The clinical implications of metal debris release from the taper junctions and bearing surfaces of metal-on-metal hip arthroplasty

JOINT FLUID AND BLOOD METAL ION CONCENTRATIONS

Aims

We wished to investigate the influence of metal debris exposure on the subsequent immune response and resulting soft-tissue injury following metal-on-metal (MoM) hip arthroplasty. Some reports have suggested that debris generated from the head-neck taper junction is more destructive than equivalent doses from metal bearing surfaces.

Patients and Methods

We investigated the influence of the source and volume of metal debris on chromium (Cr) and cobalt (Co) concentrations in corresponding blood and hip synovial fluid samples and the observed agglomerated particle sizes in excised tissues using multiple regression analysis of prospectively collected data. A total of 199 explanted MoM hips (177 patients; 132 hips female) were analysed to determine rates of volumetric wear at the bearing surfaces and taper junctions.

Results

The statistical modelling suggested that a greater source contribution of metal debris from the taper junction was associated with smaller aggregated particle sizes in the local tissues and a relative reduction of Cr ion concentrations in the corresponding synovial fluid and blood samples. Metal debris generated from taper junctions appears to be of a different morphology, composition and therefore, potentially, immunogenicity to that generated from bearing surfaces.

Conclusion

The differences in debris arising from the taper and the articulating surfaces may provide some understanding of the increased incidence of soft-tissue reactions reported in patients implanted with MoM total hip arthroplasties compared with patients with hip resurfacings.

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Cobalt chromium (CoCr) metal-on-metal (MoM) hip arthroplasty was re-introduced into clinical practice in an attempt to minimise the incidence of component loosening secondary to macrophage mediated osteolysis, which is seen in association with polyethylene debris. However, metal debris which is released from MoM devices is smaller. The immune response to wear particles is largely determined by the composition, morphology and volume of the debris.

It has become apparent that Co and Cr in ionic or particulate form is linked to a spectrum of macroscopic pathologies which have been termed “adverse reaction to metal debris” (ARMD). We therefore hypothesised that debris released from taper junctions of MoM THAs differed from bearing debris in terms of size and composition, with taper debris more representative of the base alloy. When matched for volume, we anticipated that taper debris would be more likely to be associated with greater blood and joint fluid Co concentrations.

We have published our experience with a cohort of patients implanted with Articular Surface Replacement (ASR, DePuy, Warsaw, Indiana) resurfacing and ASR total hip arthroplasties (THAs). These devices share identical articulating bearing surfaces in terms of materials and sizes, but with the THA there is an extra metallic interface – the head-neck taper junction. Compared with ASR resurfacing patients, ASR THA patients experienced double the failure rate secondary to ARMD. Additionally, those reactions usually resulted in more extensive tissue necrosis, despite THA patients generally being exposed to one third the total volume of CoCr debris as resurfacing patients.

We therefore hypothesised that debris released from taper junctions of MoM THAs differed from bearing debris in terms of size and composition, with taper debris more representative of the base alloy. When matched for volume, we anticipated that taper debris would be more likely to be associated with greater blood and joint fluid Co concentrations.
Patients and Methods

Since 2008 at our centre, all patients who experience failure of MoM hip prostheses routinely undergo pre-revision blood and hip joint synovial fluid Co and Cr ion measurements. Tissue samples excised at revision surgery from multiple sites surrounding the prosthesis are analysed by a pathologist (SN) with extensive experience in MoM cellular responses using published methodology. Subsequently, we have analysed explanted prostheses to determine the volumetric loss of material from bearing and female taper surfaces which had occurred in vivo. 3

Given that seven variables in total were being investigated (Table I), a number of which were categorical and potentially the subject of bias, we believed that we would need over 80 complete data sets in each of the resurfacing and THA groups to derive meaningful results. Approximately 40 to 50 complete data sets were received per year of the study, therefore a four-year cut off was agreed upon, covering the period from 2010 to 2014. Ethical approval was granted by Durham and Tees Valley 2 local research ethics committee and was sponsored by Newcastle upon Tyne Hospitals NHS Foundation Trust. The MoM prostheses included the ASR, Birmingham Hip Resurfacing (BHR; Smith and Nephew, Memphis, Tennessee) and the 36 mm Pinnacle THA (DePuy). All metal components were manufactured from the same high carbon content CoCr molybdenum (Mo) medical grade alloy. There were, however, differences in the manufacturing processes, which are described in Table II. The Pinnacle 36 mm THA also employs a modular acetabular system in contrast to the other designs, which are monoblock. The CoCr bearing comprised part of a THA and the stem used to support the modular head was either a Synergy (Smith & Nephew; for use with BHR) or Corail/S-ROM (DePuy; for Pinnacle and ASR devices), both of which were uncemented and manufactured from titanium alloy. A total of 199 hips in 177

<table>
<thead>
<tr>
<th>Component</th>
<th>CoCrMo alloy</th>
<th>As cast versus wrought</th>
</tr>
</thead>
<tbody>
<tr>
<td>ASR bearing surfaces</td>
<td>ASTM F75</td>
<td>As cast</td>
</tr>
<tr>
<td>ASR taper sleeve</td>
<td>ASTM F799</td>
<td>Wrought</td>
</tr>
<tr>
<td>BHR bearing surfaces</td>
<td>ASTM F75</td>
<td>As cast</td>
</tr>
<tr>
<td>BHR taper</td>
<td>ISO 5832-12</td>
<td>Wrought</td>
</tr>
<tr>
<td>Pinnacle bearing surfaces</td>
<td>ASTM F1537</td>
<td>Wrought</td>
</tr>
<tr>
<td>Pinnacle taper</td>
<td>ASTM F1537</td>
<td>Wrought</td>
</tr>
</tbody>
</table>

American Society for Testing and Materials (ASTM) F-75 and ASTM F1537 are similar alloys, with compositions by weight F-1537: chromium (Cr) 26% to 30%; molybdenum (Mo) 5% to 7%; cobalt (Co) balance and for F-75: Cr 27% to 30%; Mo 5% to 7%; Co balance. ASTM F799 has an identical composition to F-1537.

BHR, Birmingham Hip Resurfacing; ASR, Articular Surface Replacement (DePuy, Warsaw, Indiana)
patients (116 THAs and 83 hip resurfacings; 132 female hips) were analysed which had corresponding blood, joint and tissue samples. Patient demographic, clinical parameters and implant details are shown in Table III.

**Wear analysis.** Explanted prostheses were analysed using a coordinate measuring machine (Legex 322; Mitutoyo, Halifax, United Kingdom) to calculate the total amount of material that has been removed from the components in vivo. This material loss can be expressed in volumetric terms either as ‘total volumetric wear (in mm$^3$)’ or this total value can be divided by the number of years in vivo to provide a mean ‘volumetric wear rate’ (expressed in mm$^3$/year). The accuracy of such methods has been validated and is of the order of 0.5 mm$^3$ per component for bearings and 0.2 mm$^3$ for tapers.5 Throughout this paper, wear rates refer only to volumetric CoCr material loss. For resurfacings, ‘total volumetric wear rates’ refer to the bearing surface wear rates (combined head and acetabular component volumetric wear rates). For THAs, ‘total volumetric wear rates’ refer to the combined wear rates of the head, the acetabular component and the female taper surface.

**Assessment of metal particle load in tissues.** Elsewhere, it has been concluded that average particle sizes released from MoM hips range from about 30 nm to 100 nm.6 To our knowledge, it is unclear whether it is the size and morphology of particles in solution, or the particles which aggregate and are deposited in local tissues, which are the determinants of the subsequent immune response. As the primary purpose of this paper was to establish relative differences in the periprosthetic environment related to variations in wear rates and the source of wear, we believed it was legitimate for this study to report the size of precipitated aggregated particles as graded when examined by a pathologist (SN) under light microscopy. This was assessed by the method used to assess tissue iron-overload in liver biopsies (Table IV).7 We have described these approaches in detail in an earlier paper which focused on the cellular response involved in the ARMD spectrum.4

### Table III. Patient and implant details and distributions of measured parameters

<table>
<thead>
<tr>
<th>Patients and implants</th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>Mated explanted components (n)</td>
<td>199</td>
</tr>
<tr>
<td>Mean age (yrs) (range)</td>
<td>58 (21 to 84)</td>
</tr>
<tr>
<td>Male:female (n)</td>
<td>67:132</td>
</tr>
<tr>
<td>Mean (range) time to revision (mths)</td>
<td>58 (8 to 109)</td>
</tr>
<tr>
<td>Unilateral vs bilateral</td>
<td>155:44</td>
</tr>
<tr>
<td><strong>Device</strong></td>
<td><strong>(n, %)</strong></td>
</tr>
<tr>
<td>ASR</td>
<td>68 (34)</td>
</tr>
<tr>
<td>Pinnacle THA 36 mm</td>
<td>75 (38)</td>
</tr>
<tr>
<td>ASR THA</td>
<td>37 (19)</td>
</tr>
<tr>
<td>BHR</td>
<td>15 (7)</td>
</tr>
<tr>
<td>BHR THA</td>
<td>4 (2)</td>
</tr>
<tr>
<td><strong>Reason for revision</strong></td>
<td><strong>(n, %)</strong></td>
</tr>
<tr>
<td>ARMD</td>
<td>187 (94)</td>
</tr>
<tr>
<td>Loose components</td>
<td>6 (3)</td>
</tr>
<tr>
<td>Infection</td>
<td>2 (1.0)</td>
</tr>
<tr>
<td>Unexplained pain</td>
<td>4 (2.0)</td>
</tr>
<tr>
<td>Median (range) bearing surface volumetric wear rate (mm$^3$/yr): hip resurfacings</td>
<td>7.35 (0.62 to 95.5)</td>
</tr>
<tr>
<td>Median (range) bearing surface volumetric wear rate (mm$^3$/yr): THAs</td>
<td>2.02 (0.27 to 68.9)</td>
</tr>
<tr>
<td>Median taper wear rate (mm$^3$/yr) (range)</td>
<td>0.20 (0.01 to 8.34)</td>
</tr>
<tr>
<td>Median blood Co levels (μg/l) (range)</td>
<td>9.60 (0.70 to 2710)</td>
</tr>
<tr>
<td>Median blood Cr levels (μg/l) (range)</td>
<td>9.90 (1.50 to 123.2)</td>
</tr>
<tr>
<td>Median joint fluid Co to Cr ratio (μg/l) (range)</td>
<td>0.69 (0.04 to 38.5)</td>
</tr>
<tr>
<td>Median joint fluid Co levels (μg/l) (range)</td>
<td>926 (13.0 to 46433)</td>
</tr>
<tr>
<td>Median joint fluid Cr levels (μg/l) (range)</td>
<td>894.4 (12.5 to 133120)</td>
</tr>
<tr>
<td>Fluid grades (see text)</td>
<td>0 = 3 (1.5); 1 = 41 (20.6); 2 = 53 (26.6); 3 = 106 (51.3)</td>
</tr>
<tr>
<td>Particle sizes*</td>
<td>0 = 9 (4.5); 1 = 60 (30.5); 2 = 87 (43.7); 3 = 28 (14.1); 4 = 15 (7.5)</td>
</tr>
</tbody>
</table>

Particle size refers to the agglomerated particle size grade in Table IV

BHR, Birmingham Hip Resurfacing (Smith and Nephew, Memphis, Tennessee; ASR, Articular Surface Replacement (DePuy, Warsaw, Indiana); ARMD, adverse reaction to metal debris; Co, cobalt; Cr, chromium; THA, total hip arthroplasty

### Table IV. Aggregated metal particle size assessment

<table>
<thead>
<tr>
<th>Particle grade</th>
<th>Ease of observation and magnification (eyepiece x objective lens)</th>
<th>Approximate size range (microns)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>Granules absent or barely discernable x400</td>
<td>&lt; 0.5</td>
</tr>
<tr>
<td>1</td>
<td>Barely discernable x250, easily confirmed x400</td>
<td>0.5 to 1</td>
</tr>
<tr>
<td>2</td>
<td>Discrete granules resolved x100</td>
<td>1 to 2</td>
</tr>
<tr>
<td>3</td>
<td>Discrete granules resolved x25</td>
<td>10 to 20</td>
</tr>
<tr>
<td>4</td>
<td>Masses visible x10, naked eye</td>
<td>&gt; 100</td>
</tr>
</tbody>
</table>

BHR, Birmingham Hip Resurfacing (Smith and Nephew, Memphis, Tennessee; ASR, Articular Surface Replacement (DePuy, Warsaw, Indiana); ARMD, adverse reaction to metal debris; Co, cobalt; Cr, chromium; THA, total hip arthroplasty
Joint fluid Co and Cr concentrations. Prior to revision surgery, joint fluid was extracted under local anaesthetic in order to rule out infection and to analyse the Co and Cr concentrations using the same technique and equipment as previously described in the collection of blood specimens.\textsuperscript{3} Samples did not undergo acid digestion prior to analysis.

Macrosopic revision findings. The amount of fluid observed at revision was categorised as described in our previous publication;\textsuperscript{8} 0, no abnormal fluid; 1, small amount of abnormal fluid; 2, copious amounts of abnormal fluid; 3, fluid under pressure/fistulation of fluid.

Statistical analysis. Initially, the distribution of blood and joint Co and Cr concentrations, as well as volumetric wear rates from the bearing and taper surface were examined using the Shapiro-Wilk test. All values were determined to be non-parametrically distributed (p < 0.001 for all tests) and were therefore log normalised. A number of multiple regression models (including stepwise and hierarchical approaches) were constructed in order to investigate whether any differences in the outcome variables were brought about due to differences in the source of the metal debris (Table I). As can be seen in Figure 1, taper debris, generated from a smaller surface interface than that of the bearing surfaces, was generally of a lower magnitude. To account for this, the proportion of the total volumetric wear rate that was contributed by the taper surface was calculated. For example, if the total wear rate was 10 mm\textsuperscript{3}/year and the taper wear rate was 1 mm\textsuperscript{3}/year, the value entered into the regression model was 0.1. Statistical analysis was carried out using Minitab v17 software (Minitab Ltd., Coventry, United Kingdom). There is limited evidence on this subject, but we believed it reasonable to include patient age and gender for all analyses as metal ions are excreted renally and renal function may decline with age,\textsuperscript{9} and some authors have noted a link between patient gender and ion concentrations.\textsuperscript{10} The observed particle grade in retrieved tissues was also included, as the surface area of agglomerated particles may affect rate of ionic release secondary to corrosion.\textsuperscript{11} For the joint fluid analysis, fluid grade was included as a variable in order to rule in or out effects secondary to dilution.

The various multiple regression modelling approaches returned consistent findings. The models reported are those

\begin{figure}
\centering
\includegraphics[width=\textwidth]{fig1.png}
\caption{A box and whisker plot showing total cobalt-chromium (CoCr) wear volumes generated from the bearing and female taper surfaces. The box lengths represent the interquartile range, the transverse line within the box the median value and the whiskers the range of the data, beyond which points are considered outliers. Nine resurfacing values and two combined total hip arthroplasty (THA) values above 250 mm\textsuperscript{3} were removed for illustrative purposes. The median resurfacing wear volume was 29.2 mm\textsuperscript{3}, the median THA bearing volume was 6.38 mm\textsuperscript{3}, the median taper wear was 1.03 mm\textsuperscript{3} and the median combined THA wear volume (bearing+taper) was 9.74 mm\textsuperscript{3}.}
\end{figure}
that best described the response variables where all the coefficients of the explanatory variables were significantly different from zero with significance drawn at a p-value < 0.05. Beta (\( \beta \)) standardised coefficients are reported as \( \beta \) with standard error as SE.

**Results**

**Part 1 – relationship of taper debris with different agglomerated particle sizes in retrieved tissues and bearing surface wear.**

In the ordinal logistic regression model, total wear rate appeared to have the strongest influence on agglomerated particle size (\( \beta = 0.339, \ SE = 0.071 \ p < 0.001 \)), with greater wear rates being associated with larger observed particle sizes. There was a trend towards smaller observed particle sizes when a greater proportion of the total volume of wear was contributed by the taper (\( \beta = -0.194, \ SE = 0.074, \ p = 0.014 \)). These two variables accounted for approximately 21% of the overall variation. Patient gender (\( p = 0.899 \)), age (\( p = 0.184 \)), device type (\( p = 0.100 \)) and fluid grade (\( p = 0.768 \)) were not found to be significant factors.

Smaller particle grades, representing smaller agglomerates, were associated with larger Co:Cr ratios in the corresponding joint fluid samples (Fig. 2).

**Part 2 – relationship of taper debris with different metal ion concentrations in the joint fluid and bearing surface wear**

**Joint fluid Cr.** In the best fitting regression model, total wear rate was the dominant variable (\( \beta = 0.716, \ SE = 0.05, \ p < 0.001 \)). A greater proportion of wear contributed by the taper was associated with a significantly lower joint Cr concentration (\( \beta = -0.107, \ SE = 0.050, \ p = 0.03 \)), as was male gender (\( \beta = -0.150, \ SE = 0.047, \ p = 0.002 \)). These two variables, however, added relatively little to the R² value derived from the total wear rate on its own (R² = 58% versus R² = 55%). Fluid grade (\( p = 0.207 \)), device type (\( p = 0.173 \)), agglomerated particle size grade \( p = 0.660 \) and patient age (\( p = 0.266 \)) were not found to be significant.

**Joint fluid Co.** In the best fitting regression model, a greater proportion of taper wear was not associated with a significantly higher Co concentration. Total volumetric wear rate (\( \beta = 0.703, \ SE = 0.060, \ p < 0.001 \)), fluid grade (\( \beta = 0.112, \ SE = 0.04, \ p = 0.05 \)) and female gender (\( \beta = 0.114, \ SE = 0.053, \ p = 0.040 \)) were the only variables significantly associated with greater Co concentrations.

Part 3 – relationship of taper debris with different metal ion concentrations in the blood and bearing surface wear (unilateral patients only).

**Blood Cr.** A greater proportion of taper wear was associated with a lower blood Cr concentration (\( \beta = -0.128, \ SE = 0.037, \ p = 0.003 \)). As with the joint fluid model, the dominant variable to explain blood Cr concentrations was clearly total volumetric wear rate (\( \beta = 0.874, \ SE = 0.041, \ p < 0.001 \)). Female gender was associated with greater Cr concentrations (\( \beta = 0.095, \ SE = 0.037, \ p = 0.045 \)), as was fluid grade (\( \beta = 0.082, \ SE = 0.038, \ p = 0.031 \)). The presence of the ASR resurfacing device was associated with lower Cr concentrations (\( \beta = -0.148, \ SE = 0.037, \ p < 0.001 \)). However, the proportion of taper wear, female gender and the presence of the ASR resurfacing device added relatively little to the R² value derived from the total wear rate on its own (R² = 80% versus R² = 76%). Particle size (\( p = 0.250 \)) and patient age (\( p = 0.804 \)) had no significant effect.

It was observed that as total wear rates increased, the ratio between the joint fluid Cr concentrations and the equivalent Cr concentrations in blood samples also increased. This implied that Cr was retained in the joint. In order to investigate this, a further regression analysis was performed on the joint Cr:blood Cr ratio versus combined wear rates and particle sizes. The equation for this model was as follows: JointCr:bloodCr = 1.58 + 0.06 * particle grade + 0.57 * log total wear rate (R² = 24%, \( p < 0.001 \); * = multiply). This confirmed that as wear rates increased, joint Cr concentrations became proportionately larger than the blood Cr concentration. This effect was not seen with Co (Fig. 3).

**Blood Co.** A greater proportion of taper wear was not significantly associated with a higher blood Co concentration. Only total volumetric wear rate (\( \beta = 0.927, \ SE = 0.039, \ p < 0.001 \)), the ASR resurfacing device (\( \beta = -0.162, \ SE = 0.042, \ p < 0.001 \)) and fluid grade (\( \beta = 0.130, \ SE = 0.036, \ p < 0.001 \)) were found to be significant.
variables. Again, the full best fitting regression model provided only a marginal increase in the power of the model to explain the variation in blood Co compared with total volumetric wear rate on its own ($R^2 = 82\%$ versus $R^2 = 78\%$).

Does metallurgy play a role? Having established that particle size and joint fluid grade had minimal impact on measured blood Cr and Co concentrations, we then drew from a larger database of results from the Northern Retrieval Registry, which conducts wear analysis on explants retrieved from external hospital trusts across the United Kingdom. These datasets do not typically include joint fluid/particle observations, however, the advantage of this further analysis was to allow us to select only unilateral Pinnacle devices, thereby eliminating the potential confounding effect of as cast versus wrought CoCr components (Table II). This left us with 107 explanted 36 mm diameter Pinnacle wear results with corresponding blood Cr and Co concentrations, we then drew

The main findings of this investigation were as follows: As the proportion of wear debris contributed by the taper increased, there was a relative reduction in Cr concentrations in the corresponding blood and joint fluid samples.
This implied either that taper debris is composed of less Cr relative to bearing surface debris, or that taper debris may have a greater tendency to precipitate out of solution.

Agglomerated particulate sizes in tissue samples tended to increase with increases in total wear rates, although they tended to be smaller when there was a greater source contribution from the taper. This implied that taper debris may be of a different morphology to bearing surface debris.

As bearing surface wear rates increased, joint fluid Cr concentrations rose disproportionately to blood Cr concentrations, implying the accumulation of Cr in the periprosthetic environment with high rates of wear.

Patient gender, age and the presence of joint fluid effusions appear to affect the body’s ability to process Cr and Co debris.

MoM bearing surface wear. In the current study, as bearing surface wear rates increased, Co concentrations became larger than the equivalent Cr concentrations in the blood (Fig. 3). The reverse was true in the joint fluid, where Cr concentrations, in general, became larger than Co as bearing wear increased. Concomitantly, the ratio of joint fluid Cr to blood Cr concentration became larger. Furthermore, as bearing surface wear rates increased, agglomerated particulate sizes in the tissues tended also to increase, this observation being consistent with a previous simulator study14 which showed an increase in particle size as wear rates increased. Taken in combination, these findings imply that larger particles precipitate, become trapped and slowly corrode, producing massive local concentrations of Cr ions in the joint fluid. Co, with greater solubility, more easily diffuses into the blood stream. The strongest evidence of this effect is the relative clearance rate of the two elements following removal of MoM prostheses.15

Taper wear. The taper junction, which, in this study featured connections composed of a CoCr modular head press-fitted onto a titanium stem, has a smaller surface area, and thus material release is generally less in total volumetric terms.16 We have previously shown that the CoCr surface in these type of tapers is unable to sustain a Cr oxide layer of sufficient thickness, leaving the underlying bulk alloy vulnerable to wear.17 Bulk CoCr alloy is composed of material with an approximate Co to Cr ratio of 2.1: 1 by weight.11 For this reason, prior to this study we believed that it was likely the absence of a thick oxide layer at the taper surface which led to the generation of metal ion concentration was more representative of the composition of the bulk alloy (Co > Cr). In a review from
Kwon et al., some authors have suggested that an elevated Co to Cr ratio in the bloodstream is diagnostic of taper failure and due to preferential leaching of Co from the implant surface secondary to corrosion. But the apparent disproportionate release of Co beyond that present in the alloy appears to be illusory. The observed clinical effect appears to be due to a reduction in Cr release into solution from the taper junction, rather than an increase in Co release secondary to a corrosive mechanism specific to the taper. When matched for volume, taper debris shows an insignificant tendency to increase blood and synovial fluid Co ion concentrations to a greater degree than bearing debris. However, the most striking biochemical difference between the two sources of debris is the significantly lower concentration of Cr in the blood and fluid samples of prostheses with failing tapers, even after the volume of metal release has been accounted for. The most likely explanation to explain these findings therefore is the precipitation from solution of Cr (Figs 4 and 5). This finding is given support from simple practical experience, where severe taper damage is often associated with heavy deposition of solid metal debris in the area of maximal damage (Fig. 6). These deposits have been shown by ourselves and other investigators to be Cr rich, and unrepresentative of the base CoCr alloy. We speculate that this tendency to form solid precipitates is due to the absence of a thick oxide layer and the immediate reduction of Cr ions released through tribocorrosion in the restricted environment of the taper connection.

The role of titanium debris in THA patients has not been explored. We have previously shown that titanium stem tapers in general release very small amounts of material when combined with CoCr heads. However, whether the presence of titanium affects either the nature of the CoCr debris which is released, or the host’s processing of the released Co and Cr ions, is unknown. It is known that metal ions do compete for albumin binding sites, which is the main protein involved in the binding of free ions.

The question arises whether blood metal ion screening be used to detect taper failure. In general, excessively wearing bearing surfaces lead to Cr accumulation in the peri-prosthetic fluid. Excessive taper wear tends to be associated with a Co rich fluid content, likely due to the precipitation from a solution of Cr. Both situations result in an elevated Co:Cr ratio in the blood stream, rendering interpretation of blood Co:Cr ratios unreliable. In future studies, it is likely that the interpretation of blood Co:Cr ratios in a patient suffering from taper failure in metal on polyethylene will prove to be a more straightforward process.

Take home message:

Metal debris released from the taper junction appears to create a different periprosthetic environment than debris released from the bearing surfaces. This may explain the difference in failure rates between hip resurfacing devices and their stemmed equivalents.

Author contributions:

R. P. Sidaginamale: Designing the study, Data collection, Writing the paper.
T. J. Joyce: Designing the study, Data collection, Writing the paper.
J. G. Bowsher: Interpreting the results, Writing the paper.
J. K. Lord: Designing the study, Data collection, Writing the paper.
P. Avery: Interpreting the results, Writing the paper.
S. Natu: Designing the study, Data collection, Writing the paper.
A. V. F. Nargol: Designing the study, Data collection, Writing the paper.
D. J. Langton: Designing the study, Data collection, Writing the paper.

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J. G. Bowsher reports Critical Path funding received, which is related to this article.
D. J. Langton, A. V. F. Nargol, S. Natu and T. J. Joyce have received fees as expert witnesses for plaintiffs in metal-on-metal litigation which is related indirectly to the contents of this paper.
A. V. F. Nargol and D. J. Langton have a lawsuit in progress in the United States indirectly to the subject of this article.

The author or one or more of the authors have received or will receive benefits for personal or professional use from a commercial party related directly or indirectly to the subject of this article.

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References


