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Friction measurement in a hip wear simulator

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Abstract

A torque measurement system was added to a widely used hip wear simulator, the biaxial rocking motion (BRM) device. With the rotary transducer, the frictional torque about the drive axis of the BRM mechanism was measured. The principle of measuring the torque about the vertical axis above the prosthetic joint, used earlier in commercial BRM simulators, was shown to sense only a minor part of the total frictional torque. With the present method the total frictional torque of the prosthetic hip was measured. This was shown to consist of the torques about the vertical axis above the joint and about the leaning axis. Femoral heads made from different materials were run against conventional and crosslinked polyethylene acetabular cups in serum lubrication. Regarding the femoral head material and the type of polyethylene, there were no categorical differences in frictional torque with the exception of zirconia heads, with which the lowest values were obtained. Diamond-like carbon coating of the CoCr femoral head did not reduce friction. The friction factor was found to always decrease with increasing load. High wear could increase the frictional torque by 75 per cent. With the present system, friction can be continuously recorded during long wear tests, and so the effect of wear on friction with different prosthetic hips can be evaluated.

Keywords

Frictional torque, hip simulator, cobalt-chrome, diamond-like carbon, zirconia

Introduction

The biaxial rocking motion (BRM) device is a standardized, widely used hip wear simulator[1]. Friction has only occasionally been measured in commercial BRM simulators[2–4]. This may be due to the fact that a generally accepted friction measurement accessory does not exist. Usually the friction is measured in uniaxial simulators specifically designed for friction experiments of short duration[5]. The effect of wear is not included in such measurements. The friction may vary in tests of long duration due to changes in the contact by wear, creep, temperature increase and lubricant degradation. The friction not only causes heating and wear but it may mechanically contribute to the loosening of the acetabular component in vivo by causing considerable shear stresses at the bone-implant interface[6,7].

Three different methods for friction measurement in the BRM hip joint simulator were described elsewhere[8]. They were based on force measurements on levers that prevented rotation, that is, on the measurement of reaction forces generated by the frictional torque about (a) the vertical loading axis above the BRM mechanism, (b) the leaning axis of the BRM mechanism, and (c) the vertical drive axis of the BRM mechanism. With constant load it was observed that the torques showed sinusoidal variation at a frequency twice that of the cycle frequency. With respect to the observed magnitudes of the friction vectors on the axes of measurement, the drive shaft torque (c) was shown to be very close to the sum of the of (a) and (b). Specifically, (a) was only a fraction of (c), whereas the maximum of (b) was close to the average of (c). The method (b) was used continuously during long wear tests of up to 12.5 million cycle total duration, and the maximum value was recorded. The torque (c) was measured from the reaction force of the drive motor that freely hung on the lower end of the drive shaft, as was done in a subsequent study[9]. It was suggested that the frictional torque could alternatively be measured directly on the drive shaft[8]. In this way, the small positive error caused by the friction of the drive motor gear would be eliminated. In the present study, a frictional torque measurement system with a commercial rotary torque transducer on the drive shaft was implemented. The novel system was used for tests with conventional and crosslinked ultra-high molecular weight polyethylene (UHMWPE) cups against various femoral head materials (polished and roughened CoCr, alumina, zirconia toughened alumina ZTA, zirconia and diamond-like carbon DLC) and different load levels using diluted calf serum as the lubricant. To confirm the relationship between the torques measured in three different ways, the lever force measurement was redone. In addition, friction coefficients were measured with similar material combinations using the circular translation pin-on-disc (CTPOD) device[10] which is a flat-on-flat analogue to the BRM simulator.

Materials and methods

In the novel frictional torque measurement system designed for the BRM simulator, the drive shaft was run via a commercial rotary torque transducer (type DRBK, nominal torque 5 Nm, ETH Messtechnik GmbH, Germany) that was connected by bellow couplings (Figure 1). An additional vertical shaft was mounted in bearings between the transducer and the drive motor. In this way, compression, tension and bending of the transducer, which could cause measurement error and transducer damage, were eliminated. In an earlier study it was found that the friction of the ball bearings of the BRM mechanism was negligible compared with the frictional torque generated by typical total hip prostheses that had an UHMWPE acetabular cup[8]. This was proven by running a test with a water-lubricated alumina-on-alumina joint (28 mm Biolox Forte), with which the torque, including the friction of the ball bearings, was only 0.03 Nm with 1 kN load, the lowest value ever obtained with the device at this load level. The torque against an UHMWPE cup was usually of the order of 1 Nm with 1 kN load[8,9].

The motion of the BRM consisted of flexion-extension (FE) and abduction-adduction (AA) of the femoral head. As the leaning axis was at an angle of 23° to the vertical, the range of both motions was 46°, they were sinusoidal and they had a phase difference of $\pi/2$. The internal-external rotation (IER) of the head was prevented by a lever, the axis of which passed through the centre of the head. The bearing housing of the leaning axis rotated at constant velocity, one revolution per second, about the vertical axis. The head was aligned with the vertical rotation axis to an accuracy of ±0.01 mm using a dial gauge. The pneumatic, static loading was applied vertically from above. The position of the cup was horizontal. It was self-centring on the head as it was loaded through a universal joint. The joint was surrounded by an open lubricant chamber. Care was taken in the mounting of the cup that no air remained between the sliding surfaces.

The tests included two types of UHMWPE cups, conventional gamma-sterilized (ISO 5834-1/-2) and highly crosslinked (Durasul), and six types of femoral heads, polished (R_a = 0.01 µm) CoCr (ISO 5832/12), alumina (Biolox Forte), zirconia toughened alumina (ZTA, Biolox Delta), zirconia (Prozyr), DLC coated CoCr, and deliberately roughened CoCr (R_a = 0.93±0.06 µm representing severe abrasive damage, criss-cross scratching with emery paper). It was specific of the present DLC coating that it did not increase the surface roughness[11]. All heads were of 28 mm nominal diameter. The thickness of the cups (inserts) was 12 mm and they were backed by an acetabular shell made from titanium.

HyClone Alpha Calf fraction serum (SH30212.03) diluted 1:1 with distilled water, without additives, was used as the lubricant. The protein concentration of the lubricant was 21 mg/ml. It was circulated by a peristaltic pump through a heat exchanger. In this way its bulk temperature was kept at 37±1 °C. The volume of the lubricant in the system was 200 ml. At room temperature, the equatorial diameters of the cups were 28.0 mm, close to those of the heads, but the temperature increase by 15 °C in the test increased the diametral clearance to an estimated value of 0.08 mm. The internal shape of the Ti shell allowed thermal expansion of the insert. Frictional heating was likely to further increase the clearance.

The measurement sequence of the drive shaft torque T for each combination of head and cup was as follows. First, 30 min tests (1800 cycles) were run with static 1.0 kN, 1.5 kN, and 2.0 kN loads. After this, the acetabular components were dismounted and the insert was rotated 120° about the axis of symmetry within the Ti shell. The tests with three different load levels were then repeated. Finally, the insert was rotated by another 120° and the tests were repeated once again. The above 4 ½ h sequence was run without changing the lubricant. The average of the steady state T at the end of each 30 min stage was recorded. The mean T and standard deviation were calculated for the three values obtained with each load level.

The friction factor for the BRM results was calculated so that T was divided by the load and by the lever arm of the resultant friction force, that is, by the distance of the theoretical point of load application from the leaning axis, $r\sin 23°$ (see Figure 3 of Ref. 8), where r is the radius of the femoral head. With the 28 mm diameter femoral heads the lever arm was 5.47 mm.

The above materials, excluding ZTA, were additionally tested in the friction measurement circularly translating pin-on-disc (CTPOD) device[10], which is a flat-on-flat analogue to the BRM simulator with respect to the multidirectional relative motion. The three constant nominal contact pressure values (2.4 MPa, 3.6 MPa and 4.8 MPa) were chosen so that they were close to the theoretical maximum contact pressure values in the BRM tests. The running times and the lubricant were similar to those in the BRM tests. In the CTPOD, the sliding speed was 31.4 mm/s, the slide track diameter was 10 mm, and the pin diameter was 9 mm.

To study the effect of wear on friction, the roughened CoCr head (R_a = 0.9 µm) was articulated against a crosslinked polyethylene cup for 1 million cycles. The load was 1 kN, and the type of lubricant was the same as in the 4 ½ h friction tests. T was recorded continuously, whereas the wear was measured gravimetrically at intervals of four days.

For the simultaneous measurement of T, $T_{leaning}$ (the torque about the leaning axis) and $T_{vertical}$ (the torque about the vertical axis above the joint), the lever force measurement accessories were temporarily added (Figure 2). See also Figures 1 and 2 of Ref. 8.

Results

The operation of the BRM simulator with the rotary torque transducer was straightforward and trouble-free. The standard deviation values of T were low indicating a robust measurement system (Figure 3). The following general observations were made on the BRM and CTPOD tests. The friction factor and the friction coefficient always decreased with increasing load (Figures 4 to 5). There was no consistent difference between conventional and crosslinked polyethylene with respect to friction. The coating of CoCr with DLC did not result in a reduction of friction. In the BRM tests, the lowest friction against both types of polyethylene cups was obtained with zirconia heads.

In the 1 million cycle wear and friction test with a rough CoCr head against a crosslinked polyethylene cup, T increased during the running-in phase by 75 per cent from a minimum value of 1.2 Nm to 2.1 Nm, after which it remained between 1.5 Nm and 2.0 Nm (Figure 6). The wear factor, calculated from the wear rate, was 3.6×10^{-5} mm^3/Nm.

In the simultaneous measurement of T, T_{leaning} and T_{vertical} it was confirmed that with respect to the magnitudes of the friction vectors on the axes of measurement, T was close to the sum of the of T_{leaning} and T_{vertical} (Figure 7). T_{leaning} was found to be sinusoidal at a frequency of 2 Hz and its maximum value was close to the average of T. The minimum of T_{leaning} was typically larger than the maximum of T_{vertical}. Similarly, T_{vertical} was sinusoidal at a frequency of 2 Hz and its minimum was close to zero. The amplitudes of the two torques were close to each other, but they had opposite phases. Even the maximum of T_{vertical} was only a fraction of the average of T.

Discussion

A novel method for the measurement of the total frictional torque of prosthetic hips in the widely used BRM hip wear simulator[1] was introduced. Earlier friction measurements in commercial BRM simulators were based on the measurement of torque about the vertical axis above the joint[2–4], a method which was shown to sense only a minor part of the total frictional torque[8] (Figure 7). Moreover this method was shown to be quite insensitive in situations with a very small contact area, such as that with alumina-on-alumina, simply because the resultant frictional force had no lever arm relative to the vertical axis[8]. The present system instead measured the total frictional torque T on the drive shaft, which was shown to be, with respect to the magnitudes on the axes of measurement, close to the sum of the torques about the vertical axis above the joint, T_{vertical}, and the torque about the leaning axis, T_{leaning} (Figures 2 and 7). In can be stated a torque transducer above the joint measuring about the vertical axis senses only the minor part of the total frictional torque which is not sensed by the rotation prevention lever of the leaning axis. Therefore the measurement of friction in the BRM simulator with a transducer above the joint sensing only T_{vertical} cannot be recommended, since the torque values obtained are likely to be too low. This holds true irrespective of the fact whether the position of the joint in the simulator is inverted or not.

When the cup was loaded via an axial (thrust) ball bearing and its rotation was prevented by a lever, and the rotation prevention lever of the leaning axis was removed, T_{vertical} became equal to T, which was unaltered, and the head rotated slowly about the leaning axis[8]. Similarly, when the rotation prevention lever of the cup was removed, and the rotation prevention lever of the leaning axis was present, T_{leaning} became equal to T, which again was unaltered, and the cup rotated very slowly and intermittently, even reversing, about the vertical axis[8]. These two trials served as a further indication that the total frictional torque of the prosthetic hip in the BRM simulator indeed consists of two distinguishable components.

One could ask why T_{vertical} differs from T as they both are measured about the same vertical axis. This can be understood by first noting that the vertical axis is the IER axis of the cup, but there is no IER in the device. Only in the case that the axis of the rotation prevention lever does not pass through the centre of the head there is a small IER component in the motion of the head[12], which is sensed by the transducer above the joint, but this motion is likely to be of minor importance tribologically. Second, the prevention of IER (complete or partial) of the head about the leaning axis requires force, and the moment generated by this force increases the torque needed to rotate the drive shaft. Third, there is the BRM mechanism between the joint and the drive shaft transducer, and the motion of the head consists of the FE and AA. The BRM mechanism is analogous to a cradle mechanism with two axes perpendicular to each other, such as that in the HUT-4 simulator[13]. Their outcomes are similar, although their implementations differ. Note still that in the HUT-4 the range of AA is 12° which is biomechanically more realistic than the 46° AA range of the BRM[14]. This was the reason for using a cradle mechanism in the HUT-4 with an outer cradle for the FE and an inner cradle for the AA. Hence, no matter how the FE+AA-mechanism is implemented, the drive shaft(s) will sense the frictional torque(s) attributable to the FE and AA. This holds true even if the drive shaft is vertical, as in the BRM simulator, in which case the single drive shaft senses the torques about both the FE and AA (theoretical) axes. In the present BRM, the axis of the rotation prevention lever is the AA axis, whereas the leaning axis is the IER axis of the head, the rotation of which is completely prevented. The FE axis is stationary, horizontal and always perpendicular to the AA axis. The BRM design is in fact ideal for friction measurement because the moments of inertia within the mechanism are negligible in comparison with oscillating cradle mechanisms, which need to be large, so that there is space for the test chambers, but still rigid, and which therefore are heavy[13].

In the calculation of the friction factors for the BRM results it should be noted that the true, effective lever arm of the frictional force[15] is naturally unknown. The effective lever arm depends on the slide track pattern[12], contact pressure distribution[16], and dependence of μ on contact pressure with different bearing couples in serum lubrication[10]. In this study the value $r\sin23°$ (5.47 mm) was used[8]. The simplification was based on the fact that the distance of the theoretical contact point from the leaning axis was $r\sin23°$. This was therefore the lever arm of the theoretical resultant frictional force, which travelled along the circular force track on the head (67° latitude, 'polar circle'), about the leaning axis at a constant velocity of 34.4 mm/s. The observation that the friction factor and the coefficient of friction decreased with increasing load (Figures 4 and 5) indicated a mixed lubrication mechanism. This is in agreement with earlier studies[10,17]. With an UHMWPE cup, the frictional torque is relatively high, as the contact area is large, due to the low elastic modulus[18] and considerable creep[19], in comparison with hard-on-hard couples in which the hydrodynamic lubrication is possible[20]. In the present study, the wear marks on the cups, scratches and flattening of machining marks, extended to a distance of a few mm from the equator already in the 4 ½ h tests.

The observation that there was no categorical difference in friction with conventional versus crosslinked polyethylene cups (Figure 3) is in agreement with another BRM study[21], and with a study in which the torque of prosthetic hips was measured in a loading frame[22]. As for the differences in T attributable to the various femoral head materials, the results were in line with those of the earlier BRM studies using lever force measurement[8,9]. Especially the superiority of the zirconia head showing the lowest friction[8] was corroborated by the present study. However, a detrimental phase transformation phenomenon *in vivo* led to the recall of the Prozyr heads, although the problem was related to a few manufacture batches only[23]. The roughening of the CoCr head did not always result in the highest T values in the 4 ½ h tests, which was in agreement with earlier studies[8,9]. In the wear test however, roughening resulted in a substantial increase of T, and in very high wear of crosslinked polyethylene (Figure 6). The R_a value of 0.9 µm represents severe, yet clinically relevant roughening that can be caused by dislocation[24]. Even crosslinked polyethylene that shows low wear in normal conditions appears to be vulnerable to severe roughening of the counterface[9].

Neither the BRM nor the CTPOD tests showed an advantage in the coating of the CoCr with DLC from the point of view of friction. This is in agreement with an earlier BRM study that showed no advantage in the DLC coating from the point of view of UHMWPE wear[11]. The above held true also for alumina in comparison with CoCr, but the superior abrasion resistance of alumina should be borne in mind. The clinical findings on DLC coatings are controversial[25]. The CTPOD results did not fully agree with the findings of the BRM tests though. In the CTPOD, zirconia did not show the lowest friction, but polished CoCr, whereas the roughening of CoCr resulted in the largest μ values (Figure 5). The kinetic analogy between the two devices is based on their multidirectionality. In both of them, the resultant friction vector rotated about the load axis at constant velocity, one revolution per second, and so the direction of sliding changed continually relative to the UHMWPE specimen. This is of fundamental importance with respect to realistic wear mechanisms[8–13]. It can be summarized that it is advisable to measure the friction while multidirectional wear simulation of long duration is being performed, so that the effect of wear on friction can be evaluated.

As limitations of the present study the horizontal position of the cup and the static load could be mentioned. Realistic wear has nevertheless been produced under these test conditions[11]. No signs of lubricant depletion have been observed. The BRM simulator can be used even with high inclination angles of the cup[26], and various dynamic loading profiles are used in commercial BRM simulators[2–4,26]. These are no hindrances to the frictional torque measurement on the drive shaft. However, if the loading in a BRM simulator is applied from below via the drive shaft, the present principle is not readily applicable because the rotary torque transducer must not be loaded axially. Moreover the bearings of the BRM mechanism must have low friction because their friction adds to the torque sensed by the transducer on the drive shaft. This requirement is met by ball bearings (Figure 1), in which friction is two orders of magnitude lower compared with serum lubricated prosthetic joints that have an UHMWPE acetabular cup[8].

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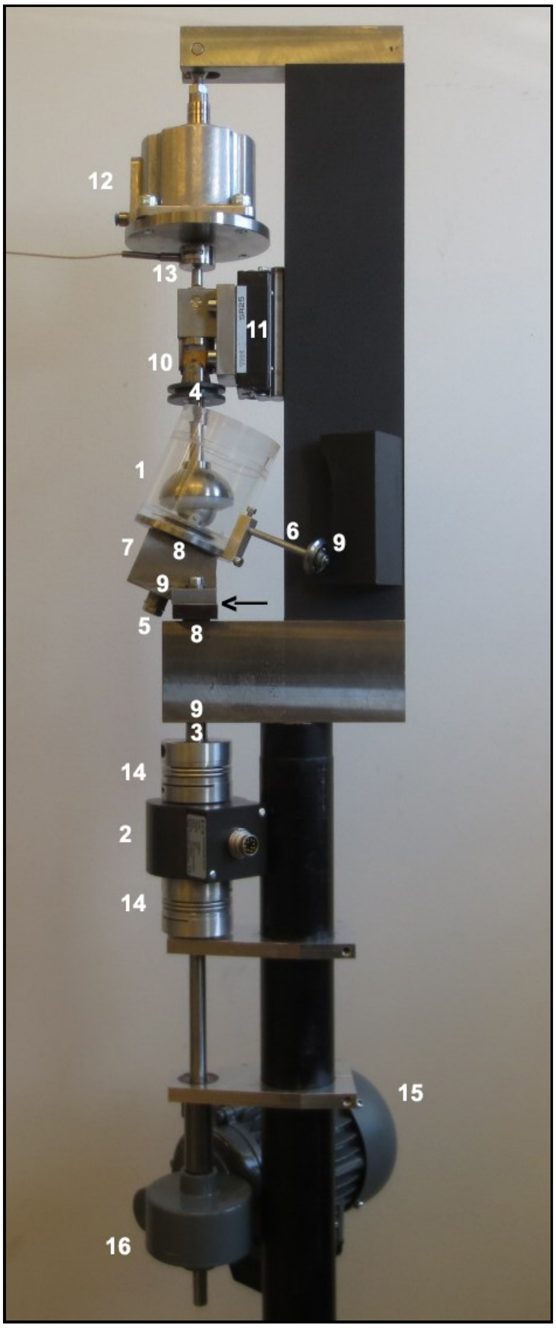


Figure 1. Biaxial rocking motion (BRM) hip wear simulator with rotary torque transducer on drive shaft for friction measurement. 1 Test chamber, 2 Torque transducer, 3 Drive shaft, 4 Vertical loading axis, 5 Leaning axis, 6 Rotation prevention lever, 7 Rotary bearing housing, 8 Angular contact ball bearing, 9 Ball bearing, 10 Universal joint, 11 Linear bearing, 12 Loading cylinder, 13 Load cell, 14 Bellow coupling, 15 Drive motor, 16 Gear. Note ball bearing on end of rotation prevention lever minimizing friction at this contact. Arrow indicates horizontal adjustment plane for femoral head alignment.

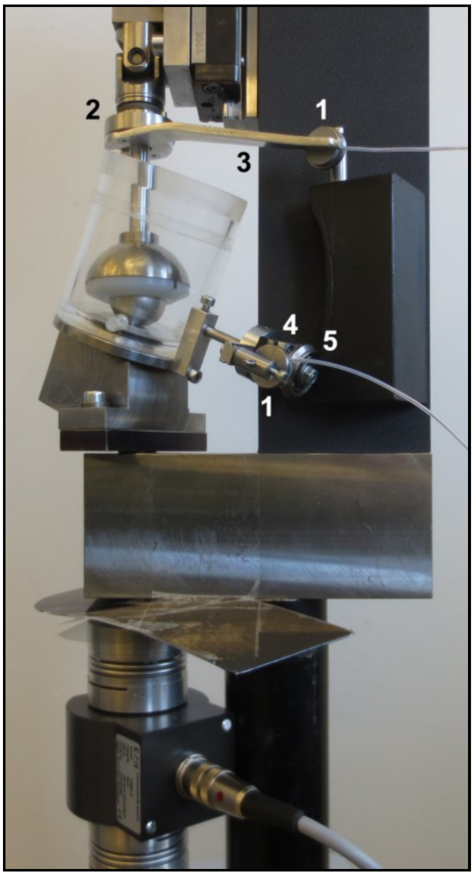
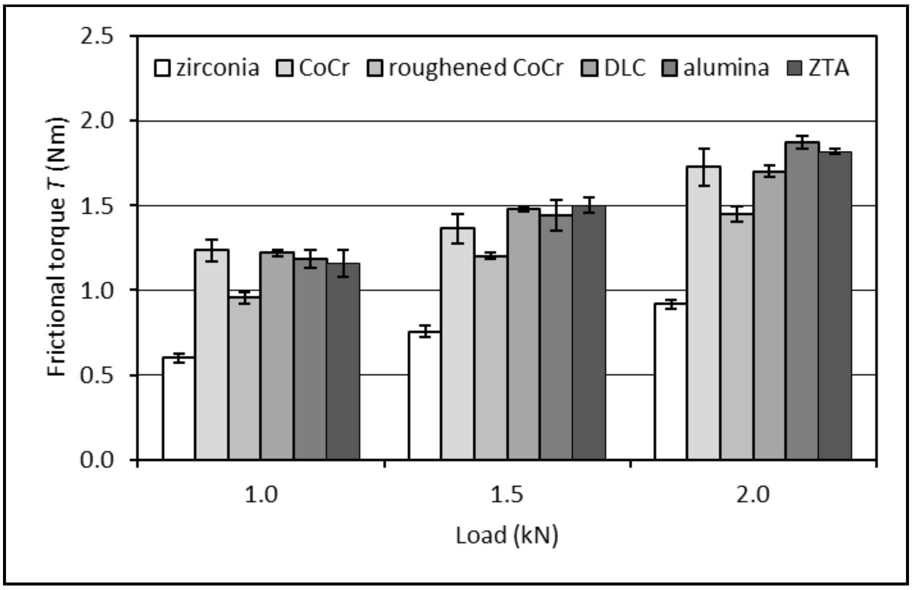


Figure 2. Accessory for simultaneous measurement of T, T_{vertical} and T_{leaning} in BRM simulator. 1 Load cell, 2 Axial ball bearing, 3 Rotation prevention lever of acetabular cup, 4 Linear bearing, 5 Ball bearing.



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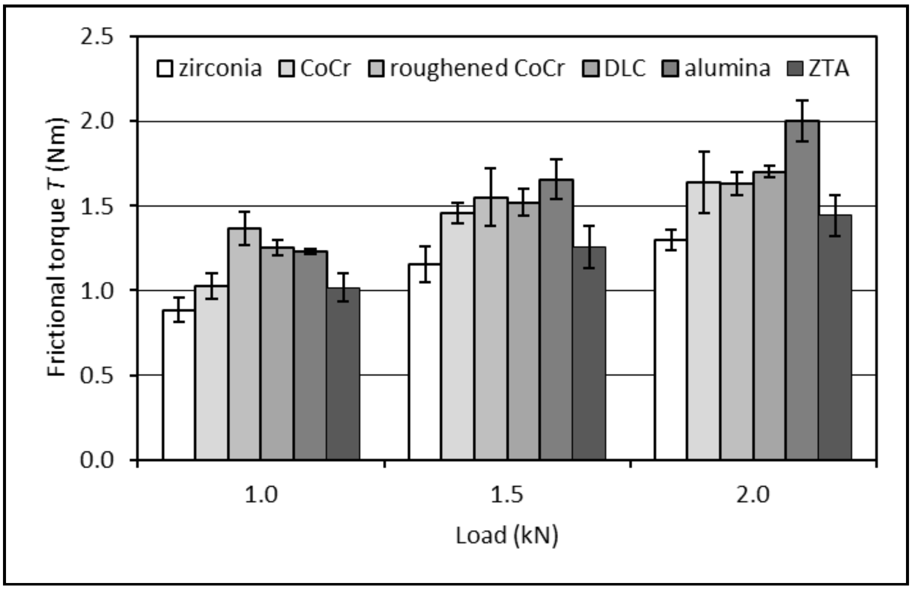
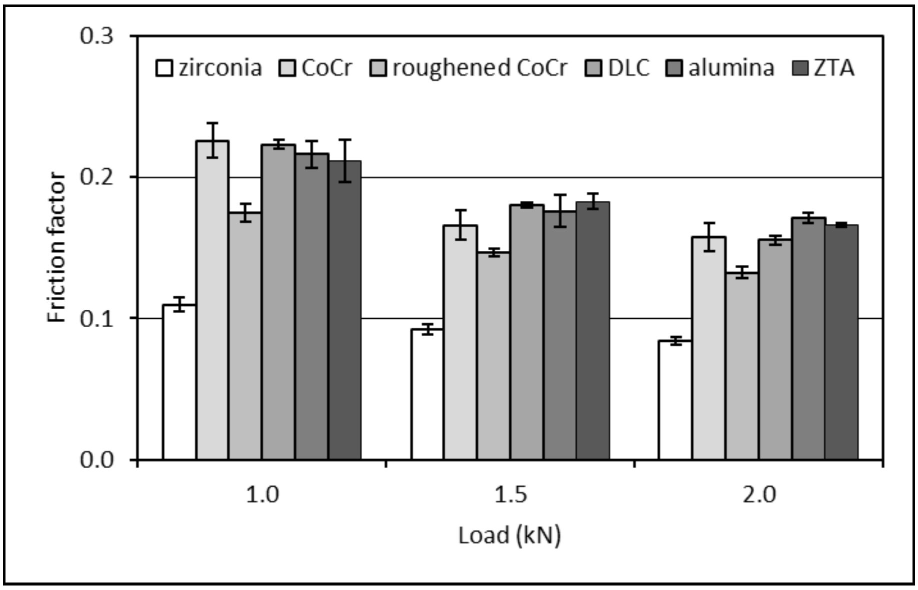


(b)

Figure 3. Frictional torque T (mean and SD) against (a) conventional and (a) crosslinked UHMWPE acetabular cups with different femoral head materials and loads measured in BRM hip simulator. Head diameter was 28 mm, and lubricant was diluted serum.



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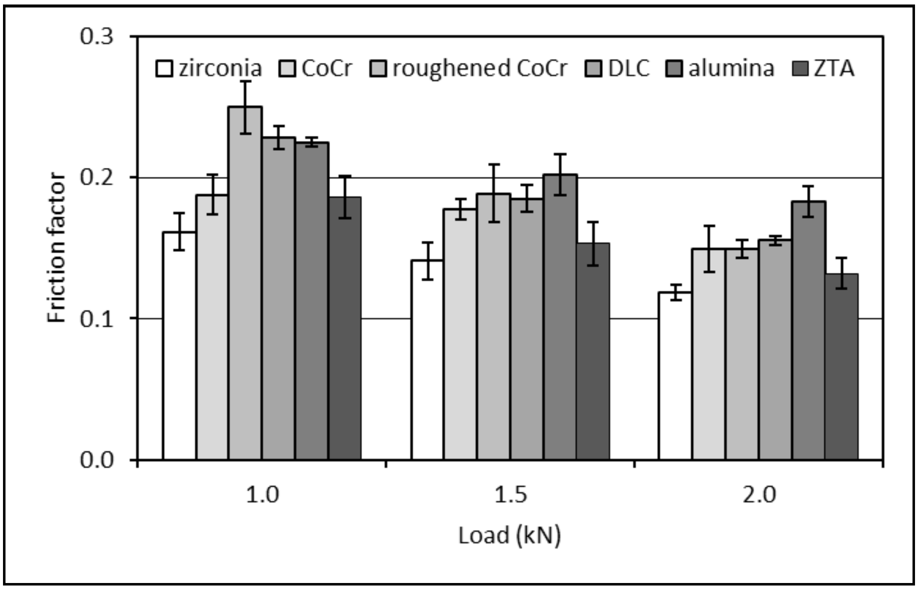
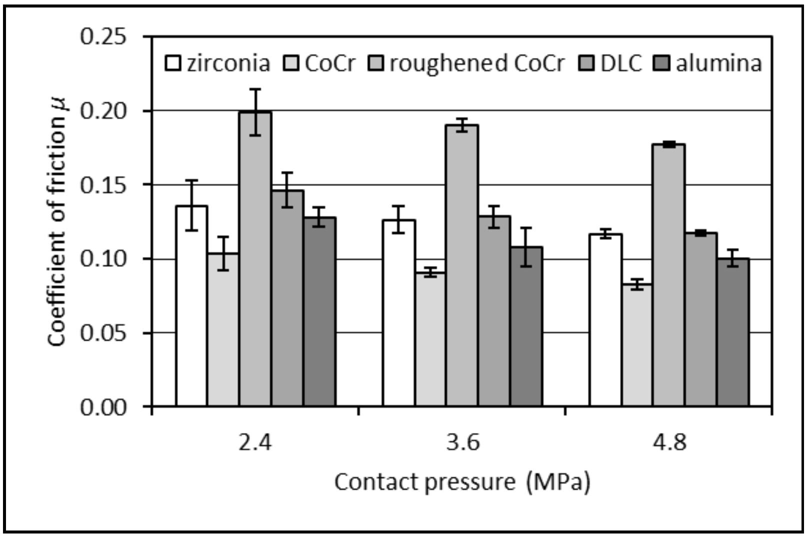
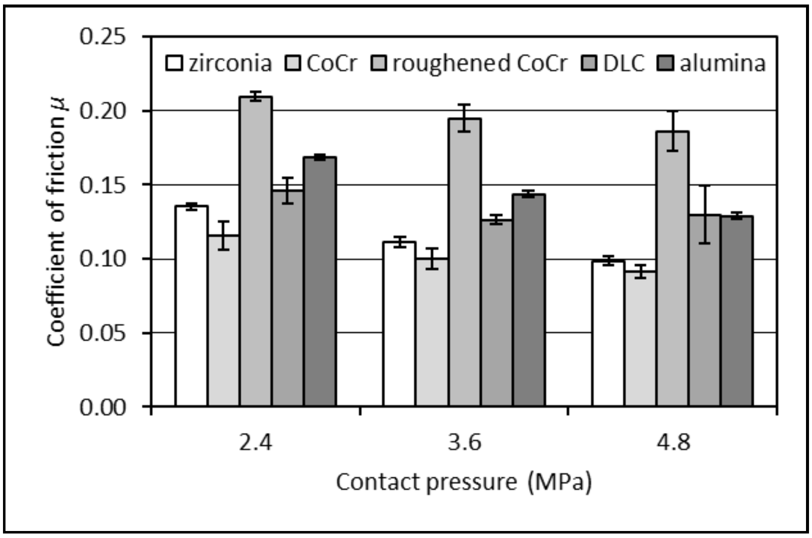


(b)

Figure 4. Friction factor (mean and SD) against (a) conventional and (a) crosslinked UHMWPE acetabular cups computed from *T* values presented in Fig. 3.



(a)



(b)

Figure 5. Coefficient of friction μ (mean and SD) against (a) conventional and (b) crosslinked UHMWPE pin with different disc materials and contact pressures measured in CTPOD device. Contact was flat-on-flat, and lubricant was diluted serum.

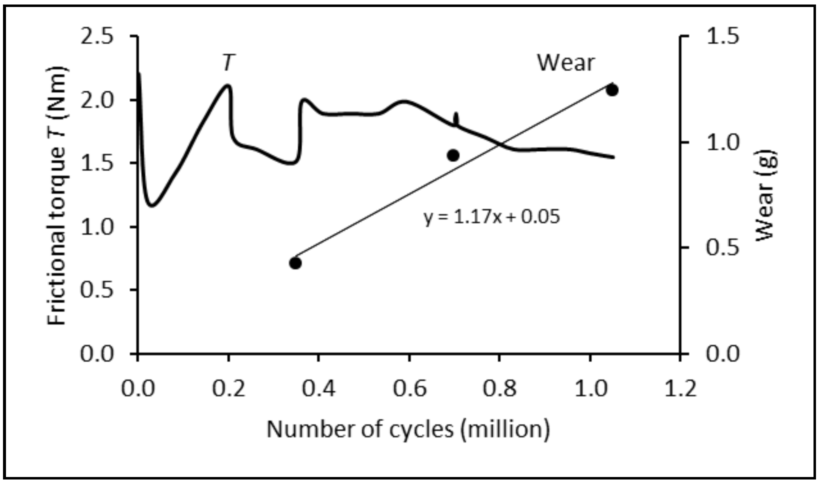


Figure 6. Frictional torque T and wear of crosslinked UHMWPE cup against roughened (R_a = 0.9 μm) 28 mm CoCr head. Load was static 1 kN, and lubricant was diluted serum.

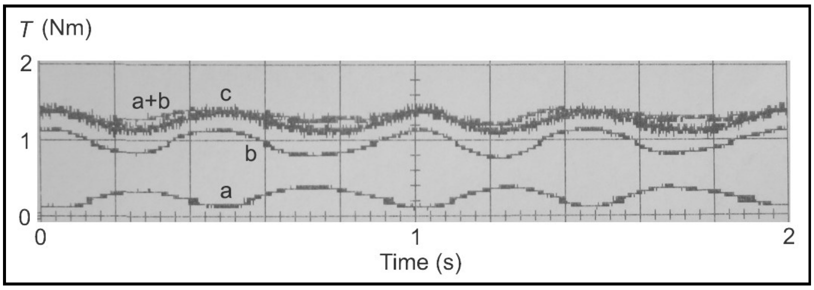


Figure 7. Simultaneously measured (a) $T_{vertical}$, (b) $T_{leaning}$, and (c) total frictional torque T in BRM simulator with roughened CoCr against UHMWPE, static 1 kN load and diluted serum lubricant. Note that with respect to magnitudes, T is very close to $T_{vertical}$ + $T_{leaning}$.