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Effect of wear, acetabular cup inclination angle, load and serum degradation on the friction of a large diameter metal-on-metal hip prosthesis

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Abstract

Background: The large-scale clinical problem caused by unacceptable tribological behaviour of certain large diameter metal-on-metal prosthetic hips has directed attention to adverse condition testing. High metal-on-metal wear is connected with adverse reaction to metal debris. Friction is important because high friction may be associated with high wear, risk the fixation of the cup, and cause detrimental heating of periprosthetic tissues.

Methods: A friction measurement system was added to a multidirectional, established hip joint wear simulator, and its functionality was evaluated. In preliminary tests, a 50 mm diameter metal-on-metal prosthesis was tested in an optimal acetabular cup inclination angle (48°) and in a steep angle (70°) using a normal peak load (2 kN) and an increased peak load (3 kN). The test length was 100 h. Long-term adverse condition tests of 3 million cycles were run for three 52 mm metal-on-metal prostheses. The lubricant was diluted calf serum at 37 °C.

Findings: In the 100 h tests, metal-on-metal frictional torque was not highly sensitive to the angle, load and serum degradation, and it was close to that of a conventional 28 mm prosthesis with a polyethylene cup, mostly below 5 Nm. However, a manyfold higher frictional torque (10 to 20 Nm) was observed in long-term metal-on-metal tests with substantial wear.

Interpretation: To obtain a realistic prediction of the frictional behaviour of a hip design, long-term, multidirectional wear tests are necessary. The friction should preferably be measured during the wear test. In addition to normal conditions, adverse condition testing is strongly recommended.

Keywords: joint mechanics; artificial joint; metal-on-metal; acetabular cup position; frictional torque; hip simulator

1. Introduction

Certain types of large-diameter CoCr-on-CoCr (metal-on-metal, MoM) hip prostheses, modular and resurfacing, have failed clinically due to unexpectedly high wear (Langton et al., 2011a,b; Scholes et al. 2017; Wienroth et al., 2014). Consequently, more attention has recently been paid to adverse condition testing so that the tribological behaviour in conditions that represent sub-optimal clinical reality would be covered in the test programme (Lee et al., 2010; Saikko, 2015). In this way, future implant wear disasters may possibly be avoided, provided that negative test results are not neglected offhand on the basis that they are merely an outcome of ‘incorrect testing’, as probably has been done in the past. Two highly relevant aspects from the tribological point of view are the inclination angle of the acetabular cup (Bernasek et al., 2011; Esposito et al., 2010; Lewinnek et al., 1978) and the magnitude of the peak load (Bergmann et al., 2001a; Bergmann et al., 2010). In addition to wear, friction is also interesting in the failing MoM prostheses in comparison with the optimally functioning MoM that may be hydrodynamically or elastohydrodynamically lubricated (Dowson, 2006; Gao et al., 2016; Hu et al., 2011; Liu et al., 2005; Smith et al., 2001). High friction may cause detrimental heating of joint fluid and of periprosthetic tissues (Bergmann et al., 2001b), risk the fixation of the acetabular component (Janssen et al., 2010), and exacerbate the wear at the taper fixation of the femoral head, which is known to be a clinical problem in modular large diameter MoMs (Brock et al., 2015; Langton et al., 2017a,b). High friction is also likely to coincide with high wear. Therefore, friction measurement was added to an established, multidirectional HUT-4 hip joint simulator used earlier in MoM wear studies (Saikko, 2005; Saikko et al., 2013; Saikko, 2015; Saikko, 2016a).

The most commonly used lubricant in orthopaedic tribology is serum, usually bovine, because it is practically the only lubricant, available in sufficient quantities and at a reasonable price, with which a realistic wear simulation can be produced. The serum

lubricant, a biological fluid, inevitably degrades during a long test run, which is known to have an effect on the polyethylene wear rate (Harsha and Joyce, 2011; Wang et al., 1999). There is no consensus of the optimal lubricant change interval. Half a million cycles appears to be common (ISO 14242-1, 2014). In the present study it was hypothesized that (a) the serum degradation has a measurable effect on MoM friction, (b) the large-diameter MoM in an optimal position of the cup is efficiently lubricated with low friction, as originally intended, and (c) the steep inclination angle of the cup with a large-diameter MoM leads to a metal-on-metal contact and to a considerable increase in friction, which is exacerbated by high wear. The present tests were designed to evaluate the functionality of the new friction measurement accessory and to show that friction should be measured as a part of a wear test in order to reflect the true frictional behaviour of a prosthesis design that follows the wear process.

2. Methods

A torque transducer (DRBK 5 Nm, ETH-Messtechnik GmbH, Gschwend, Germany) was added to the drive shaft of the 12-station HUT-4 hip joint simulator (Fig. 1) described in detail elsewhere (Saikko, 2005). According to the manufacturer, the maximum measurement error of the transducer was 0.5 per cent of full scale. The motion consisted of flexion-extension (FE, range 46°) and abduction-adduction (AA, range 12°) of the femoral head. The sinusoidal waveforms had a phase shift of $\pi/2$ to produce a multidirectional relative motion, which is essential in the reproduction of clinical wear mechanisms (Wang et al., 1996). The vertical load, implemented proportional-pneumatically, was applied through a universal joint to the acetabular cup that thus was self-centering on the femoral head. The cycle frequency was 1.06 Hz.

In addition to the frictional torque of the prosthesis tested, the transducer reading was

affected by the following effects: the friction of the bearings and their seals and the inertia of the mechanism and cradles, and the cam mechanism with a spring for the generation of the load control signal. Therefore, a reference signal had to be measured first with a joint that is known to have extremely low friction (Fig. 2). In an earlier study it was found that a 28 mm polished alumina-on-alumina (AoA) joint has a frictional torque of 0.03 Nm with 1 kN static load and water lubrication (Saikko, 2009). This value was considered ‘negligible’ for the present purpose. The femoral heads naturally needed to be carefully aligned. In the HUT-4 simulator, the positioning of the femoral head relative to the machine origin could readily be done so that the alignment error was ± 0.01 mm at most in the medial-lateral, anterior-posterior and superior-inferior directions. All the structures of the HUT-4 simulator were designed to be rigid so that the increase of the alignment error due to the loading was minimal (Saikko, 2005).

The point of consideration was chosen to be at 25 per cent of the cycle (Fig. 1b). At this point, the load L was close to its maximum, the relative motion was instantaneously purely extension and the sliding velocity was at its maximum (67.1 mm/s with a 50 mm head diameter). The moment arm correction factor of the crank mechanism at 25 per cent was $1/(46.9/120 \times \cos 6^\circ) = 2.57$ (see caption of Fig. 1). The value of the reference signal at this point was 0.40 Nm with 2 kN peak load and 0.87 Nm with 3 kN. The reference value was subtracted from the measured torque value with the prosthesis in question and the remainder was multiplied by 2.57 to obtain the resultant frictional torque T . With the varying correction factor taken into account, the largest T values occurred near 10 per cent of the cycle, i.e., soon after the maximum flexion and at 55 to 60 per cent of the cycle, soon after the maximum extension. At these points, the abduction-adduction (AA) was important and the sliding velocity was close to its minimum, which was 17.5 mm/s with a 50 mm head diameter. There were however large variations in these peaks, partly because the reference value increased

steeply between 50 and 60 per cent of the cycle (Fig. 2). Therefore, only the more steady situation at 25 per cent was systematically considered in the present preliminary study. At 0 and at 50 per cent of the cycle the correction factor was at its maximum, $1/((46.9/446/\cos 23^\circ)) = 8.75$, that is, the transducer was at its most insensitive to the frictional torque T .

A lubricant chamber with a heat exchanger (Fig. 3) was designed so that the tests could be run at a controlled lubricant bulk temperature of 36.5 °C to 37.5 °C. The thin inner tube was made from stainless steel and the outer tube from polyvinyl chloride (PVC). Between them was a space for the circulating heating or cooling water. The lubricant was HyClone (Logan, Utah, USA) Alpha calf fraction serum SH30212.03, diluted 1:1 with Milli-Q grade ultrapure deionized water. The protein concentration of the lubricant after the dilution was 20 mg/ml. No additives were used. The volume of the lubricant was 260 ml. This type of lubricant is widely used in orthopaedic tribology (Wang et al., 1996; Wang et al., 1999; Essner et al., 2005).

Two different load profiles were used so that the peak load value was either 2 kN or 3 kN (Saikko, 2015; Saikko, 2016a). The former was considered a normal value, and the latter an increased value caused by obesity (Bergmann et al., 2001; Bergmann et al., 2010). The test programme consisted of preliminary tests of 100 h duration and long-term tests of 800 h duration (Table 1). Hence the test lengths were 382 000 cycles or 3.056 million cycles.

In the 100 h tests, a 50 mm diameter Metasul (Protasul-21WF, Centerpulse Orthopedics Ltd, Winterthur, Switzerland) MoM bearing with a diametral clearance of 0.16 mm was chosen to represent a large-diameter MoM articulation. It was manufactured in the early 2000s. The cup was sub-hemispherical (165°). The cups were cemented in a mould in two positions so that the anatomical inclination angles (Murray, 1993) were 48°, ‘optimal’ (45° operative inclination, 20° operative anteversion), and 70°, ‘steep’ (66° operative inclination, 32° operative anteversion) (Langton et al., 2011b). A 2 kN test was run first, and then, after

cleaning and with fresh lubricant, a 3 kN test with an unchanged positioning of the components was run. In addition, a 3 kN steep test was run with the severely worn couple no. 2 of test 2 from (Saikko, 2016a). Hence, the test was a continuation of the earlier, adverse condition, 3 million cycle wear test with a 3 kN peak load, in which the wear rate was as high as 150 mg/10⁶ cycles. A wear pit with a depth of 0.4 mm formed on the cup. Scratching occurred especially on the head but the wear pit on the cup was mostly burnished. For the quantification of wear in mg, a method based on the measurement of the Co and Cr concentrations of the used lubricant by atomic absorption spectroscopy (AAS) was developed earlier (Saikko et al., 1998). The prosthesis of comparison was a conventional 28 mm alumina-on-ultrahigh molecular weight polyethylene (AoP) with a 48° anatomical inclination angle, and a diametral clearance of 0.1 mm at the operating temperature. The test procedure was similar to that of the MoM tests. One specimen under each condition was tested. The lubricant was not changed during the 100 h tests.

In the long-term tests of 3 million cycles, three similar MoMs were tested in adverse conditions. They were similar to the MoM of the preliminary tests, but they were of 52 mm diameter. These tests were a continuation of the steep 70° test 1 of (Saikko, 2016a), but with increased peak load of 3 kN. This arrangement was necessitated by the unavailability of unused components of this recalled design. The earlier 2 kN tests resulted in moderate wear, from 36 µm to 72 µm after 3 million cycles (combined linear wear of the head and the cup). The wear was measured and lubricant changed at intervals of 382 000 cycles (100 h). The wear was measured with a Talyrond apparatus as described in (Saikko, 2016a). In all tests, the friction was recorded twice a day.

3. Results

The separation of the reference torque signal from the measured torque signal was straightforward (Fig. 2). No squeak occurred in any test. In the 100 h tests the most conspicuous observation was the high and stable T , $15 \text{ Nm} \pm 0.4 \text{ Nm}$, of the worn MoM specimen (Fig. 4). In the other 6 tests, the frictional behaviour during 100 h was more complex, but the differences were moderate, and T was mostly below 5 Nm. With an optimal angle MoM and 3 kN load, T was lower than with 2 kN for the most part of the test. In the optimal angle tests, the contact did not extend to the edge of the cup, but in the steep angle tests, it did (Fig. 5). Multidirectional scratches were observed especially on the MoM heads (Fig. 6). More darkening of the lubricant occurred in the steep angle tests (Fig. 7).

In the long-term tests that were a continuation of test 1 of (Saikko, 2016a) with couples no. 1, 2 and 3 (52 mm MoMs, 70° inclination and 3 kN peak load), T varied widely from 1 Nm to 20 Nm (Fig. 8), which influenced the darkening of the lubricant (Fig. 9). The results with couples no. 1 and 3 were similar. The linear wear of the cup was almost $100 \text{ }\mu\text{m}$ and T remained below 10 Nm and showed a decreasing trend. With couple no. 2, the cup wear was 8-fold higher, and T showed an increasing trend and rose up to 20 Nm towards the end of the test. Still the head wear, $28 \text{ }\mu\text{m}$, was as low as with couple no. 1. Scratching took place mainly on the heads, whereas the cups remained mostly shiny (Fig. 10).

4. Discussion

The present friction measurement system was designed to be a simple, easy-to-use accessory to an established, multidirectional hip joint wear simulator, not a separate, high-precision friction measurement device. The advantage of the present system is that long-term effects such as wear and lubricant degradation can be included in the study using multidirectional motion. The present torque transducer was successfully used earlier with the biaxial rocking

motion hip simulator in wear tests (Saikko, 2016b). Similarly, the present system proved practical and useful. Dedicated friction measurement simulators typically are single-axis devices (Bishop et al., 2008; Bishop et al., 2013; Brockett et al., 2007; Saikko, 1992; Scholes and Unsworth, 2000) and therefore they cannot be used to produce a realistic wear simulation that requires multidirectional motion (Wang et al., 1996). Hence the information that can be obtained with them is limited because the in vivo motion is not uniaxial, and because wear inevitably takes place in vivo and affects the friction, as indicated by multidirectional, long-term hip simulator friction and wear studies (Saikko, 2009; Saikko, 2016b). Uniaxial motion results in unidirectional grooving that strongly affects friction and wear mechanisms but it is never seen in retrieved prosthetic components. Uniaxial friction measurement hip simulators typically are used to test new, unworn prostheses, which gives only a limited picture of the frictional behaviour of the design in question.

Friction measurement has been added to three-axis hip joint simulators by (Haider et al., 2016; Kaddick et al., 2015; Sonntag et al., 2017). In these sophisticated, complex designs the three orthogonal components of T can be obtained separately, as the femoral head support is instrumented. The resultant values of the present study measured with a more simple, inexpensive method were still in agreement with those reported in (Haider et al., 2016; Kaddick et al., 2015; Sonntag et al., 2017). Haider et al. (2016) obtained a friction factor of 0.08 ± 0.01 for the 44 mm diameter AoP. The corresponding values in the present study can be calculated by dividing the T values, $2.6 \text{ Nm} \pm 1.0 \text{ Nm}$ (Table 1), by the peak load, 3 kN, and by the radius, 14 mm, which results 0.06 ± 0.02 . Kaddick et al. (2015) obtained a mean frictional torque value of 2.4 Nm for the 28 mm diameter AoP. Sonntag et al. (2017) obtained a mean value of 3.8 Nm for the 28 mm diameter AoP after 1 000 cycles, which is close to the early values measured in the present study (Fig. 4). These above three studies used a peak load of 3 kN in the measurement of these friction values. The present AoP results were also in

agreement with those measured in vivo with instrumented hip joint implants (Damm et al., 2017). The in vivo variation of the friction factor of the 32 mm diameter AoP during the first 4 steps of the subjects after standing still on both legs for 12 s and then starting to walk with the ipsilateral leg was 0.02 to 0.12 during a cycle, and the mean peak load was 2.4 kN. It may therefore be assumed that the present MoM results also reflect the clinical reality in that high friction indeed is connected with high wear. High friction may cause heating of joint fluid and of periprosthetic tissues above 40 °C (Bergmann et al., 2001b) and risk the fixation of the acetabular component (Janssen et al., 2010).

When the change of friction during the 100 h tests with previously unworn bearing surfaces is considered, it is admittedly difficult to completely separate the effect of wear from the effect of serum degradation. They are likely to interact, e.g., as the metal-on metal contact takes place, the contact temperature increases, which accelerates the degradation and further reduces the capability of the serum to lubricate and prevent wear. A bedding-in phenomenon has been suggested to be an explanation for the reduction of wear and friction after the early stage of the test (Lee et al., 2010). In the present study, the variation of friction during the 100 h tests was more complex and did not follow simple patterns neither with MoM nor with AoP (Fig. 4). In the earlier steep angle (70°) wear tests with 50 mm MoM and 3 kN peak load, there was no high wear limited to the early test stage only (the so-called running-in wear), and a lower steady state wear rate thereafter, but the wear rate was very high and linear (mean 167 mg/10⁶ cycles) during the entire 3 million cycle test (Saikko, 2016a). Expectedly, the effect of this wear on friction was strong. The specimen that had undergone severe wear earlier was the only one that showed a stable, albeit very high torque value, 15 Nm. The worn surfaces had already accommodated to each other during 3 million cycles and the present 382 000 cycle test obviously did not change the contact mechanics much, which may explain the stable frictional behaviour. The value of 15 Nm that was manyfold to those measured with

previously unworn specimens is assumed to be so high that it would have clinical significance regarding frictional heating (Bergmann et al., 2001b) and loosening of acetabular fixation (Janssen et al., 2010). Recent clinical findings show that the taper fixation of the femoral head is problematic in modular large diameter MoMs (Brock et al., 2015; Langton et al., 2017a,b), and a high T is likely to be an exacerbating factor in the fretting wear of the taper surfaces (Kaddick et al., 2015). The observed torque values (Table 1) support hypotheses (b) and (c). However, the values with previously unworn specimens, up to 4 Nm, were not low enough to indicate a hydrodynamic fluid film lubrication, but rather a mixed lubrication mechanism (Dowson, 2006).

The mean wear rate $\pm$ SD of the present 50 mm MoM was 0.89 ± 0.13 mg/ 10^6 cycles with a 48° inclination and a 2 kN peak load (Saikko, 2005), and 167 ± 15.7 mg/ 10^6 cycles with a 70° inclination and a 3 kN peak load (Saikko, 2016a). Such a dramatic increase in the wear rate indicates a fundamental change in the lubrication mechanisms, presumably from mixed lubrication to boundary lubrication with metal-on-metal contact, but still no galling. In fact, the wear pit with a depth of 0.4 mm was surprisingly burnished, whereas the head was scratched (Saikko, 2016a). Obviously, the degraded, blackened lubricant was still capable of providing boundary lubrication and preventing galling. Water lubrication led to galling and seizure within minutes (Saikko, 2015). Since measurable but very low wear did occur in the optimal position (48°), the possible hydrodynamic fluid film did not separate the bearing surfaces during the entire gait cycle (Saikko, 2005). When the contact point was close to the cup edge in the steep position (70°), the formation of any fluid film was geometrically difficult, or impossible, resulting in disastrous wear (Saikko, 2016a). The above earlier assumptions regarding lubrication mechanisms were only partly corroborated by the present friction study. The steep MoM tests with previously unworn bearing surfaces did not show exceptionally high friction, apparently because the test lengths were not sufficiently long to

produce a severe wear damage. Further, it was surprising that in the optimal position MoM tests the frictional torque with the 3 kN peak load was lower than with 2 kN for the most part of the test. It seems that the conditions were not favourable to hydrodynamic lubrication, primarily due to the relatively low velocity and viscosity and the large relative clearance (3.2×10^{-3}), and the more probable elastohydrodynamic mechanism required a sufficiently high load to produce optimal contact mechanics.

In the 3 million cycle tests with the 52 mm MoMs, the variability of the results was conspicuous as the behaviour of one the prostheses differed from that of the other two. This indicated that the tribological endurance limit, below which the tribological behaviour shows high variation, was not exceeded. The tribological endurance limit unquestionably was exceeded when the same test conditions (70°, 3 kN) but 50 mm MoMs of the same design were used (Saikko, 2016a). In that test, the wear was high and its variation was low. A possible explanation for the discrepancy was the 2 mm difference in the head diameter. It may appear small, but may still be decisive for the Metasul MoM under the present test conditions. Similar erratic wear behaviour typical of specifically MoM tests has been observed by other researchers as well (Essner et al., 2005; Vassiliou et al., 2006). Nevertheless, the present tests leave little doubt that high MoM wear is accompanied by high and most probably detrimental friction. The largest T values well exceeded the lever-out moments, 4.0 Nm to 13.5 Nm, computed for press-fit acetabular shells by Janssen et al. (2009). The variation of T during each 382 000 cycle stage did not show a repetitive trend but still the observed large variation apparently was caused by serum degradation. Hence, hypothesis (a) was supported. In general, the degradation behaviour of serum and its effect on friction, wear and lubrication in orthopaedic tribology with different bearing couples appear too complex to be satisfactorily explained by contemporary knowledge.

One noteworthy difference with the earlier wear study (Saikko, 2016a) was the absence

of squeak, which was likely to be due to the fact that the frictional heating above 40 °C of the serum bulk temperature was prevented by the heat exchanger. Earlier HUT-4 wear tests showed that the uncontrolled bulk temperature of the diluted serum lubricant with a volume of 500 ml can vary from a couple of degrees above the room temperature (50 mm MoM and 28 mm MoP in the optimal position with 2 kN) to 44 °C (50 mm MoM in the steep position with 3 kN) (Saikko, 2005; Saikko et al., 2013; Saikko, 2015; Saikko, 2016a). The temperature naturally affects the behaviour of the lubricant, mainly by protein denaturation (Harsha and Joyce, 2011; Wang et al., 1999), and the clearance, most notably with a plastic cup. Frictional heating above 40 °C is known to occur in vivo (Bergmann et al., 2001b), but its realistic simulation together with a continuous replenishment of fresh lubricant, corresponding to the biologic regeneration of the small amount of synovial fluid present within the joint capsule, is difficult to arrange in a tribological hip joint simulator.

The principal limitations of the present study were considered to be the following. First, the 12-station simulator was run with one test station only and so the production of results was a slow process. The number of specimens and test duration were therefore limited. Note also that the historic, recalled, large-diameter MoM specimens are difficult to source. The idea was that the frictional behaviour of specimens previously worn in multi-station tests can separately be studied using the novel system described. This holds true for unworn specimens as well. Second, the focus was on the resultant value of the frictional torque. The resultant, instead of its components, was nevertheless considered the principal issue, as most of the hip simulator wear testing is still done with no friction measurement at all. The simplicity of the measurement system may also be considered an advantage from points of view such as the cost, introduction and user-friendliness. Third, although the friction of AoA in water is known to be very low (Saikko, 2009), it still somewhat affected the reference signal. Therefore, the present T values were slightly underestimates due to the minute AoA frictional torque in the

reference signal measurements.

Conclusions

Friction measurement was successfully added to an existing multidirectional wear simulator. In the 100 h tests for previously unworn specimens, friction was not particularly sensitive to the acetabular cup inclination angle, load or serum degradation. The frictional torque values were mostly below 5 Nm. A manyfold higher frictional torque, 15 Nm, was observed for a MoM specimen that was severely worn in an earlier, multidirectional, adverse condition wear test performed with the present simulator. Values of T up to 20 Nm and wide variation of T due to serum degradation were observed in 3 million cycle adverse condition MoM tests.

To obtain a realistic picture of the frictional behaviour of a prosthetic hip design, it is suggested that the friction be measured simultaneously with long-term, multidirectional wear tests. The friction measurement system should be a part of the structure of the wear simulator. In addition to normal condition tests, adverse condition tests are strongly recommended. Since dedicated, uniaxial friction measurement hip simulators cannot produce realistic wear, their usefulness is limited.

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Conflict of interest statement

The author has nothing to declare.

References

- Bergmann, G., Deuretzbacher, G., Heller, M., Graichen, F., Rohlmann, A., Strauss, J., Duda, G.N., 2001a. Hip contact forces and gait patterns from routine activities. *J. Biomech.* 34, 859–871. [http://dx.doi.org/10.1016/S0021-9290\(01\)00040-9](http://dx.doi.org/10.1016/S0021-9290(01)00040-9).
- Bergmann, G., Graichen, F., Rohlmann, A., Bender, A., Heinlein, B., Duda, G.N., Heller, M.O., Morlock, M.M., 2010. Realistic loads for testing hip implants. *Bio-Med. Mater. Eng.* 20, 65–75. <http://dx.doi.org/10.3233/BME-2010-0616>.
- Bergmann, G., Graichen, F., Rohlmann, A., Verdonschot, N., van Lenthe, G.H., 2001b. Frictional heating of total hip implants. Part 1. measurements in patients. *J. Biomech.* 34, 421–428. [http://dx.doi.org/10.1016/S0021-9290\(00\)00188-3](http://dx.doi.org/10.1016/S0021-9290(00)00188-3).
- Bernasek, T., Fisher, D., Dalury, D., Levering, M., Dimitris, K., 2011. Is metal-on-metal squeaking related to acetabular angle of inclination? *Clin. Orthop. Relat. Res.* 469, 2577–2582. <http://dx.doi.org/10.1007/s11999-011-1900-4>.
- Bishop, N.E., Waldow, F., Morlock, M.M., 2008. Friction moments of large metal-on-metal hip joint bearings and other modern designs. *Med. Eng. & Phys.* 30, 1057–1064. <http://dx.doi.org/10.1016/j.medengphy.2008.01.001>.
- Bishop, N.E., Hothan, A., Morlock, M.M., 2013. High friction moments in large hard-on-hard hip replacement bearings in conditions of poor lubrication. *J. Orthop. Res.* 31, 807–813. <http://dx.doi.org/10.1002/jor.22255>.
- Brock, T.M., Sidaginamale, R., Rushton, S., Nargol, A.V.F., Bowsher, J.G., Savisaar, C., Joyce, T.J., Deehan, D.J., Lord, J.K., Langton, D.J., 2015. Shorter, rough trunnion surfaces are associated with higher taper wear rates than longer, smooth trunnion surfaces in a contemporary large head metal-on-metal total hip arthroplasty system. *J. Orthop. Res.* 33, 1868–1874. <http://dx.doi.org/10.1002/jor.22970>.
- Brockett, C., Williams, S., Jin, Z., Fisher, J., 2007. Friction of total hip replacements with different bearings and loading conditions. *J. Biomed. Mater. Res. B.* 81, 508–515. <http://dx.doi.org/10.1002/jbm.b.30691>.
- Damm, P., Bender, A., Duda, G., Bergmann, G., 2017. In vivo measured joint friction in hip implants during walking after a short rest. *PLoS ONE* 12, e0174788. <http://dx.doi.org/10.1371/journal.pone.0174788>.
- Dowson, D., 2006. Tribological principles in metal-on-metal hip joint design. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 220, 161–171. <http://dx.doi.org/10.1243/095441105X63255>.

- Esposito, C., Walter, W.L., Campbell, P., Roques, A., 2010. Squeaking in metal-on-metal hip resurfacing arthroplasties. *Clin. Orthop. Relat. Res.* 468, 2333–2339.
<http://dx.doi.org/10.1007/s11999-010-1344-2>.
- Essner, A., Sutton, K., Wang, A., 2005. Hip simulator wear comparison of metal-on-metal, ceramic-on-ceramic and crosslinked UHMWPE bearings. *Wear* 259, 992–995.
<http://dx.doi.org/10.1016/j.wear.2005.02.104>.
- Gao, L., Dowson, D., Hewson, R.W., 2016. A numerical study of non-Newtonian transient elastohydrodynamic lubrication of metal-on-metal hip prostheses. *Tribol. Int.* 93, 486–494. <http://dx.doi.org/10.1016/j.triboint.2015.03.003>.
- Haider, H., Weisenburger, J.N., Garvin, K.L., 2016. Simultaneous measurement of friction and wear in hip simulators. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 230, 373–388.
<http://dx.doi.org/10.1177/0954411916644476>.
- Harsha, A.P., Joyce, T.J., 2011. Challenges associated with using bovine serum in wear testing orthopaedic biopolymers. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 225, 948–958. <http://dx.doi.org/10.1177/0954411911416047>
- Hu, X.Q., Wood, R.J.K., Taylor, A., Tuke, M.A., 2011. The tribological behaviour of different clearance MOM hip joints with lubricants of physiological viscosities. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 225, 1061–1069.
<http://dx.doi.org/10.1177/0954411911419061>.
- ISO 14242-1:2014, Implants for surgery — Wear of total hip-joint prostheses — Part 1: Loading and displacement parameters for wear-testing machines and corresponding environmental conditions for test, International Organization for Standardization, Geneva, Switzerland.
- Janssen, D., Zwartelee, R.E., Doets, H.C., Verdonschot, N., 2010. Computational assessment of press-fit acetabular implant fixation: the effect of implant design, interference fit, bone quality, and frictional properties. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 224, 67–75.
<http://dx.doi.org/10.1243/09544119JEIM645>.
- Kaddick, C., Malczan, M., Buechele, C., Hintner, M., Wimmer, M.A., 2015. On the measurement of three-dimensional taper moments due to friction and contact load in total hip replacement. *ASTM Special Technical Publication STP 1591*, 336–350.
<http://dx.doi.org/10.1520/STP159120140146>.
- Langton, D.J., Jameson, S.S., Joyce, T.J., Gandhi, J.N., Sidaginamale, R., Mereddy, P., Lord, J., Nargol, A.V.F., 2011a. Accelerating failure rate of the ASR total hip replacement. *J. Bone & Joint Surg.* 93-B, 1011–1016. <http://dx.doi.org/10.1302/0301-620X.93B8.26040>.

- Langton, D.J., Joyce, T.J., Jameson, S.S., Lord, J., Van Orsouw, M., Holland, J.P., Nargol, A.V., De Smet, K.A., 2011b. Adverse reaction to metal debris following hip resurfacing. The influence of component type, orientation and volumetric wear. *J. Bone & Joint Surg.* 93-B, 164–171. <http://dx.doi.org/10.1302/0301-620X.93B2.25099>.
- Langton, D., Ahmed, I., Avery, P., Bone, M., Cooke, N., Deehan, D., Duffy, P., Foguet, P., Green, S., Holland, J., Jafri, A., Longstaff, L., Lord, J., Loughhead, J., Meek, R.M.D., Murray, H., Nanu, A., Nargol, A.V.F., Scholes, S.C., Sidaginamale, R.P., Waller, S., Joyce, T., 2017a. Investigation of taper failure in a contemporary metal-on-metal hip arthroplasty system through examination of unused and explanted prostheses. *J. Bone & Joint Surg.* 99-A, 427–436. <http://dx.doi.org/10.2106/JBJS.15.00917>.
- Langton, D.J., Sidaginamale, R.P., Joyce, T.J., Meek, R.D., Bowsher, J.G., Deehan, D., Nargol, A.V.F., Holland, J.P., 2017b. A comparison study of stem taper material loss at similar and mixed metal head-neck taper junctions. *Bone & Jt J.* 99-B, 1304–1312. <http://dx.doi.org/10.1302/0301-620X.99B10.BJJ-2016-1005.R1>.
- Lee, R.K., Longaray, J., Essner, A., Wang, A., 2010. Metal-on-metal bearings: the problem is edge-loading wear. *Surg. Technol. Int.* 20, 303–308.
- Lewinnek, G.E., Lewis, J.L., Tarr, R., Compere, C.L., Zimmerman, J.R., 1978. Dislocations after total hip-replacement arthroplasties. *J. Bone & Joint Surg.* 60-A, 217–220.
- Liu, F., Jin, Z.M., Hirt, F., Rieker, C., Roberts, P., Grigoris, P., 2005. Effect of wear of bearing surfaces on elastohydrodynamic lubrication of metal-on-metal hip implants. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 219, 319–328. <http://dx.doi.org/10.1243/095441105X34356>.
- Murray, D.W., 1993. The definition and measurement of acetabular orientation. *J. Bone & Joint Surg.* 75-B, 228–232.
- Saikko, V., 1992. A simulator study of friction in total replacement hip joints. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 206, 201–211. http://dx.doi.org/10.1243/PIME_PROC_1992_206_292_02.
- Saikko, V., 2005. A 12-station, anatomic hip joint simulator. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 219, 437–448. <http://dx.doi.org/10.1243/095441105X34419>.
- Saikko, V., 2009. Friction measurement in the biaxial rocking motion hip joint simulator. *J. Tribol.* 131, 011201. <http://dx.doi.org/10.1115/1.2991121>.
- Saikko, V., 2015. Adverse condition testing with hip simulators. *Biotribol* 1–2, 2–10. <http://dx.doi.org/10.1016/j.biotri.2015.02.001>.

- Saikko, V., 2016a. Effect of increased load on the wear of a large-diameter metal-on-metal modular hip prosthesis with a high inclination angle of the acetabular cup. *Tribol. Int.* 96, 149–54. <http://dx.doi.org/10.1016/j.triboint.2015.12.032>.
- Saikko, V., 2016b. Friction measurement in a hip wear simulator. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 230, 366–372. <http://dx.doi.org/10.1177/0954411915610602>.
- Saikko, V., Nevalainen, J., Revitzer, H., Ylinen, P., 1998. Metal release from total hip articulations in vitro: Substantial from CoCr/CrCr, negligible from CoCr/PE and alumina/PE. *Acta Orthop. Scand.* 69, 449–454. <http://dx.doi.org/10.3109/17453679808997777>.
- Saikko, V., Ahlroos, T., Revitzer, H., Ryti, O., Kuosmanen, P., 2013. The effect of acetabular cup position on wear of a large-diameter metal-on-metal prosthesis studied with a hip joint simulator. *Tribol. Int.* 60, 70–76. <http://dx.doi.org/10.1016/j.triboint.2012.10.011>.
- Scholes, S.C., Unsworth, A., 2000. Comparison of friction and lubrication of different hip prostheses. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 214, 49–57. <http://dx.doi.org/10.1243/0954411001535237>.
- Scholes, S.C., Hunt, B.J., Richardson, V.M., Langton, D.J., Smith, E., Joyce, T.J., 2017. Explant analysis of the Biomet Magnum/ReCap metal-on-metal hip joint. *Bone & Jt Res.* 6, 113–122. <http://dx.doi.org/10.1302/2046-3758.62.BJR-2016-0130.R2>.
- Smith, S.L., Dowson, D., Goldsmith, A.A.J., 2001. The effect of diametral clearance, motion and loading cycles upon lubrication of metal-on-metal total hip replacements. *Proc. Inst. Mech. Eng. Part C. J Mech. Eng. Sci.* 215, 1–5. <http://dx.doi.org/10.1243/0954406011520454>.
- Sonntag, R., Braun, S., Al-Salehi, L., Reinders, J., Mueller, U., Kretzer, J.P., 2017. Three-dimensional friction measurement during hip simulation. *PLoS ONE* 12, e0184043. <http://dx.doi.org/10.1371/journal.pone.0184043>.
- Vassiliou, K., Elfick, A.P.D., Scholes, S.C., Unsworth, A., 2006. The effect of ‘running-in’ on the tribology and surface morphology of metal-on-metal Birmingham hip resurfacing device in simulator studies. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 220, 269–277. <http://dx.doi.org/10.1243/09544119JEIM63>.
- Wang, A., Stark, C., Dumbleton, J.H., 1996. Mechanistic and morphological origins of ultra-high molecular weight polyethylene wear debris in total joint replacement prostheses. *Proc. Inst. Mech. Eng. Part H. J. Eng. Med.* 210, 141–155. http://dx.doi.org/10.1243/PIME_PROC_1996_210_407_02.

- Wang, A., Polineni, V.K., Essner, A., Stark, C., Dumbleton, J.H., 1999. Quantitative analysis of serum degradation and its effect on the outcome of hip joint simulator wear testing of UHMWPE, 45th Annual Meeting of the Orthopaedic Research Society, Feb 1–4, 1999, Anaheim, Ca, USA.
- Wienroth, M., McCormack, P., Joyce, T.J., 2014. Precaution, governance and the failure of medical implants: the ASR((TM)) hip in the UK. *Life Sci. Soc. Policy* 10, 19.
<http://dx.doi.org/10.1186/s40504-014-0019-2>.

Table 1. Tests run with HUT-4 simulator in diluted serum at 37 °C. Previous wear tests described in (Saikko, 2016a).

Type of prosthesis	Head dia. (mm)	No. of tests	Cup angle	Peak load (kN)	Test length (10 ⁶ cycles)	Previously worn	T (Nm) mean ± SD
AoP	28	1	48°	2.0	0.382	no	2.6 ± 1.1
AoP	28	1	48°	3.0	0.382	no	2.6 ± 1.0
MoM	50	1	48°	2.0	0.382	no	2.1 ± 1.8
MoM	50	1	48°	3.0	0.382	no	1.4 ± 1.1
MoM	50	1	70°	2.0	0.382	no	1.9 ± 1.6
MoM	50	1	70°	3.0	0.382	no	3.1 ± 1.3
MoM	50	1	70°	3.0	0.382	severely	15 ± 0.4
MoM	52	3	70°	3.0	3.056	moderately	3.6 ± 2.6
							14 ± 4.6
							4.3 ± 2.6

Figure captions

Fig. 1. Drive mechanism of HUT-4 hip joint simulator with torque transducer added to drive shaft, shown in four different positions of cycle. -shaped ‘push-pull-lean part’ moved by crank was connected to inner (AA) cradle by spherical rod ends. Its vertical motion produced FE (range 46°) and its leaning motion produced AA (range 12°). Their phase shift was $\pi/2$. Crank offset = 46.9 mm, height of push-pull-lean part = 446 mm, and its lateral distance (positions b and d) from centre of femoral head = 120 mm. (a) 0 per cent of cycle, FE = 23°, AA = 0°, load L began to increase rapidly from minimum value, (b) 25 per cent, FE = 0°, AA = -6°, L was close to maximum of first peak, symmetry axis of push-pull-lean part was perpendicular to crank, and so transducer sensed 38.9 per cent of frictional torque T , since $(46.9/120)\cos 6^\circ = 0.389$, (c) 50 per cent, FE = -23°, AA = 0°, L was at maximum of second peak, symmetry axis of push-pull-lean part was aligned with crank, and so transducer sensed 11.4 per cent of T , since $(46.9/446)/\cos 23^\circ = 0.114$, (d) 75 per cent, FE = 0°, AA = 6°, L was 1/3 of maximum. 1 torque transducer, 2 bellow coupling, 3 motor/gear, 4 crank, 5 push-pull-lean part, 6 inner (AA) cradle, 7 outer (FE) cradle, 8 bearing housing, 9 cam mechanism for generation of load control signal, 10 counterweight, 11 prosthesis (inside cradles).

Fig. 2. Separation of reference torque signal from measured torque signal with 28 mm AoP. L is load, FE is flexion-extension, and AA is abduction-adduction. Cycle time was 0.94 s. Sign did not change because torque was measured at unidirectionally rotating drive shaft.

Fig. 3. Front view of stations 1 to 3 of HUT-4 hip simulator showing lubricant chamber with temperature control in test station no. 2. Position is as in Fig. 1c. There was space between stainless steel inner tube and PVC outer tube where heating (or cooling) water was circulated. Abduction-adduction of stations 1 and 3 was disconnected to reduce bearing seal friction.

Fig. 4. Variation of frictional torque T at 25 per cent of gait cycle with testing time in HUT-4 simulator. Lubricant was 1:1 diluted Alpha calf serum at 37 °C.

Fig. 5. MoM acetabular cups (50 mm diameter) after 100h tests with 3 kN peak load. Left: optimal inclination angle 48°, right: steep inclination angle 70°. On optimal inclination angle cup, contact did not extend to edge but on steep inclination angle cup, it did (arrows). Serum residue has not been removed.

Fig. 6. Multidirectional scratches at apex of elliptical slide track on 50 mm MoM head after 100h steep 70° test with 3 kN peak load.

Fig. 7. Samples of used, 1:1 diluted Alpha calf serum lubricants after 100 h MoM tests. From left to right: optimal 48° and 2 kN, optimal 48° and 3 kN, steep 70° and 3 kN.

Fig. 8. Variation of frictional torque T and linear wear with number of cycles in long-term tests with 52 mm MoMs, (a) couple no. 1, (b) couple no. 2 and (c) couple no. 3. Cup position 70°, peak load 3 kN. Initial values of wear equal end values of wear in test 1 described in (Saikko, 2016a).

Fig. 9. Samples of used, 1:1 diluted Alpha calf serum lubricants from 3 million cycle adverse condition MoM tests. Symbols as in Fig. 8. Samples are in test sequence from left to right.

Fig. 10. Specimens of 52 mm MoM couple no. 2 after 3 million cycle adverse condition test, (a) border between worn and unworn bearing surface of cup, after removal of serum residue, caused by superior edge wear, depth of which is almost 0.8 mm, but still worn surface is shiny, (b) elliptical scratches on head, and (c) superior gap of 0.8 mm that illustrates magnitude of linear wear, shown by moving head to its original location within cup.

(a)

(b)

(c)

(d)

Fig. 1

Fig. 4

Fig. 5

Fig. 6

Fig. 7

Fig. 8a

Fig. 8b

Fig. 8c

Fig. 9a

Fig. 9b

Fig. 9c

Fig. 10a

Fig. 10b

Fig. 10c