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Torsional friction moments in artificial hip joint bearings

- A biomechanical *in-vitro* study

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1 Introduction

1.1 Background

The reduction of the friction moment in the artificial hip joint bearing initiated the success of artificial hip replacements (Charnley, 1979). However, small metal heads running against UHMWPE bear a high risk of luxation, which is not acceptable for young and/or active patients. This, together with PE wear-related problems has led to the revision of large-diameter metal-on-metal bearings. Large diameter metal-on-metal bearings (e.g. McKee-Farrar hip implants in 1965) have generally been less successful in the past and the failure rate was high However, some of these bearings survived for very long time periods (Higuchi et al., 1997).

The loosening of large diameter Me-Me bearings in the past is speculated to be due to interface friction or jamming rather than to metallosis. This problem has been overcome by new designs and modern manufacturing methods. The problem of elevated metal ion content in the blood is still not completely solved. Presently several new implants have been introduced onto the market (Figure 1).

The most wear-resistant material combination, however, is still the ceramic-on-ceramic pairing. Material deficiencies and fracture risk have been progressively resolved in the last decade (Bierbaum, 2002, Santavirta, 2003). A new trend seems to be the pairing of metal and ceramics, combining the good fixation possibilities of a metal cup with the advantages of a ceramic head.

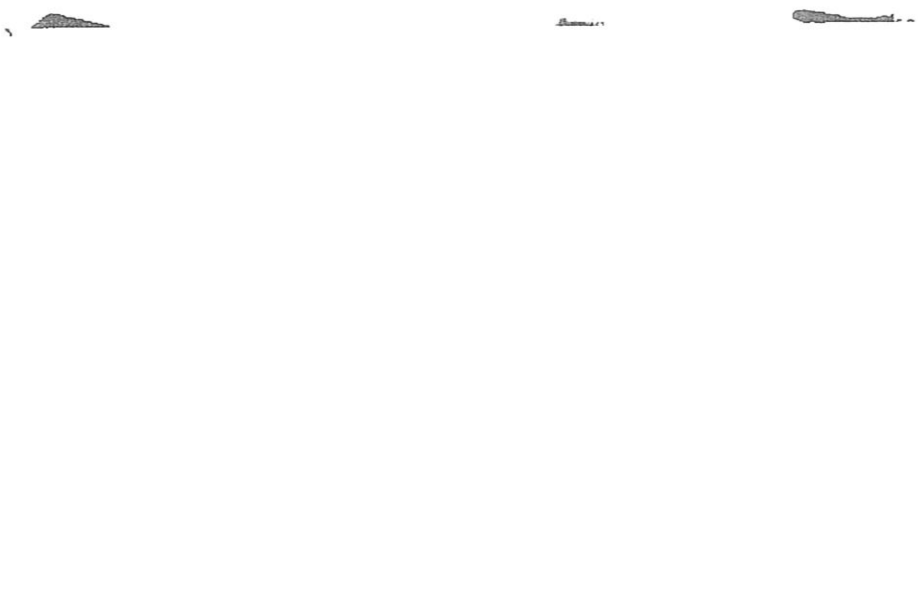

Figure 1: Contemporary large diameter Me-on-Me surface hip replacements.

1.2 Previous study

A previous study was conducted to assess the moment magnitudes acting in large diameter metal-metal bearings, and to make comparison with other contemporary components ([redacted] et al 2004, [redacted] internal report for [redacted]). A custom hip-simulator was constructed to measure moments acting in the saggital plane, which coincides with the plane of motion of the leg during normal activities, and is therefore expected to be the plane of the maximum moment magnitude.

Higher moments for large diameter metal-on-metal bearings were observed for water vs. serum, start-up vs continuous motion, unworn vs. worn bearings, lower oscillation frequency, longer resting period, higher radial clearance, and diametral deformation of the cup. These results were consistent with theoretical models of hydrodynamic fluid film lubrication in such bearings. The moment was most sensitive to the fluid medium, for which water increased moments by factor 2-3 relative to serum.

The same trends were observed for other bearing pairs, but magnitudes were more than 2.5 times less than those for the large diameter me-me bearings. Nevertheless, it was concluded that the current design for the [redacted] implant under normal patient conditions and activities was well within safety zone regarding moment magnitudes, based on comparison with higher failure loads for such implants.

1.3 Purpose of the study

It has been proposed that the torsional component of the moment acting between cup and head may be responsible for some failures of both components, particularly since torsional rotation has been observed in failed components ([redacted] et al, 2005). It is also conceivable that friction is high under torsional loading because there is no relative translation at the pole, where most of the load is transferred. This could preclude the development of a fluid layer in large diameter bearings, which is presumed to reduce friction under rolling motions (as tested in the simulator). Furthermore, torsional testing of components is perceived as a more simple mode of testing, which could be used as a substitute for the more complicated simulator tests.

Therefore, this study replicates the previous simulator experiments, but measures the polar torque, rather than the moment in the saggital plane. The aim was to compare magnitudes with previous moment measurements from the simulator and with measured failure loads, particularly for large diameter bearings.

2 Materials and Methods

2.1 Study design

Four bearing pairs were tested, comprising metal (me), polyethylene (pe) and ceramic (ce) components. The dynamic torque profile was measured for a constant cyclic torsional rotation amplitude of +/- 20°, similarly to the physiological range of motion assumed for gait in the simulator. A cyclic axial force was applied in phase with the motion, to simulate the joint load during gait. The following variables were considered:

- bearing pair (55mm me-me, 36mm me-me, 28mm me-pe, 28mm me-ce) 4 levels
- history (unworn, worn) 2 levels
- lubrication fluid (water, bovine serum) 2 levels
- resting duration (10s, 30s) 2 levels
- movement frequency (0.5 Hz, 1 Hz, 2 Hz) 3 levels
- dynamic load curve variation (minimum loads of 100N, 250N, 500N) 3 levels

The study thereby resulted in a full matrix of variable permutations, apart from the metal-polyethylene bearing pair, which was tested only in its new state (not worn). Each bearing pair was tested 3 times for each permutation.

2.2 Bearings

2.2.1 Bearing pairs

The following implants were used for the tests:

#1 [redacted] metal-metal resurfacing implant	Ø 55 mm
#2 [redacted] metal-metal implant	Ø 36 mm
#3 [redacted] metal-polyethylene implant	Ø 28 mm
#4 [redacted] metal-ceramic implant	Ø 28 mm

2.2.2 Bearing history

Bearing pairs labelled 'worn' were subjected to 5 million loading cycles in a hip simulator ([redacted]) prior to testing.

2.3 Test apparatus

An MTS Bionix 858.2 servohydraulic testing machine was used to apply cyclic torsional rotation to each bearing, with a superimposed cyclic axial force, which was isolated by incorporation of an X-Y table perpendicularly to the machine axis (Figure 2). Each cup was cemented into a 75mm diameter cylindrical depression in a steel block with methyl-methacrylate. The cup was able to rest on the metal surface. The block was mounted in a bath of lubricating fluid, which was mounted on the X-Y table. The ball was mounted to the vertical ram of the testing machine by press fit on a conical stub or by cementation (55mm heads: internal fixation of a tubular cylinder over the pin).

2.2.3 Lubrication medium

As lubricating media distilled water and distilled water with 15 % calf serum were used.

2.4 Loading

Implants were rotated sinusoidally with an amplitude of 20°, similarly to the simulator study ([redacted] *et al*, 2004). A dynamic force was superimposed axially, according to joint loads measured for normal walking ([redacted], 2001) (Figure 3). Loading was designed to simulate walking ([redacted] *et al*

2001) from stand-still, to measure the start-up torque peak. and also the mean peak torque over 20 cycles. Therefore the displacement was started 100ms before the load cycle in order to simulate the start of walking from stand-still and a preload F_{start} = 650N was applied to simulate the joint load for 2 legged stance. The magnitude of the peak applied force was 2000N (Figure 3).

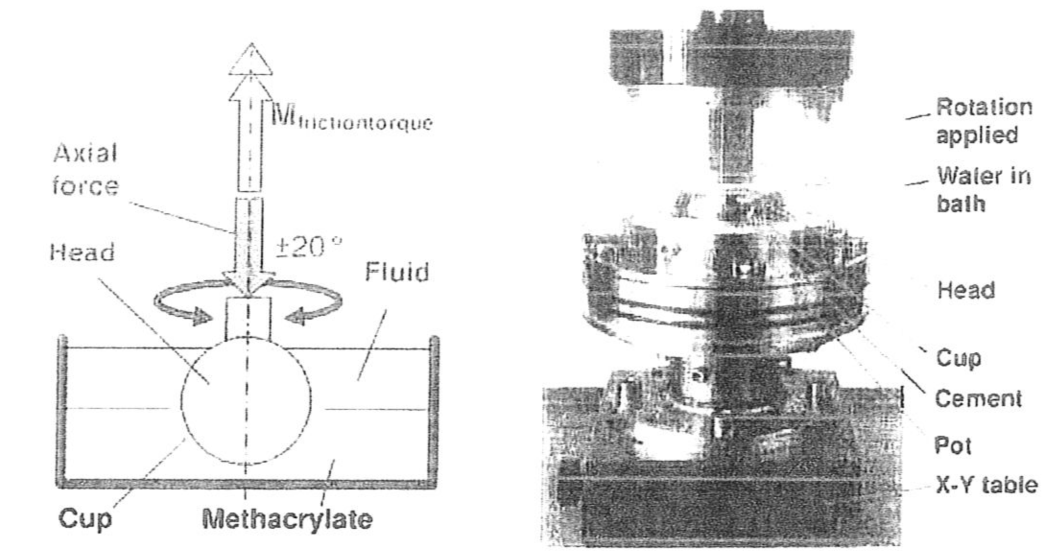

Figure 2: Torque testing apparatus mounted on an MTS universal testing machine.

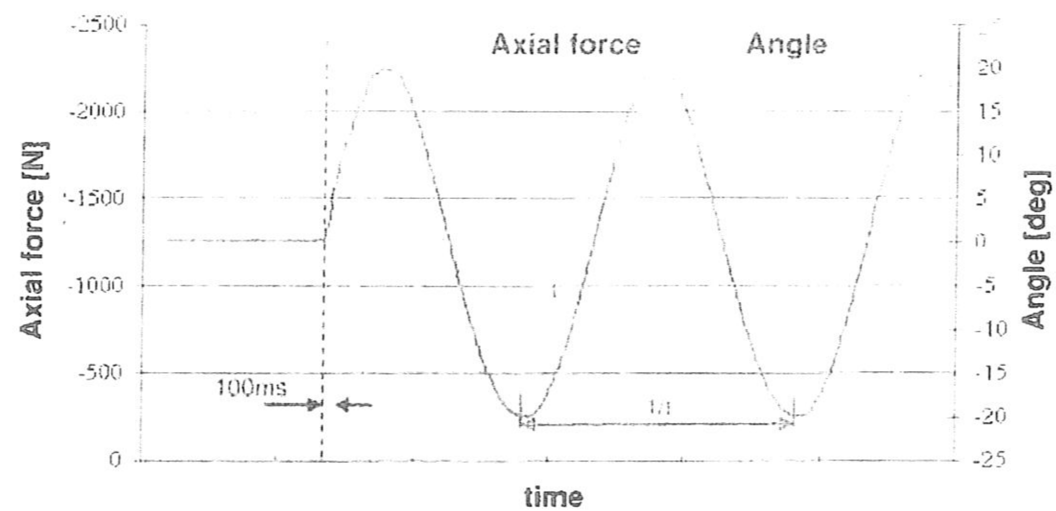

Figure 3: Time course of the dynamic load and the rotation angle.

2.2.4 *Load frequency*

A loading frequency **f** (Figure 3) of 1Hz is generally assumed for walking. This was also halved and doubled to investigate the effect of loading rate for 0.5Hz, 1Hz, 2Hz.

2.2.5 *Swing load*

The minimum load applied during the cycle (F_{swing}, Figure 3) represents the swing phase of gait and has been found to influence the wear characteristics of bearings ([redacted] *et al*, 2004). The magnitude measured for walking is 250N ([redacted] *et al*, 2001) and one value on either side was also applied to investigate 3 levels: 100N, 250N, 500N.

2.2.6 *Resting periods*

The coefficient of friction has been shown to increase after a resting period, which is thought to be due to loss of lubrication fluid and consequently the direct contact of the frictional partners ([redacted] *et al* 2003). For this experiment the resting durations (t_{rest}, Figure 4) of 10s and 30s were compared

2.5 Testing procedure

After fixation of the implant inside the simulator and addition of the lubrication fluid to the pot (Figure 2), the test procedure shown in Figure 4 was applied as a series of loading cycles for each set of test conditions. Preloading of 650 N was applied prior to each loading cycle to simulate two legged stance. Each test was repeated three times for each implant. The head was briefly lifted off the cup after each step to assure replenishment of the lubricating fluid. Data acquisition (torque) was performed continuously throughout each test series at 400 Hz.

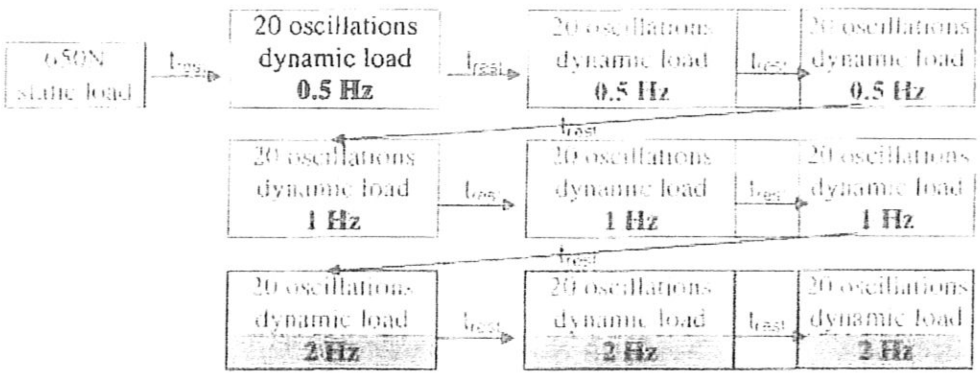

Figure 4: A single test series procedure.

2.6 Analysis

The resulting torque for each motion cycle was analysed with respect to start-up and peak cyclic magnitude. The start-up peak was determined at the beginning of the first movement cycle (Figure 5) and cyclic peak torque was determined as the mean of the torque peaks for the middle 10 loading cycles (cycles 6-15). Statistical analysis was performed using One-way Analysis of Variance and Regression Analysis with $\alpha = 0.05$ (confidence level 95%, SPSS for Windows).

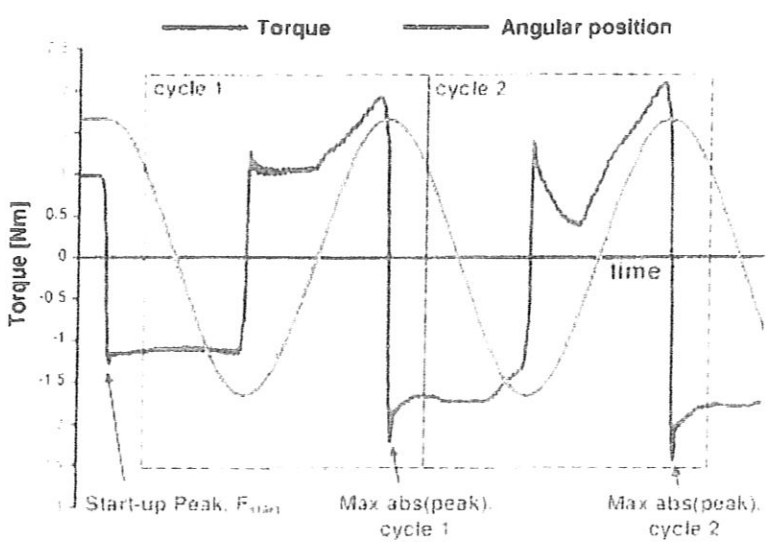

Figure 5: Definition of the analysed distinct time instances on the torque curve.

4 Discussion

4.1 General

A torsion test was conceived based on a similar previous study in a hip simulation system. A full matrix of data was collected for 5 variables for 4 bearing pairs. The level of spread within each test series (3 backswing loads at 3 frequencies. 3 times each) was generally quite low. One exception was a new 36mm me-me bearing tested in water with 10 rest period (See (?), Figure 8). It seems that a small defect or a third body particle may have increased the friction initially, which then becomes either embedded, crushed, dissolves or leaves the bearing space, since it was observed that the magnitudes decreased to more consistent magnitudes during the series test (not shown).

4.2 Torsion versus simulator testing

The torque measurements in this study were made with similar input variables to those used in the previous study using a custom hip simulator. In that study the device displacement of the bearing was made in a rolling mode, according to physiological conditions. The peak moment magnitudes for both loading modes were compared for the 55mm me-me bearing, shown in Figure 12.

Figure 12: Comparison of maximum positive and negative cyclic moment magnitudes for the current torsion test and the previous hip simulator test.

Positive and negative values are shown, since they were found to have different magnitudes in the simulator study (Figure 12), probably due to the asymmetric loading conditions applied. It can be seen that this effect is much less pronounced under torsional loading It is clear from Figure 12 that the moment magnitudes are much lower for the torsion study than for the simulator, especially for adverse lubricant conditions (simulated by the use of water as lubricant). Magnitudes range between 2 and 10 times lower in torsion. Furthermore, the trends observed within each variable level are inconsistent between the two testing modes for the simulator moments are much lower for serum than for water (about factor 2), while moments are greater for serum than water under torsional loading. This could be an artefact of the testing order since water was always tested prior to lubricant. Since the magnituide of the differences between water and serum are rather minor, the increase could be attributed to wear of the earlier tests or to some unknown phenomenon (which are speculated on in the next section). In the simulator moments were found to reduce with increasing frequency, while under torsion the frequency had no clear effect. In the hip simulator, the moment was less for longer resting durations, while no influence was found under torsion loading. For both loading modes, no effects of backswing load were observed. Therefore, the two loading modes cannot be compared and it would appear that different load transfer mechanisms are responsible for different trends.

An effect of the backswing load was not observed in either of the studies, which may be due to the relatively low number of cycles tested. A reduced backswing load has been proposed to allow better replenishment of the fluid layer and consequently lower wear rates (V[redacted]t al, 2004). We found no corresponding decrease in joint moment. Besides, the swing phase load is not really possible to control in a patient.

4.3 Mechanisms

Observation of the cup and head after torsion testing showed wear marks at the pole of both components, with very little evidence of contact elsewhere over the surfaces, particularly in the me-me bearings (Figure 13: Point contact and "wear" in the 55mm me-me bearing..Figure 13). This polar point contact occurs due to the designed radial mismatch (clearance) between cup and head. Geometry measurements using a Mitutoyo BHN 305 (Resolution: 0.5µm, accuracy: < 3µm), however, did not show measurable deviations from the original shape of the implants. Since there is no relative motion at the polar point of contact, there is little chance for fluid film lubrication to evolve. Furthermore, there is no shear motion of the head surface in and out of the cup so that there is little possibility for fresh fluid to be induced into the bearing. The moment peaks appeared to increase somewhat in the first few cycles of every 20 cycle sequence, which indicates that a small amount of trapped fluid was squeezed out in the first few cycles, leaving direct contact of the solid surfaces. With such a small area of contact, very high pressures could be expected. This, combined with continuous rotation around one point would be likely to result in substantial wear

Figure 13: Point contact and "wear" in the 55mm me-me bearing..

Some rather unusual results were observed for the torsional loading mode. In particular, the serum as a lubricant had the effect of increasing friction in the joint in comparison with water. It seems that the effect of the serum being twisted between the bearing surfaces on one contact point increases the friction properties. The high pressure, and lack of fluid exchange may serve to denature, or even dry out the serum, perhaps generating a layer with a higher friction coefficient. This would not occur for distilled water. In contrast, serum reduces friction under rolling conditions because hydrodynamic lubrication can be developed. In this case the serum reduces friction due to higher viscosity, which results in a thicker fluid layer.

A lower friction coefficient was observed for a worn joint than for a new bearing pair under torsion loading. This could be explained by bedding-in, in which a steady wear state is achieved once a sufficient contact area has been attained. This is a well documented phenomenon for metal-metal bearings ([redacted] et al. 2004)

In contrast to the torsion measurements, a fluid film is thought to develop under normal rolling or sliding motion (simulator) in large diameter me-me bearings, due to relative motion between the surfaces. This explains the decrease in moments measured for increasing loading frequency in the simulator, in comparison with frequency-independent behaviour under torsional loading.

4.3 Absolute effects

The effect of moment transfer over the hip joint has been proposed as a reason for failure of either the cup or head components ([redacted] *et al*, 2005). Sufficiently high moments could cause interface overload or cyclic micromotion, which can prevent bony ingrowth and possible osteolysis However, the comparison of the measured moment magnitude with some experimental values for the cup holding capacity in bone re. PU foam performed in unsubmitted testing indicated that the [redacted] maintains a safety factor of at least 2 or 3 with regard to overload (column 2 in Table 2). In the present torsional test, the bearing moment magnitudes are clearly lower in torsion than in simulated gait (Simulator) and, furthermore, the failure loads in Torsional direction are clearly higher (column 3 in Table 2). Therefore, failure in torsion loading is clearly less probable than in simulator loading. In fact, failure under torsional rotation is particularly unlikely, since this mode of loading would not be easy to achieve physiologically.

Table 2: Moments measured for 14 different press-fit cups under torsion and roll-out failure in foam 'bone' specimens ([redacted] al, 2003, internal data). Torsion relates to the loading applied in the current study; Roll-out relates to the moment applied in the hip simulator study. Failure occurs at lower moments in the hip simulator mode. See also [redacted] et al, 1989, [redacted] et al 1991, [redacted] et al 1983 for similar values.

	Failure Moment [Nm]	
Cup type	Torsion	Roll-out
[redacted]	[redacted]	[redacted]

Although failure appears unlikely due to friction moments, some early failures have indicated a roll-out failure mechanism for the cup, judging by its final position radiographically. The two critical conditions for such a failure are a high load and a weak interface. Perhaps a single high-load event could create such high instantaneous moments (for example during stumbling), in combination with a poor press fit of the cup component.

4.4 Limitations

Although a large number oftests were undertaken, with three measurements for each permutation of factor (5) for each bearing pair (4). it must be noted that only one bearing combination was tested for each series. Therefore we have a sample size of n=1. Furthermore, each test series (3 backswing loads at 3 frequencies, 3 times each = 27 measurements) was conducted in series on the same head. This could cause the conditions to change progressively throughout a series test. However, very little variation was found in any of the [redacted] torque measurements made within each continuous series, indicating that there is not a progressive effect of measurement sequence.

It should also be noted that the torque test employed in the current study does not appear to be a physiological loading condition. Although the femoral head rotates approximately in torsion relative to the cup, the joint load does not act through the torsion (polar) axis of the joint, as is modelled in the current study Figure 14 The actual joint load acts in a more vertical direction, which loads the top side of the cup Off-axis loading would lead to a higher effective moment arm (greater distance to the torsional axis) and also relative motion of the bearing surfaces at the point of loading. This would result in considerably different joint mechanics and for this reason the joint simulator was constructed and should probably be used in future for such assessment of bearings.

The torsional component of the joint moment measured using the hip simulator was found to be less than [redacted]n for any cup combination in a previous study ([redacted], 2004) and would not be considered as dangerous as the possibility of roll-out.

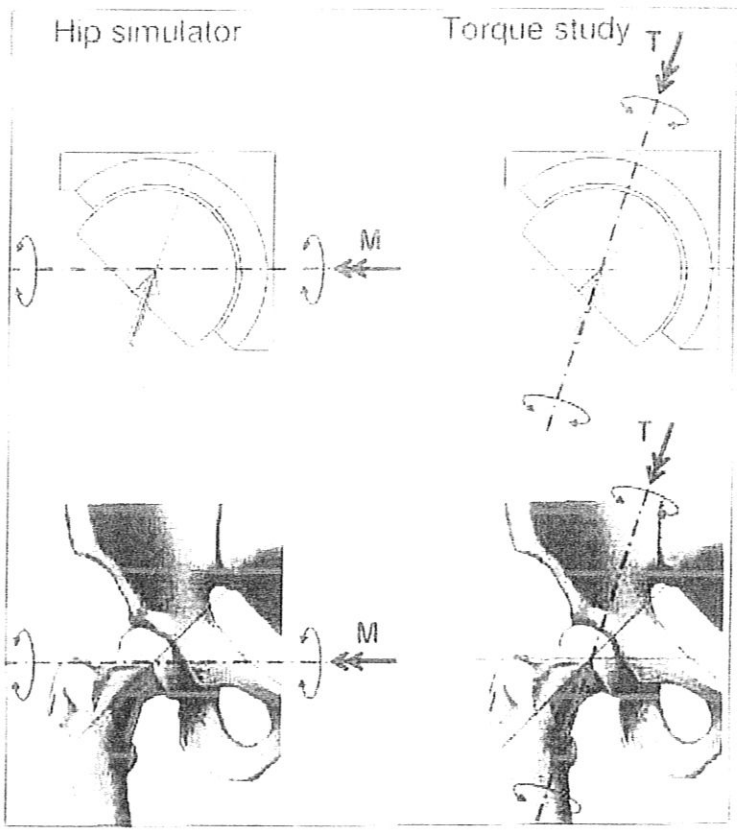

Figure 14: Diagram of the implanted hip, showing the relative orientations of joint load and rotational axis in two different testing modes. The hip simulator was designed to represent the gait cycle. The torsion test aligns the joint load axis with the rotation axis, thus effectively simulating a nearly vertical twist of the femur relative to the hip. This is rather un-physiological.

5 Conclusions

Large diameter metal-on-metal bearings tested under torsional rotation were found to result in joint torques of between [redacted] Nm. These values are an order of magnitude lower than failure torques measured in *in vitro* tests for a series of uncemented cups. These values are also lower than moments measured for simulated gait (roughly factor 2) and, since roll-out (failure) moments for this loading mode are less than those for torsion, failure would be less likely in torsion.

Because the torsion test does not appear to correspond to physiological conditions within the human hip (Figure 14) it is suggested that measurements in the hip simulator would be preferable. The only consistency between the torsion and simulator loading modes was that large diameter metal-metal bearings still have higher moment transfer than smaller diameter conventional bearings (metal-polyethylene), with minimum moments achieved using the ceramic head in a metal cup. The higher moments in the metal-metal bearings are not considered to be dangerous, but could be related to higher wear. Such factors can be offset by other advantages, such as increased stability and maintenance of bone stock.

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