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A COMPARISON OF FRICTION IN 28MM CONVENTIONAL AND 55 MM RESURFACING METAL-ON-METAL HIP REPLACEMENTS

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ABSTRACT

Total hip replacement has been a common, successful surgical intervention for many years. However, it has long been considered unsatisfactory for younger, more active patients due to the limited survivorship of conventional implants employing polyethylene. Larger resurfacing implants were developed to preserve bone stock and improve stability, however, early metal-on-polyethylene implants failed due to high wear. More recent developments, utilising a metal-on-metal bearing, have encouraging short- to medium term clinical performance. Concerns exist regarding the increased sliding distance and frictional torque generated within a larger diameter bearing. A large diameter metal-on-metal surface replacement was contrasted with a conventional 28mm diameter implant with the same bearing material combination using a pendulum friction simulator. Studies were performed under different swing-phase load and lubrication conditions. The larger diameter bearing exhibited the lower friction factor under all test conditions, although the measured frictional torque was higher. Increasing swing phase load was shown to cause an increase in friction factor in all tests. Increased serum concentration resulted in a reduction in friction factor for both bearings. Variation of the friction factor with the head diameter suggested that fluid film lubrication contributed to the reduction of friction.

KEYWORDS

Friction, Total Hip Replacement, Lubrication, Surface Replacement, Metal-on-metal

1. Introduction

Metal-on-ultra-high molecular weight polyethylene total hip replacements continue to be the most clinically successful long-term bearing, with a survivorship of over 75% at 20 years post-implantation in bearings where the cup and stem were cemented into the acetabulum and femur respectively [1]. However, loosening of the prosthesis, and ultimate failure of the implant, caused by the osteolytic response to polyethylene wear debris means alternative bearings continue to be explored [2, 3]. Metal-on-metal bearings have a long clinical history, with Phillip Wiles introducing a stainless steel bearing in 1938 [4]. The McKee-Farrar cobalt-chrome alloy metal-on-metal total hip replacement yielded variable early results, with premature loosening, believed to be due to high frictional torque, however, some successful implants had long lifetimes (in excess of 20 years) with very low wear rates [5, 6]. Recent improvements in manufacturing techniques, yielding superior surface finishes and accurate forms, has led to a resurgence in the use of metal-on-metal bearings. In-vitro studies have demonstrated the low-wearing behaviour of the metal-on-metal bearing [7, 8], and clinical studies have also demonstrated favourable results [9]. The frictional torques generated within the new generation bearings are less than those required to cause mechanical loosening [10, 11].

Although total hip replacements are considered clinically successful, it has long been considered that this procedure is not satisfactory for younger patients, due to the limited survivorship of the conventional replacement. Sir John Charnley noted his metal on polyethylene prosthesis was producing good results within the elderly, inactive population, but he cautioned against the use of the prosthesis in the younger patient, primarily due to their higher activity levels[12-14]. Additionally, smaller diameter prostheses resulted in reduced stability and increased dislocation rates with respect to the normal hip joint [15, 16]. Hip resurfacing prostheses replace only the damaged articulating surfaces and retain much of the bone within the joint. The first resurfacing prostheses employed Teflon as a bearing material, which failed due to osteolysis. Subsequent prostheses used the popular metal-on-polyethylene bearing of the conventional implant, but failure rates were reported to be as high as 34% failure after just 2 ½ years [17-24] These poor performances led to the abandonment of the resurfacing concept; at the time, it was considered to be a fault with the design, rather than the material coupling, that was causing the failures. It has since been widely accepted that the likely cause of failure of these prostheses was the material selection and not geometric design. The large diameter prostheses would give rise to large sliding distances, and subsequently high wear of the polyethylene component, resulting in polyethylene wear induced osteolysis and loosening. Renewed interest in the resurfacing concept arose in the early 1990s, with McMinn and Wagner pioneering the first metal-on-metal resurfacing prostheses[18, 24]. The initial short and mid-term results have appeared positive, with a high short-term survival rates, and rapid rehabilitation of the patients, with many returning to high levels of activity[24, 25]. Benefits of the resurfacing replacement, in addition to the bone conservation, include more natural loading of the femur [26]. Studies have shown that bone is often resorbed around the femoral component of a conventional replacement due to the stress shielding effect, recent studies exploring the behaviour within the resurfacing replacement have shown a more natural transfer of load, with another study suggesting bone mineral density is likely to be preserved within the femoral head post implant [27].

Early studies, examining the tribology of a metal-on-metal total hip replacement using a free pendulum machine, indicated that the bearings operate within a mixed lubricating regime [28]. Further friction studies examined the influence of bearing material and lubricant, and demonstrated that metal-on-metal bearings exhibited higher friction during testing than other material combinations, such as metal-on-polyethylene and ceramic-on-ceramic[29, 30].

Additionally, it was found that friction in metal-on-metal bearings would decrease with increasing protein concentration in the lubricating fluid.

Direct measurement of the contact between the head and cup, via an electrical resistance method, has also demonstrated that metal-on-metal bearings operate within the mixed lubricating regime. Complete separation of asperities was possible during the swing phase of a wear simulator cycle, but asperity contact always occurred during the stance phase of the cycle [31]. Theoretical analysis has also indicated that the mixed regime is the dominant lubricating mode in metal-on-metal bearings [32]. Application of the Hamrock-Dowson equation appears to indicate that increasing the diameter of the bearing should result in an increased film thickness, and hence improved lubrication and reduced friction [33]. However, an increase in bearing diameter would also increase the sliding distance, making the role of lubrication critical in the wear performance of the bearing.

A study by Wimmer et al, investigating the stick-slip phenomenon in metal-on-metal bearings, assumed that the friction factor was the same for different diameter metal-on-metal bearings and used this simple relationship to extract the torque generated for large diameter metal-on-metal bearings [10]. This study did not consider the influence of bearing size upon the lubrication regime. It was proposed that moments of up to 15Nm could be generated, and concluded that though this was insufficient to cause immediate loosening, it may contribute to progressive loosening, due to bone-implant interface failure. However, there have been no experimental studies comparing friction in standard and large diameter metal-on-metal bearings.

Joint laxity may vary following hip replacement surgery, and this may influence the load occurring during the swing phase of the walking cycle. A previous study has indicated, through in-vitro wear testing and theoretical lubrication modelling, that swing phase load has a significant effect of the wear of metal-on-metal bearings, possibly due to the depletion of lubricating film thickness[34].

This study aimed to compare the friction of a clinically available conventional metal-on-metal total hip replacement with a large diameter surface replacement under different swing-phase load and lubricant conditions.

2. Materials and Methodology

2.1 Materials

Four samples of two different bearing combinations, a 28mm-diameter total hip replacement, and a 55-mm diameter surface replacement, supplied by DePuy International Ltd (Leeds, UK) were tested. Both bearings are currently in clinical use. Geometric and surface parameter data were measured prior to test using a co-ordinate measuring machine (Kemco, UK) and two dimensional contacting profilometry (Form Talysurf Series, Taylor-Hobson, Leicester, UK) respectively. The mean measurements for each bearing are shown in Table 1.

2.2 Friction Simulator Methodology

Friction testing was performed using a friction simulator (Simulator Solutions, Manchester, UK); a single-station servo-hydraulic machine, controlled by a personal computer, which applies dynamic loading cycles, similar to that experienced by implants in-vivo, (Figure 1). A fixed frame mounted on two pressurised hydrostatic bearings formed the friction carriage. This allowed the friction of the machine to be considered negligible with respect to the friction within the bearing, as it was over 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, and the cup was not angled. 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 ±25° (±1°), at a frequency of 1Hz. A simple sinusoidal waveform was used through 60% of the cycle to apply a dynamic load with a peak of 2 ±0.1 kN, and different swing phase loads of 25N, 100N and 300N. The ISO standard for hip simulator testing (ISO 14242-1) recommends a swing phase load of 280N. Previous work [35] demonstrated a decrease in wear rate and friction with decreasing swing phase load. The further load of 25N was selected to represent a "high level of joint laxity" in vivo. This was at the limits of the friction simulator and was used to consider the "most favourable" lubrication regime. An example loading-motion profile is shown in Figure 2. Lubricants used were 25% (w/v) bovine serum (with 0.1% sodium azide) and 100% bovine calf serum (Harlan Sera-Lab, Loughborough, UK) . Protein concentration was 15.51g/l in the 25% bovine serum, and 62.04g/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. 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:

$$T_t = \frac{(T_f - T_r)}{2}, \qquad (1)$$

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:

$$f = \frac{T_t}{RL_p}, \tag{2}$$

The friction factor defined is similar in magnitude to the co-efficient of friction, but variable with the finite contact area, (R) is the bearing radius (m), and (L_p) is the peak load (N). Tests were run for a minimum of 180 cycles, to allow analysis of friction once steady-state had been achieved, and data were logged at 30 cycle intervals, for 5 cycles. Frictional torque was measured over ten points around the peak load, high velocity phase of the cycle. Data was analysed after 120 cycles, when the friction had stabilised.

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 ceramic-on-ceramic bearing tested in a thick silicone oil (viscosity =5000cs). 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.

28mm diameter metal-on-UHMWPE (MoP) and ceramic-on-ceramic (CoC) bearings were used as control samples to assess the repeatability of the friction measured. The results achieved, shown in Table 2, indicate 95% confidence limits of 0.004 and 0.005 respectively for MoP and CoC.

Prediction of lubricant film thickness and friction under transient loading conditions is complex [36]. However a simple estimation of the lubricant film thickness can be obtained using average load and speed for two bearings using the Hamrock and Dowson formula [33] shown below;

$$h_{min} = 2.798R\left(\frac{\eta u}{E'R}\right)^{0.65}\left(\frac{L}{E'R^2}\right)^{-0.21} \tag{3}$$

where h_{min} is the minimum film thickness, η 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, R, was calculated from:

$$\frac{1}{R} = \frac{1}{R_1} - \frac{1}{R_2} \qquad (4)$$

where R_1 and R_2 are the radii of the femoral head and acetabular cup respectively.

The predicted minimum film thickness was used to predict the lambda ratio, as shown below;

$$\lambda = \frac{h_{min}}{\left[(Ra_h)^2 + (Ra_c)^2\right]^{1/2}} \qquad (5)$$

where Ra_h and Ra_c are the measured surface roughness values for the head and cup respectively. It has been shown that a lambda ratio of greater than 3, indicates fluid film lubrication, whereas a value below 1, where the predicted minimum film thickness is less than the surface roughness indicates that boundary lubrication prevails. A lambda ratio between 1 and 3 indicates that the bearing is operating in a mixed lubrication regime.

3. Results

A comparative study examining the influence of bearing size, swing phase load and lubricant was performed. The bearing size appeared to have an effect upon friction factor. Figure 3 demonstrates the friction factor for the bearing combinations, tested with a swing phase load of 100N. The mean friction factor for the surface replacements in 25% serum was 0.098 (±0.02), with a mean friction factor of 0.121 (±0.02) for the 28mm MoM bearings. While the surface replacements exhibited lower friction factors than the 28mm bearings, this was not found to be statistically significant (ANOVA $p>0.05$). Furthermore, the peak frictional torque measured during testing was found to be greater for the larger diameter bearings (Table 2).

Protein concentration was shown to have a marked effect upon the friction factor. Figure 3 shows the friction factors calculated during tests performed in both 25% and 100% serum under 100N swing phase load conditions. Both metal-on-metal bearings exhibited a reduction in friction factor with increasing protein concentration, however, this was only found to be statistically significant in the 28mm bearing (ANOVA, $p<0.05$). A reduction in friction factor from 0.121 (±0.02) in 25% serum to 0.095 (±0.011) in 100% serum in the 28mm bearings, and from 0.098 (±0.02) to 0.079 (±0.011) in the surface replacements was observed.

The results shown in Figure 4 demonstrate the effect of swing phase load, and were achieved from tests performed in 25% serum. It may be seen that the increase in swing phase load results in an increase in friction factor, and the trends are similar for both bearings. The increase was not found to be statistically significant (ANOVA $p>0.05$).

Theoretical analysis, shown in Table 3, indicated that the conventional 28mm total hip replacement had a lambda ratio of less than one, suggesting a boundary lubricating regime. The larger diameter resurfacing bearing exhibited a higher lambda ratio, suggesting improved lubrication. Comparison of these predictions with the experimental data shows the larger diameter bearing to have reduced friction factor with respect to the conventional total hip replacement, also indicating an improved lubrication regime.

4. Discussion

This study has examined the influence of bearing size, lubricant and swing phase load upon the friction of hip bearings. Although friction no longer appears to be a significant contributory factor in the failure of hip replacements, studies of friction are still useful. A low friction factor indicates improved lubrication, which may reduce the wear of the prosthesis. Friction studies are a useful analytical tool to allow evaluation of the tribology of the hip replacement.

The present study employed a uni-directional motion, and a simplified loading profile whilst testing the head-cup arrangement in an inverted position, with respect to in-vivo, with no cup inclination. The loading and motion conditions applied within this study were a simplified model of a standard gait cycle, to allow comparison of the factors influencing the friction of the bearing. Additionally, new, unworn samples were tested for short periods, hence the influence of wear, and the related geometric and surface changes, were not considered within this study.

This study examined the influence of bearing size, contrasting two metal-on-metal bearings of 28mm and 55mm nominal diameter. It was shown that the larger diameter bearing generated a lower friction factor than the 28mm bearing under all test conditions, although this was not significant. This differs from Wimmer et al [10] who assumed a similar friction factor for metal-on-metal bearings with different diameters. Such a discrepancy can be readily explained from the fluid film lubrication contribution. According to equation (4), an increase in the bearing diameter leads to an increase in the equivalent bearing radius, and this in turn should lead to an increase in the wear factor under boundary lubrication conditions.

Therefore one of the potential factors to reduce the friction factor as the head diameter increased from 28mm to 50mm would be the fluid-film lubrication contribution. This is consistent with the fluid film lubrication analysis based on the Hamrock and Dowson formulae. However, it was also noted that the frictional torque generated within the bearings was higher for the large diameter implant under all loading and lubricant conditions. However, the highest frictional torque, measured in the surface replacements in 25% serum at a swing phase load of 300N was approximately 6Nm and was lower than the minimum frictional torque expected to cause immediate loosening [11].

The influence of head diameter on tribology shown within this study appears to correlate well with previous friction studies, and the outcomes of in-vitro wear simulator studies. O'Connor et al demonstrated femoral head size had an influence on friction using a pendulum comparator [37]. Simulator studies have demonstrated the importance of lubricating regime with increased head diameters. In small diameter bearings, where the prevalent lubricating regime is boundary, an increase in head size, from 16mm diameter to 22mm diameter resulted in an increase in volumetric wear, as a result of the increased sliding distance [38]. When increasing the sample diameter above 28mm, an increased fluid film contribution to the mixed lubricating regime resulted in reduced volumetric wear [39] .

The friction studies were performed in 25% and 100% bovine serum. The friction factor was seen to reduce with increasing serum concentration for both bearings. Past studies examining the behaviour of metal-on-metal implants have identified the presence of tribo-chemical films on the metallic surfaces of explants and after in-vitro wear testing [40]. Although it is unlikely that a tribo-chemical film would be generated within a short term friction study, others report the presence of proteins upon the surface of the metal bearings after low-cycle tests [41]. Electrical resistance measurements during a wear simulator study also indicated that protein-rich lubricants provided better lubricating conditions than those without proteins

[42]. It is proposed that these proteins adhere to the metallic surfaces acting as a solid phase lubricant at the bearing interface, reducing the adhesive forces between the metal surfaces and thus reducing friction. However, the precise mechanism of this boundary action has yet to be established.

Swing phase loads of 25N, 100N and 300N were tested within this study. A previous study has predicted a reduced film thickness with increasing swing phase load through theoretical analysis [35], and demonstrated increased friction and wear within metal-on-metal bearings at increased swing phase loads. The current study showed increased friction with increasing swing phase loads for both bearings. It is proposed that as the swing phase load is increased, there is less efficient entrainment of fluid into the contact region, hence the film thickness becomes depleted. This would increase asperity contact and hence generate higher friction, as observed within this study.

5. Conclusions

Large diameter MOM surface replacements are being used increasingly to address the needs of younger and more active patients. This study compared the friction of a large diameter metal bearing with a conventional metal-on-metal total hip replacement, examining the influence of lubricant and swing phase load. The large diameter bearing had a lower friction factor than the 28mm bearing under all test conditions. The peak frictional torque occurred within the large diameter bearing, although it was lower than that required to cause frictional loosening. However, increasing the bearing diameter will increase sliding distance, thus lubrication is an important factor governing wear performance. An increase in serum concentration reduced the friction in both bearings, and therefore appeared to improve the lubrication. An increase in swing phase load was shown to increase friction factor.

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List of Tables

Table 1 Sample Data

	No of Samples	Nominal Diameter/mm	Mean Radial Clearance/mm	MeanRa/µm
Surface Replacement	**4**	**55**	**0.046 ±0.001**	**0.004±0.001**
Total Hip Replacement	**4**	**28**	**0.029±0.002**	**0.010±0.002**

Table 2 Frictional Torque during testing in 25% serum

	Frictional Torque/Nm	
Swing Phase Load/N	**Surface Replacement**	**Total Hip Replacement**
25	4.71	2.98
100	5.33	3.38
300	5.81	3.81

Table 3 Theoretical Minimum Film Prediction

Bearing	**Predicted h_{min} (µm)**	**λ ratio**
Surface Replacement	**0.17**	**>3**
Total Hip Replacement	**0.07**	**<1**

List of Illustrations

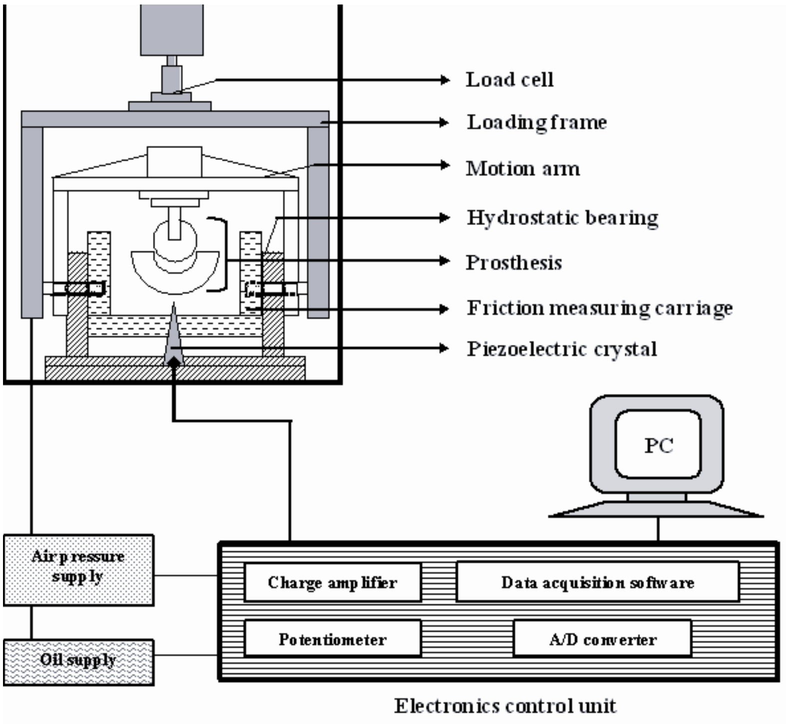

Figure 1: Schematic of the Friction Simulator

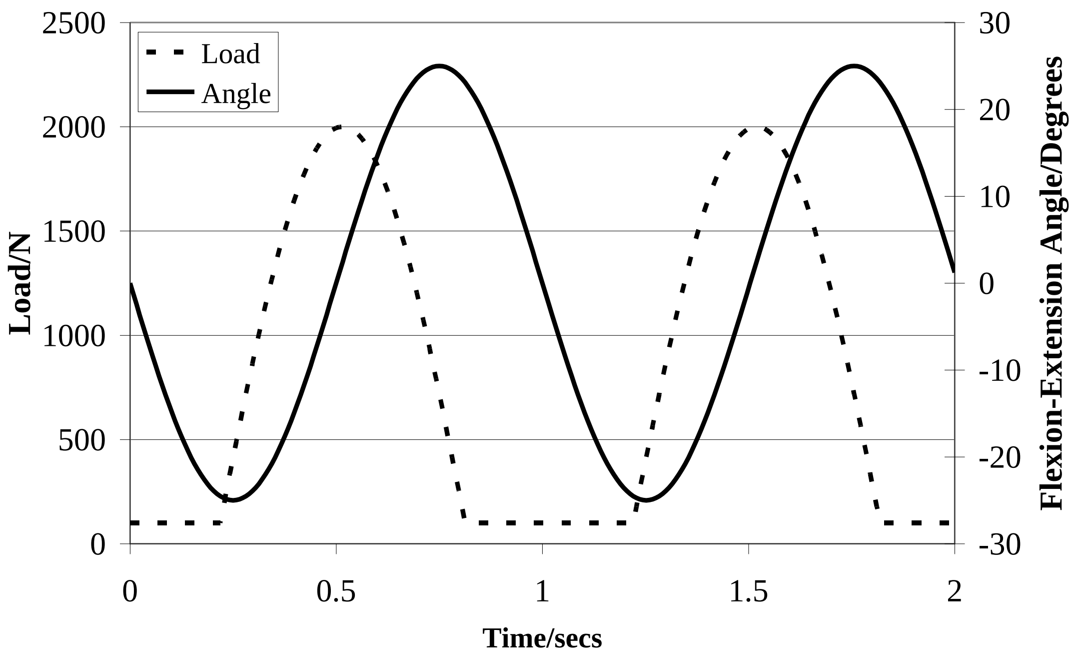

Figure 2: Loading and Motion Test Profile

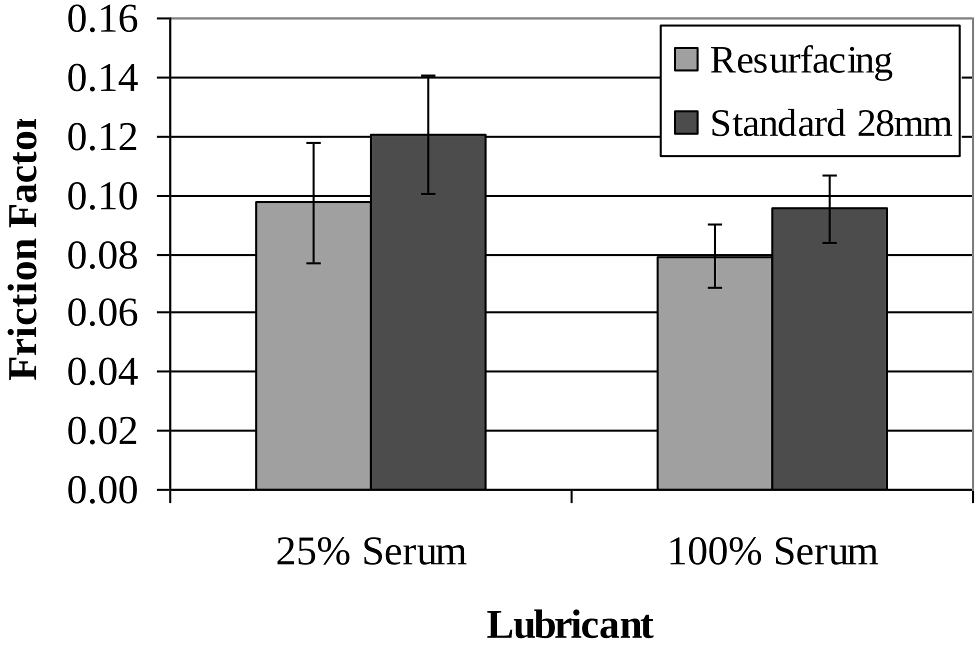

Figure 3: Effect of Serum Concentration upon Friction Factor

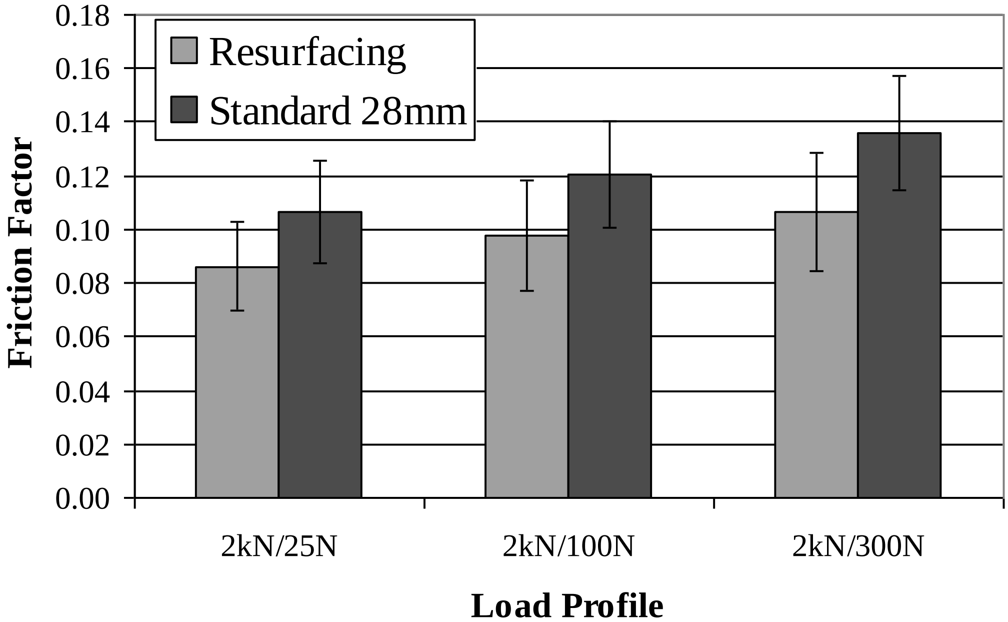

Figure 4: Effect of Swing Phase Load upon Friction Factor

Appendix 1: List of Notation

E'	Equivalent Youngs modulus
f	Friction factor
h_{min}	Minimum film thickness
L	Applied Load
L_p	Peak load
λ	Lambda ratio
η	Lubricant viscosity
R	Equivalent radius
R_1	Femoral head radius
R_2	Acetabular cup radius
Ra_c	Surface roughness of acetabular cup
Ra_h	Surface roughness of femoral head
T_f	Frictional torque in forward direction
T_r	Frictional torque in reverse direction
T_t	True frictional torque
u	Entraining velocity