Friction of Total Hip Replacements With Different Bearings and Loading Conditions

Claire Brockett,1 Sophie Williams,1 Zhongmin Jin,1 Graham Isaac,2 John Fisher1

1 Institute of Medical and Biological Engineering, School of Mechanical Engineering, University of Leeds, Leeds LS2 9JT, United Kingdom
2 DePuy International Limited, Leeds, United Kingdom

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Abstract: Metal-on-ultra-high molecular weight polyethylene (UHMWPE) total hip replacements have been the most popular and clinically successful implants to date. However, it is well documented that the wear debris from these prostheses contributes to osteolysis and ultimate failure of the prosthesis, hence alternative materials have been sought. A range of 28 mm diameter bearings were investigated using a hip friction simulator, including conventional material combinations such as metal-on-UHMWPE, ceramic-on-ceramic (CoC), and metal-on-metal (MoM), as well as novel ceramic-on-metal (CoM) pairings. Studies were performed under different swing-phase load and lubricant conditions. The friction factors were lowest in the ceramic bearings, with the CoC bearing having the lowest friction factor in all conditions. CoM bearings also had low friction factors compared with MoM, and the trends were similar to CoC bearings for all test conditions. Increasing swing phase load was shown to cause an increase in friction factor in all tests. Increased serum concentration resulted in increased friction factor in all material combinations, except MoM, where increased serum concentration produced a significant reduction in friction factor.

INTRODUCTION

Metal-on-ultra-high molecular weight polyethylene (UHMWPE) (MoP) hip replacements are currently the most commonly implanted and most clinically successful long-term bearings, with a survivorship of over 75% at 20 years follow-up for cemented bearings.1 However, it is well documented that polyethylene wear debris contributes to osteolysis, resulting in loosening of the prosthesis and ultimate failure.2,3 The onset of osteolysis is closely linked to both the volume and size of the wear debris produced. Particles produced from the UHMWPE bearing surface are predominantly in the biologically active size range (0.1–1 μm), hence alternative materials have been sought.4 Additionally, the volume of debris generated is inversely proportional to the hardness of the material, hence hard-on-hard bearings have also been investigated.5 In vitro wear studies have shown metal-on-metal (MoM) and ceramic-on-ceramic (CoC) bearings to have a lower wear rate compared with the UHMWPE bearings, and the wear debris particles generated are significantly smaller than those of UHMWPE.5

Early MoM implants experienced variable results, with some implants surviving up to 30 years in vivo with exceptionally low wear rates, yet some failing due to high frictional torques.7,8 It is generally accepted that the failure was due to the manufacturing methods at the time, leading to poor tolerances and surface finish. Improvements in design and manufacture led to a resurgence in the use of MoM bearings, with improved wear performance seen both in vitro and in vivo.9–13 The frictional torques generated within the new-generation well-designed MoM implants are significantly less than that required to cause mechanical loosening. Wimmer et al. highlighted several cadaveric studies which demonstrated that the torque required to incur instant loosening of the acetabular cup ranged between 6.8 and 170 N m, dependent upon the design and fixation of the cup.14 CoC bearings, with superior wear resistance and surface finish, have lower wear rates than both MoM and MoP implants, but despite improvements in mechanical properties they are still susceptible to fracture. Ceramic wear debris has significantly less toxicity than MoM debris, and is less biologically active than UHMWPE particles.2,4,15 An in vitro study of novel differential hardness ceramic-on-metal (CoM) implants have shown reduced wear rates compared with MoM bearings.16

Keywords: friction; total hip replacement; lubrication
Theoretical lubrication studies have proposed that MoP bearings operate towards the boundary end of the mixed lubricating regime\(^{17}\), suggesting that the load is primarily carried by solid asperity contact. Further studies examining MoM and CoC bearings have shown that both operate towards the fluid film lubrication end of the mixed lubricating regime, where the bearing surfaces are mostly separated by the lubricant. Fluid film lubrication is more likely to be achieved in CoC bearings\(^{18,19}\), with MoM operating in a mixed lubrication regime. Direct measurement of the contact between the head and cup during in vitro wear testing, using an electrical resistance method, demonstrated that complete separation of the two bearing surfaces was possible, such that there was no contact between surface asperities during part of the cycle with both MoM and CoC hip replacements\(^{20,21}\). The MoM study demonstrated that separation of the head and cup surfaces was possible during the swing phase of the gait cycle; however, asperity contact always occurred during the stance phase when the load was applied, indicating that a mixed lubricating regime was prevalent. Recent theoretical and experimental studies examining the effect of swing phase load, which may vary in vivo due to joint laxity following surgery, on the lubrication of MoM bearings have suggested that increased swing phase load may result in reduced film thickness\(^{22}\). Theoretical predictions indicated that a higher swing phase load would reduce the fluid film, which was consistent with the experimental studies that indicated higher wear and friction with increased swing phase load. The theoretical lubrication studies are generally difficult to validate directly and alternative experimental studies of friction have been carried out.

Studies of friction in total hip replacements have been carried out using both free and driven pendulum machines. The head and cup of a hip replacement act as the fulcrum of a pendulum within the free pendulum studies, with the desired load attached to the pendulum. The pendulum swing is initiated from a known amplitude, and the decay of the swing monitored. Frictional damping forces occurring between the head and cup resist the motion of the pendulum, causing the decay in swing amplitude and ultimately halting the motion. Chamley determined that a linear decay indicated boundary lubrication, with increasing time to decay indicating improved lubrication\(^{23}\). Early studies demonstrated MoP implants had superior frictional properties compared with MoM\(^{23,24}\). Further studies, using the driven pendulum, have examined the effects of material combination and lubricant upon the friction within a bearing. The driven pendulum includes a head/cup assembly, where a known load and displacement are applied cyclically to the implant. Rather than initiating the swing and monitoring the decay, motion and loading profiles, representative of in vivo conditions, are maintained for a required number of cycles, and the frictional forces acting within the implant are measured directly. CoC bearings were shown to have the lowest friction factors, with MoM experiencing the highest friction\(^{25,26}\). Previous studies have used synthetic lubricants with artificially elevated viscosity to demonstrate the conditions that may be achieved during full fluid film lubrication. However, although the exact constituents of synovial fluid may vary, they do not contain these synthetic molecules, hence such studies do not replicate the conditions that the implant would operate within in vivo. The viscosity of the biological lubricants used within simulator studies and found within the body has been found to be higher than water\(^{27}\). Furthermore, one of the key variables in vivo is the protein concentration, and this has been shown to be important in determining friction and wear\(^{12}\). Additionally, as all bearing combinations are operating in different regions of the mixed regime, the role of the protein as boundary lubricant is critical. Indeed, the ability of the improved fluid lubrication to enhance the availability of boundary molecules has not yet been investigated.

The aim of this study was to compare the friction of several clinically available 28 mm diameter bearing combinations under different swing-phase load and lubricant conditions, and contrast these results with the novel CoM hip replacements.

### Materials and Methods

**Materials**

Five material combinations, of a nominal 28 mm diameter, supplied by DePuy International Ltd, UK, were tested. The component bearings are readily available and in current clinical use, with the exception of the CoM bearing, which is currently being evaluated clinically. Prior to testing, each sample was measured for dimension and surface roughness (Ra) using a coordinate measuring machine (Kemco, UK) and two dimensional contacting profilometry (Form Talysurf Series, Taylor-Hobson, Leicester, UK), respectively. The bearing combinations tested and the results for these initial measurements are shown in Table I.

<table>
<thead>
<tr>
<th>Head</th>
<th>Cup</th>
<th>Bearing Combination</th>
<th>No. of Samples</th>
<th>Mean Radial Clearance (mm)</th>
<th>Head Ra (μm)</th>
<th>Cup Ra (μm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>CoCrMo</td>
<td>CoCrMo</td>
<td>MoM</td>
<td>6</td>
<td>0.029</td>
<td>0.011</td>
<td>0.009</td>
</tr>
<tr>
<td>CoCrMo</td>
<td>UHMWPE</td>
<td>MoP</td>
<td>4</td>
<td>0.132</td>
<td>0.010</td>
<td>0.752</td>
</tr>
<tr>
<td>Alumina</td>
<td>UHMWPE</td>
<td>CoP</td>
<td>4</td>
<td>0.123</td>
<td>0.004</td>
<td>0.752</td>
</tr>
<tr>
<td>Zirconia-toughened alumina</td>
<td>CoCrMo</td>
<td>CoM</td>
<td>4</td>
<td>0.034</td>
<td>0.003</td>
<td>0.009</td>
</tr>
<tr>
<td>Alumina</td>
<td>Alumina</td>
<td>CoC</td>
<td>4</td>
<td>0.030</td>
<td>0.004</td>
<td>0.005</td>
</tr>
</tbody>
</table>
Friction Test Method

Friction testing was performed using a pendulum friction simulator (Simulator Solutions, Manchester, UK), a single-station servo-hydraulic machine controlled by a personal computer via a graphic user interface, which can apply a dynamic loading cycle, similar to that experienced by implants in vivo (Figure 1). A fixed frame mounted on two pressurized hydrostatic bearings formed the friction carriage. This allowed the friction within the carriage to be considered negligible, as it was two orders of magnitude smaller than the friction within the implant, hence all measured frictional torque can be assumed to be between the bearing surfaces of the implant. A piezoelectric transducer connected to the front of friction carriage determined the frictional torque within the system, by measuring the forces transferred between the fixed frame and the carriage. Load and displacement were applied through the femoral head via the loading frame and motion arm, respectively. The implants were inverted with respect to anatomical position. It was essential to eliminate experimental inaccuracies, hence the centre of rotation of the head and cup were carefully aligned with the centre of rotation for the motion arm. The friction carriage was self-aligning to ensure correct positioning of the implant during test.

Tests were performed with a flexion-extension of \( \pm 25^\circ \) (\( \pm 1^\circ \)), at a frequency of 1 Hz. A simple sinusoidal waveform was used through 60\% of the cycle to apply a dynamic load with a peak of 2 \( \pm 0.1 \) kN, and different swing phase loads of 25, 100, and 300 N. An example of loading-motion profile is shown in Figure 2. Lubricants used were distilled water, 25\% (w/v) bovine serum (with 0.1\% sodium azide), and 100\% bovine calf serum (Harlan Sera-Lab, Loughborough, UK). Protein concentration was 15.46 g/L in the 25\% bovine serum and 62.43 g/L in the 100\% bovine serum, with albumin being the primary protein constituent. Each test was performed in a forward and reverse direction, using the same loading conditions, to eliminate any residual error from misalignment of the bearing components.\(^{26,29}\) All tests were performed at room temperature. Lubricant was changed and the bearings were cleaned between each test.

The frictional torques during the peak load, peak velocity phase of the cycle in forward and reverse directions were used to calculate the true frictional torque (Eq. (1)). The frictional torque was given by

\[
T_t = \frac{(T_f - T_r)}{2},
\]

where \( T_t \) is the true torque, while \( T_f \) and \( T_r \) are the frictional torques measured during the forward and reverse directions of swing, respectively. The friction factor was calculated from the true frictional torque as follows:

\[
f = \frac{T_t}{R L_p},
\]

The friction factor defined before is similar in magnitude to the coefficient of friction, but variable with the finite contact area, \( R \) is the bearing radius in m, and \( L_p \) is the peak load in N. Tests were run for a minimum of 180 cycles, and the data were logged at 30 cycle intervals for 5 cycles. Frictional torque was measured over five points around the peak load, high velocity phase of the cycle. Data was analyzed after 120 cycles, where the friction had stabilized.

Initial studies to determine the sensitivity of the friction simulator and the repeatability of the data were performed prior to this study. The sensitivity of the simulator was assessed by measuring the friction factor of the low friction CoC bearing tested in a thick silicone oil (viscosity = 5000 cs). The friction factor measured was 0.005. This has been the lowest level of friction measured with this system. At lower levels, frictional torque induced through changes in load distribution and load vector may influence the measurement.

MoP and CoC bearings were subject to repeated testing to assess the repeatability of the friction measured. The results achieved, shown in Table II, indicate 95\% confidence limits of 0.004 and 0.005, respectively, for MoP and CoC.
Theoretical Analysis of Lubricating Regime

The prediction of lubricant film thickness and consequently friction under transient operating conditions are generally quite complex. As a first approximation, the predicted minimum film thicknesses were calculated using the Hamrock and Dowson equation based on the average load and average speed, as shown later:

\[ h_{\text{min}} = 2.798R \left( \frac{\eta u}{E' R} \right)^{0.65} \left( \frac{L}{R^2} \right)^{-0.21} \]  

where \( h_{\text{min}} \) is the minimum film thickness, \( \eta \) is the lubricant viscosity, \( u \) is the entraining velocity, \( R \) is the equivalent radius for a ball-on-plane model, \( E' \) is the equivalent Young’s modulus for the material pair, and \( L \) is the applied load. The equivalent radius was calculated from:

\[ \frac{1}{R} = \frac{1}{R_1} - \frac{1}{R_2} \]  

where \( R_1 \) and \( R_2 \) are the radii of the femoral head and acetabular cup, respectively.

The predicted minimum film thickness, together with the surface roughness measurements of the bearing surfaces were used to predict the \( \lambda \) ratio, a dimensionless parameter that indicates the lubricating regime of the system.

\[ \lambda = \frac{h_{\text{min}}}{\left( Ra_h \right)^2 + \left( Ra_c \right)^2}^{1/2} \]  

where \( Ra_h \) and \( Ra_c \) are the \( Ra \) values for the head and cup, respectively. It has been shown that a \( \lambda \) ratio is greater than 3, then fluid film lubrication is predominant. A value below 1 indicates that boundary lubrication prevails, and between 1 and 3, the system is operating in a mixed lubrication regime.

RESULTS

A comparative study independently examining the influence of three variables, material, load, and lubricant, was performed. Example traces of frictional torque, in forward and reverse test directions, during a test cycle for MoP are shown in Figure 3(a,b), respectively. The combination of material tested appeared to have a significant effect upon the friction factor. Figure 4 shows the friction factor calculated for each material combination with a swing phase load of 100 N. The mean friction factor for the MoM bearings in 25% serum was 0.12 (±0.020), significantly higher than the other bearing combinations (ANOVA, \( p < 0.05 \)). There was no significant difference found amongst the other bearing materials, though CoC bearings exhibited the lowest friction factors during testing [0.04 (±0.007)]. The polyethylene bearings had higher friction factors than the CoM and CoC, though not significantly different, with the MoP exhibiting a higher friction factor than the CoP bearing. The CoM bearing had a significantly lower friction factor than the MoM bearing (ANOVA, \( p < 0.05 \)), and the behavior of CoM most closely matched that of CoC bearing, with a mean friction factor of 0.05 (±0.010).

Protein concentration was shown to have a marked effect upon the friction factors of the different bearings. Figure 4 shows the friction factors for the different bearings tested with a 100 N swing phase load in different lubricants. For all material combinations, except MoM, increasing the protein concentration resulted in an increase in friction factor. However, this was only found to be a significant increase (ANOVA, \( p < 0.05 \)) for the CoC and CoM bearings when changing the lubricant from water to 25% serum. The increase in friction factor with increase in from 25 to 100% protein concentration was shown to have a marked effect upon the friction factors of the different bearings.
serum was not found to be significant. For CoC bearings, increasing the protein concentration by changing the lubricant from water to 100% bovine serum resulted in the friction factor increase by over 100%. Conversely, the MoM bearings exhibited a statistically significant reduction in friction factor with increasing protein concentration (ANOVA $p < 0.05$), with a reduction of friction factor from 0.17 (±0.015) in water to 0.10 (±0.011) in 100% serum. Again, the behavior of the CoM bearing most closely matched the CoC material and was significantly different to the MoM hip replacement (ANOVA, $p < 0.05$).

Friction measurements were performed on all bearing combinations under swing phase loads of 25, 100, and 300 N. The results shown in Figure 5 were achieved using 25% bovine serum as a lubricant. It is clearly shown that increasing swing phase load resulted in increased friction factor for all bearing combinations; however, these results were not found to be statistically significant (ANOVA $p > 0.05$).

Table III shows the maximum frictional torque measured during the friction simulator study for each bearing combination and indicates the test conditions in which it was achieved. It should be noted that the highest frictional torque was measured within the MoM implant, tested in water with a swing phase load of 300 N. It is clear that this value, 3.6 N m, is lower than the minimum level found to cause immediate loosening.14

Theoretical analysis, shown in Table IV, indicated that both the CoC and CoM bearings had $\lambda$ values of greater than 3, suggesting both were capable of full fluid film lubrication. The $\lambda$ values for the MoM, MoP, and CoP bearings were all less than 1, indicating a boundary lubrication regime prevalence. Comparison of these predictions with the experimental results shows that the CoC and CoM bearings have the lowest friction factors, also suggesting better lubricating regimes.

**DISCUSSION**

This study has examined the friction of total hip replacements with a fixed 28 mm diameter, manufactured from different materials. Friction measurements may be used to determine the lubricating mechanism prevalent within the implant. Although friction no longer appears to be a major factor in implant failure, a low friction factor indicates improved lubrication, which may result in lowered wear. Measurement of friction is a useful tool for evaluating the tribological performance of replacement bearings.

Previous studies have shown friction to be dependent upon several factors. Wang et al.32 examined the influence of contact stress upon the friction of MoP implants, under serum-lubricated conditions, and found the friction factor varied with clearance and peak load. A reduction in friction factor was seen to occur with increased contact stress, resulting from either increased radial clearance or increased peak load. Further studies examining the influence of lubricant upon the friction factor have generated varying results. O’Connor et al.33 used a pendulum comparator to explore the influence of implant size, and lubricant upon the “number of swings” of the pendulum. Though finding the head size influenced the friction, the lubricants, human joint fluid, saline, and bovine serum, were found to have little

**TABLE III. Peak Frictional Torque Data**

<table>
<thead>
<tr>
<th>Bearing Combination</th>
<th>Test Conditions</th>
<th>Frictional Torque (N m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>MoM</td>
<td>300 N/water</td>
<td>3.6</td>
</tr>
<tr>
<td>MoP</td>
<td>300 N/100% serum</td>
<td>2.3</td>
</tr>
<tr>
<td>CoP</td>
<td>300 N/100% serum</td>
<td>2.2</td>
</tr>
<tr>
<td>CoM</td>
<td>300 N/100% serum</td>
<td>1.7</td>
</tr>
<tr>
<td>CoC</td>
<td>300 N/100% serum</td>
<td>1.4</td>
</tr>
</tbody>
</table>

**TABLE IV. Theoretical Film Thickness Data**

<table>
<thead>
<tr>
<th>Bearing Combination</th>
<th>$h_{max}$ (µm)</th>
<th>$\lambda$ Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>MoM</td>
<td>0.07</td>
<td>&lt;1</td>
</tr>
<tr>
<td>MoP</td>
<td>0.14</td>
<td>&lt;1</td>
</tr>
<tr>
<td>CoP</td>
<td>0.15</td>
<td>&lt;1</td>
</tr>
<tr>
<td>CoM</td>
<td>0.05</td>
<td>&gt;3</td>
</tr>
<tr>
<td>CoC</td>
<td>0.04</td>
<td>&gt;3</td>
</tr>
</tbody>
</table>
impact upon the outcome of the test, contrasting with the results anticipated. Conversely, studies with the Durham Hip Function Simulator, a similar design to that used within this study, have shown the lubricant to have a significant effect upon the friction in the implant, and this effect to vary with the materials tested. It was found in the Durham studies that testing in protein-containing lubricants, such as bovine serum and synovial fluid, resulted in reduced friction factor in MoM implants, whereas the friction factor increased for MoP, CoC, and MoP. However, the Durham study did not investigate the CoP or CoM bearings.

During the present study, a unidirectional motion and a simple loading profile were applied to the femoral head and tested in an inverted arrangement with respect to the anatomical position. These loading and motion cycles were not fully representative of the standard gait cycle, but a simplified model to allow comparison of the variables influencing the friction of the bearing. New implants were tested for short intervals, hence the effect of potential wear, and subsequent dimensional and surface finish changes, has not been considered.

This study examined the effect of material using implants of nominally 28 mm diameter. It was shown that the MoM implants generated significantly higher friction than all other material combinations. CoC implants had the lowest friction, although the CoM implants exhibited similar friction factors during all test conditions. This result compares well with the findings of Firkins et al., which demonstrated significantly lower wear rates in the CoM bearings when compared with the MoM. According to the elastohydrodynamic lubrication theory, the predicted lubrication regime depends on radial clearance, surface roughness, and the Young’s modulus of the bearing materials, and therefore all these can potentially affect the friction. Comparing the bearing materials under the same loading and lubricant conditions (25% serum, swing phase load of 100 N), the CoC bearing has the lowest friction factor, with the highest friction shown in the MoM bearings. The hard-on-soft bearings of CoP and MoP may benefit from enhanced fluid film thickness due to the lower Young’s modulus of UHMWPE, compared with the harder materials. However, the higher surface roughness associated with the UHMWPE bearing surface would act to break the fluid–film lubrication, and the hard-on-soft bearings would therefore experience a predominantly boundary-to-mixed lubricating regime. The surface roughness of the hard-on-hard bearings is lower than the UHMWPE bearings, potentially leading to an improved lubricating regime. The radial clearances in the hard-on-hard bearings were also lower than those of the hard-on-soft bearings (~0.03 and 0.12 mm respectively). There have been theoretical and experimental studies that demonstrate that a reduction in radial clearance would result in improved fluid–film lubrication. Hence, it would be expected that the hard-on-hard bearings would experience improved lubrication with the reduced clearance and the smooth bearing surfaces compared with the hard-on-soft bearings. Nevertheless, a resultant mixed lubrication is still dominant, as evidenced by the level of the measured friction coefficients. The friction under such a mixed lubrication regime depends on both the viscous shearing of the lubricant and the direct asperity contacts.

Studies of dry friction have shown metal–metal contacts to have a higher friction than predicted by simple friction theory, with experimental measurement of dry friction coefficient within a chromium–chromium contact to be ~0.4. Metal–metal contacts often result in adhesive junction forming where the asperities meet. An increased shear stress required to break the bond results in the higher friction coefficient. This phenomenon is not seen in ceramic contacts, nor does it occur with metal surfaces articulating with alternative materials. Hence, it is proposed that the high friction observed within the MoM bearings in water is due to the shear stresses required to break the adhesive junctions, formed through asperity contact in the mixed lubricating regime.

Studies examining the wear behavior of MoM implants have shown the presence of tribo-chemical films on the metal surfaces after wear simulator testing and on explants. However, it is unclear whether this type of film may form during a limited-cycle friction test. A similar friction study by Scholes et al. was able to determine the presence of proteins, particularly bovine serum albumin, on the surfaces of both metal and ceramic implants after short-term studies. It is proposed that these proteins act as a solid-phase lubricant at the metal–metal interface, thus reducing the adhesive forces between the two metallic surfaces. Additionally, electrical resistance studies on MoM implants, measured during in vitro wear simulator testing, indicated that those lubricants that contained no proteins produced a less effective lubrication than those with a protein content. It was shown that of the protein-containing lubricants, those at the higher concentration demonstrated a greater extent of separation than those at the lower concentrations. This would indicate that the proteins had some influence upon the lubrication of the implant, but the exact mechanism of boundary or fluid film has yet to be established.

Conversely, in the CoC bearing, the adherence of proteins to the surface effectively increases surface roughness, and reduces the effectiveness of lubricant film thickness. In addition, the force required to shear the proteins results in an increase in friction factor. It was noted that the CoM implants behaved in a similar way to the CoC implants, rather than the MoM implants.

Friction measurements were performed with swing phase loads of 25, 100, and 300 N. A previous study including theoretical analysis has predicted a reduced fluid film thickness with increasing swing phase load, which may be associated with less effective entrainment of fluid during the swing phase of the gait cycle. Studies with MoM implants demonstrated increases in both friction and wear with increase in swing phase load. This study achieved similar results for the MoM implants, and the trend for increasing friction factor with increasing swing phase load was exhibited in all the other material combinations under all lubricant
conditions. As the film thickness is reduced due to the increased swing phase load, asperity contact would increase in a mixed lubrication regime, thus resulting in the increased friction found in this study.

**CONCLUSION**

This study has examined the friction of a number of total hip replacements with a fixed 28 mm diameter, investigating the influence of material combination, lubricant, and swing phase load. MoM bearings had significantly higher friction than the other bearings under all test conditions (ANOVA, \( p < 0.05 \)). Ceramic bearings were shown to have lower friction than either MoM or metal-on-polyethylene implants under all test conditions. Increasing serum concentration was shown to increase the friction factor for all bearing combinations, with the exception of MoM, where the increased concentration of proteins appeared to improve the lubrication within the bearing. However, the effect of protein concentration was statistically significant in just the CoM, CoC, and MoM bearings (ANOVA, \( p < 0.05 \)). An increase of swing phase load was also shown to increase the friction factor, although this was not statistically significant (ANOVA, \( p > 0.05 \)).

There have been no previous studies examining the friction of CoM implants. This study has shown that the friction of such a novel bearing combination compares well with CoC bearings and exhibits trends close to these hip replacements. The friction was found to be significantly less than the MoM implants.

This study showed that test conditions, such as swing phase load and lubricant, may have a substantial effect upon the outcome of the friction study, hence it is important to evaluate this information when comparing materials or implant design.

The implants tested within this study were supplied by DePuy International Ltd (Leeds, UK).

**REFERENCES**


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