w/ Calciect	Ti Alloy	189.11
54	14.01	NA	F	NA	8	Instability	26	Zimmer	Versys Heritage	CoCr	352.29
55	15.25	67	NA	42.97	6	Loosening	28	Zimmer	Versys Cemented Collarless	CoCr	355.02
56	15.75	65	M	30.99	NA	Instability	22	Zimmer	Versys beaded full coat stem	CoCr	365.43
57	16.26	41	F	NA	8	Infection	28	Zimmer	Versys Fiber Metal	Ti Alloy	202.34
58	17.25	66	M	24.14	8	Loosening	28	Depuy	Ultimate-C	CoCr	360.20
59	17.25	68	M	32.49	4	Loosening	28	Depuy	Ultimate-C	CoCr	361.24
60	18.75	39	M	20.23	5	Loosening	32	Dow Corning Wright	McCutchen	Ti Alloy	183.64

Table 2

Clinical and Device Information Corresponding to the Smooth Stem Taper Cohort

Case number	Implantation time (years)	Age at insertion (years)	Sex	BMI (kg/m ²)	Maximum UCLA score	Revision reason	Head size (mm)	Manufacturer	Stem type	Stem Alloy	Flexural rigidity (10 ⁹ N/m ²)
1	0.25	NA	F	34.18	4	NA	32	Stryker	Accolade II	Ti Alloy	123.59
2	0.50	72	M	25.84	1	Loosening	36	Stryker	Secur-Fit Max	Ti Alloy	188.55
3	0.25	75	F	33.13	NA	Infection	32	Stryker	Restoration HA	Ti Alloy	190.22
4	0.25	54	F	NA	5	Infection	32	Stryker	Accolade C	CoCr	222.77
5	0.25	50	M	34.14	5	Infection	40	Stryker	Accolade C	CoCr	223.98
6	0.25	50	M	NA	2	Infection	40	Stryker	Accolade C	CoCr	224.47
7	1.50	42	M	26.94	5	Infection	32	Stryker	Accolade C	CoCr	225.44
8	1.50	59	F	43.40	5	Loosening	28	Stryker	Definition PM	CoCr	259.44
9	1.50	60	F	37.69	NA	Infection	32	Stryker	Premise	CoCr	330.95
10	1.75	71	F	33.07	4	Loosening	28	Stryker	Omnifit EON	CoCr	351.94
11	2.00	NA	F	22.36	4	Loosening	28	Stryker	Accolade C	CoCr	218.92
12	2.25	74	M	25.75	3	Infection	40	Stryker	Accolade C	CoCr	224.71
13	2.50	71	F	31.18	4	Loosening	28	Stryker	Reliance CM	CoCr	230.61
14	2.50	49	M	39.99	NA	Infection	28	Stryker	Citation AT	CoCr	235.87
15	2.25	70	M	24.68	NA	Loosening	36	Stryker	Reliance PF	CoCr	237.39
16	2.50	74	F	28.35	5	Loosening	28	Stryker	Reliance CM	CoCr	261.07
17	2.25	61	M	27.47	NA	Loosening	28	Stryker	Citation AT	CoCr	261.07
18	3.25	NA	F	NA	NA	NA	28	Zimmer	Harris Precoat Plus	CoCr	154.48
19	3.25	71	F	22.46	6	Infection	28	Zimmer	Harris Precoat Plus	CoCr	166.98
20	3.75	68	F	32.92	6	PE Wear	28	Depuy	PFC Cemented	CoCr	185.86
21	3.75	81	F	25.06	5	Loosening	28	Biomet	Answer	CoCr	223.98
22	3.75	73	F	26.63	NA	Loosening	28	Stryker	Definition PM	CoCr	243.54
23	3.25	84	F	28.71	4	Loosening	28	Stryker	Definition PM	CoCr	260.26

Case number	Implantation time (years)	Age at insertion (years)	Sex	BMI (kg/m ²)	Maximum UCLA score	Revision reason	Head size (mm)	Manufacturer	Stem type	Stem Alloy	Flexural rigidity (10 ⁻⁹ /Nm ²)
24	3.75	54	M	33.90	NA	Infection	28	Stryker	PCA E-Series Textured	CoCr	332.90
25	4.25	66	F	19.57	NA	Loosening	28	Zimmer	Harris Precoat Plus	CoCr	165.23
26	4.75	51	M	25.77	6	Other	32	Howmedica	Osteolock	Ti Alloy	184.72
27	4.25	72	F	23.03	NA	Loosening	28	Zimmer	Harris Precoat Plus	CoCr	214.65
28	4.00	56	M	34.18	8	Loosening	28	Biomet	Integral Centralizer	CoCr	278.68
29	4.50	76	F	23.78	NA	Loosening	32	Stryker	PCA E-Series Textured	CoCr	332.90
30	4.25	55	M	24.37	NA	Infection	32	Zimmer	Beaded Stem	CoCr	362.29
31	5.00	75	M	27.28	5	Loosening	32	Stryker	Secur-Fit Plus	Ti Alloy	193.01
32	5.75	41	F	28.32	7	Loosening	28	Stryker	Definition PM	CoCr	259.44
33	5.75	47	F	NA	5	Loosening	32	Biomet	Integral Centralizer	CoCr	303.82
34	5.00	51	M	NA	NA	Loosening	28	Stryker	HNR	CoCr	331.92
35	5.25	69	M	NA	NA	Infection	36	Depuy	Prodigy	CoCr	358.12
36	6.25	46	F	34.78	NA	Infection	32	Stryker	Meridian ST	CoCr	256.20
37	6.25	29	F	NA	NA	Infection	32	Stryker	PCA	CoCr	330.62
38	6.25	59	F	35.61	NA	Loosening	26	Stryker	Precision	CoCr	332.90
39	7.25	NA	F	NA	NA	Loosening	22	Biomet	Tapertloc	Ti Alloy	123.59
40	7.50	50	M	26.36	9	Other	28	Zimmer	Harris Precoat Plus	CoCr	216.78
41	7.50	65	M	18.54	NA	Loosening	28	Zimmer	Harris Precoat	CoCr	227.65
42	8.00	58	M	20.34	4	PE Wear	28	Biomet	Mallory-Head Interlok	CoCr	275.26
43	9.50	72	F	27.17	4	Other	28	Zimmer	Harris Precoat Plus	CoCr	191.63
44	10.00	57	M	25.85	NA	Loosening	28	Zimmer	Harris Precoat Plus	CoCr	156.14
45	10.25	66	M	NA	6	Lysis	28	Biomet	Integral Centralizer	CoCr	295.06

Case number	Implantation time (years)	Age at insertion (years)	Sex	BMI (kg/m ²)	Maximum UCLA score	Revision reason	Head size (mm)	Manufacturer	Stem type	Stem Alloy	Flexural rigidity (10 ⁹ Nm ²)
46	10.75	NA	M	22.32	6	Loosening	32	Stryker	Precision	CoCr	330.95
47	10.25	32	M	24.41	7	Fracture	28	Stryker	HNR	CoCr	334.87
48	10.75	59	M	30.52	9	Fracture	28	Stryker	Precision	CoCr	335.86
49	10.50	73	F	31.71	6	Loosening	32	Stryker	Omnifit Eon	CoCr	344.84
50	10.50	45	F	NA	NA	Loosening	28	Stryker	Omnifit	CoCr	398.42
51	11.76	58	M	NA	NA	Loosening	32	Stryker	PCA E-Series Porous	CoCr	329.97
52	12.76	78	F	NA	NA	Loosening	32	Zimmer	Harris Precoat Plus	CoCr	211.14
53	12.50	62	M	25.53	8	Loosening	26	Stryker	Omnifit	CoCr	400.29
54	14.25	49	M	27.05	10	Pain	28	Kinamed	Reality	CoCr	257.82
55	15.25	50	F	38.73	NA	Infection	32	Stryker	PCA E-Series Porous	CoCr	334.87
56	15.00	60	M	29.83	3	Infection	28	Biomet	Ranawat-Burstein	CoCr	384.75
57	16.25	47	M	32.49	4	Other	28	Zimmer	Harris Precoat	CoCr	167.57
58	17.75	55	M	30.68	4	Loosening	32	Stryker	PCA	CoCr	336.85
59	18.50	81	F	25.66	8	Loosening	28	Stryker	Definition PM	CoCr	336.85
60	19.01	63	F	NA	NA	Fracture	28	Zimmer	Harris Precoat Plus	CoCr	219.64

Table 3

Summary of matching criteria between smooth and micro-grooved cohorts.

Cohort	Average $\pm$ Stdev implantation time (years)	Average $\pm$ Stdev flexural rigidity (10 ⁹ /Nm ²)	Average $\pm$ Stdev apparent length of engagement (mm)	Median head size (mm)
Smooth	6 $\pm$ 5 (Range: 0.25 to 19)	262 $\pm$ 72 (Range: 124 to 400)	12.9 $\pm$ 3.2 (Range: 8 to 25)	28 (Range: 22 to 40)
Micro-grooved	6 $\pm$ 5 (Range: 0.25 to 19)	270 $\pm$ 84 (Range: 181 to 496)	11.9 $\pm$ 1.8 (Range: 7 to 19)	32 (Range: 22 to 44)
p - value	p = 0.68	p = 0.73	p = 0.18	p = 0.04