THE EFFECT OF CLEARANCE UPON FRICTION AND LUBRICATION OF LARGE DIAMETER HIP RESURFACING PROSTHESES USING BLOOD AND COMBINATIONS OF BOVINE SERUM WITH AQUEOUS SOLUTIONS OF CMC AND HYALURONIC ACID AS LUBRICANTS

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THE EFFECT OF CLEARANCE UPON FRICTION AND LUBRICATION OF LARGE DIAMETER HIP RESURFACING PROSTHESIS USING BLOOD AND COMBINATIONS OF BOVINE SERUM WITH AQUEOUS SOLUTIONS OF CMC AND HYALURONIC ACID AS LUBRICANTS

Friction and lubrication behaviour of 50mm diameter Birmingham Hip Resurfacing implants with diametral clearances 80 to 300µm, using blood (clotted and whole blood), a combination of bovine serum with hyaluronic acid and carboxymethyl cellulose (CMC) adjusted to a range of viscosities (0.001-0.2 Pas), and bovine serum with CMC adjusted to a similar range of viscosities, have been investigated.

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STATMENT OF ORIGINALITY

To the best of my knowledge, the material or the contents presented in this thesis are original except where otherwise noted within the text. None of this research has been submitted in whole or in part for any degree at this or any other University.

Saeed Afshinjavid

Abstract

In real life, immediately after joint replacement, the artificial joint is actually bathed in blood (and clotted blood) instead of synovial fluid. Blood contains large molecules and cells of size ~ 5 to 20 μm suspended in plasma and considered to be a non-Newtonian (pseudoplastic) fluid with density of 1060 Kg/m^3 and viscosity ~ 0.01 Pas at shear rates of 3000 s^{-1} (as obtained in this work). The effect of these properties on friction and lubrication is not fully understood and, so far to our knowledge, hardly any studies have been carried out regarding friction of metal-on-metal bearings with various clearances in the presence of lubricants such as blood or a fluid containing macromolecules such as hyaluronic acid (HA) which is a major component of synovial fluid increasing its viscosity and lubricating properties. In this work, therefore, we have investigated the frictional behaviour of a group of Smith and Nephew Birmingham Hip Resurfacing implants with a nominal diameter of 50mm and diametral clearances in the range ~ 80 μm to 300 μm, in the presence of blood (clotted and whole blood), a combination of bovine serum (BS) with hyaluronic acid (HA) and carboxymethyl cellulose (CMC, as gelling agent) adjusted to a range of viscosities (~0.001-0.2 Pas), and bovine serum with CMC adjusted to a similar range of viscosities.

These results suggested that reduced clearance bearings have the potential to generate high friction especially in the presence of blood which is indeed the in vivo lubricant in the early weeks after implantation. Friction factors in higher clearance bearings were found to be lower than those of the lower clearance bearings using blood as the lubricant. Similar trends, i.e. increase in friction factor with reduction in diametral clearance, were found to be also the case using a combination of BS+CMC or BS+HA+CMC as lubricants having viscosities in the range 0.1-0.2 and 0.03-0.14 Pas, respectively. On the other hand, all the lubricants with lower viscosities in the range 0.001-0.0013 and 0.001-0.013 Pas for both BS+CMC and

BS+HA+CMC, respectively, showed the opposite effect, i.e. caused an increase in friction factor with increase in diametral clearance.

Another six large diameter (50mm nominal) BHR deflected prostheses with various clearances (~ 50-280μm after cup deflection) were friction tested in vitro in the presence of blood and clotted blood to study the effect of cup deflection on friction. It was found that the biological lubricants caused higher friction factors at the lower diametral clearances for blood and clotted blood as clearance decreased from 280μm to 50μm (after deflection).

The result of this investigation has suggested strongly that the optimum clearance for the 50 mm diameter MOM BHR implants to be ≥150μm and <235μm when blood lubricant used, so as to avoid high frictions (i.e. avoid friction factors >0.2) and be able to accommodate a mixed lubrication mode and hence lower the risk of micro- or even macro-motion specially immediately after hip implantation. These suggested optimum clearances will also allow for low friction (i.e. friction factors of <0.2-0.07) and reasonable lubrication (dominantly mixed regime) for the likely cup deflection occurring as a result of press-fit fixation.

Keywords: Friction; Lubrication; 50mm diameter metal-on-metal Birmingham hip resurfacing (BHR) prosthesis; Diametral Clearances (80-300μm); Blood; Clotted Blood; Bovine Serum (BS); BS+CMC; BS+HA+CMC

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Nomenclature

ω	The angular velocity of the femoral head representing flexion and extension
μ	Coefficient of friction
μm	micrometer
a	contact surfaces
BS	bovine serum
C_d	The diametral clearance between the head and the cup (
CMC	carboxy methylcellulose
Co–Cr–Mo	cobalt-chromium- molybdenum
D	head diameter
d	Femoral head diameter
E'	Equivalent Young's Modulus of the two materials
E'	The equivalent Young's Modulus
E_c	Young's Modulus of the acetabular cup material
E_f	Young's Modulus of the femoral stem material
f	friction factor
F	Frictional force
GPa	Giga Pascal
h_{min}	The lubricating film thickness
K	Wear factor, (mm^3/Nm)
K_1	Wear coefficient
kN	kilo Newton
m	Meter
mm	Millimetre
MoM	Metal on Metal
N	Newton
Nm	Newton meter
P	The normal load (N)
P/t	penetration rate
Pa	Pascal
R	reduced radius
R	The composite surface roughness

$\mathbf{r}$	Radius
$\boldsymbol{R_1}$	femoral head radius
$\mathbf{R_1}$	The surface roughness of the femoral component
$\boldsymbol{R_2}$	outside radius of the acetabular cup
R_2	inner radius of the acetabular cup
$\mathbf{R_2}$	The surface roughness of the acetabular component
R_3	outer radius of the acetabular cup
$\mathbf{R_a}$	The composite surface roughness
$\mathbf{R_{a1}}$	The surface roughness of the femoral component
$\mathbf{R_{a2}}$	The surface roughness of the acetabular component
rad	radian
R_l	radius of the femoral head
s	time
T	frictional torque
$\mathbf{u}$	The entraining velocity, $(u_l+u_2/2$
UHMWPE	ultra high molecular weight polyethylene
$\boldsymbol{V}$	volume of the material removed from the pin
$\mathbf{V}$	Sliding speed
$\boldsymbol{V_2}$	the initial '*running-in*' wear
$\mathbf{W}$	The load at the hip
$\mathbf{W}$	Applied load
$\boldsymbol{w_a}$	asperity contacts
$\boldsymbol{X}$	The sliding distance (m)
$\boldsymbol{Z}$	Sommerfeld number
η	viscosity of the lubricant
$\boldsymbol{\lambda}$	Lambda, ratio for lubricating mode
ν_c	Poisson's ratio of the acetabular cup material
ν_f	Poisson's ratio of the femoral stem material
$\boldsymbol{\upsilon}$	Poisson's ratio
φ	spherical coordinates

INTRODUCTION TO THE PRESENT WORK

The major objectives in the design of joint prostheses are the development of stable articulations, low friction and wear, solid fixation into the bone, and normal range of motion. However, the demands presented by highly active patients with longer life expectancy have challenged the orthopaedic companies to improve both design and materials of joint implants. The current trend in total hip arthroplasty (THA) is, therefore, moving away from a conventional metal-on-polyethylene THA to metal-on-metal (MOM) hip resurfacing for the treatment of younger and more active patients with arthritis and other advanced hip diseases [Amstutz *et al,* 1996]. This is mainly due to a remarkable improvement in the prosthetic design of metal-on-metal resurfacing devices including improved sphericity and excellent tolerances, use of large sized joints (>35-60mm diameter head and cup) to lower risk of dislocation, good lubrication between the articulating surfaces, and high carbon and carbide content to reduce wear. To improve the stability and osteointegration, [McMinn *et al,* 1996, McMinn 2009] some prostheses have been modified to a cemented femoral component and with a hydroxyapatite coated cup. A survival rate of 99.8 % at four years along with short rehabilitation periods allowing patients to return to their preoperative levels of activity has been reported for the McMinn BHR implants Conventional THAs have shown to encourage stress shielding around the femoral head causing bone resorption (migration) and consequently loosening or failure of the implant. On the other hand, hip resurfacing improves the load distribution *in vivo*, resulting in more natural loading of the femur [Ebied *et al,* 2002, McMinn 2009] and thus reducing bone resorption. Other researchers have also demonstrated preservation of the bone mineral density postoperatively within the femoral head [Thompson *et al,* 2000b]. However, it has been shown [Watanabe *et al,* 2000] that there is a potential for stress concentration around the base of the femoral component in hip resurfacing devices which could consequently result in femoral neck fracture. Also, hip replacement studies

[McMinn 2009, Itayem *et al,* 2005] on postoperative implant migration have postulated that the stability and therefore long-term success of the implants may be predicted from the levels of migration within the first two years after implantation. Radiostereophotogrammetric Analysis (RSA) has therefore been used to measure the migration of the prosthesis with respect to the bone and its stability *in vivo*. RSA studies carried out on McMinn BHR prostheses have shown negligible migration of the implant [Itayem *et al,* 2005] suggesting long-term stability *in vivo* for the hip resurfacing prosthesis. Another important feature of the hip resurfacing devices is the improved bony ingrowth or ongrowth due to their roughened backing via hydroxyapatite coating, resulting in improved rotational stability of the prosthesis. Also, use of mechanical fixations such as fins has been suggested and implemented for improving the initial rotational stability of the prosthesis in the early weeks and months after implantation [Thompson *et al.* 2000a].

Notably, it is well established that metal-on-metal bearings produce far less wear (or ~ 50 times less wear particles) than conventional metal on polyethylene bearings and offer the prospect of lower failure rates and that the conventional 28mm diameter MOM THRs have shown higher wear rates than the MOM resurfacing prostheses [http://www.totaljoints.info/metal_on_metal_total_hips.htm, Smith et al. 2001a, Smith et al. 2001b, Smith et al. 2001c, Liu et al. 2006, Dowson et al. 2004, Rieker et al. 2005]. Tribology theories and hip joint simulator studies have also predicted that friction, lubrication and wear within these bearing systems are affected by several factors including load applied, material hardness, surface roughness, bearing diameter, sliding speed, radial clearance and the viscosity of the lubricant [Liu et al. 2006, Dowson et al. 2004, Rieker et al. 2005, Udofia et al. 2003, Liu et al. 2005]. It is certain that *in vitro* studies will continue to determine the optimum clearance for a given bearing diameter (as in this work being one of the main objectives) and that lubrication plays an important role in maintaining a low friction (and wear) performance for MOM bearings.

However, some hip/knee friction studies have employed bovine serum only as the lubricant with added carboxymethyl cellulose (CMC), since this combination simulates the viscosity and other characteristics of the *in vivo* lubricant, i.e. synovial fluid [Scholes et al. 2000], but this combination does not contain hyaluronic acid (HA) which is a lubricating substance with good shock absorption properties present as a major component in cartilage and the synovial fluid in joints and also distributed widely throughout connective, epithelial (skin), and neural tissues where it has a protective, structure stabilizing and also shock-absorbing role [Brown et al. 2005, http://www.raysahelian.com/hyaluronic-acid.html]. To be noted also is that hyaluronic acid is a naturally occurring polyanionic, polysaccharide substance that consists of N-acetyl-d-glucosamine and beta-glucoronic acid [Brown et al. 2005]. The unique viscoelastic nature of hyaluronic acid along with its biocompatibility and non-immunogenicity has led to its use in a number of clinical applications including the supplementation of joint fluid in arthritis, as a surgical aid in eye surgery, and to facilitate the healing and regeneration of surgical wounds [Brown et al. 2005, http://www.raysahelian.com/hyaluronic-acid.html]. Related to this work, is the fact that hyaluronic acid has large molecules of glucosaminoglycan (or special mucopolysacharide) and a high but variable molecular weight (in the range 10^4 - 10^7 Da) and viscosity and hence these molecules are likely to get entrained into the bearing and the shearing forces generated are expected to raise the friction factor causing increased bearing friction. This is especially true in lower clearance bearings. It is therefore important to add HA to the serum in order to investigate its effect on friction and lubrication behaviour for any implant and this was another main aims of this present research programme.

Most modern cementless joints depend on a press fit primary fixation which stabilises the component in the early weeks. This allows bony ingrowth and ongrowth to occur which in turn provides durable long term fixation. Increased bearing friction in the early weeks and months after implantation can lead to micromotion and has the potential to prevent effective bony ingrowth from occurring. Therefore, friction in the early postoperative period can be critical to the long-term success of the fixation. This has been one of the concerns raised in a recent clinicoradiological study of metal-metal bearings with reduced and closely controlled clearance [McMinn et al. 2006]. A progressive radiolucent line at the periphery of the socket component was evidenced in a few of these cases at follow-up, as shown in Figure 1 [Itayem et al. 2005] and raised the possibility that increased friction is affecting component fixation.

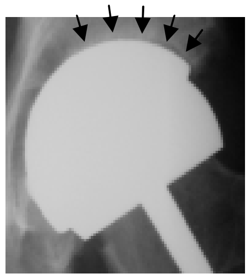


Figure 1. A 1-year radiograph of a patient with a low clearance (86μm) Birmingham Hip Resurfacing device, showing a progressive radiolucent line around zones 1 and 2 of the acetabular component, suggesting increased friction and micromotion resulting in poor fixation [Itayem et al. 2005, McMinn et al. 2006].

It follows therefore that one of the main reasons for this increased friction is that of low clearance and that as mentioned earlier, immediately after joint implantation, the artificial joint is actually bathed in blood for couple of weeks or even months and not in synovial fluid [McMinn et al. 2006]. Blood contains large molecules and cells of size ~ 5 to 20 μm and the effect of these on friction and lubrication is not yet fully understood. So far, we have been unable to find any study on friction of metal-on-metal bearings with varying clearances in the presence of blood or a fluid containing macromolecules (e.g. both HA and blood) as lubricants. It became one of the most important objectives in this work to use blood and serum with added HA as lubricants so as to investigate their effects on friction of large diameter hip resurfacing implants with various clearances.

Historically, there have been very few incidences of mechanical failures with metal-on-metal total hip replacements causing dislocation. While the optimal clearance to achieve elastohydrodynamic lubrication and avoid equatorial seizing is still being studied and debated, tribologists recommend that the diametral clearance be as small as possible in large-diameter bearings [Dowson et al. 2004, Rieker et al. 2005]. This requirement must be balanced against practical limitations of manufacturing tolerances and also must take into account the possibility that deformation of the acetabular cup may occur when it is implanted into the acetabulum with a press-fit of 1 to 2 mm. Initial stability can be influenced by the method of fixation (press-fit), the surgical technique, the quantity and quality of the bone structure, bearing geometry and applied loading conditions. Press-fit fixation involves inserting an acetabular cup into an under reamed acetabulum, where the primary stability is gained through the frictional compressive forces generated about the acetabular periphery. The press-fit procedures have moderate influence on the contact mechanics at the bearing surfaces, but produce remarkable deformation of the acetabular cup. Further deformation of the acetabular cup, and subsequent reduction of the effective clearance, may also occur with physiological loading. The effect of cup deflection on clearance has been studied experimentally in cadaver pelvis and with the use of finite-element modelling [Jin et al. 2006]. The wall thickness of the cup showed to be the most important factor influencing deformation of the acetabular cup in both studies, although diametral clearance and bearing diameter were also important. Therefore, another aim in this work was to study the effect of cup deflection (deformation) on friction, particularly in the presence of blood and clotted blood which are the main lubricants immediately after implantation. Design and manufacturing parameters such as diametral clearance, femoral head/cup diameter, surface finish or roughness, have therefore, shown to significantly influence the contact mechanics and tribology at the bearing surfaces of hip resurfacing arthroplasty. Hence, orthopaedic manufacturers must ensure that deformation of the component does not adversely affect clearance since this would lead to increased friction and hence joint dislocation [Muller et al. 2003]. It is, therefore, postulated that if the cup is deflected by press fitting, this may result in increased contact at bearing surfaces around the equatorial rib of the cup and result in higher frictional torque which can increase the risk of dislocation and hamper fixation. This has been the case for some early loosening of the implants after few weeks of implantation [McMinn 2009].

AIMS AND OBJECTIVES OF THE PRESENT WORK

The aims of this work were, therefore, to investigate the frictional and lubrication behaviour of a group of S&N Birmingham Hip Resurfacing (BHR) implants with a nominal diameter of 50mm and a range of different clearances ranging around 80µm to 306µm. The testing was carried out in the presence of the following lubricants using a friction hip simulator to obtain frictional torques and then friction factors: i) Blood (clotted and whole blood); ii) A combination of bovine serum, hyaluronic acid and carboxymethyl cellulose (CMC) adjusted to a range of physiological viscosities; and iii) Bovine serum with CMC adjusted to a similar range of viscosities. Stribeck analyses were then carried out by plotting friction factor versus Sommerfeld number for each clearance using the above lubricants with different viscosities. The Stribeck curves allowed comparison amongst different clearances and lubricants which in turn made it possible to obtain and suggest the optimum results for the 50 mm diameter BHR implants in terms of clearance and the lubricating mode. Other main objective was to study the generated dynamic motion profiles in terms of frictional torque, friction factor and applied load versus number of cycles for an extension-flexion of ±24º in order to obtain the exact torque applied to each joint during friction tests at various dynamic loadings. This was a very important objective since frictional torques of less than 10 Nm are expected during normal gait and for properly aligned artificial joints and it is known that excessive amount of torque (>50-100Nm) will cause fixation impairments leading to implant loosening requiring revision surgery. Other main objectives were to investigate the rheological characteristics of all the main lubricants including blood and clotted blood which were carried out for the first time in this study as well as that for pure and diluted bovine serum using a typical rheometer.

Since clearance plays a unique role in squeezing lubricants between contact surfaces allowing the formation of a fluid film, deflection of a cup during surgery may result in negative action during articulation. The aim of this part was, therefore, to investigate the effect of cup deflection (initially ~25-30μm and finally up to ~70 μm) on friction of large diameter (50mm) metal on metal Birmingham Hip Resurfacing (BHR) prosthesis with various original clearances (80-306μm) using blood and clotted blood as lubricants. This is another original work carried out in this work for the first time due to the fact that orthopaedic surgeons have become aware of the fact that the implant is soaked in blood for at least couple of month and hence any deviation from an optimum clearance will result in high frictions leading to micro-motion and hence impaired fixation. The use of blood and clotted blood as the only lubricants for the deflected cups will give the necessary friction data in order to assess the effect of cup deformation on friction and lubrication which were other main objectives in this work.

CHAPTER ONE

1.0 Literature Review

1.1 INTRODUCTION TO HIP JOINT PROSTHESES

As a result of recent advancements in biomedical technology, artificial joints can now replace many joints of the body in case of disease or injuries. Joint replacement, called arthroplasty, was first developed in the late 1930's. The goal of arthroplasty surgery is to remove the two damaged and worn parts, for example of the hip or knee joint and replace them with artificial implants to reproduce the form and function of the natural joint, relief pain, restore function and correct deformity.

The major objectives in the design of joint prostheses are the development of stable articulations, low friction and wear, solid fixation into the bone, and normal range of motion. New synthetic replacement materials are being designed [Dowson 2006] by biomedical engineers to accomplish these objectives. Total Hip joint replacements (THR) are usually composed of cobalt chrome alloys in combination with modern plastic [Charnley 1982], Ultra-High Molecular Weight Polyethylene (UHMWPE).

In recent years, the proportions of younger and more active patients undergoing arthroplasty have increased, perhaps due to the increased confidence in the operation techniques. However, due to their higher activity levels compared to the inactive population, the elderly, and the limited survivorship of the conventional replacement [Charnley 1982], there are still concerns about the use of metal on polyethylene prosthesis for the younger generation. Despite the fact that, as Charnley noticed, the metal on polyethylene prosthesis would produce superior results within the elderly with less activity, the use of these prostheses are not favourable for more active patients [Kobayashi et al., 1997; Torchia et al., 1996]. Highly active patients could potentially generate extremely high wear rates, which would in turn result in premature failure of the implant. The demands presented by highly active patients with longer life expectancy have challenged the orthopaedic companies to improve the designs and materials of THRs. However, the most commonly quoted survivorship rates [Charnley 1982] followed by revision for a particular implant or treatment is 10 – 15 years. Hence, it would be advantageous to develop a replacement that could either survive the patients' lifetime, or use a replacement that would conserve bone stock for the eventual revision [Mogensen et al, 1981].

One of the hip replacement procedures in which the head of the femur is retained resulting in minimum bone removal is called hip resurfacing. Instead of removing the head completely, it is shaped to accept an anatomically sized metal sphere. There is no large stem to go down the central part of the femur and the surface of the acetabulum is also replaced with a metal implant, which is wedged directly into the bone.

The modern resurfacing components are made of cobalt chrome, which is finely machined to produce a very high quality surface with a low friction finish, resulting in low wear. Some of the early works on hip resurfacing processes resulted in the Smith-Peterson hip resurfacing device and the Judet prosthesis. Sir Charnley developed this procedure further by using Teflon bearing components. Unfortunately due to aseptic loosening and excessive wear this design failed. Later on, the development of the metal-on-metal resurfacing concept by Muller in 1967 replaced the existing metal on polyethylene material combination.

Various designs were considered during the 1970s; Freeman and Furuya introduced the metal acetabular component [Freeman et al.,1978b] with the polyethylene head (Figure 1.1), while Wagner [Wagner 1978] and Amstutz [Amstutz et al., 1981] used polyethylene cups and metallic heads.

Figure 1.1. Freeman's Resurfacing Replacement [Freeman *et al.,* 1978b].

All these designs faced a high failure level (34%) after a short period of time (two and half years), which led to the abandonment of the resurfacing concept [Bell *et al.,* 1985; Head *et al.,* 1982; Howie *et al.,* 1990a; Tanaka *et al.,* 1978; Wagner and Wagner *et al.,* 1996; Wiadrowski *et al.,* 1991]. Nowadays, it is believed that the most likely cause of failure within these prostheses was the material selection whereas, at the time, the design was considered to be the main issue.

Resorption of periprosthetic bone has shown a major problem in the development of satisfactory in total hip arthroplasty [Kobayashi *et al,* 1997]. Substantially, osteolysis can lead to bone loss around artificial implant and aseptic loosening of the implant [McGee *et al,* 2000]. Osteolysis and aseptic loosening have been noticed in the conventional hip replacement to be caused by the high wear level of the polyethylene component. The large diameter prostheses would give rise to large sliding distances, and subsequently high wear of the polyethylene component [Howie *et al.,* 1990b]. The poor performance of the existing prosthesis and the interest of use of these components for younger/more active patients led to reconsideration of resurfacing in the early 1990s. The first metal-on-metal resurfacing prostheses were established by McMinn and Wagner [McMinn *et al.,* 1996; Wagner *et al.,* 1996]. These prostheses were made from cobalt chrome and were initially cementless. To improve the stability and osteointegration, the McMinn prosthesis has been modified to a cemented femoral component and with a hydroxyapatite (HA) coated cup. McMinn *et al.* (1996) have reported a survival rate of 99.8 % at four years for the McMinn prosthesis and also short rehabilitation periods allowing patients to return to their preoperative levels of activity. Total hip replacements have been shown to encourage stress shielding around the femoral head causing bone resorption and consequently failure of the implanted femoral head. Hip resurfacing improves the load distribution *in vivo*, resulting in more natural loading of the femur [Ebied *et al.,* 2002]. Other researchers have also demonstrated preservation of the bone mineral density postoperatively within the femoral head [Thompson *et al.*, 2000b]. However, [Watanabe *et al.* 2000] have shown in a study that there is a potential for stress concentration around the base of the femoral component in hip resurfacing devices, which could consequently result in femoral neck fracture.

Roentgen Stereophotogrammetric Analysis (RSA) has been used to measure the migration of the prosthesis with respect to the bone and its stability *in vivo.* RSA studies have been carried out on McMinn prosthesis [McMinn *et al.,* 1996] showing negligible migration of the implant. Hip replacement studies on postoperative implant migration have postulated that that the stability and therefore long-term success of the implants may be predicted from the levels of migration within the first two years after implantation. Hence, this suggests long-term stability *in vivo* for the McMinn prosthesis [McMinn *et al.,* 2009]. Another important feature in the hip resurfacing devices is the improved bony ingrowth or ongrowth due to their roughened backing, resulting in improved rotational stability of the prosthesis. Other mechanical fixation, such as fins could also optimise the initial rotational stability of the prosthesis in the early weeks and months after implantation [Thompson *et al.,* 2000a & 2000b].

One of the major complications after total hip replacement is femoral head dislocation. [Philips *et al.* 2003] reported a 3.9% incidence of dislocation in a study of 58,000 Medicare patients within the first six months after primary total hip replacement. Furthermore, the incidence of dislocation increases after revision surgery to 9-12% [Amstutz *et al.*, 2004]. One approach in producing greater joint stability and thereby reducing dislocation is to use a larger head diameter. Traditionally this was done at the expense of increased wear of polyethylene as the larger femoral heads produce increased sliding distances. The primary factor affecting the longevity of total joint replacements is the wear of the components and the resultant wear debris. Wear debris has been shown by many authors to trigger an osteolytic reaction by causing adverse cellular reactions which lead to bone resorption and implant loosening [Amstutz *et al.*, 1992; Harris 1995; Hailey *et al.*, 1996; Green *et al.*, 1998; Ingham *et al.,* 2000; Fisher *et al.*, 2001 and 2004 a; Ingham and Fisher, 2005]. Hence, with the improved wear performance and joint stability of metal on metal hip resurfacing, this approach has become popular. The restoration of normal anatomy has also been noted as a benefit of this replacement method. The size of the replacement is similar to that of the natural hip, resulting in a lower dislocation risk than a standard total hip replacement. Also in hip resurfacing, due to the large size of the head, little adjustment is required to ensure that the lengths are maintained in comparison to the four adjustments required in total hip replacements. However, some researchers have reported this as a disadvantage, as there is little or no potential to correct limb length with the resurfacing replacement [Kilgus *et al.,* 1991; Knecht *et al.,* 2004; Silva *et al.,* 2004; Vale *et al.,* 2002].

Despite the fact that satisfactory short to medium term clinical results of the resurfacing replacements have been reported, long term results will be required to examine other concerns such as the effect of resurfacing replacement upon the outcomes of a revision implant and also the effect of metallic wear debris on the long term survival of the prosthesis. Further studies are required to fully understand the effects of other factors such as clearance, implant design, manufacturing and surgical procedures upon the performance of the prosthesis. It is also important to point out that the effectiveness and suitability of hip resurfacing depends on the bone quality and therefore may not be an option for all patients [Vale *et al.*, 2002].

1.2 NATURAL SYNOVIAL JOINT

1.2.1 Introduction

Through the centuries, the joints in the human body have been classified into two main groups, synarthroses and diarthrodial (synovial joints). In synarthroses joints, the bones are linked by fibrous tissue or cartilage, which may be replaced by bone later. Only synovial joints will be discussed here. These joints are different from synarthroses joints in that they allow for a large degree of relative motion between the opposing bones. Some examples of this type of joint are the shoulder, elbow, hip, knee and ankle.

The natural synovial joint is a remarkable bearing. It is expected to perform its task without any service for at least 70 years whilst transmitting dynamic loads of large magnitude and yet accommodating a wide range of movement. The main purpose of the synovial joint is to allow for movement. The synovial joint is an encapsulated system which encloses its articulating surfaces and lubricant, as shown in Figure 1.2 The lubricant is called synovial fluid which allows considerable movement with ease between articulating bones. The end of each bone is covered with a protective layer of articular cartilage, which serves to reduce contact stresses in the joint, protect bone surfaces from impact loads, and minimise friction and wear in the joint [Dowson *et al*., 1981].

A synovial joint has an outer layer which has a similar shape to a sleeve and is made of strong, collagen tissue. Ligaments are also part of this sleeve and account for the primary stability of the joint. The sleeve is oversized to allow the joint to move. This sleeve is fed by blood vessels and can repair itself in case of injury. There is a tissue lining attached to the inner side of the sleeve which is called 'synovium'. This membrane secretes synovial fluid into the synovial cavity, which fills the joint space and is the prime lubricant and source of food for the joint. Tendons attach muscle to bone. Allows the movement and acts as the second joint stabilizer. Muscles contract to provide the force for movement. Muscles are critical for shock absorption around a joint. Bursa is tiny, fluid-filled sacs located at strategic points to cushion ligaments and tendons and protect them against friction, wear and tear.

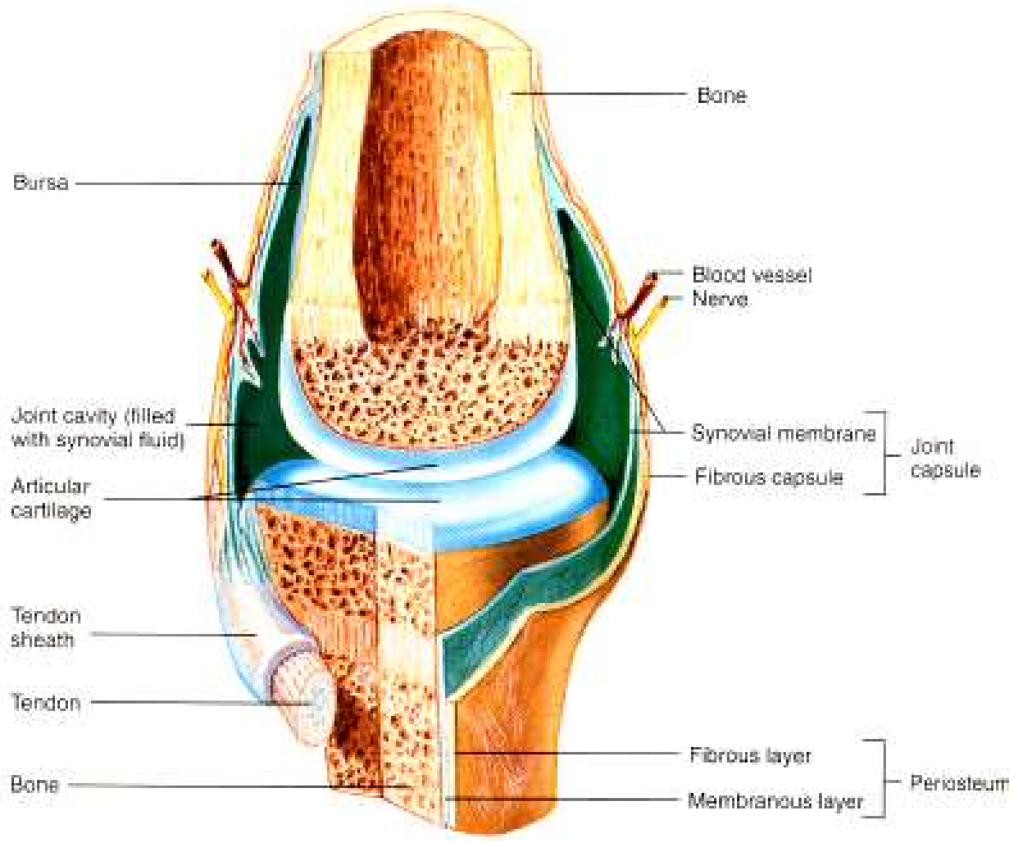


Figure 1.2. Anatomy of a synovial joint [Seeley *et al*., 1998].

The synovial membrane, which surrounds the joint, serves several purposes:

- it regulates the amount and content of the synovial fluid,
- it removes waste materials from the synovial fluid,

- it allows nutrients to enter the synovial capsule,
- it secretes synovial fluid and other macromolecules for lubrication of the joint [Mow *et al.*, 1993].

1.2.2 Hip anatomy

The hip joint is a connection between the lower limb and the pelvic girdle. Hip joint is a strong, sturdy synovial ball-and-socket joint (Figure 1.3). Acetabulum, femur, head of femur, neck of femur, articular cartilage, synovial fluid and ligament of head of femur are structures of the hip joint.

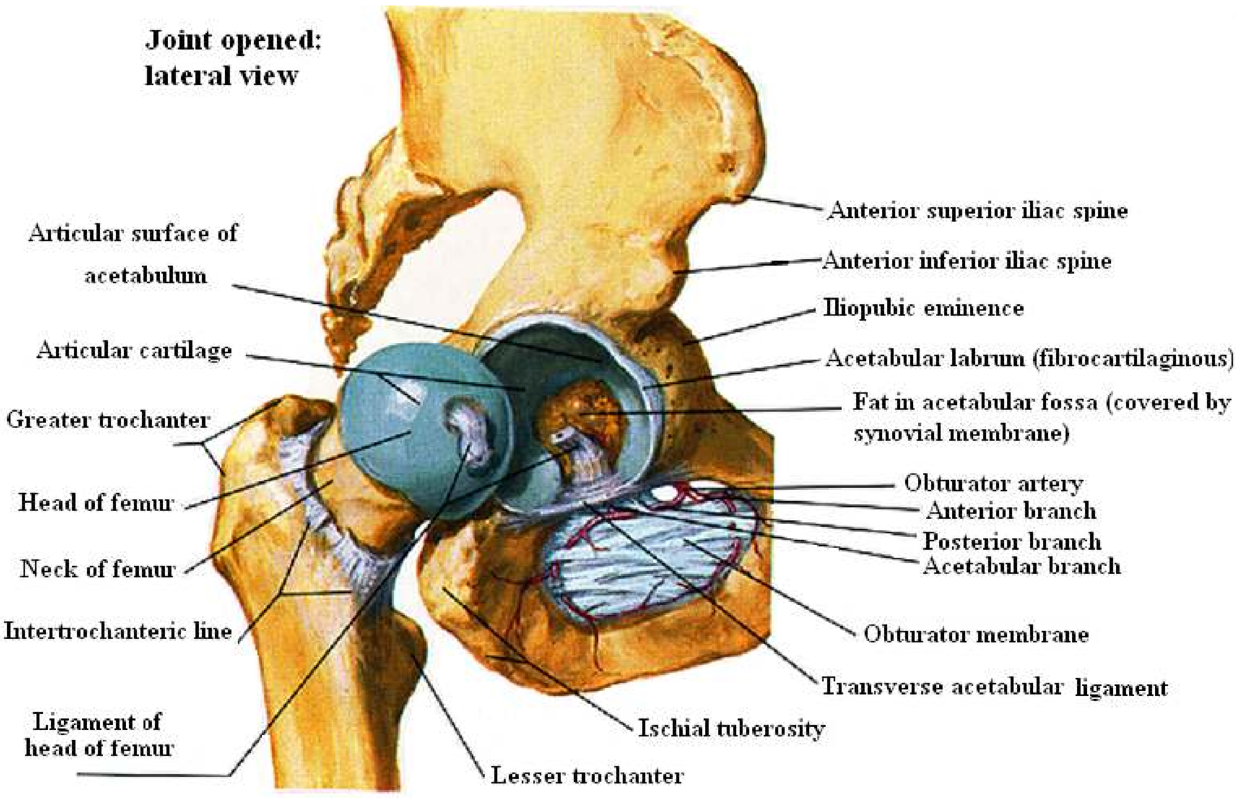


Figure 1.3. Hip Anatomy [Netter *et al.*, 2003].

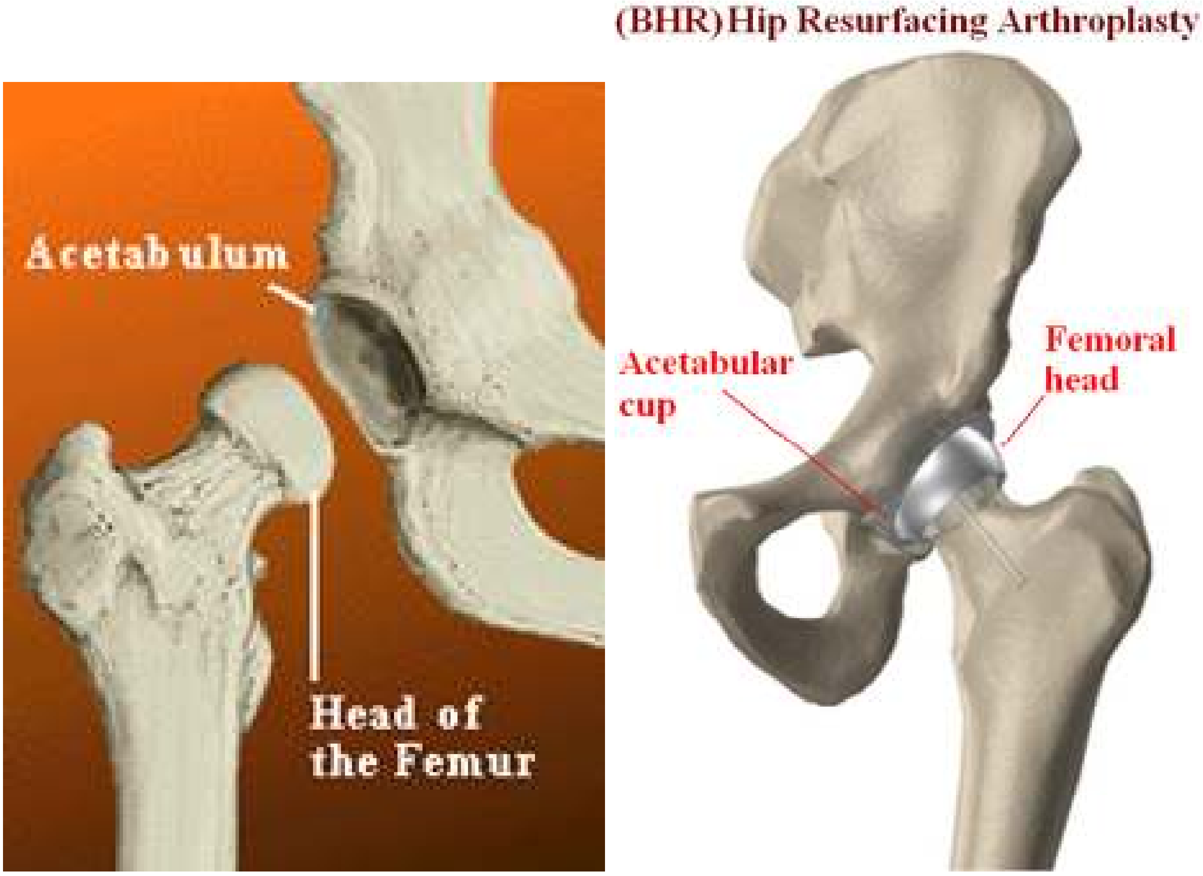
More than half of the rounded head of the femur (ball) fits and articulates within the acetabulum (socket). This articulation permits flexion and extension, adduction and abduction, circumduction and rotation, whilst maintaining stability. The stability of the hip joint is determined by the shape of the articular surfaces, the strength of the joint capsule and associated ligaments, and the insertion of muscles crossing the joint, which tend to be at the same distance from the centre of the movement.

1.2.3 Femur

The femur is the longest, strongest and heaviest bone in the body that articulates with the os coxae at the hip joint. The anterior surface of the femur consists of the femoral head and a short neck, shaft, the greater and lesser trochanter and the intertrochanteric line. The femoral head articulates with the pelvis at the acetabulum (see Figure 1.3). The femoral head is attached to the acetabulum by a ligament at the fovea capitis (posterior surface). The shaft of femur is almost cylindrical in most of its length and is jointed to the neck at the angle of about 125^{o}. This angle varies throughout the adult life cycle and is called the inclination angle [Gray, 1997; Nordin 2001; Palastanga *et al.,* 1998]. The grater and lesser trochanters are large, rough projections that extend laterally from the junction of the neck and shaft. On anterior surface of the femur, the raised interochanteric line makes the edge of the articular capsule.

1.2.4 Femoral head

The femoral head forms two-thirds of a sphere, being slightly compresses in an anteroposterior direction. The femoral head is covered in articular cartilage, except for a small area superolaterally adjacent to the neck and at the fovea capitis (a pit on the posteromedical part of the head). Anteriorly the cartilage extends on to the femoral neck for a short distance. The cartilage is thickest on the superior surface of the head and that of acetabulum as the greatest contact pressure occurs in this area. Generally the femoral head diameter in adults ranges between 45-56mm and is angled medially, anteriorly and superiorly [Palastanga *et al.,* 1998].



(a) **(b)**

Figure 1.4. Natural Human hip joint (a) and Hip Resurfacing Arthroplasty (b) [www.fauxpress.com, www.eorthopod.com].

1.2.5 Acetabulum

On the lateral aspect of the hip bone, acetabulum is a hemispherical large cup shaped cavity on the outer surface of the innominate. The acetabulum is the fusion of its three component parts: the anterior one-fifth of the acetabulum is formed by pubis, the superior posterior two-fifths formed by the body of the ilium and the inferior posterior two-fifths formed by the ischium. These bones meet at a Y-shaped cartilage forming their epiphyseal junction. The prominent rim of the acetabulum is deficient inferiorly as the acetabular notch. The heavy wall of the acetabulum consists of a semilunar articular part, covered with hyaline cartilage, which is open below, and the acetabular fossa which is a deep central non-articular part (see Figure 1.3). The acetabular fossa is formed mainly from the ischium and its wall is frequently thin. The acetabulum is orientated anteriorly, laterally and inferiorly. The articular surface of the acetabulum is covered with articular cartilage and is semi-lunar in shape. The cartilage layer is thickest on the upper portion of the articular surface where the highest forces are applied [Dowson *et al.,* 1981; Levangie 2001].

1.2.6 Articular cartilage and synovial fluid

Articular cartilage is a complex material consisting of both solid and fluid components. The solid portion is composed primarily of a network of collagen fibres and brush-like proteoglycan molecules. This network traps water in the material and stores it as a gel; this gel becomes pressurised upon application of a load to the joint, and enables the cartilage to support relatively high loads. In addition to providing the framework for the material, the collagen network provides an ideal surface for sliding. Cartilage is more flexible than the subchondral bone that supports it, and is therefore well suited for padding the bone surfaces to reduce contact and impact stresses [Dowson *et al*., 1981]. The thickness of articular cartilage varies from one joint to another, and often from one position to another on a single joint surface. In the larger joints of young men and women, it may be about 2 - 3 mm in depth.

The natural lubricant, synovial fluid, is a clear, yellowish, and viscous substance. A normal human synovial joint contains only about 2 ml of synovial fluid [Dowson *et al*., 1981], but this small amount filling the space between the articulating surfaces of the joint, and enclosed within the synovial membranes serves several purposes: it lubricates the articulating surfaces, carries nutrients to the cartilage cells, or chondrocytes, transports waste products away from the cartilage, and also protects the joint surfaces against degenerative enzymes [Seeley *et al.*, 1998]. Synovial fluid is a dialysate of blood plasma, which consists of a complex mixture of polysaccharides, proteins and lipids.

1.2.7 Diseased synovial joints

One of the most common diseases that affect the joints is osteoarthritis (OA) [Jones *et al.*, 1995]. OA cannot be cured at the present time, but it can be treated by various methods such as weight loss and exercise, nonsteroidal anti-inflammatory drugs (NSAID) and injection of corticosteroids directly into the affected joint. Eventually if none of these treatments is effective, joint replacement is the only option, where the articulating surfaces of the joint are surgically removed and replaced with prostheses. OA is the most common form of arthritis, affecting millions of people, as shown in Table 1.1. Synovial fluid plays a functional role in nutrition and removal of waste from the joint. It also aids in a proportion of joint movement. Movement of the joint acts to "pump" the synovial fluid from the synovial membrane to apply a lubricant cover to all the joint surfaces while also flushing waste from the synovial joint in the process.

Table 1.1. Diagnosed total prevalent of OA [Adopted from Wieland *et al.*, 2005]

Osteoarthritis Epidemiology (in millions)

Country	2002	2007	2012
United States	13.2	14.4	15.5
Europe	14.5	15.2	15.8
Japan	6.6	6.9	7.2
OA total prevalent cases	34.3	36.5	38.6
RA total prevalent cases	6.6	6.9	7.2

OA, Osteoarthritis; RA, Rheumatoid arthritis

Other researchers have predicted similar OA trends, expecting the prevalence to increase to 40 million individuals in the year 2020 [Shadick, 1999].

OA may begin as a molecular abnormality in articular cartilage, with heredity and normal "wear and tear" of the joint important contributing factors. A slowed metabolic rate with increased age also seems to contribute to OA. Inflammation is usually secondary in this disorder. It tends to occur in the weight-bearing joints such as the hip and is more common in overweight individuals [Seeley *et al*., 1998]. Although OA tends to be much more common among the elderly, joint trauma or various other factors can cause an early onset of degenerative joint disease.

A stiff and painful hip, due to arthritis of the hip joint, can prevent a patient from performing even the simplest of activities. Initially arthritis can be dealt with symptomatically, with oral medications, exercise programs, weight reduction and occasionally braces or ambulatory assistance devices. When the pain and disability increases to the point where simply standing, walking, and climbing stairs causes pain, it is time to consider surgery [Wright *et al.*, 2001]. The procedure is called Total Hip Arthroplasty (THA), which in general involves replacing the damaged cartilage of the hip joint with prosthesis.

1.3 IMPLANT BEARING MATERIALS

1.3.1 Introduction

Most artificial hip joints have one highly polished metallic femoral part sliding over an UHMWPE cup, which is supported by a metallic shell. Two tribological principles support the selection of these materials. First, the combination provides low friction and, second, the hard femoral component is highly polished so that, provided it remains undamaged, low wear rates of the UHMWPE surface are produced [Fisher, 1994]. There are also other requirements which must be fulfilled with all biomaterials such as biocompatibility, durability, high static and fatigue resistance, high fracture toughness and high corrosion resistance.

1.3.2 Cross-linked Ultra-High Molecular Weight Polyethylene

For more than 30 years, ultra-high molecular weight polyethylene (UHMWPE) has been used as a bearing material in orthopaedic implants due to its outstanding wear properties, biocompatibility, ductility, chemical stability, effective impact load damping and low coefficient of friction against metallic femoral components, such as Cobalt Chromium (CoCr) alloys [Li *et al.,* 1994; Landy *et al.,* 1998; McKellop *et al.,* 1995]. UHMWPE is a linear polyethylene with mechanical properties linked to its chemical structure, molecular weight (1,000,000 to 10,000,000 Da), crystalline organisation, and thermal history. All these factors, in turn, affect the morphological, chemical, and mechanical processes, which may influence wear and performance after implantation.

UHMWPE varies greatly in its consistency and its physical properties, not only between medical grades of polyethylene, but also within single batches of UHMWPE. For this reason, ASTM F648 stipulates minimum requirements for the mechanical properties of UHMWPE which is to be employed in orthopaedics. These are outlined in the Table 1.2 [ASTM F648].

Table 1.2. Requirements for UHMWPE fabricated forms for surgical implants [ASTM F648].

Property	Test method	Requirement		
		Type 1	**Type 2**	**Type 3**
Density (Kg/m^3)	**ASTM D 792 or D 1505**	930-940	927-938	927-944
Ultimate Tensile Strength, 23°C,min, MPa (Minimum)	ASTM D 638,Type IV, 5.08 cm/min	35	27	27
Tensile Yield Strength, 23° C, min, MPa (Minimum)	ASTM D 638,Type IV, 5.08 cm/min	21	19	19
Elongation, min, % (Minimum)	ASTM D 638,Type IV, 5.08 cm/min	300	300	250
Impact Strength, min, kJ/m^2 (Minimum)	Annex A1	140	73	30
Deformation under load, max % After 90 min recovery	ASTM D621 (A) (7 MPa for 24 h)	2	2	2
Hardness, Shore D, min	ASTM D 2440	60	60	60

Some methods used to make orthopaedic components from UHMWPE powder, are as shown in Figure 1.5.

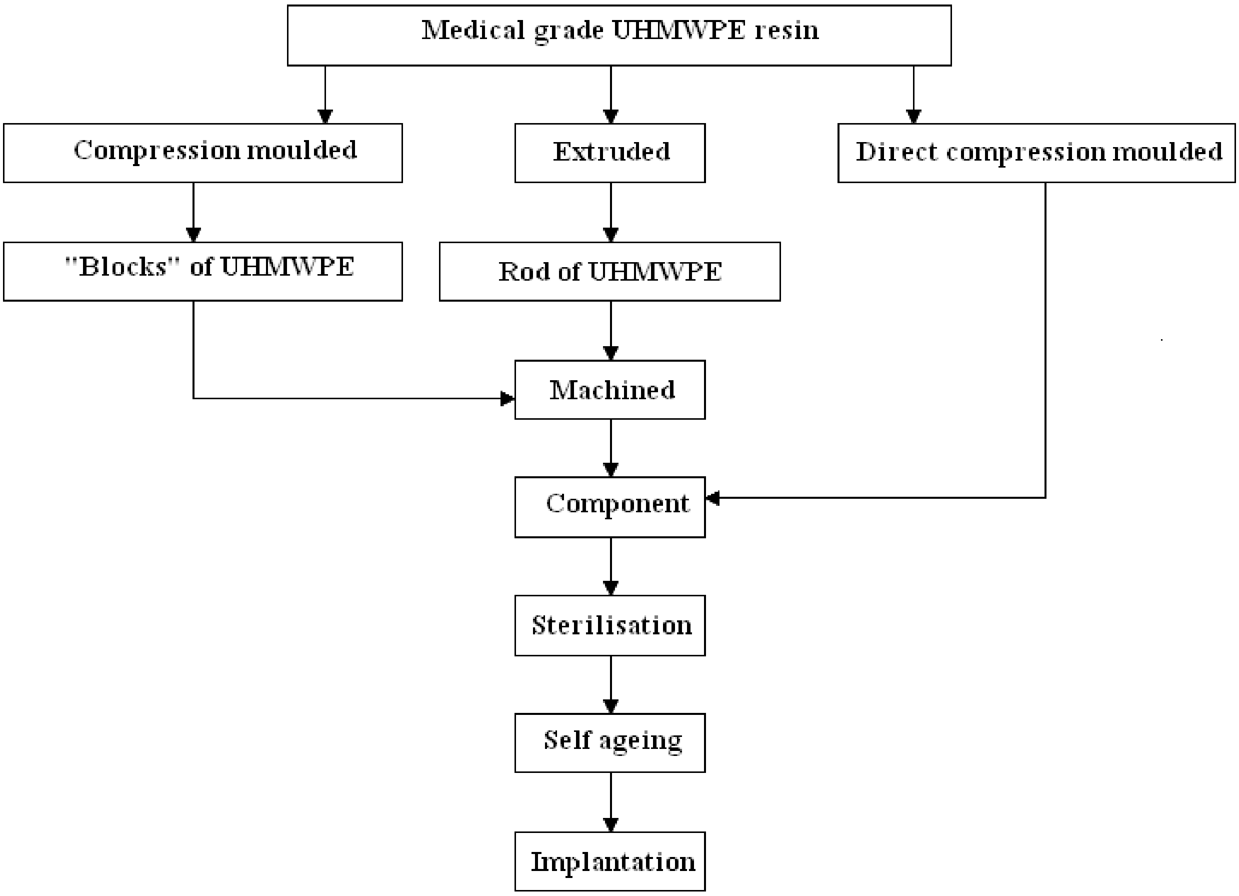


Figure 1.5. Different methods for making orthopaedic components from UHMWPE powder [Adapted from Goldman *et al*., 1998].

Although UHMWPE has very low wear compared to other polymers, still wear is a major concern in total hip replacement. Sliding of UHMWPE components against the metal or ceramic counterface result in production of wear debris leading to complications such as tissue inflammation, bone resorption (osteolysis) and consequently aseptic implant loosening, affecting the longevity of hip prostheses [Dannemaier *et al.,* 1985; Howie *et al.,* 1990; Harris *et al.,* 1996; Sochart *et al.,* 1999]. This is of particular concern for young and/or active patients, who may face one or more revisions, with cumulative bone loss, in their lifetime. As a consequence, wear resistance of UHMWPE must be improved to extend the clinical life span of joint-replacement prostheses.

With the goals of reducing creep and wear rates, in the past, some formulations of UHMWPE components have been developed; for example, blending UHMWPE with carbon fibres to get total-joint-replacement fabrication components known as Poly Two (Zimmer Inc., Warsaw, IN) [Wright *et al.,* 1981]. However it was not a complete success, due to the decrease in fatigue resistance, and as no improvement in wear resistance was observed the material was ultimately discontinued for further use in joint-replacement devices. The other failed improving method was high-pressure crystallization to produce UHMWPE components with an increase in mechanical properties such as yield stress and Young's modulus which was known as Hylamer (DePuy-Dupont Orthopaedics, Newark, DE) [Li *et al.,* 1991]. However, Hylamer has again not shown any improvement in laboratory wear tests, despite enhanced creep resistance and an increase in resistance to fatigue crack growth [Mckellop *et al* 1992]. It was also indicated that Hylamer does not demonstrate increased resistance to wear in total-hip-replacement prostheses compared with conventional UHMWPE [Chmell *et al.,* 1996; Liningston *et al.,* 1997].

Nowadays, it is believed that instead of using novel processing methods such as high-pressure crystallization or physical blending, it is better to modify UHMWPE components via chemical methods. Cross-linking of UHMWPE macromolecules has been performed using cross-linking agents such as peroxides [Shen *et al.,* 1996], and gamma [Oonishi *et al*., 1996; Oonishi *et al.,* 1997; Clarke *et al.,* 1997] or electron beam (EB) irradiation [Premnath *et al.,* 1996; Muratoglu *et al.,* 1995].

Wear properties of UHMWPE have been improved considerably since 1995. In the past, due to sterilization with gamma irradiation in the presence of air; the molecules in the long polyethylene chains were broken by gamma irradiation, giving rise to free radicals. This method was made at doses between 25 and 40 KGy. The presence of oxygen in the polyethylene during irradiation or the diffused oxygen into the polyethylene during shelf storage and/or *in vivo*, could react with the free radicals, causing oxidative degradation which would in turn lower the molecular weight, increase the density, stiffness and brittleness, and also reduce the fracture strength and elongation to failure of polyethylene [Costa *et al.,* 1998; Kurtz *et al.,* 1999]. These changes could adversely affect the polyethylene wear resistance. However, irradiation of polyethylene can also lead to cross-linking and when carried out in the absence of oxygen it will markedly improve the wear resistance of polyethylene. Due to the improvement made in the properties of polyethylene in the absence of oxygen, by 1998, all of the major orthopaedic manufacturers in the United States were either sterilising UHMWPE using gamma radiation in a reduced oxygen environment or sterilising without ionising radiation, using ethylene oxide (EtO) or gas plasma [Kurtz *et al*., 1999].

The post treatment of the acetabular cup and irradiation must be optimised in order to minimize the degradation and to achieve cross-linking. The material is cross-linked as pre-pegs due to oxidation of the top layer, and then during machining this oxidised top layer is machined off; leaving only the underlying cross-linked material. Higher irradiation doses are used for cross-linking (about 50-100 KGY) in comparison with sterilization. Residual radicals in the cup are quenched by heat treatment before sterilisation [Kurtz *et al.*, 1999], either by irradiating in an inert atmosphere or without using irradiation (i.e. with ethylene oxide or gas plasma process).

There are at the present time six different types of cross-linked cups on the market. They are all made using a variation of irradiation doses and sources and are heat treated in different ways. Table 1.3 lists the now commercially available cross-linked cups and how they are made.

Table 1.3. Present available cross-linked cups and methods of manufacturing them [Kurtz *et al.*, 1999].

Name and manufacturer	Radiation type and dose	Thermal stabilization	Final stabilization	Total cross-linking Dose and Type
Marathon™ DePuy, Inc.	γ radiation to 50 KGY at room temperature	Remelted at 155°C for 24 hours followed by annealing at 120°C for 24 hours.	Gas plasma	50 KGy gamma
XLPE™ Smith & Nephew-Richards, Inc.	γ radiation to 100 KGy at room temperature	Remelted at 155°C for two hours	Ethylene oxide	100 KGy gamma
Longevity™ Zimmer, Inc.	Electron beam radiation to 100 KGy at room temperature	Remelted at 150°C for about six hours	Gas plasma	100 KGy electron beam
Durasul™ Sulzer, Inc.	Electron beam radiation to 95 KGy at 125°C	Remelted at 150°C for about two hours	Ethylene oxide	95 KGy electron beam
Crossfire™ Stryker-Osteonics-Howmedica, Inc.	γ radiation to 75 KGy at room temperature	Annealed at about 120°C for a proprietary duration	Gamma at 25 to 35 KGy while packaged in nitrogen	*100 to 110 KGy of gamma
Aeonian™ Kyocera, Inc.	γ radiation to 35 KGy at room temperature	Annealed at 110°C for 10 hours	Gamma at 25 to 40 KGy while packaged in nitrogen	*60 to 75 KGy of gamma

*For crossfire™ and Aeonian™, the total crosslinking does will depend on how much irradiation is used for terminal sterilization. The allowable range is 25 to 40 KGy.

It is important to point out that the decrease in wear is accompanied by a decrease in other mechanical properties, such as fatigue strength. Table 1.4 lists some material properties and how they change upon cross-linking [Lewis *et al.*, 2001].

Table 1.4. Material properties and how they change upon cross-linking [Adapted from Lewis *et al.*, 2001].

Property	Uncross UHMWPE - linked	Cross-linked UHMWPE
% Crystallinity	53.6 ± 6.2	45.3 ± 5.3
Melting temperature (°C)	13.9 ± 3.3	135.8 ± 5.6
Yield strength (MPa)	25.6 ± 3.3	21.1 ± 2.5
Ultimate tensile strength (MPa)	48.7 ± 7.5	29.3 ± 7.7
Tensile modulus of elasticity (MPa)	915 ± 423	860 ± 206
Tensile elongation at fracture (%)	317 ± 140	212 ± 61

Change in fatigue resistance [from Lewis *et al.*, 2001] due to irradiation:

- Unirradiated 1.37 ± 0.06
- Irradiated Electron Beam 100 KGy 0.74 ± 0.01
- Irradiated Electron Beam 100 KGy, melted 0.56 ± 0.02

1.3.3 Metal-on-metal hip prostheses

Osteolysis, causing wear particles inducing aseptic loosening, is now thought to be one of the major contributing factors to the failure of hip resurfacing [Howie *et al.,* 1998, Schmalzried *et al.,* 1992]. There are many different types of hip prosthesis incorporating various material combinations on the market today but there is a renewed scientific interest in particular over the use of metal-on-metal bearings for hip arthroplasties. Laboratory tests by [Smith *et al.* 2001] has verified that metal-on-metal implants demonstrate much lower volumetric wear compared to that of metal-on-polyethylene joints. However, the wear volume is not the only factor that predicts the biological response to wear particles. Studies have shown that biological response is also influenced by the type of material and the shape and size of the wear particles [Green *et al.,* 2000; Ingham *et al.,* 2000]. It is important to point out that it has been acknowledged that the low-wear characteristics of metal-on-metal implants are related to the generation of full or partial lubricating films throughout part of the walking cycle. It has been suggested by lubrication theory and the joint simulator tests that the lubrication and thereby the wear rate within the bearing system is affected by femoral head diameter and a variation in radial clearance between the femoral head and acetabular cup. In addition, the results of the study of steady-state wear by [Smith *et al.* 2001] clearly indicate the merits of larger-diameter femoral heads and small well-controlled clearances. Metal-on-metal also permits the use of thinner acetabular cups and larger-diameter femoral heads, which potentially enjoy the advantage of reduced dislocations without incurring the risk of fracture associated with ceramic implants. Release of metallic ions and the long-term local and systemic effects of nanometer-scale wear debris are still the main issues, which lead to essential minimization of volumetric wear.

1.3.3.1 History

Thomas Gluck developed the first ball-in-socket joint in 1890, which consisted of a femoral head made of smooth, hard ivory, and fixed with nickel screws. It was followed by few other attempts during the successive years to progress this idea [Ott *et al.,* 2002]. The preliminary challenge was to enable the separation of the two bone surfaces, and to smooth the counterface of the joint in order to reduce pain caused by the degeneration of the cartilage.

In 1917, William Baer (John Hopkins Medical School) employed sheets of pig's bladder in order to achieve an interposing membrane. However, due to the high stresses acting within the joint this method soon failed. In the early 1920s a hollow glass hemisphere that fitted over the femoral head to provide a new smooth joint surface was introduced by Marius Smith-Peterson (Boston, USA). Unfortunately, there was a failure of the glass components due to the generated stresses in the normal gait cycle. Although Pyrex was later introduced, this procedure still proved to be inadequate.

Ernest Hey-Groves (Bristol, UK) challenged the first total femoral replacement involving an ivory ball inserted with a short peg in 1927. Whilst Smith-Peterson was further developing the idea of mould-arthroplasty, in 1936 a new cobalt-chromium alloy, Vitallium, started to be applied to this principle. Despite the fact that the material was found to be excellent for its application, the design was found to be insufficient.

Philip Wiles introduced the first total hip replacement in London in 1938. The implant was composed of matching femoral and acetabular components manufactured from stainless steel. Concurrently, Preston and Albee in the USA developed the first metallic acetabular cups made from Vitallium.

Edwarc Haboush (New York, USA) introduced in early 1940, a hollow formed implant that could fit around the femoral head and neck. He also found out that Vitallium prosthesis against an acrylic acetabular component would result in excessive wear.

Jean and Robert Judet (Paris, France, 1946) failed to accept the current methodology and started employing acrylic materials to create short-stemmed femoral replacement. Although in the beginning the implants were extensively used throughout Europe, these implants soon failed due to the brittle nature of the materials. However this design was further developed by introducing metallic materials. Despite the fact that this new material combination reduced the fatigue and wear characteristics of the prosthesis, unfortunately this design failed due to loosening [Ott *et al.,* 2002].

Smith Peresen [Smith *et al.,* 1948] introduced contemporary designs that progressed directly from the original mould arthroplasty. There were some long term survivals, regardless of being a hemi-arthroplasty with no means of stable fixation to the femoral head.

The femoral head component replacement was further developed by Frederick Thompson (New York, USA) and Austin Moore (South Carolina, USA) in the 1950s. This hemi-arthroplasty failed to achieve the ideal result as the design failed to defeat the further degeneration of the acetabulum. Despite the failure of these designs, they encouraged the development of the first intra-medullary implants, with metallic stems being press-fitted into the canal of the femur.

Haboush further developed the use of metal implants in 1951, and he also introduced the use of a dental cement to secure the device in position. During this period, the first design of metal-on-metal (stainless steels) total hip replacement was introduced by George McKee and John Watson-Farrar (UK). However, due to loosening of these implants, McKee adopted the Thompson design for the femoral component and employed a new prosthesis design, in which the femoral component was articulating against a cup screwed into the acetabulum. He determined that stainless steel and titanium did not meet the criteria and were not suitable options for the metal-on-metal pairing. As a result, he employed cobalt chrome alloy to manufacture the components. However, these components displayed premature failures due to loosening. This was believed to be due to high frictional torques generated between the articulating surfaces [August *et al.,* 1986; Jacobsson *et al.,* 1996; McKee and Watson-Farrar *et al.,* 1966].

Sir John Charnley [Charnley *et al.,* 1961; Charnley *et al.,* 1963] developed the first total resurfacing arthoplasty using Teflon-on-Teflon in the early 1950s. This method purely replaced the damaged bone surface rather than resection of the entire head, which was associated with high early failure (within two years) due to the poor wear characteristics of Teflon.

Charnley (Lancashire, UK) attempted his first metal on polymer bearing in 1958. He determined that the friction in the natural joint was significantly lower than that generated in prosthesis. Therefore, he introduced an implant in which the two bearing materials would articulate freely, without generating great frictional torques. He employed a Polytetrafluoroethylene (PTFE) cup and stainless steel head. This selection was due to the PTFE's low friction and its biological inertness. However, these bearings failed due to the high level of wear within approximately two years. As a result, he introduced a new bearing material using polyethylene. Due to its good clinical results, Charnley`s 'Low Friction Arthroplasty' (Figure 1.6) was widely accepted by 1961. These clinical outcomes resulted in the Charnley's concept to become the 'gold standard' within the field of orthopaedics, and even today there is no other implant design as successful with a clinical history [Charnley *et al.,* 1970; Charnley *et al.,* 1982]. In order to firmly secure the joint into the bone, Charnley also introduced polymethylmethacrylate (PMMA). Consequently, due to the clinical success of the Charnley prosthesis and the early failures of the McKee-Farrar's metal-on-metal implants due to loosening, the metal-on-UHMWPE became the bearing couple of choice.

Figure 1.6. Charnley's Low Friction Arthroplasty [Charnley *et al.,* 1961].

A self-locking metal-on-metal total hip replacement (Figure 1.7) was developed by Peter Ring (Surrey, UK) in 1960s, which did not require the use of cement [Ring *et al.,* 1971; Ring *et al.,* 1974].

Figure 1.7. Ring's Implant [Ring *et al.,* 1974].

In 1968, Muller [Muller *et al.,* 1995] designed a cementless cobalt-chromium metal-on-metal articulation, and resurrected the resurfacing principle first introduced by Charnley in 1967, which was an implantation of 18 surface replacements in young patients in addition to 35-stemmed prostheses. Despite the excellent early clinical results, Muller abandoned the use of metal-on-metal in favour of metal-on-polyethylene. Six of these metal-on-metal articulations were revised after functioning for up to 25 years.

In 1970, Gerard introduced a bipolar metal-on-metal resurfacing device. This system consisted of a Luck cup inserted into an Aufrane Vitallium cup with movement occurring between the prostheses and between the outer cup and the bony socket [Gerard *et al*., 1978]. Wagner (Germany) and Freeman (UK) introduced a metal-on-polymer prosthesis in 1970s. Despite the initial success these bearings failed mainly with femoral neck fracture, after a short period of time (within six years) [Freeman *et al.,* 1978b; Wagner *et al.,* 1978]. At the time, the nature of this failure was not apparent, but nowadays it is believed that it was due to Osteolysis caused by the polymer wear debris [Freeman 1978a].

A porous-coated cementless prosthesis that consists of a CoCr femoral component, a modular liner, and a Ti-6Al-4A hemispherical acetabular component was introduced by Harlan Amstutz in 1988 [Amstutz *et al.,* 1991]. He achieved similar results to Wagner and Freeman while further developing metal on polymer resurfacing in the 1980s. At the same time, he developed a resurfacing hemi-arthroplasty, where the femoral head was resurfaced with a metallic component. He also determined that the common neck fracture resulting in the failure of metal-on-polymer resurfacing did not occur when there was no polymer coupling [Amstutz *et al.,* 1982; Amstutz *et al.,* 1986; Amstutz and Graff-Radford *et al.,* 1981; Amstutz *et al.,* 1984].

[Amstutz *et al. 1*996] and [Jacobsson *et al.* 1996] reported that there was very low wear rate in metal-on-metal prostheses, thereby ensuring a long term survivorship of these total hip replacements. It was understood that factors such as superior manufacturing processes, as well as better tolerances and smoother surface finishes, had a significant effect on the outcome of these prostheses. Hence there was a renewed interest in metal-on-metal bearings after the late 1980s.

1.3.3.2 The renewed interest in metal-on-metal hip arthroplasty

In 1988, metal-on-metal articulation for total hip arthroplasty showed new interests [Amstutz *et al.,* 1996]. An accurately engineered Metasul bearing with high carbon wrought Co-Cr alloy and outstanding wear characteristics was developed by Weber and Sulzer [Weber *et al.,* 1996]. In 1991 Heinz Wagner [Wagner *et al.,* 1996] introduced a cementless second-generation hip arthroplasty by employing the Weber and Sulzer's durable low-wear bearing procedure. The acetabular cup was a titanium alloy shell with a Metasul inlay. The implantation of this method faced some difficulties due to thickness of the construct and the extensive macro features on its external surface.

As cast CoCr alloy hip resurfacing arthroplasty was introduced by McMinn [McMinn *et al.,* 1996] in cooperation with Corin Medical (Cirencester, United Kingdom) in 1991. This device was a smooth surface and press-fit design on both sides. Unfortunately, this design soon failed due to aseptic loosening of both components. In the following year he introduced a system in which both components were cemented. The central peg and peripheral fins were removed to modify the original acetabular design while the femoral head was kept in the original phase for cementing. This system failed due to high incidence of early acetabular loosening due to cement-cup debonding, which led to introduction of a hybrid system in 1994 with a cementless HA-coated acetabulum, this system was abandoned due to manufacturing problems in 1996. Therefore, there was an introduction and development of two different resurfacing systems; The Coemet-2000 that was developed by Corin Medical and the Birmingham Hip Resurfacing (Midland Medical Technologies, Birmingham, United Kingdom; now Smith & Nephew, Memphis, Tennessee). The Conserve Plus hybrid hip resurfacing (Weight Medical Technology, Arlington, Tennessee), with both components made of cast, heat-treated, solution-annealed CoCr alloy were introduced by Amstutz in 1996.

1.4 EXISTING HIP RESURCAING

1.4.1 Introduction

Metal-on-metal hip resurfacing systems were introduced by most of the main implant manufacturers by the end of 2004. Table 1.5 lists the currently marketed hip resurfacing systems.

Table 1.5. Currently marketed hip resurfacing systems [Adopted from Grigoris *et al.,* 2005].

Bearing				Acetabulum			Femur		
System	**Introduced**	**process**	**Heat treatment**	**Size increments (mm)**	**shape**	**surface**	**Size increments (mm)**	**Cement mantle (mm)**	**Stem**
Converse Plus (Wright Medical Technology, Arlington, Tennessee)	**1996**	**Cast**	**HIP, SHT**	**2**	**Truncated hemisphere**	**Sintered CO-Cr beads +/- HA**	**2**	**1**	**+ / - Load bearing**
Birmingham Hip Resurfacing (Smith & Nephew, Memphis, Tennessee)	**1997**	**Cast**	**None**	**2**	**Equatorial expansion**	**Co-Cr beads; cast-in + HA**	**4**	**0**	**Not defined**
Cormet Resurfacing Hip System (Corin Medical, Cirencester, UK	**1997**	**Cast**	**HIP, SHT**	**2**	**Truncated hemisphere**	**Ti- VPS + HA**	**4**	**0**	**Not defined**
Durum (Zimmer, Winterthur, Switzerland)	**2001**	**Wroug ht**	**Not applicable**	**2**	**Truncated hemisphere**	**Ti- VPS**	**2**	**1**	**Non-load bearing**
ASR (Articular Surface Replacement; Depuy Orthopaedics, Warsaw, Indiana	**2003**	**Cast**	**HIP**	**2**	**Truncated hemisphere**	**Sintered CO-Cr beads + HA**	**2**	**0.5**	**Non-load bearing**
Recap (Biomet, Warsaw, Indiana)	**2004**	**Cast**	**None**	**2**	**Hemispher e**	**Ti- VPS +/- HA**	**2**	**0.5**	**Not defined**
Icon Hip Resurfacing (International Orthopaedics GMBH, Bromsgrove, UK	**2004**	**Cast**	**None**	**2**	**Hemispher e**	**Co-Cr beads; cast-in + HA**	**4**	**0**	**Not defined**

Abbreviations: HIP, hot isostatically pressed; SHT, solution heat treated; Ti- VSP, titanium vacuum plasma sprayed.

As it can be observed in Table 1.5, despite the important differences between these systems (e.g. particularly relating to the metallurgy and geometry of the bearing and the fixation of the femoral and acetabular components) there are also some common factors between them. These include (1) the femoral component of cemented fixation, (2) the acetabular component of cementless fixation, and (3) a high carbon containing CoCr alloy bearing.

1.4.2 Bearing materials

The metallurgy of the bearings is the most contentious concern in the existing metal-on-metal resurfacing. Despite the fact that all the manufacturers use high carbon containing CoCr alloy, there is a difference in the processing of the alloy. This alloy can be Cast (in which components may undergo post-casting heat treatments such as hot isostatic pressing or solution heat treatment) or Wrought. Post-casting heat treatments have been an important issue over the last six years. Although this treatment results in depletion of the surface carbides, the outcomes of hip simulator studies have not shown any significant difference in the wear behaviour between as-cast and heat treated alloys [Bowsher *et al.,* 2003].

1.4.3 Acetabular fixation

The surface used for bone in-growth is the main difference between the various current resurfacing acetabular components. CoCr beads and Titanium vacuum plasma sprays are the contemporary materials in use. Despite a satisfactory performance in conventional total hip replacements, there are some concerns about the extreme temperature involved in the sintering process of CoCr beads. It is suggested that the process may alter the metallurgy of the monobloc component, which in turn could have a deleterious effect on the bearing surface.

1.4.4 Femoral fixation

The optimal cement mantle thickness and the degree of the cement pressurization are the major issues. Diametric difference between the implant and the corresponding reamer determine the thickness of the cement mantle. Systems that do not allow escape of cement during the implantation of femoral heads may result in excessive penetration of cement into the cancellous bone. Also, femoral neck fracture may be caused due to the high force required to fully seat such implants.

1.5 DEFINITION OF TRIBOLOGY

Tribology is a relatively new word in the vocabulary of the scientific world and is derived from the Greek word "Tribein" meaning rubbing and Friction [Oxford English Dictionary], which was introduced into the United Kingdom in 1966. Tribology is the study of friction, wear and lubrication, and design of bearings, science and technology of interacting surfaces in relative motion and of related subjects and practices. In other words, tribology deals with lubrication, friction and wear, which can be involved with a number of basic engineering subjects such as solid mechanics, fluid mechanics, lubricant chemistry, material science, heat transfer, etc. Understanding of tribology is a very important matter since relative motion is taking place in all engineering components and systems. Tribology in practice is a main issue in the design process for machine elements and the formulation of lubricants. Typical examples in which tribology is one of the key factors include: ball bearings, gears, tyres, clutches, brakes, cams and Followers, constant Velocity and Universal Joints and biomedical Implants, e.g. replacement human hip joints. Tribology is also playing an important role in biological systems, Dowson and Wright introduced the term bio-tribology in 1973 that cover all aspects of tribology related to biological systems. Some of the common examples related to the field of bio-tribology are as follows:

- numerous studies of natural synovial joint lubrication,
- the design of various forms of total joint replacements,
- manufacture and performance of various forms of total joint replacements.

Other examples of tribology applied to biology include:

- friction of skin and garments, affecting the comfort of clothes, socks and shoes [Dowson *et al.,* 1998; Sivamani *et al.,* 2003;], and slipperiness [Gronqvist *et al.,* 2001; Maynard *et al.,* 2002]
- tribology of contact lenses and ocular tribology [Holly *et al.,* 1994]
- tribology at micro-levels—inside cells, vessels and capillaries such as lubrication by plasma of red blood cells in narrow capillaries [Secomb *et al.,* 2001]
- the wear of replacement heart valves [Reul *et al.,* 2002]
- the lubrication of the pump in total artificial hearts [Walowit *et al.,* 1997]
- the wear of screws and plates in bone fracture repair [Shahgaldi *et al.,* 2000]
- lubrication in pericardium and pleural surfaces [Gouldstone *et al.,* 2003]
- tribology of natural synovial joints and artificial replacements [Mow *et al*., 1993; Dowson *et al.,* 2001]

1.5.1 Tribology of the natural hip joint

In order to achieve an optimal function of hip replacement, it is essential to understand the tribology of the natural joint. The universal perceptive is based on the fact that the hip joint can function successfully for approximately 70 years under low load and low friction. In an average adult, in severe condition (e.g. high fluctuating load and low velocity), the hip joint could undergo two million cycles per year. Studies on the natural joints are used by the orthopaedic companies to develop successful hip replacements.

Lubrication (synovial fluid) in the natural joint is playing an important role in reducing friction and wear, and promoting optimal function of the joint. Synovial fluid, as it has been mentioned previously, is a plasma transudate, which contains proteins, and many other constituents such as glucose and lubricin. Hyaluronic acid is also a constituent of synovial fluid, as it contributes to the viscosity of the fluid. Synovial fluid performs two functions, providing nutrition to the cartilage and assisting lubrication [Mow *et al*., 1997].

There have been many studies exploring the effect of fluid viscosity on the lubrication of the joint [Dowson 2006; Dowson *et al,* 2005; Ingham *et al,* 2000; Jin *et al,* 2006]. Also, studies on determining the lubrication regime within the joint via filtering the fluid or separation of the fluid by adding fat solvents have taken place. The outcome of these experiments has shown that the boundary lubrication is the most likely mode within the natural joint (with lubricin contributing significantly to the lubrication).

1.5.2 Tribology of the artificial hip joints

Full understanding of the tribology of the hip joint is required in order to enable optimisation of current prosthesis designs. Some of the studies have been only concentrating in either wear or friction of the artificial joint, and have not directly evaluated the lubrication regime. Simulators have been used to carry out *in vitro* tests in which the implants undergo the loading and motion profiles which estimate those generated *in vivo*. Wear behaviour of an implant could be examined by employing a wear simulator. This can be done by screening weight, dimensional and surface roughness changes at numerous intervals throughout the test period. The frictional torque during a cyclic test can be measured by use of a friction simulator, and then the friction factor can be calculated. Furthermore, Stribeck analysis using these results takes place in order to establish the lubricating mode. It should be pointed out that there are several ultrasound techniques that could be used for determination of lubrication mode and measurements of contact pressure [Quinn *et al*., 2002]. 'Time of flight' is one of these procedures in which film thickness is being determined by using ultrasound. An alternative method is the 'Continuum Model Approach' which measures the response of the arrangement to waves of a known frequency and amplitude. The measurement of thin film thicknesses is based on the maximum operating frequency, however it should be pointed out that this process could fail at thicknesses below 10μm [Dwyer-Joyce *et al*., 2003].

1.5.3 Factors influencing tribology

Studies have shown that in order to achieve optimal performance of hip prostheses, some design features like femoral head radius, cup thickness, clearance and surface finish, as well as material type should be seriously considered [Goldsmith *et al.,* 2000b; Saikko *et al.,* 1998; Smith *et al.,* 2001b]. In particular these features have been shown to have a significant effect upon the lubricating regime between the articulating surfaces of hard-on-hard bearings.

The outcome of recent studies have shown that in order to achieve superior lubrication and consequently low wear of the hard-on-hard prostheses, an improved surface finish, larger head diameter and a reduced radial clearance are essential [Jin 2002; Liu *et al*, 2006]. It should be noted that the increase of the head size would result in an increase of the sliding distance.

Therefore the resultant wear of prostheses would increase. However in hard-on-hard bearings, due to their inherent wear resistance and improved lubrication, this effect would be small.

Another factor resulting in better lubrications and reduced wear would be a low clearance device [Farrar *et al*, 1997; Liu *et al*, 2006; Tipper *et al,* 2005]. However, the manufacturing precision becomes critical with low clearance bearings, as deviation in sphericity and tolerance may cause the joint to seize resulting in increase in both friction and wear.

In contrast, researchers have shown that in metal-on-polymer bearings, wear would increase below a critical minimum clearance. It is postulated that at low clearances some articulating areas may become depleted of lubrication. Also the increase in head size would increase the sliding distance in metal-on-polymer prostheses resulting in an increase in the generation of wear particles [Cooper *et al.,* 1992; Edidin *et al.,* 2001; Elfick *et al.,* 1999; Fisher *et al*., 1991].

1.5.4 Friction

Friction is resistance to motion encountered when a solid object moves tangentially with respect to the surface of another which it touches, or when an attempt is made to produce such motion [Rabinowicz *et al*., 1965]. Friction generates between two interacting surfaces in regions where they are in contact. Although the shearing at the contacting areas between two surfaces generates the principal resistance, roughness and ploughing of the surfaces also contribute to the overall frictional force.

Depending upon the situation, friction may be advantageous or detrimental. The invention of the wheel, in order to reduce friction, and the production of fire from the heat generated by rubbing two sticks together are positive use of high friction. However, there have been attempts to overcome the unfavourable effects of friction by pouring a liquid that is believed to be oil for lubrication, on to the ground in front of the sedge to facilitate the pulling of the Colossus (Figure 1.8). This suggests an early appreciation of the benefits of lubrication. Figure 1.8 shows the transportation of an Egyptian colossus from a painting in the tomb of Tehuti-Hetep dated about 1800 BC.

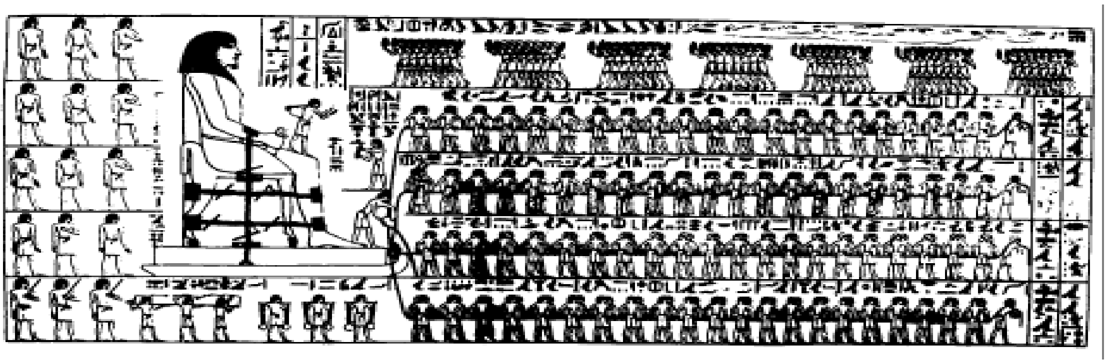


Figure 1.8. Transporting an Egyptian colossus.

Friction was first studied by Leonardo da Vinci (AD 1452-1519) in the fifteenth century and further developed by Coulomb (1736-1806) [Bowden *et al.*, 1974].

There are three laws of dry friction:

I. The force of friction (F) is directly proportional to the applied load (W)

II. The force of friction (F) is independent of the apparent area of contact

III. The kinetic force of friction (F) is independent of the sliding speed (V).

The first law gives rise to the definition of Coefficient of friction (μ) which is a non-dimensional ratio, as shown in Equation 1.1.

$$\mu = \frac{F}{W}$$ **Equation 1.1**

Where F is the frictional force (N) and W is the applied load (N).

The second law is being counterintuitive with friction apparently independent of the area of contact. That is until one notes that it is the apparent area of contact that is referred to, not the real area of contact.

The third law was introduced by Coulomb in the 18th century. It has a much smaller range of applicability than the first two and should therefore be treated with caution when considering real engineering systems.

Nowadays it is believed that friction can be caused by two factors, deformation and adhesion. It is understood that adhesion contributes significantly towards the friction between articulating surfaces. This theory is based on the fact that when rough surfaces are in contact, the real area of contact is at the tip of their asperities. Hence, the load is predominantly supported by the deformation of these asperities [Rabinowicz *et al,* 1965]. Due to the fact that the real area of contact is very small, it is assumed that the pressure acting at the asperity contacts is high enough to cause them to deform plastically. Figure 1.9 considers one such asperity contact.

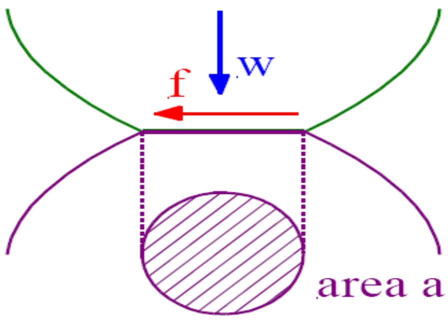


Figure 1.9. An Adhesive Asperity Contact.

Adhesive bonds are taking place, when the asperities of two surfaces are in contact. These adhesive bonds can either be chemical or physical. The movement of the surfaces with respect to each other can be achieved by brakeage (sheared) of these adhesive bonds.

Friction between the articulating surfaces could be influenced by the surface roughness of materials. `Run in` or `wear in` is a period in which wear and frictional properties of material change due to the transformation of the surfaces. As the asperities are worn down and the surfaces become deformed, the frictional forces tend towards a steady state value. Factors such as adhesion and third body wear could increase the surface roughness of the articulating surfaces and consequently increase the friction to a higher steady-state value.

Introduction of a lubricant between two sliding surfaces would result in a reduction in friction value. This is due to the lubricant's lower shear strength in comparison with either of the surface materials. The relative shearing motion therefore occurs within the lubricant as it requires much less force in order to produce any motion.

Concerns over the generation of high friction between bearing surfaces has played an important role in the development of artificial hip joints. Sir John Charnley used a simple pendulum experiment to demonstrate superior frictional properties of metal on PTFE with respect to the metal on metal hip replacements in his early hip replacement design (Table 1.6) [Charnley *et al*., 1966]. It is important to note that despite the low friction of the metal on PTFE bearing couples, these implants failed very quickly due to wear.

Table 1.6. Typical coefficients of friction for clean material in dry contact in the presence of air [Adopted from Dowson et al., 1981].

Material combination	Coefficient of friction
Steel on steel	**0.6-0.8**
Polyethylene on steel	**0.3**
Polyethylene on polyethylene	**0.2-0.4**
PTFE on PTFE	**0.04-0.2**
PTFE on steel	**0.04-0.2**

The majority of early hip implants using a metal-on-metal articulation also failed rapidly, largely due to the equatorial contact and resultant high friction and generated frictional torque. A free pendulum machine with a hydrostatic frictionless carriage was used by Unsworth in 1975, to determine the friction and predict the lubricating mode of various prostheses [Unsworth *et al.*, 1976]. These tests demonstrated a mixed lubrication for both metal-on-metal and metal-on-polyethylene bearings, and also showed that static conditions made formation of a lubricating film difficult.

Further studies by Unsworth (1978) on the effects of the viscosity of the lubricant on friction showed that viscosities above 0.1 Pa s enabled fluid film lubrication in both metal on metal and metal on polyethylene bearings [Unsworth *et al.*, 1978].

Studies by Scholes and Unsworth also investigated the effects of bearing material combinations and various lubricant types with different viscosities on the lubrication regimes [Scholes and Unsworth *et al.*, 2000; Scholes *et al.,* 2000a & b]. In their work, the researchers compared the experimental results with theoretical calculations. These studies demonstrated friction factors of

0.001 – 0.06 in the ceramic-on-ceramic bearings which appeared to be lower than that generated by the metal-on-metal (0.16 – 0.35) and metal-on-polyethylene bearing couples (0.02 – 0.07). The authors carried out Stribeck analysis (to give an indication of the lubrication regime) for each arrangement in synthetic fluid, demonstrating fluid film lubrication in the ceramic-on-ceramic joint and mixed lubrication for both the metal-on-metal and metal-on-polymer implants. It is important to note that all material combinations were shown to be operating within a mixed lubrication regime when physiological fluid was used as a lubricant. Although a reduction in friction factor in metal-on-metal bearings in biological fluid was observed, both ceramic-on-ceramic and metal-on-polymer bearing couples displayed an increase in friction factor. It is believed that the adsorption of proteins from the biological fluid onto the articulating surfaces may have been the cause of the increase in friction. It is also postulated that in metal-on-metal bearings, the adsorbed proteins may reduce the friction by protein-on-protein contact. In contrast, the effect of protein adsorption onto the ceramic bearings may result in an increase of the counterface surface roughness and consequently adversely affecting the operating conditions.

1.5.5 Wear

Wear is a process in which progressive damage, involving material loss occurs on the solid surface of a body as a result of its motion over another [Rabinowicz *et al.*, 1965]. Wear usually occurs in a very small amount making the protection and prevention hard to achieve. However, indications such as, mass loss (weight loss), volume loss, and changes in roughness, waviness and form suggest that wear is occurring. It should be pointed out that such changes can also result from plastic deformation of the surface with no material loss. Despite the fact that there is no correlation between wear and friction, introduction of a lubricant could result in reduction in wear and friction. Studies have also shown that, factors such as sliding distance, applied load and softness of the sliding surface have an increasing effect on material wear [Ahlroos *et al,* 1997]. Nowadays, it is believed that the rough nature of real surfaces and the influence of asperities are the major origin of wear between dry surfaces in relative motion.

The wear is an important factor relating not only to the decreased function and replacement cost of bearing materials, but also the adverse effects of wear particles. For example, in synovial joints, it has been shown that the wear particles could cause adverse tissue reactions leading to Osteolysis and consequently loosening of the implants [Anissian *et al,* 2001]. Similarly, adverse effects on the quality of magnetic recording systems may be caused by the wear debris.

Five types of wear mechanism are described below;

Adhesive wear is the transference of material from one surface to another during relative motion by the process of solid-phase welding (Figure 1.10).

Figure 1.10. Adhesive Wear at a Single Asperity Contact.

Adhesive wear is the most common phase of wear, in which generally materials of similar hardness are involved and is taking place by shearing the junctions formed between surface asperities. It is important to point out that load and sliding distance have a proportional effect on adhesive wear, whereas surface roughness [Elfick *et al,* 1999; Edidin *et al,* 2001]and hardness of the wearing surfaces have an inverse effect.

Abrasive wear is the removal of material from one surface by the other. It is proposed that abrasive wear results from the ploughing by a hard asperity which has penetrated into a softer counterface or where loose particles scratch the surfaces of the material in third body wear. In other words abrasive wear is the displacement of materials by hard particles.

The ploughed materials from the grooves often form loose wear particles and eventually contribute to third body wear. Abrasive wear is affected by surface finish of the materials and is inversely proportional to the hardness of the surface [Rabinowicz *et al*., 1965].

Corrosive wear is a process in which chemical or electrochemical reactions with the environment are the predominant factors, e.g. oxidative wear. During sliding, the resultant product of these reactions could be worn off and cause further corrosion. The formation of oxide layers on metal surfaces and subsequent removal by rubbing is conceivably the most common example of corrosive wear.

Fatigue wear is the removal of materials as a result of cyclic stress variation over a long time periods. The number of cycles and the magnitude of the applied stresses are playing an important role in this failure [Rabinowicz *et al*., 1965]. Unlike adhesive and abrasive wear, fatigue wear can occur without direct contact between the surfaces. Fatigue wear also can occur in well-lubricated rolling contacts.

Erosive wear is a process in which material loss from a solid surface occurs due to relative motion in contact with a fluid which contains solid particles. The basic mechanism of erosive wear is damage to a surface caused by the impact of hard solid particles carried by a fluid. The wear rate is found to be directly proportional to the kinetic energy of the particles [Bowden *et al*., 1974]. Erosive wear is often subdivided into impingement erosion and abrasive erosion. Erosion could still take place even if solid particles are not present in the fluid, e.g. rain erosion and cavitations [Bhushan *et al*., 1999].

It should be pointed out that some of the above wear types (adhesive, abrasive, fatigue, erosive wear) are taking place due to mechanical actions whereas corrosive wear is due to chemical action. In addition it should be said that some of the above wear types could occur simultaneously or sequentially, i.e. wear particles, which may be produced as a result of adhesive wear, can then act as third bodies causing abrasive wear. Studies have shown that adhesive, abrasive and fatigue wear can contribute to the overall wear in polymeric bearing surfaces. Some wear terms often described for artificial joints can be related to the above mechanisms. For example, pitting and delamination are specific forms of fatigue wear, while burnishing and scratching are different degrees of abrasive wear. Understanding the wear mechanism also helps to achieve an appropriate design strategy to reduce wear [Jin *et al,* 2006; Mckellop *et al,* 1992]. For example, using hard and smooth bearing surfaces such as ceramics could result in minimised abrasive wear. Designs of prostheses and material selection have an important role in fatigue wear, hence, it is important to minimise the contact stresses in order to avoid short-term fatigue failure. An effective lubrication regime is one of the predominant factors minimising adhesive wear. In addition, wear may result in failure of total hip replacement hence consideration of wear behaviour is an important issue to optimise the design and consequently improve the longevity of implants. Although wear is a complex interaction of different mechanisms, a number of common factors have emerged as being influential in determining the rate of wear. These include; Kinetics, kinematics, sliding speed, temperature, oxides and contaminant surface films, compatibility of surface materials, surface treatments and coatings and lubrication [Landy *et al*, 1998; Mow *et al*, 1997; Goldsmith *et al,* 2000a; Dowson 2001]. There are also a wide range of laboratory equipments, test methods and measuring systems that have been employed to measure wear and to study wear mechanisms in artificial hip joints.

The most popular implant wear testing machines are as follow:

- pin-on-disc machine
- pin-on-plate machine
- hip wear simulator

Pin on plate/disc tests are carried out for evaluating the wear properties of combinations of materials that are being considered for use as bearing surfaces of artificial joints (Figure 1.11).

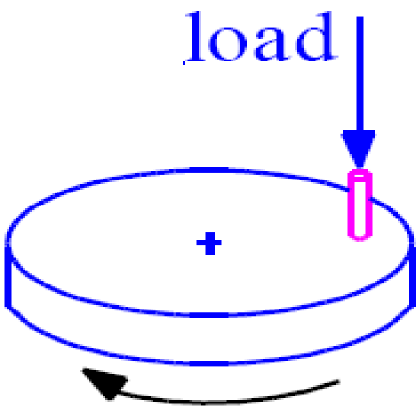


Figure 1.11. Pin on disc machine.

The purpose of these test methods is to rank materials according to their wear rate under physiological conditions.

Gravimetric assessments of the pins are carried out in order to determine the wear. The average mass of each pin is calculated and compared with the previous results to calculate the weight loss.

The wear results are analysed in terms of the wear factor, K (mm^3/Nm). In order to calculate K, the volume loss is calculated by dividing the weight loss by the density of the pin and then the wear factor is calculated according to the relationship:

$$K = \frac{V}{P \times X}$$ **Equation 1.2**

Where V is the volume of the material removed from the pin (mm^3), P is the normal load (N) and X is the sliding distance (m). The wear factor results for the individual test pins are grouped to give a mean result for each set of tests. Typical wear coefficients for various materials combinations are shown in Table 1.7.

Table 1.7. Representative wear coefficient, K_1, for various material combinations [Dowson *et al.,* 1981]

Material combination	Wear coefficient
PTFE-on-steel	**10^{-4}-10^{-5}**
High-density, high molecular weigh Polyethylene on steel t	**10^{-7}-10^{-8}**

However, the results must be viewed with some caution, since the conditions under which the materials are tested are drastically simplified. These test methods, therefore, represent only an initial stage in the full wear characterisation of a candidate material. After the selection of the material, more expensive and time consuming hip wear simulator tests should be carried out allowing a close approximation to the *in vivo* situation.

The hip wear simulator studies have shown that the hard-on-hard bearings have a lower volumetric wear rate than a metal-on-polymer bearings [Ahloors and Saikko *et al*., 1997; Chan *et al.,* 1999; Clarke *et al.,* 2000; Dowson 2001; Goldsmith *et al.,* 2000a]. It should be pointed out that hard-on-hard bearings generate a relatively high wear in the early stages of the test, known as the bedding-in period, and would thereafter shift towards a low wear rate, known as steady-state period. In contrast, metal-on-polymer bearings have shown a reasonably steady wear rate throughout the duration of the wear tests [Annissian *et al.,* 2001; Campbell *et al.,* 1999; Scholes *et al.,* 2001].

Several researchers have investigated the lubrication regimes between the metal-on-metal articulating surfaces in hip wear simulator studies [Goldsmith *et al.,* 2000a; Dowson *et al.,* 2001; Annissian *et al.,* 2001]. The authors reported a self-polishing phenomenon between the articulating surfaces resulting in wear reduction, asperity contact reduction, and superior lubrication.

Lubricant is used, in order to achieve an *in vivo* environment while testing the wear of implants. Although the lubricant used in the environment created within the wear simulator is 25% newborn calf serum, with an antibacterial additive, e.g. sodium azide, the fact that there is circulation and re-generation of the fluid within the synovial joint, produces an unfavourable difference between the natural hip joint, and the environment created within the wear simulator [Cooke *et al.,* 1978; Dowson *et al.,* 2003; Saikko *et al*., 2003]. In order to achieve an optimal lubricant performance and due to the high level of serum protein concentration (affecting serum life limitation), it is essential to replace the serum after two to three days (at a rate of one Hz, this is approximately ¼ million cycles) [Bell et al., 2000; Liao et al., 1999].

It should be noted that wear evaluations can also be carried out on components that are either in patients or have been in patients. Measurements of the *in vivo* or *ex vivo* components are certainly very interesting, as these components have been subjected to an extended exposure to the environment of the body [Clarke *et al,* 1997; Cooper *et al,* 1992; Edidin *et al,* 2001; Goldsmith *et al,* 2000a; Howie *et al,* 1990a; Landy *et al,* 1998; Chmell *et al,* 1996]. This means that the effects of chemical degradation and the interaction of these effects with wear processes will be more likely to be evident.

1.5.6 Lubrication

The main reason for introducing a lubricant between articulating surfaces is to reduce friction and/or wear. Commonly regimes or types of lubrication may be considered in the order of increasing severity or decreasing lubricant film thickness (Figure 1.12):

1. Hydrodynamic lubrication (fluid film lubrication – no contact between articulating surfaces)
2. Elastohydrodynamic lubrication (fluid film lubrication – no contact between articulating surfaces)
3. Transition from hydrodynamic and elastohydrodynamic lubrication to boundary lubrication (mixed lubrication – stage between full fluid film and boundary lubrication)
4. Boundary lubrication (significant contact between the asperities of the articulating surfaces)

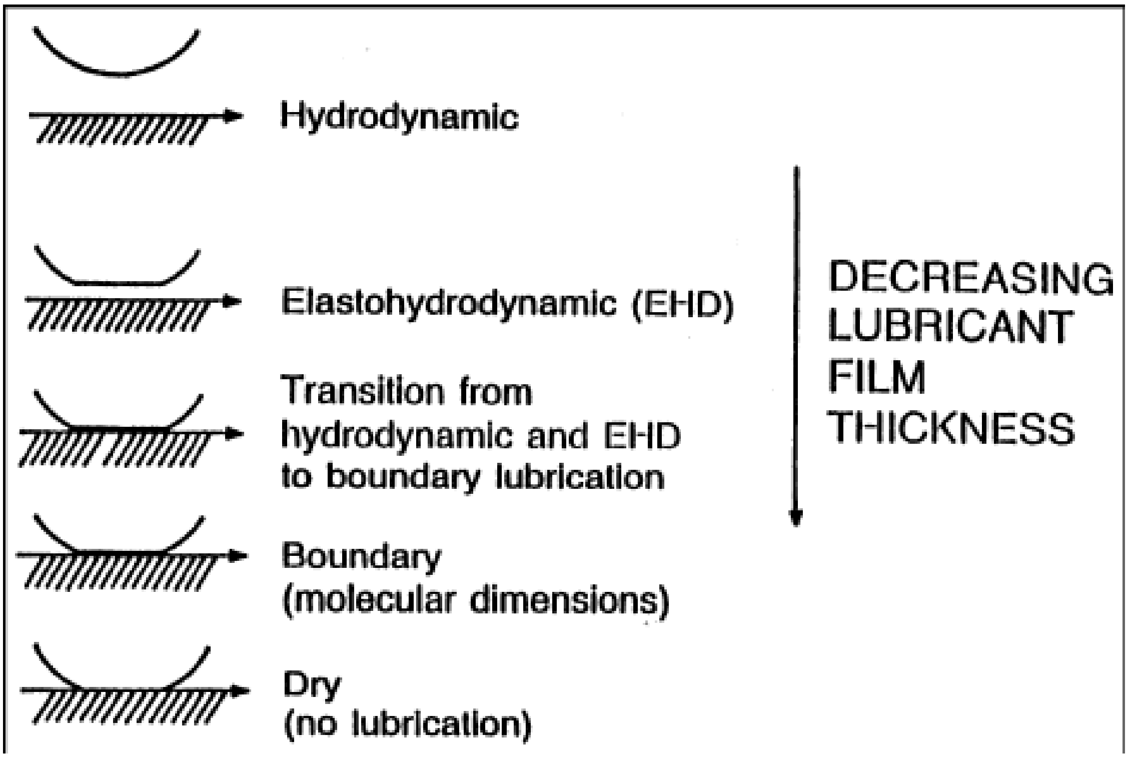
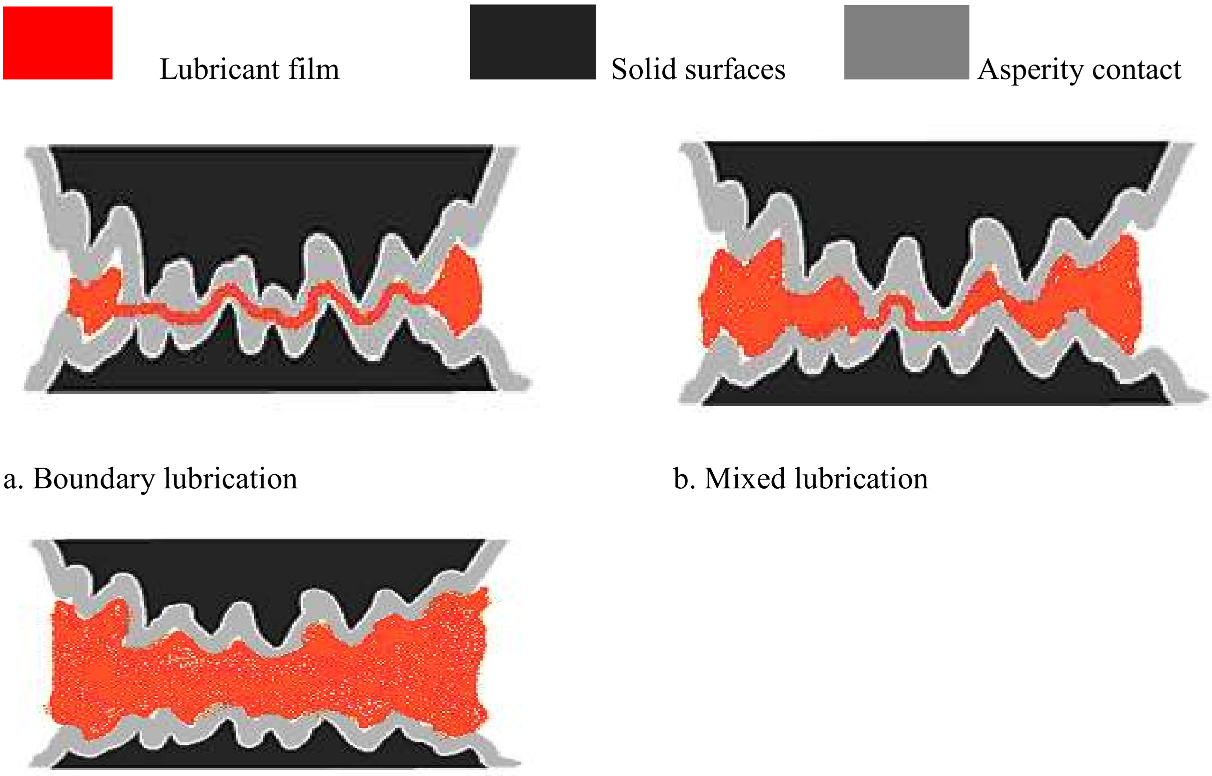


Figure 1.12. Types of lubrication [Adopted from Furey *et al.*, 2000].

Lubrication is mainly influenced by sliding speed and load. High sliding speed will promote the generation of fluid film lubrication which may result in reduction of friction and wear of the articulating surfaces. However, it is important to note that extremely high sliding speed alternation may cause the lubricant depletion to fail. In addition, excessive loading could result in an increase in the real area of contact between the counterfaces and consequently may cause breakthrough of the lubricant. Furthermore, it should be mentioned that, the lubrication mode may also vary depending on lubricant properties and surface roughness of the contacting surfaces, i.e. high lubricant viscosity and smooth counterfaces will promote lubrication. Although, it is well known that wear cannot be eliminated completely, in order to minimise wear, the ideal lubrication mode to generate is full fluid film. The characteristics of each lubrication mode in terms of friction and wear are summarised in Figure 1.13.



c. Fluid film lubrication

Figure1.13. Schematic drawings of asperity contacts between articulating surfaces in various lubrication modes [Dowson and Jin, 2005].

It can be seen in Figure 1.13 a-c that in boundary lubrication, where boundary films are protecting the surfaces, significant asperity contact is present which in turn would result in a significant increase in wear and friction. In mixed lubrication mode, there is a mixture of characteristic between boundary and fluid film lubrication and in fluid film lubrication a complete separation between the articulating surfaces can be observed.

1.5.6.1 Boundary Lubrication

Boundary lubrication is a condition in which a fluid lubricant does not separate the counterfaces and contact takes place over an area similar to that which develops in dry contact. In boundary lubrication the friction and wear characteristics are determined by the properties of the:

- surface materials
- lubricant films at their common interfaces
- physical and chemical properties of thin surface films

It should be pointed out that boundary lubricants reduce friction and wear principally by minimising adhesion and abrasion. Boundary lubricants on most metals are attached to a thin oxide layer, which is formed on their surfaces due to the rapid oxidation of metals in the presence of oxygen. The thickness of this oxide layer is usually one to five nanometres. At slow speed and high contact pressure between the articulating surfaces, boundary lubrication is the predominant lubrication mode. It has been reported that in boundary lubrication, the coefficients of friction are lower than those for dry bearings but much larger than those in full fluid film lubrication [Bhushan *et al*., 1999; Rabinowicz *et al*., 1965].

1.5.6.2 Mixed Lubrication

This is the region in which lubrication goes from the desirable full fluid film with no contact between the articulating surfaces to the less acceptable boundary condition, where increased contact usually leads to higher friction and wear. This lubrication regime is referred to as mixed lubrication. It should be noted that both the physical properties of the bulk lubricant and chemical properties of the boundary lubricant are important factors affecting the wear and friction between the counterfaces.

In the mixed lubrication regime, there are regions between the articulating surfaces which are separated by a lubricant film and also regions with contact between asperity peaks on the counterfaces. Therefore, the load is distributed between the contacting asperities and the lubricating film separating the counterfaces in some areas [Nordin *et al*., 2001].

1.5.6.3 Fluid Film Lubrication

In fluid film lubrication, there is no contact between the counterfaces and the load is supported by the pressure developed due to relative motion and the geometry of the system. As the film thickness depends on the bulk physical properties of the lubricants, the most important factor is the viscosity of the lubricants. In full fluid lubrication mode, friction would solely generate from shearing of the viscous lubricant.

Fluid film lubrication is generated by two major methods, hydrodynamic and squeeze-film action. Hydrodynamic action occurs where the lubricant converges into a wedge, increasing the pressure within the lubricant. The pressure can cause the bearing surfaces to deform elastically and consequently increase the separation between the counterfaces. This mechanism is called elastohydrodynamic lubrication. Squeeze-film lubrication occurs in conditions where counterfaces approach each other under rapid loading conditions. This mode often maintains a fluid film between the two surfaces for short periods of time.

It is important to understand and to determine the lubrication mechanism in artificial joints, as it has been mentioned previously, such an understanding may help to optimise the bearing surfaces to minimise friction and wear. Conventional engineering methods of assessing lubrication regimens can be applied to artificial hip joints. These can be generally classified into two categories, experimental measurements and theoretical predictions. The experimental methods involve either friction measurements using hip friction simulators and relating the results to the Stribeck curve, or the detection of separation between the two bearing surfaces using a simple resistivity technique. The resistivity technique is particularly useful for conducting metal-on-metal bearings. Theoretical predictions are based on the dimensionless parameter λ (lambda) ratio and it is also a ratio of film thickness (h_{min}) to the composite average surface roughness (R_a) of the counterfaces [Dowson *et al.,* 2001]. The calculations of these parameters are as follows:

$$\lambda = \frac{h_{min}}{\sqrt{R^2{}_{a_1} + R^2{}_{a_2}}}$$

Equation 1.3

Where R = the composite surface roughness (m)

h_{min} = the lubricating film thickness (m)

R_{a1} = the surface roughness of the femoral component (m)

R_{a2} = the surface roughness of the acetabular component (m)

$$R_a = \sqrt{R_{a_1}^2 + R_{a_2}^2}$$

Equation 1.4

Where R_a = the composite surface roughness (m)

R_{a1} = the surface roughness of the femoral component (m)

R_{a2} = the surface roughness of the acetabular component (m)

The lubricating mode is then determined by the calculated value for λ:

- boundary lubrication is achieved for a ratio below one ($\lambda \leq 1$)
- fluid film lubrication is generated for ratios greater than or equal to three ($\lambda \geq 3$)
- mixed lubrication is achieved between boundary and fluid film values ($1 < \lambda > 3$) [Williams *et al.*, 1996].

In order to conduct theoretical assessment for fluid film lubrication the accurate measurement of surface roughness (R_a) and the calculation of a representative film thickness for the bearing (h_{min}) are required. A typical film thickness equation is [Jin *et al.,* 1997]:

$$\frac{h_{min}}{R} = 2.8\left(\frac{\eta u}{E'R}\right)^{0.65}\left(\frac{W}{E'R^2}\right)^{-0.21}$$ **Equation 1.5**

Where h_{min} = the film thickness (m)

R = the reduced radius (m)

η = the viscosity of the synovial fluid (Pas)

u = the entraining velocity, (u_l+u_2/2 (m/s)

E' = the equivalent Young's Modulus (Pa)

W = the load at the hip (N)

Where equivalent radius (R) depends on:

- the femoral head diameter (d)
- the diametral clearance between the head and the cup (C_d)

The equivalent radius is then calculated from:

$$R = \frac{R_1 R_3}{R_3 \pm R1}$$ **Equation 1.6**

Where R_l = radius of the femoral head (m)

R_2 = inner radius of the acetabular cup (m)

R_3 = outer radius of the acetabular cup (m)

R = reduced radius (m)

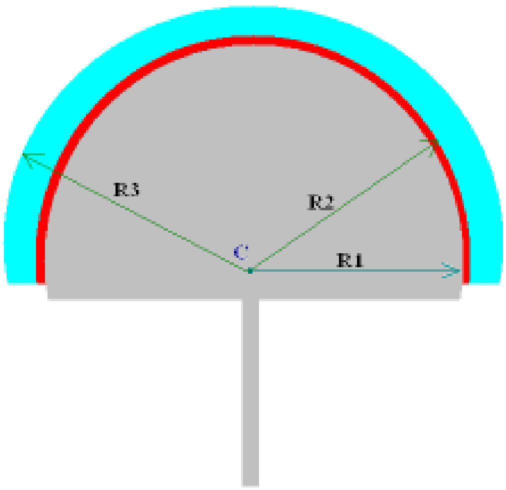


Figure 1.14. Radius variation for Metal head and acetabular cup.

The entraining velocity (u) can be calculated from the angular velocity of the femoral head (ω)

$$u = \frac{\omega d}{4}$$ **Equation 1.7**

Finally, the equivalent elastic modulus (E`) is a function of the Young's modulus of each of the two materials and is given by:

$$\frac{1}{E'} = \frac{1}{2}\left[\frac{1-\nu_C^2}{E_C} + \frac{1-\nu_f^2}{E_f}\right]$$ **Equation 1.8**

Where ν_c = Poisson's ratio of the acetabular cup material

E_c = Young's Modulus of the acetabular cup material (Pa)

ν_f= Poisson's ratio of the femoral stem material

E_f= Young's Modulus of the femoral stem material (Pa)

E' = Equivalent Young's Modulus of the two materials (Pa)

The variation in the friction factor against the Sommerfeld number (z) can further indicate the mode of lubrication (Figure 1.14).

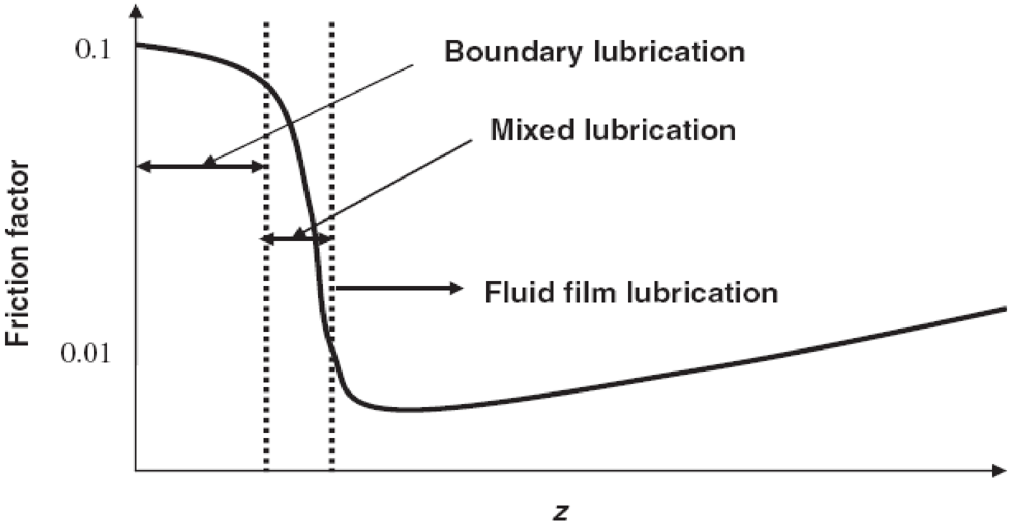


Figure 1.15. Typical friction factors and associated lubrication regimens [Adopted from Jin *et al.,* 2006].

Figure 1.15 is known as a Stribeck curve, which is an acknowledgment of the great contribution to studies of journal bearing lubrication by this German engineer in the 1920s. The Sommerfeld number (z) in Figure 1.14 is expressed as:

$$z = \frac{\eta u r}{L}$$ **Equation 1.9**

Where η = viscosity of the lubricant (Pa s)

u = the entraining velocity of the bearing surfaces (m/s)

r = the radius of the femoral head (m)

L = the applied load (N)

The friction factor (*f*) in Figure 1 is expressed as:

$$f = \frac{T}{rL}$$ **Equation 1.10**

Where T = the measured frictional torque

L = the applied load

r = the femoral head radius

As it can be observed in Figure 1.14, the constant friction factor with increasing Sommerfeld number indicates boundary lubrication. A reducing friction factor with increasing Sommerfeld number is indicative of a mixed lubrication and increasing friction factor with increasing Sommerfeld number indicates fluid film lubrication. Typical friction factors in various hip joints and determination of lubrication in typical metal-on-metal hip implant are shown in Table 1.8 and Table 1.9 respectively.

Table 1.8. Typical friction factors for various bearings for artificial hip joints in presence of bovine serum [Jin *et al.,* 2006].

Lubrication regimes	**Friction factor**
Boundary lubrication	**0.1-0.7**
Mixed lubrication	**0.01-0.1**
Fluid film lubrication	**0.001-0.01**

Table 1.9. Determination of lubrication in typical metal-on-metal hip implants [Jin *et al.,* 2006].

Femoral hear diameter	28 mm
Diametral clearance	0.06 mm
Elastic modulus (Co-Cr)	210 GPa
Poisson's ratio (Co-Cr)	0.3
Load	2.5 kN
Angular velocity	1.5 rad/s
Viscosity	0.0025 Pa s
Composite R_a	0.014μm
Calculation	
Equivalent radius	6.55 m
Entraining velocity	0.0105 m/s
Equivalent elastic modulus	230 GPa
Minimum film thickness	0.024 μm
λ ratio	1.7
Lubrication regime	Mixed lubrication regime

As it can be observed in Table 1.10 the lubrication regimes that exist within natural hips. Comparison of Tables 1.9 and 1.10 demonstrate that lubrication regime at natural hip joint showed totally different to that of replaced joint with a metal-on- metal 28mm implant.

Table 1.10. Material Properties and Lubrication Regimes for Natural Hip

Parameter	Natural Hip
Young's Modulus, E_{bone}	300MPa
Poisson's Ratio, E_{bone}	Considered to be rigid
Young's Modulus, E_{cart}	16 MPa
Poisson's Ratio, v_{cart}	0.4
Reduced radius, R	0.35 m
Load, W	3.7 KN
Entraining velocity, u	0.0191 m/s
Viscosity, η	0.005Pas
Surface roughness, R_a	3.25 μm
Film thickness, $h_{\min}$	1.12 μm
Lambda ratio, λ	0.345
Lubrication regime	Boundary

CHAPTER TWO

2.0 Review of Hip Resurfacing Implants

2.1 History of Metal-on-Metal Articulation

Metal-on-Metal (MoM) hip joints were first used in the UK in ~ 1953 and Figure 2.1 shows some typical examples. This experience was therefore acquired ~ 40-50 years ago with McKee-Farrar MoM prostheses as the commanding interest [Clarke *et al*., 2005]. Many other implants became loose in the body and some exhibited wear whereas McKee did not observe any undesirable effects of metal wear on soft tissues or the surrounding bone. It is well documented that the loosening of several of these early MoM implants was frequently due to equatorial binding and this performance is commonly as a result of insufficient sphericity and clearance and probable elastic deformation under load. Aseptic loosening was, however, observed after 14-20 years for only ~18 % of McKee- Farrar joints requiring adjustment in a group of 511 implants. It has been argued [Amstutz; *et al*., 1996] that some early failures of the McKee-Farrar prostheses were also ascribed to high frictional torque due to equatorial binding and many failures were wrongly attributed to this cause [Clarke *et al*., 2005].

Figure 2.1. McKee-Farrar (left), Huggler (middle), and Müller (Right) Metal-on-Metal Total Hip Joint Replacement (THJR) prostheses [Jin *et al*., 2006].

More recently since early 90's, the metal-on-metal combination using cobalt-chromium-molybdenum alloy heads and cups has been widely reintroduced as the bearing couple for artificial hip joints, following on the long-term success of early McKee-Farrar implants. It has been illustrated from both simulator studies and clinical trials that correct manufacturing of the prosthesis leading to excellent sphericity, tolerances, and optimum radial clearance is the main reason for their success. Use of larger diameter bearings (>35-50mm diameter) and hip resurfacing prostheses have the advantages of increased range of motion and decreased incidence of dislocation for younger and more active patients. It has been shown experimentally via simulator studies [Jin *et al*., 2002] that an increase in the femoral head diameter from 16 to 28mm led to an increase in wear as also predicted from the classical Lancaster equation, but a further increase from 28 to 36mm resulted in the improved lubrication and formation of fluid film due to the elastohydrodynamic lubrication action. The clearance between the articulating components is size-dependent, i.e. the larger the diameter the higher the gap/clearance between the components. The range for the entire family of various diameters is from 90 to 200 microns of diametral clearance, with each bearing size having an optimized gap for maximum fluid film thickness. The diametral clearances between articulation components play a major role in the generation of wear debris which is probably the most influential factor in wear behaviour. The proper clearance is essential for entrapping the synovial fluid between the articulating surfaces. This fluid is largely responsible for separating the surfaces while the joint is in motion and, thereby, reducing wear. If the gap between components is too small or too large (see Figure 2.2) there will be a sharp increase in wear rates. Wrought and as-cast components with various clearances have been investigated [Liu *et al*., 2006] via hip simulator studies which strongly indicated that clearance plays a major role in generation of high friction and wear, and that wear appears to be relatively insensitive to changes in materials that have similar chemical compositions but different microstructures [Liu *et al*., 2006].

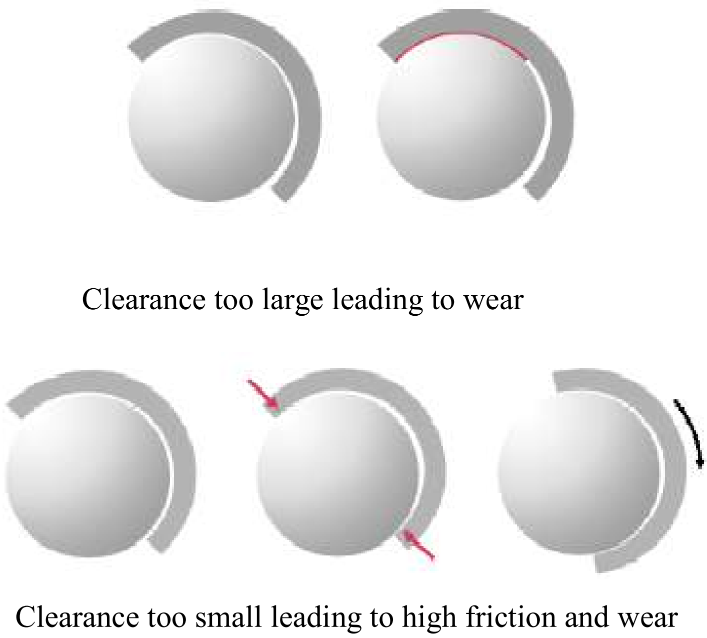


Figure 2.2. Effects of Clearance on wear and lubrication [Liu *et al*., 2006].

2.2 Hip Resurfacing Prostheses

First-generation metal-on-metal (MoM) hip resurfacing prostheses with larger bearing diameters than conventional total hip joint replacements were introduced for younger, more active and demanding patients since the early 1970's. However, in the 1970s, there were high rates of failure due to wear-debris-induced osteolysis as a result of insufficient wear resistance of the materials available in that time, loosening due to poor fixation methods and lack of standardization of operative and technical approaches altogether leading to the abandonment of the first-generation hip resurfacing prostheses. In contrast, the second generation with improved longevity and better resurfacing procedures (during which the femoral head is resurfaced and articulates against the acetabulum cup), improved manufacturing and bearing materials [McMinn *et al*, 1996; McMinn 2009] have been accepted widely as a better option for primary hip arthroplasty, particularly for young patients who otherwise most likely require a revision surgery with the conventional total hip joint replacements [Tipper *et al,* 2005]. There are many other advantages of using hip resurfacing arthroplasty including bone conservation, improved function due to retention of the femoral head and neck and better biomechanical restoration, decreased morbidity at the time of revision arthroplasty, reduced dislocation rates and stress-shielding, less infection, and reduced occurrences of thromboembolic phenomena (due to not using any tools/stems in the femur). Typical examples of such devices include the Birmingham Hip Resurfacing prostheses from Smith and Nephew Orthopaedic Ltd [Itayem *et al,* 2005], ASR from DePuy International, and DUROM from Zimmer. Following their promising short to medium term clinical results, second-generation metal-on-metal hip resurfacing prostheses have, therefore, been extensively introduced since the 1990's by almost all major orthopaedic companies. It is interesting to note that the introduction of the second-generation metal-on-metal hip resurfacing prostheses has been based on extensive laboratory simulator testing and design optimization leading to optimization of the diametral clearance and hence lower friction and wear as a result of improved lubricity. Initially, however, a larger diametral clearance of around 300μm was mainly adopted in the first-generation of MoM hip resurfacing prostheses. This was optimized for the second-generation hip resurfacing bearings to smaller clearances, typically between 100 and 150μm. Furthermore, the thickness of the acetabular cup wall in the second-generation prostheses is also smaller in order to reduce bone reaming and achieve greater bone stock saving. From simulator studies and compared with conventional 28mm diameter metal-on-metal total hip replacements, the MoM resurfacing bearings have shown lower wear rates. Lubrication has been recognized as an important factor in ensuring the remarkably low wear performance of these resurfacing bearings. It is generally accepted that the femoral head size is important, and this has been demonstrated in simulator studies [Feng *et al*., 2006]. Head diameter is becoming increasingly important as metal-on-metal resurfacing prostheses gain popularity with surgeons and younger patients. This type of prosthesis has the advantage of conserving bone on the femoral side, and is less invasive. Resurfacing prostheses cover the femoral head and therefore have large diameter femoral components, the average being in the region of 54 mm. Others [Tipper *et al*., 2005] have considered the effect of increasing head diameter on the wear of metal-on-metal hip prostheses. These authors tested 16, 22.225 and 28 mm diameter CoCrMo alloy femoral heads against CoCrMo alloy acetabular cups in a hip-joint simulator and found that with increasing head diameter, volumetric wear rate increased firstly and then decreased. Wear volumes were highest for the smallest diameter heads at 4.85 and 6.30 $mm^3/10^6$ cycles, respectively, for the 16 and 22.225 mm diameter heads. There was a marked decrease in wear exhibited by the 28 mm diameter heads, with bedding in wear of 1.60 $mm^3/10^6$ cycles and a steady-state wear of 0.54 $mm^3/10^6$ cycles. The effect of increasing head diameter on the wear of metal-on-metal bearings have been investigated [Hutchings *et al.*, 1992, Dowson and Jin, 2005] and thus testing 28 and 36 mm conventional hip prostheses in comparison to 54 mm diameter hip resurfacing prostheses in a hip-joint simulator was carried out. Stable running-in surfaces were established quickly as the head diameter increased from 28 to 36 mm and then to 54 mm. In agreement with previous studies, as head diameter increased wear volume decreased markedly, with steady-state values of 0.17 $mm^3/10^6$ cycles for the 54 mm diameter bearings. The bedding in wear rates for all prostheses were substantially higher at 3.23 $mm^3/10^6$ cycles for the 54 mm bearings. Note that diametral clearance is defined as the diameter of the acetabular cup minus the diameter of the femoral head (see Figure 2.3). There is a direct relationship between clearance and lubrication, and since metal-on-metal bearings are lubrication sensitive, clearance has a direct effect on wear. It has been reported [Dowson and Jin, 2005] that for both 36 and 54 mm bearings as diametral clearance increased, bedding in wear of the metal-on-metal components increased significantly. For the resurfacing components, those couples with smaller diametral clearances (83–129μm) with a head diameter of 54.5mm exhibited running in wear rates that were four-fold lower and steady-state wear rates that were two-fold lower, than those components with larger clearances (254–307μm) with a head diameter of 54mm. However, there appear to be an optimum band of clearance, which produces favourable wear rates. Farrar and colleagues were the first to show reducing wear rates with reducing clearance down to approximately 80μm with 28 mm metal-on-metal hip prostheses. However, reduction of diametral clearances to below 30μm caused wear to increase substantially. This was thought to be due to geometrical errors, which are inevitable with any manufactured part. Where small clearances approached the order of the cumulative geometrical errors, contacts may develop much closer to the equator and the possibility of a local negative clearance exists. These authors found that it was possible to simulate the wear of equatorial bearing devices, such as those described for the pre-1970 McKee Farrar and Ring prostheses, with modern metal-on-metal prostheses in a hip simulator by having negative or very low clearances. During testing these devices with low clearances reached approximately 20,000 cycles, and exhibited extremely high wear, before seizing completely [Tipper *et al.*, 2005].

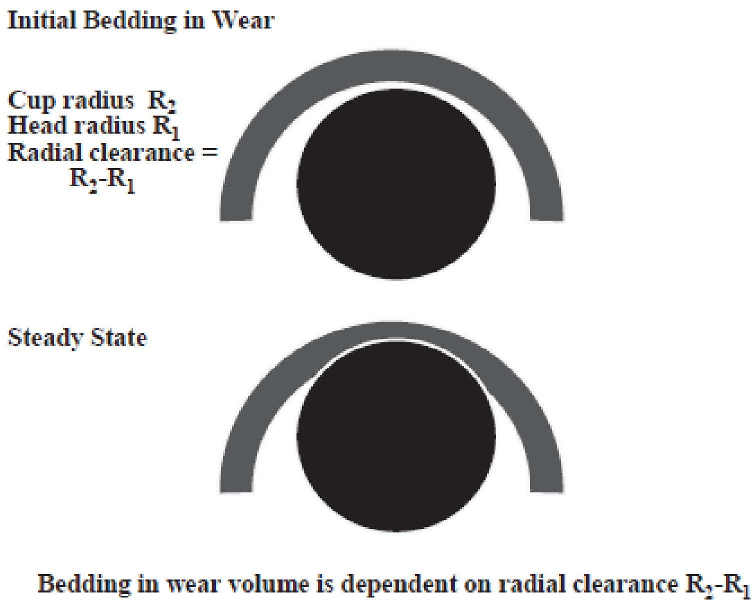


Figure 2.3. The effect of radial clearance on bedding in and steady state wear [Tipper *et al.*, 2005].

In summary, therefore, diametral clearance is one of the most important geometrical features of metal-on-metal hip replacements, and if the clearance is too small and the sphericity as well as surface finish is inadequate, equatorial binding can occur owing to deformation under load. As discussed earlier, it is widely believed that this accounted for most failed McKee–Farrar implants after only few years, while some survived and performed well for 20 or 30 years. It is interesting to note that McKee wrote that it is very important that the two components be correctly lapped in, so that they articulate freely without any binding, and they are paired and numbered to ensure this accuracy of fit. It is also clear now that mixed lubrication has been shown to control the tribological behaviour of such implants that the running-in wear and the long term steady state wear rates will be related to the ability of effective lubricating films to support some of the applied load. This implies that the severity of solid contact and the wear process will be minimized if the thickest possible effective lubricating films can be generated. Geometrical factors, such as diameter and clearance, should thus be optimized to maximize the effective film thickness and to minimize wear in metal-on-metal joints.

2.2.1 Effect of joint diameter on lubrication

It has been shown [Dowson *et al*., 2006] that the head diameter plays a major role in determining the mode of lubrication, the volumetric wear and wear rate in metal-on-metal hip replacements. The same authors have investigated this effect extensively using hip joint simulators under conditions of simulated walking. They have reported that for relatively small heads of diameters 16 and 22.225 mm, boundary lubrication prevails such that the applied load is carried directly by metal-to-metal contact. In this regime, the volumetric wear increased linearly with increasing head diameter as the sliding distance per unit time increased. For heads of diameter 28mm and greater, a mixed mode of lubrication prevailed. The proportion of load supported by fluid-film lubrication grew as the head diameter and hence the entraining velocity increased. This resulted in a dramatic reduction in steady-state wear until wear rates of the order of 0.1mm^3 per 10^6 cycles were achieved. This dramatic evidence of either boundary or mixed lubrication in metal-on-metal hip joints, depending on the head diameter, reflects the findings from studies of other engineering journal bearings in the UK, USA and in Germany. A representation of the well-known Stribeck curve showing the boundary, mixed and fluid-film lubrication regimes is shown in Figure 5 and without any exceptions these researchers have measured friction using joint simulators and found it useful to present their data in the form of Stribeck diagrams for a wide range of prostheses, including metal-on-metal, UHMWPE-on-metal, and ceramic-on-ceramic joints and their combinations.

2.3 Lubrication mode in metal-on-metal hip joints

Natural synovial joints such as hips and knees are remarkable bearings. These bearings are expected to function in the human body for a lifetime while transmitting large dynamic loads and yet accommodating a wide range of movements. However, diseases such as osteoarthritis and rheumatoid arthritis often require these natural bearings to be replaced by artificial joint prostheses. Total joint replacement has been the most successful surgical treatment for hip joint diseases in the last 40-50 years. Currently, about 70,000 hip joint replacements are carried out every year in the UK alone. One of the best approaches is to promote lubrication, and hence minimize wear, with metal-on-metal and other material combinations. The essential features of the three regimes of lubrication are shown in Figure 1.13. Furthermore, the purpose of presenting this section might be to show the lubrication regime generated between articulating surfaces using various biological lubricants.

Its significance lies not only in the indication of representative values of the coefficient of friction in the various lubrication regimes but also in the trends in friction as the Sommerfeld number is varied (see Figure 1.13). However, since such bearings do wear over time, it is clear that any fluid-film lubrication developed between the opposing surfaces must break down at some stage. Protection of the cartilage surfaces then depends upon the efficacy of boundary lubrication, and much effort is now made on the identification of the effective constituents of synovial fluid that act as boundary lubricants, with attention being focused upon proteinaceous matter. While boundary lubrication mechanisms have to be effective in synovial joints [Scholes *et al,* 2000; Scholes *et al,* 2006] if they are to survive the totality of lifetime activity, laboratory measurements of friction and theoretical analyses confirm that in normal gait there is every possibility of effective fluid-film lubrication. A major factor in this process is the elastic deformation of the articular cartilage under load. Metal-on-metal total hip replacement has been shown experimentally and theoretically to operate in the mixed and maybe fluid-film regimes during normal gait [Jin 2001; Jin *et al,* 2006]. If fluid-film action is capable of supporting some of the applied load in a mixed lubrication regime, the severity of contact between opposing asperities and hence friction and wear will be substantially reduced.

In summary, therefore, Metal-on-metal hip prostheses can be lubricated in three ways: boundary lubrication, mixed lubrication and full fluid-film lubrication, either alone or in combination. Lubrication is generally related to friction and wear and hence can play an important role in wear of particle generation in metal-on-metal bearings. From equation (1.9), it becomes clear that [Jin *et al,* 2006] lubrication is dependent upon the viscosity of the lubricant, the sliding speed, the diameter of the femoral head, clearance and surface roughness of the components. One of the experimental methods of studying lubrication is through measuring friction. The coefficient of friction is commonly plotted against the Sommerfeld parameter, which is the product of the velocity, viscosity of the lubricant and the diameter of the femoral head, divided by the load. This type of plot is referred to as "Stribeck curve" as shown in chapter one (Figure 1.15). The trend of the curve indicates the modes of lubrication. The effect of head diameter on lubrication has been investigated [Tipper *et al*., 2005], where head diameters of 16 and 22.225mm were shown to have contact between the bearing surfaces at all times during the simulator tests, and hence a boundary lubrication regime was found to be prevalent. Alternatively, a mixed lubrication regime involving significant asperity contact may have prevailed. As head diameter increased to 28 mm a mixed lubrication regime was found to be prevalent; however, as only limited asperity contact occurred occasional fluid-film lubrication was indicated. As further increases in head diameter occurred to 36mm [Dowson *et al,* 2004] and beyond, the lubricating film alleviates metallic contact between the articulating surfaces and the volumetric running in wear and steady state wear fall dramatically. As head diameter increases, the articulation is more likely to promote fluid-film lubrication [Dowson *et al,* 2006] and the benefits to the joints are subsequently seen in the wear characteristics. It was also shown [Tipper *et al*., 2005] that by increasing head diameter to 54mm, the volumetric wear of metal-on-metal articulations decreased, but if clearance was optimised further reductions in wear could be achieved as shown in 2.4. For the older designs of hip replacement prostheses, the λ ratio was between one and two, indicating a mixed lubrication regime (see Figure 2.4). However, as prosthesis design was improved and clearances optimised, the λ ratio approached three, indicating that full fluid-film lubrication was possible in these newer devices. These studies clearly demonstrate the benefits of optimising the design of metal-on-metal hip prostheses, and that improvements in design such as optimising clearances for surface replacement prostheses and improving surface finishes of all components can have a significant effect on the wear of metal-on-metal devices.

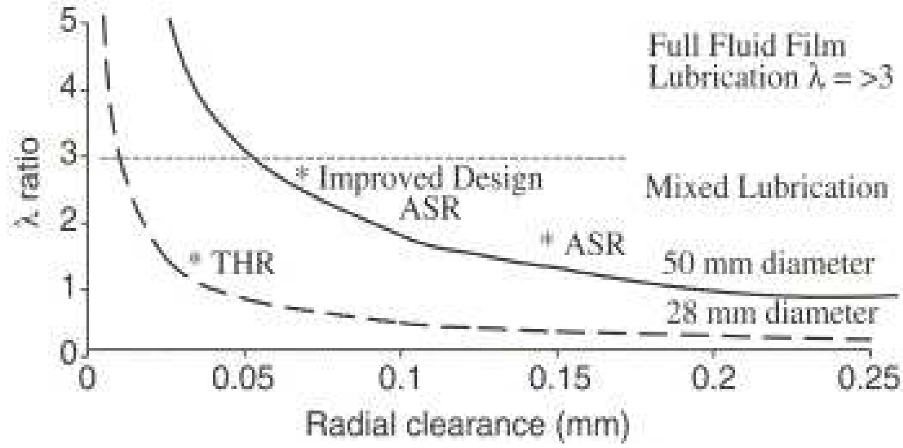


Figure 2.4. The effect of radial clearance (half of diametral clearance) upon lubrication and λ ratio in metal-on-metal total hip implants and resurfacing prostheses (ASR, DePuy Int.) [Tipper *et al*., 2005].

Experimental evidence from joint simulators in which friction tests and wear measurements are made helps to indicate the mode of lubrication in metal-on-metal prostheses. Alternatively, direct measurement of the existence, or otherwise, of a film of protective lubricant separating the head from the cup can be attempted. All these approaches have yielded evidence that relatively high friction and wear occur in an initial running-in period of about one million cycles, followed by a relatively low friction and wear in a steady state condition. It further appears that, under the severe bearing operating conditions experienced in normal gait, mixed lubricious is generally the main mode of operation. It is thus imperative to design and manufacture metal-on-metal hip joint replacements capable of operating with maximum benefit from the fluid-film lubrication element of the mixed lubrication process. The bearing operating conditions encountered in replacement hip joints, even in steady walking, place them in the category of dynamically loaded bearings. They are essentially reciprocating bearings in which the load varies dynamically, typically with a double peak and the loads vary from a few tens or hundreds of Newton's in the swing phase to several thousands of Newton's in the stance phase [Dowson and Jin, 2005].

2.3.1 Numerical solution to the problem of fluid film lubrication in metal-on-metal hip replacements

A full numerical solution to the lubrication problem of artificial hip joints requires simultaneous solutions to the Reynolds equation of fluid-film lubrication and the elasticity equation considering the full anatomical structure of bone under transient walking conditions of load and speed. However, such a task in which spherical coordinates have to be adopted in the solution of the Reynolds equation is complex and time consuming, and only a limited number of solutions have been obtained, often with several simplifications. The appropriate form of Reynolds equation in spherical coordinates (θ, φ) is shown in equation (1) under steady state operating conditions for the model shown in Figure 2.5a.

$$\sin\theta \frac{\partial}{\partial\theta}(h^3 \sin\theta \frac{\partial p}{\partial\theta}) + \frac{\partial}{\partial\varphi}(h^3 \frac{\partial p}{\partial\varphi}) = 6\eta(R_1)^2 \omega \sin^2\theta \frac{\partial h}{\partial\varphi} \qquad (1)$$

As a first approximation, the ball-in-socket model of a hip joint shown in Figure 2.5a was represented by a geometrically equivalent sphere-on-plane configuration as shown in Figure 2.5b. The effective radius of the equivalent sphere, (*R)*, was determined from the diameter of the femoral head, (*d)*, and the diametral clearance, (C_d), using the following equation:

$$R = \frac{d(d + c_d)}{2c_d} = \frac{d}{2}(1 + \frac{d}{c_d}) \qquad (2)$$

The entraining velocity was calculated from:

$$u = \frac{\omega d}{4} \qquad (3)$$

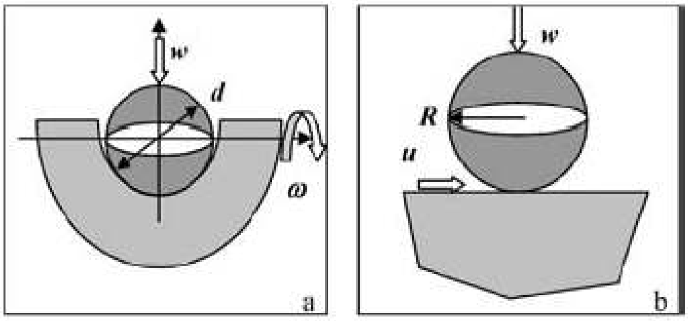


Figure 2.5. Models for lubrication analysis of hip implants: (a) ball-in-socket; (b) equivalent ball-on-plane geometry with effective radius *R* [Dowson and Jin, 2005].

Where ω is the angular velocity representing flexion and extension. The materials currently used in the manufacture of metal-on-metal hip replacements are usually cobalt–chrome-molybdenum alloys. If the radius of the contact area under load is smaller than the cup wall thickness, it is acceptable to treat both the femoral head and the cup as semi-infinite solids for the purpose of the calculation of the elastic deformation of the bearing surfaces. The lubricant in healthy natural joints is synovial fluid, but in total joint replacements it is periprosthetic synovial fluid, similar to that obtained from patients with osteoarthritis. The lubricant used for simulator testing is usually bovine serum diluted to various concentrations; typically at 25 per cent BS with 75% distilled water. All these biological lubricants [Dowson 2006] exhibit non-Newtonian characteristics of shear thinning, i.e. pseudoplastic flow behaviour, particularly under relatively low shear rates. Under the very high shear rates and the relatively low contact pressures experienced in various forms of hip implants, the viscosity of the fluid varies very little, and hence it is reasonable to represent the lubricant as an isoviscous, incompressible, low-viscosity Newtonian fluid. Typical viscosities of 0.0025 Pas and 0.0009 Pas have been suggested for periprosthetic and 25% synovial fluid, respectively. The load and speed experienced in hip joints during walking are transient in nature, not only in magnitude but also in direction. However, the major load component is roughly in the vertical direction [Dowson and Jin, 2005], while the sliding and entraining speeds arise around a horizontal axis associated with flexion–extension, as schematically shown in Figures 2.5 and 2.6 [Dowson *et al*., 2005]. Figure 2.6 shows the transient variation in load and speed during one walking cycle. An average load of between 1346 N (average for a complete cycle) and 2500N (average in the stance phase), equivalent to about 3 times body weight of 750N, and an average resultant angular velocity of about 1.5 rad/s have been suggested for quasi steady state lubrication analysis under in vivo conditions. The local elastic deformations of the metal components in the load-bearing regions of metal-on-metal joints can readily be shown by Hertzian contact theory to be of micron proportions. It will also be shown that the film thicknesses in metal-on-metal joints are calculated to be a few tens of nanometres. The elastic deformations thus greatly exceed the film thicknesses, and fluid-film lubrication will be elastohydrodynamic in nature. Since it has been argued above that the lubricant essentially displays a constant viscosity under the conditions encountered in metal-on-metal implants, an isoviscous elastic analysis is required. Under the above assumptions, the minimum film thickness formula equation (4) developed by Hamrock and Dowson [Dowson *et al*., 2005] was used to estimate the lubricant film thickness in metal-on-metal hip implant where the equivalent elastic modulus for the present problem becomes $E^{'} = \frac{E}{(1-v^2)}$.

$$\frac{h_{\min}}{R} = 2.8\left(\frac{\eta u}{E^{'}R}\right)^{0.65}\left(\frac{w}{E^{'}R^{2}}\right)^{-0.21} \qquad (4)$$

The elastic modulus for the cobalt– chrome–molybdenum alloy is about 210 GPa and Poisson's ratio is 0.3 [Dowson *et al*., 2005].

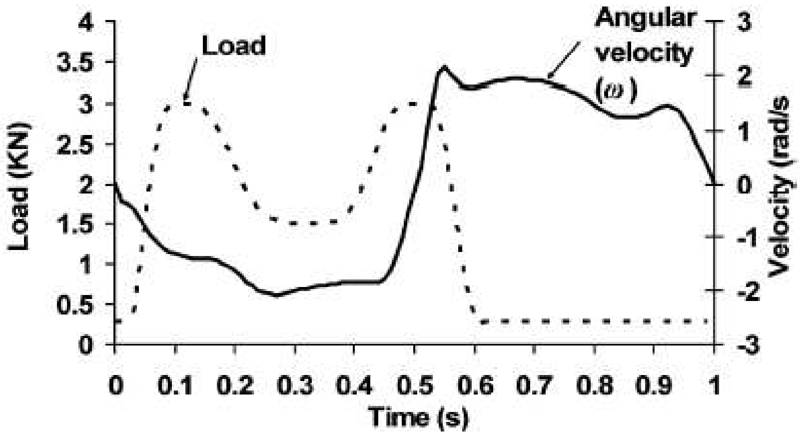


Figure 2.6. Typical variation in transient load and angular velocity with time in hip joints during walking [Dowson *et al.*, 2005].

2.3.2 Minimum film thickness predictions

A full numerical solution to the elastohydrodynamic lubrication problem for a ball-in-socket configuration representative of the conditions shown in Figure 2.6 and for a joint of 28mm diameter and 120μm diametral clearance lubricated by a fluid of viscosity 0.01 Pas is shown in Figure 2.7. It is evident that the minimum film thickness fluctuations throughout the cycle are modest for such extreme variations in speed and load. This is a consequence of the powerful squeeze-film action, which restrains the film from exhibiting rapid and major changes under quite severe dynamic conditions. The quasi-static minimum film thickness computed for mean values of load and speed throughout the cycle is also shown in Figure 2.7. These results indicate that a simple application of the isoviscous/elastic EHL equation (4) provides a fair prediction of the effective film thickness under full dynamic conditions. Further comparisons between the predictions of equation (4) and the minimum film thicknesses revealed by full numerical solutions to the elastohydrodynamic problem were then made for the operating conditions in the three typical metal-on-metal hip implants documented in Table 2.1.

Table 2.1. Three typical metal-on-metal hip implants considered for the purpose of comparison [Dowson and Jin, 2005].

Reference	Diameter (mm)	Diameteral Clearance (μm)	Cup Wall Thickness (mm)
Jagatia and Jin	28	60	9.5
Liu *et al.*	28	120	7
Udofia and Jin	50	300	4.8

Various loads, angular velocities, and viscosities were selected, and the minimum film thicknesses revealed by solutions for eight cases are shown in Table 2.2.

Table 2.2. Comparison of the predicted minimum film thickness between full numerical solutions and the Hamrock–Dowson formula [Dowson and Jin, 2005].

Reference	ω (rad/s)	η (Pas)	*w* (N)	Full Numerical Solutions	Equation (4)
Jagatia and Jin	2	0.01	500 1500 2500	0.09 0.06 0.04	0.1 0.08 0.072
Liu *et al,*	2	0.02 0.05 0.1	2500 2500 2500	0.03 0.1 0.17	0.066 0.12 0.19
Udofia and Jin	2	0.2 0.5	2500 2500	0.6 0.1	0.52 0.95

This wider range of comparisons again confirms the merit of equation (4) for a wide range of conditions. The values of the viscosities considered ranged from 0.5 Pas down to 0.01Pas. This range is higher than the value of about 0.0009 Pas thought to be representative of the 25 per cent bovine serum used in many hip joint simulators and the value of 0.0025Pas proposed by others [Dowson and Jin 2005, Dowson *et al.*, 2006] for periprosthetic fluid. However, there is no indication in Table 2.2 that the predictions by the simple equation (4) become less accurate as the viscosity decreases. The numerical analysis becomes more difficult and time consuming when such low-viscosity fluids are considered and therefore the higher range of viscosity values noted above was adopted. To provide realistic film thickness predictions in total hip replacements for example, if the periprosthetic fluid viscosity of 0.0025Pas is adopted rather than the value of 0.01Pas used, the calculated minimum film thickness for a mean load of 1500 N would be 24nm rather than 60nm. Likewise, for joint simulators using 25% bovine serum the predicted film thickness would be 12.5nm rather than the 60nm associated with a viscosity of 0.01Pas. Numerical procedures are now being developed to deal with such low-viscosity conditions.

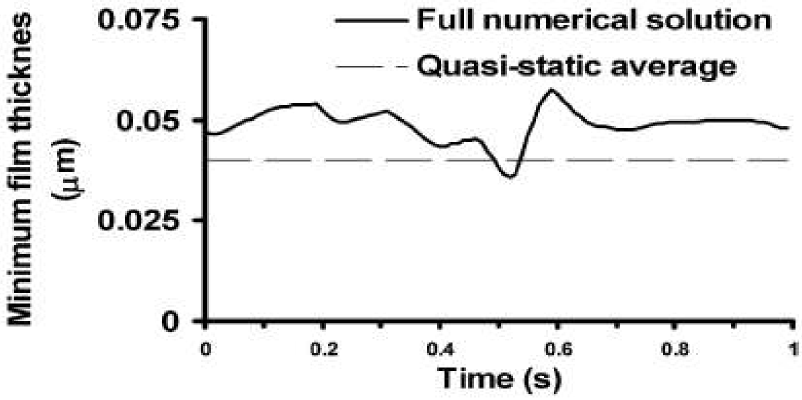


Figure 2.7. Comparison of the quasi-static average minimum film thickness with the full transient numerical solution (C_d) =120µm, η= 0.01Pas [Dowson and Jin 2005].

2.3.3 Mixed Lubrication Model

The total load experienced between the two bearing surfaces in the mixed lubrication regime is shared between the asperity contact and the lubricant as shown schematically in Figure 2.8. It is generally expected that the asperity contacts contribute mainly to the wear of the bearing surfaces. The interaction between the asperities where R_1 and R_2 are the surface roughness, then and $R= (R_1^{\ 2} + R_2^{\ 2})^{1/2}$ the average lubricant film thickness, h, can be characterized by the well-known film thickness ratio (λ) as defined by equation (2) earlier.

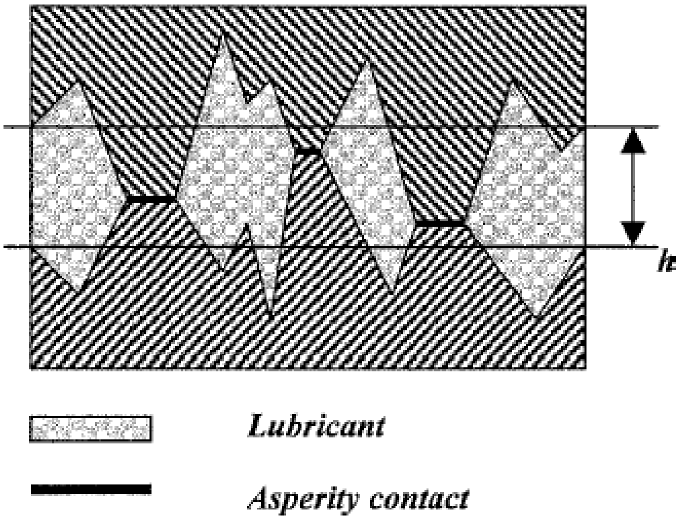


Figure 2.8. Simple schematic diagram for load sharing between asperities and fluid film in a mixed lubrication regime [Jin *et al.*, 2001].

The theoretical determination of f (λ) would require a coupled solution to both the lubrication equation under the elastohydrodynamic condition and the asperity contact model. Such an analysis can be very difficult to perform for artificial hip joint replacements. The following assumptions were made in this preliminary analysis for metal-on-metal hip prostheses employing cobalt– chrome alloys.

1. Only steady state conditions with an average load and an average speed were considered.
2. The lubricant (synovial fluid) was assumed to be isoviscous, incompressible and Newtonian.
3. The average lubricant film thickness in the mixed lubrication regime *(h)* was estimated by using the same methodology outlined earlier [Jin 2001]. For an assumed ratio of the load due to the elastohydrodynamic action to the total load, *(γ),* the average lubricant film thickness for the rough surface was estimated by modifying the equivalent elastic modulus and load, and then

using the 'smooth' surface central film thickness formulae developed by Hamrock and Dowson [Dowson *et al*., 2005] as follows:

$$h = 5.29R[\frac{\eta u}{(E'/\gamma)R}]^{0.665}[\frac{w/\gamma}{(E'/\gamma)R^2}]^{-0.22} \text{..................................}(7)$$

Where the effective radius *R* was determined from the femoral head radius R_1 and the radial clearance (C_d), $R = (R_1 + C_d)\frac{R_1}{C_d}$ and the entraining velocity *u* was calculated from the angular velocity *ω* and the femoral head radius, $u = \frac{wR_1}{2}$. The equivalent elastic modulus *E'* was given by $\frac{E}{(1-v^2)}$ for metal-on-metal combinations (*E* and *υ* are the elastic modulus and Poisson's ratio, respectively for cobalt–chrome alloys).

4. The asperity contact load was calculated from the model developed for general engineering surfaces by Greenwood, Tripp, Patir and Cheng [Jin *et al*., 2001]. Also, this calculation indicates that load ratio (*γ)* depends on the reflection between asperity contacts and lubricant film thickness while under boundary lubrication regime load ratio increases with the increased femoral head radius. This load ratio in mixed lubrication regime toward full fluid film decreased when diametral clearance increased. For a given ratio, the asperity contact pressure was estimated from the following equation in presence of synovial fluid lubricant:

$$P_a = 1.323\, X 10^{-7}(4 - \frac{h}{R})^{6.804} E' \rightarrow \frac{h}{R} \prec 4 \text{.......... ....}(8)$$

$$Pa = 0 \rightarrow \frac{h}{R} \geq 4$$

The load due to asperity contacts (w_a) was found by the integration of the asperity contact pressure over the contact area determined from the Hertzian contact analysis:

$$w_a = p_a \pi (\frac{3wR}{2E'})^{2/3} \text{.......... ....(9)}$$

The input parameters required for the mixed lubrication analysis for a nominal metal-on-metal hip joint replacement are given in Table 2.3 and the radius of the femoral head was varied between 28 and 16 mm, to cover the whole lubrication regime. In addition, two more cases were considered by either reducing the R.M.S. surface roughness from 0.02 to 0.01 μm or increasing the diametral clearance from 50 to 80 μm [Jin *et al*., 2001].

Table 2.3. Parameters for mixed lubrication analysis of a nominal metal-on-metal hip prosthesis [Jin *et al*., 2001].

Elastic modulus for cobalt-chrome alloys (MPa)	210000
Poisson s' ratio for cobalt-chrome alloys	0.3
Viscosity for synovial fluid or serum (Pas)	0.005
Average load (N)	2500
Average angular velocity (rad/s)	1
Nominal R.M.S surface roughness for cobalt alloy (µm)	0.02
Nominal diametral clearance (µm)	50

2.4 Wear performance

The general form of the wear or penetration characteristic of metal-on-metal total hip replacements shown in Figure 2.9 has been widely reported in the literature. There are two distinctive regions of the wear trace. Initially the femoral and acetabular components show a relatively rapid but decreasing wear rate over the first $(1–2) \times 10^6$ cycles in a hip joint simulator, or for 1 or 2 years *in vivo*. This feature is variously attributed to a '*bedding-in*' or '*running-in*' process. Once this process has been completed, the rate of wear becomes reasonably steady and generally relatively small. These two distinctive regions are shown in Figure 2.9 a. Clinical assessment of wear in metal-on-metal joints has to rely upon successive measurements of the very small penetrations of the head into the cup. Furthermore, it is not uncommon to record wear or penetration rates as the ratio of the penetration *(p)* at any time *(t)* following implantation to the post-operative time *(t)*. This measure yields neither the initial '*running-in*' wear nor the subsequent '*steady state*' wear rate. It has frequently been reported that this snapshot of penetration rate decreases with increasing time of implantation, but this often leads to confusion and it does not imply a decreasing real

Steady state wear rate. This point is demonstrated in Figure 2.9 b, where the measure of (p/t) derived from Figure 2.9 a is clearly much larger than the steady state wear rate until the implantation period achieves very large times. The analysis of wear rates presented later is thus related to the initial '*running-in*' wear V_2 and the long-term measure of '*steady state*' wear [Dowson *et al*., 2006].

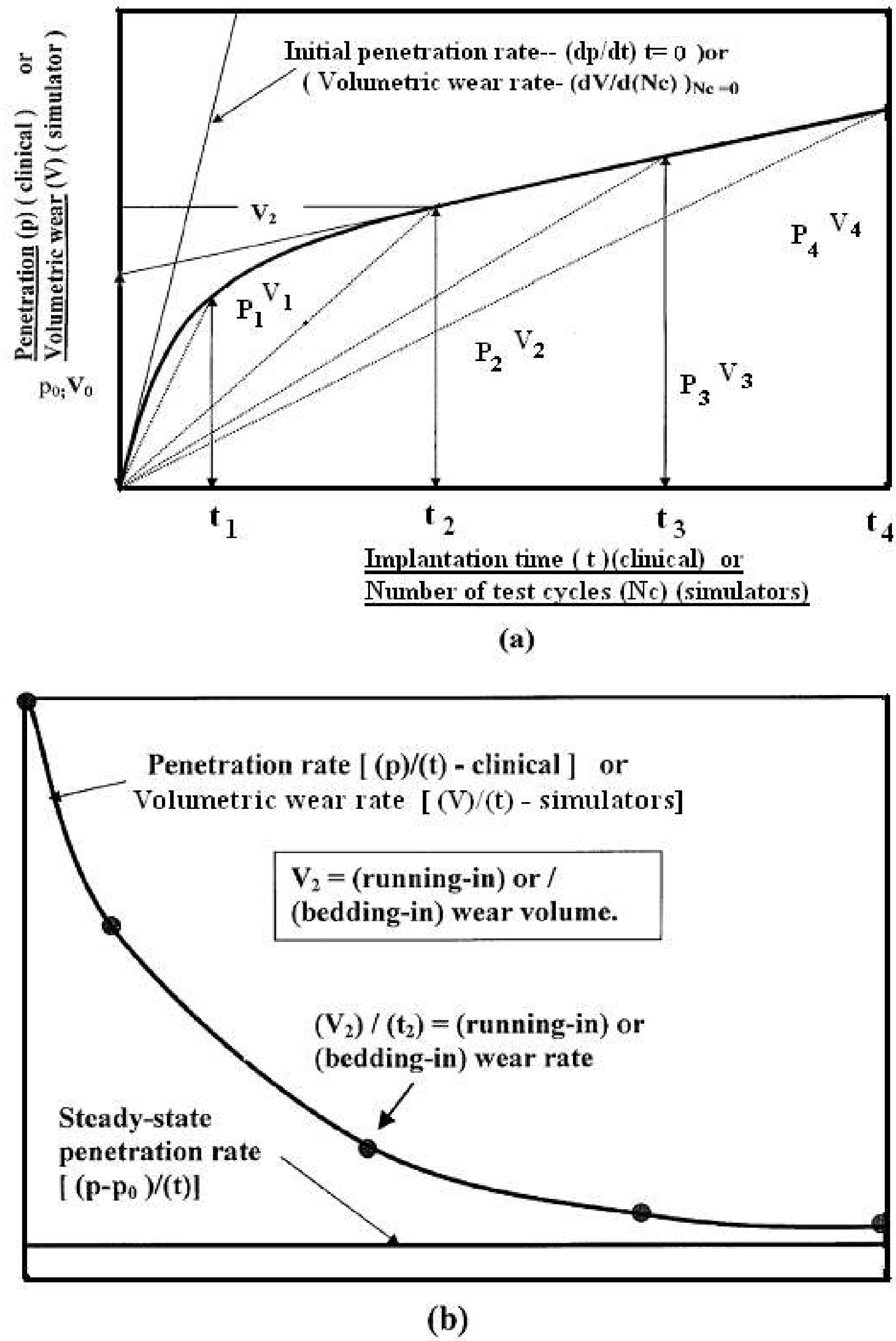


Figure 2.9. Representations of metal-on-metal hip replacement wear characteristics: (a) volumetric wear (or penetration) versus time; (b) volumetric wear rate (or penetration rate) versus time [Amstutz and Grigoris 1996].

A low wear rate is believed to be critical for extending the implant life of a prosthetic joint, and wear volumes produced by metal-on-metal articulations have been estimated to be 40–100 times lower than metal-on-polyethylene bearings [Hernandez *et al*., 2005]. The wear of metal-on-metal prostheses is known to be highly dependent upon the materials, tribological design and finishing technique. Clinical studies of retrieved first and second-generation metal-on-metal hip prostheses have shown linear penetrations of approximately 5μm/year [Hernandez *et al*., 2005]. This is equivalent to a wear volume of approximately 1 mm^3/year, two orders of magnitude lower than conventional polyethylene acetabular cups. The wear of hard-on-hard bearings such as metal-on-metal hip prostheses has two distinct phases. Initial elevated "bedding in" wear period occurs during the first million cycles or first year in vivo. This is followed by a lower steady-state wear period once the bearing surfaces have been subjected to the self-polishing action of the metal wear particles, which may act as a solid-phase lubricant. Hip-joint simulators have generally shown steady-state wear rates to be lower than those reported in vivo and that wear simulators represent ideal articulation conditions during the walking cycle.

Currently, the factors that influence wear such as particle size, concentration, and subsequent levels of ions released from metal-on-metal hip bearings are not well understood. Some studies show an influence (positive and negative) in ion levels with time and exercise, while others show an influence (positive and negative) of head diameter. In addition, a review of 12 clinical and laboratory debris studies showed a mean (Co–Cr–Mo) particle diameter of 79nm and 45nm, respectively (see Table 2.4), therefore suggesting possible differences [Bowsher *et al*., 2005]. It was also noted that laboratory wear studies did not generate the larger particles observed clinically, greater than 1000 nm. Reasons for this apparent difference be it a lack of severe conditions in laboratory testing, differences in bearing design parameters, measurement or imaging techniques, or other has yet to be established. The use of metal-on-metal hip resurfacing prostheses has been recommended for younger and more active patients with advanced hip disease. These active patients, however, are likely to be at greater risk of ion release from metal-on-metal bearings than less active patients.

Table 2.4. Summary of average Co–Cr wear particle sizes from metal-on-metal hip bearings, generated either *in vivo* or in a hip joint simulator (ranked chronologically in each group) [Bowsher *et al*., 2005].

Study	Particle Origin	MOM head diameter (mm)	Mean wear particle diameter(nm)	Large particles(nm)
Shahgaldi *et al.*, 1995	Retrieved tissue	-	-	4000
Soh *et al.*, 1996	Retrieved tissue	-	-	4000
Doorn *et al.*, 1998	Retrieved tissue	45	60	-
Doorn *et al.*, 1998	Retrieved tissue	28	120	-
Catelas *et al.*, 2004	Retrieved tissue	-	57	-
Overall mean size(nm)			79	-
Firkins *et al.*, 1999	Hip simulator	28	25-36	-
Tipper *et al.*, 1999	Reciprocating, Pin-on-disk	-	60-90	-
Fisher *et al.*, 2000	Hip simulator	28	30±5	-
Catelas *et al.*, 2003	Hip simulator	-	52	-
Catelas *et al.*, 2004	Hip simulator	28	43	-
Brown *et al.*, 2004	Hip simulator	28	43	100
Williams *et al.*, 2004 Overall mean size (nm)	Hip simulator	28	<40 45	-

To date, the performance of metal-on-metal bearings under 'severe' conditions is not well understood. Using a more severe-wear condition, such as intermittent loading and micro separation, studies have successfully created higher metal-on-metal wear, with most studies [Semlitsch *et al,* 1997; Dowson 2001] reporting similar maximum wear rates of approximately

2.0 mm³ per 10^6 cycles for 28 mm Co–Cr–Mo hip bearings. These studies are certainly an improvement for simulating highly active patients compared with normal-walking cycles, but they still do not explain the high clinical wear rates seen for 28 mm bearings (approximately 10 mm³/year). In an effort to create higher bearing wear and improved discrimination in behaviour, the same authors [Bowsher *et al*., 2005] have introduced fast-jogging cycles. So far, this model represents the most severe hip simulator-testing regime published; creating between twofold and fourfold greater wear than other studies using undamaged bearings (see Table 2.5). Therefore, this model could be used to represent high-demand patients in accelerated-wear simulations. However, the influence of this severe-wear protocol on wear particle sizes and surface areas is unknown [Bowsher *et al*., 2005].

Table 2.5. Summary of wear rates for 28 mm Co–Cr–Mo hip bearings from hip simulator studies under 'severe'-wear conditions (ranked chronologically) [Bowsher *et al*., 2005].

Study	MOM wear model type	Condition of wear surfaces	MOM wear rate (mm³ per 10^6cycles)
Chan *et al.*, 1999	Intermittent loading	Pristine	Mean, 1.0; Maximum,1.2
Flrkins et al,2001	Eccentric wear paths	Pristine	Mean,1.64;Maximum,1.8
Butterfield *et al.*, 2002	Microseparation	Pristine	Mean, 1.5; Maximum,2.0
Williams *et al.*, 2004	High swing-phase load	Pristine	Mean, 0.6; Maximum,2.0
Lu *et al.*, 1996	150 third-body particles	Rough	1.8 and 15
Bowsher *et al.*, 2003	Fast-jogging cycles	Pristine	Mean, 4.0; Maximum, 8.0
Liao *et al.*, 2004			Mean, 0.4
Maximum wear rate			8mm³ per 10^6 cycle

2.4.1 Influence of elastohydrodynamic film thickness upon steady state wear rate

Simulators are now widely used to ascertain the wear characteristics of metal-on-metal hip joint replacements. A simulated walking cycle is generally applied and volumetric wear is deduced from periodic measurements of dimension, or weight changes in both the cup and the head. The results follow the form shown in Figure 2.9a. The wear tests are generally of long duration, typically extending over $5X10^6$ cycles. This is necessary to provide reliable measurements of the very low steady state wear rates achieved after the running-in period. The latter generally extends from about $0.5X10^6$ cycles to $(2-3)X10^6$ cycles, depending upon the materials of construction and the joint geometry. A representative period for running-in is roughly 10^6 cycles, equivalent to about 1 year's service in the body. A compilation of data from measurements of steady state wear rates has been reported earlier. At the time, some 70-test results were available from five laboratories in the UK, Canada, and the USA. More data have now become available and the analysis has been extended to include 103 data points for steady state wear rates recorded in eight laboratories. The results are shown in Figure 2.10 [Dowson *et al*., 2006].

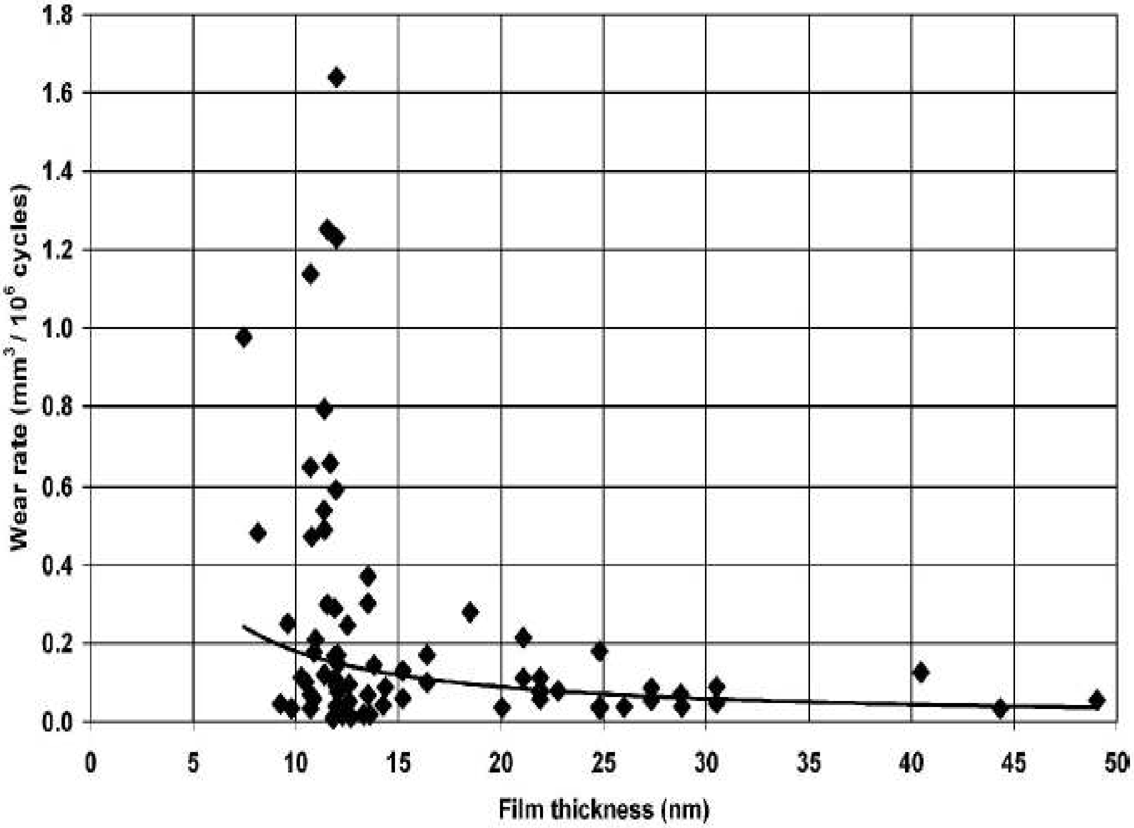


Figure 2.10. Steady state wear rates in the film thickness range 7–60 nm [Dowson *et al*., 2006].

Two distinct regions are evident: a high wear rate at small values of film thickness which decreases rapidly as the film thickness increases from about 7 to 12nm and a sensibly steady but low wear rate for film thicknesses in excess of about 20–25nm. The low average wear rate in the film thickness range 20–60nm was found to be about 0.07 mm^3 per 10^6 cycles, while the wear rate for a film thickness of 7nm was more than 20 times greater than this. Since the steady state wear rate varies so little for film thicknesses in excess of about 20–25nm, the wear rate for these and higher film thicknesses can be approximated by steady state volumetric wear rate (20–60μm) ≈0.068 mm^3 per 10^6 cycles using equation (11).

If relatively small effective film thicknesses are calculated, it may be deemed desirable to take account of the associated variation in the steady state wear rate. The trend line for the relatively low steady state wear rates measured after running in is not generally statistically significant, but if $h_{\min}$ is expressed in nanometres, it can be written as steady state volumetric wear rate (7–50nm) as follows:

$$\text{Steady State Volumetric Wear Rate} = \frac{1.871}{[h_{\min}(nm)]^{1.016}} \, mm^3 \, per \, 10^6 \, cycles \; .....(11)$$

2.4.2 Influence of elastohydrodynamic film thickness upon running-in wear

Initially only the steady state wear rates were analysed, but since most of the wear took place during running in, the analysis has been expanded to cover the volumetric wear in this important initial period. The derived results covering the film thickness range 0–140 nm are shown in Figure 2.11 [Dowson *et al*., 2006].

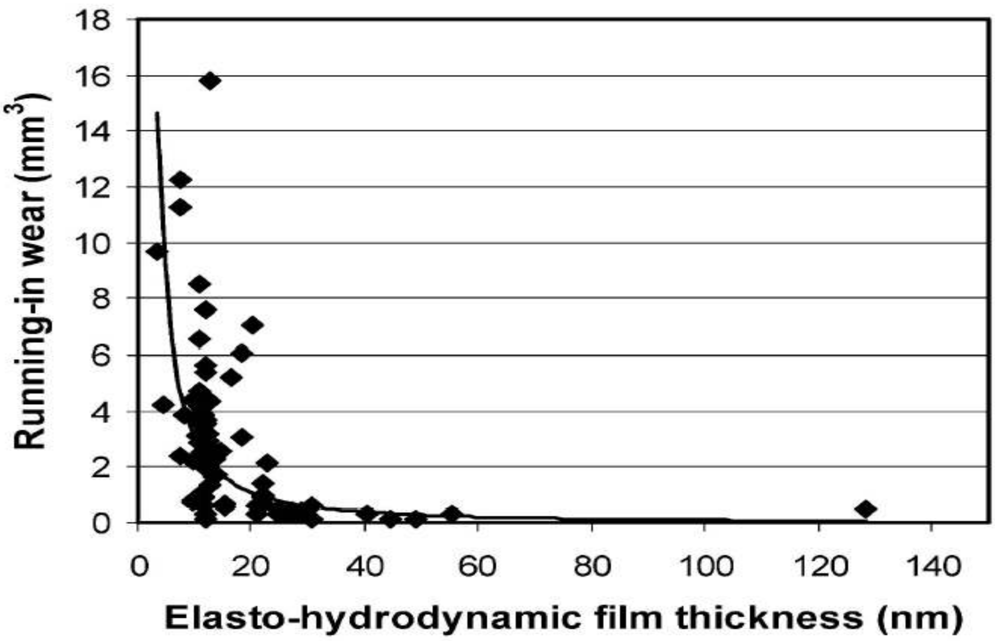


Figure 2.11. Running-in wear versus Elasto-hydrodynamic film thickness (predicted film thicknesses 0–120 nm) [Dowson *et al*., 2006].

The results showed in Figure 2.11 exhibit the same characteristics for running-in volumetric wear (mm^3) as for steady state volumetric wear rate (mm^3 per 10^6 cycles) as evident in Figure 2.11. The best fit power relationship (R_2 =0.459) over the film thickness range 0–140 nm in Figure 2.11 was found to be running-in volumetric wear

$$V(mm^3) = \frac{93.97}{[h_{\min}(nm)]^{1.492}} \quad \text{........(12)}$$

It is evident that the steady state wear regime would have to operate for many years before a volume of wear equivalent to the running-in wear could be generated. For example, for a film thickness of 20 nm the running-in wear equation (12) is about 0.9 mm^3, while the steady state wear rate subsequently achieved is only about 0.08 mm^3 per 10^6 cycles using equation (11). Figures 2.10 and 2.11 strongly suggest that both the steady state wear rate and the running in wear are intimately linked to the extent of load support from elastohydrodynamic films in current forms of metal-on-metal hip replacements. The rates of change in wear and wear rates with elastohydrodynamic film thickness are so great at film thicknesses less than about 15–20 nm that tests designed to investigate the influence of other factors upon wear should carefully replicate the film thicknesses in each test. This observation might well contribute to a better understanding of the apparent conflict in some findings from earlier simulator studies. Great advances have been reported in recent years in metal-on-metal hip joint replacement technology. The work presented confirms that wear in such implants is minimized if the largest possible diameter is adopted and the clearance is minimized as much as is practicable. The former requirement encourages the use of articular surface replacement rather than traditional monolithic femoral heads while the latter calls for precision manufacture and careful evaluation of the potential distortion of the acetabular shell in the pelvis [Dowson *et al*., 2006]. A range of clearances of joints of 28 and 45 mm diameter have been tested on a simulator with bovine serum as the lubricant [Scholes *et al.*, 2006]. For most of the joints tested, they found a biphasic wear pattern with an initial high wear bedding-in phase which then dropped to a lower steady state wear. However, this behaviour was not observed in the joint with the largest clearance (head diameter of 45mm and diametral clearance of 315µm) for which the wear rate remained high for the duration of the test [Scholes *et al.*, 2006]. The effect of diametral clearance on the wear of metal-on-metal hip joints of 28mm in diameter and diametral clearances in the range 74 to 161µm were also studied [Scholes *et al.*, 2006] and bovine serum was used as the lubricant, but it was not stated at what concentration. The results clearly demonstrated lower wear at the lower clearances but with diametral clearances below 16.5 µm an increase in wear was observed. The two joints with the negative diametral clearances reached about 20,000 cycles, exhibiting the highest wear but then the components seized. The same authors [Scholes *et al.*, 2006] reported using five low-carbon wrought CoCrMo against itself and four high-carbon as-cast CoCrMo against itself, both of 45mm diameters with diametral clearances ranging from 44.5 to 99µm and from 5 to 315µm, respectively. For all joints, there was an increase in wear with increase in clearance. In addition to this, the low-carbon wrought material gave lower wear (0.25 mm^3 per 10^6 cycles) than the high-carbon cast material (0.6 mm^3 per 10^6 cycles). The effect of radial clearance on the wear of wrought CoCrMo pairings (eight joints) were also studies and reported [Scholes *et al.*, 2006] with diametral clearances ranging from 7 to 141µm. These were tested in 50 per cent bovine serum. Again, all components experienced a period of initial bedding-in followed by a lower steady state wear rate. In fact, all but one of the joints had no detectable wear from $0.5X10^6$ cycles onwards. From the results, it was clear to see that an increase in clearance led to an increase in wear rate. They concluded that the optimum diametral clearance for a joint of 28 mm diameter was 20–80μm. In a hip simulator study reported by the same authors [Scholes *et al*., 2006] joints of 22mm, 28mm, and 35mm diameters were tested with 12 diametral clearances in the ranges 5–75μm, 7–161μm, and 5–75μm, respectively. Bovine serum was used as the lubricant but the concentration was not quoted. For all diameters, clearances below 7.5μm resulted in increased wear. For the joint of 28mm diameter, diametral clearances above 7.5μm gave a positive correlation between clearance and wear. However, for joints with the other two diameters, no strong correlation was found. This work found that the heads of different diameters did not show a statistically significant difference between the wear rates produced; this is in contrast with the findings of other workers. Most other workers have found that larger-diameter metal-on-metal joints produce lower wear, although this has not been statistically proven. These authors reported on ten joints of 28mm diameter made from wrought CoCrMo with average wear rate of 0.12±0.07 mm³ per 10^6 cycles. This wear volume is similar to that found by other workers. However, the radial clearances were not specified. A thorough investigation into the comparative wear of joints of 28mm and 36mm diameter have found [Scholes *et al*., 2006] an initial bedding-in wear rate to be apparent for the first 1–2 million cycles and a lower steady state wear rate was seen thereafter. The mean steady state volumetric wear rate was found to be 0.45 mm³ per 10^6 cycles and 0.36mm³ per10^6 cycles for the bearings of 28mm and 36mm diameters respectively. This may imply that more wear occurred in the smaller-diameter joint; however, the joint of 36mm diameter had a larger range (0.03–1.62mm³ per 10^6 cycles) than the joint of 28 mm diameter (0.12–0.77mm³ per 10^6 cycles). It was noted, however, that the wear rate was generally lower for the joint of 36mm diameter than for the implants of 28 mm diameter. A wear particle analysis study has also been carried out using a hip simulator [Tipper et al., 2005] and involved the testing of three CoCrMo-on-CoCrMo prostheses with 28mmin diameter tested in 25 per cent bovine serum. The initial bedding-in wear was 3.1mm^3 per 10^6 cycles and the steady state wear was 1.23 mm^3 per 10^6 cycles. Also wear tests on low-carbon CoCrMo joints of 28mm diameter with two different diametral clearances (22 and 40 µm) using 25 per cent bovine serum as the lubricant have been carried out [Scholes *et al.*, 2006] and the wear of the acetabular cups was measured and, again, two distinct wear phases were found: an initial bedding-in wear phase and the lower steady state wear (after about $1X10^6$ cycles, which is the equivalent of approximately 1 year *in vivo*). Similar wear was found for both radial clearances with the smaller radial clearance giving slightly lower wear**.** Friction tests were also performed on these joints and showed that the friction significantly decreased post-wear testing, although there was not a significant difference in the friction factors produced by the two radial clearance joints. The reduction in friction is thought to be due to the self-polishing of the materials during the wear test. Others [Smith *et al.*, 2004] have tested joints of 16, 22.225, 28, and 36mm diameter in 25 per cent bovine serum. In an attempt to analyse the influence on the wear rate by joint diameter as a single variable, all prostheses were manufactured to the same, clinically relevant standards. However, although the joints of 16, 22.225, and 28mm diameter had radial clearances of approximately 30µm, the joints of 36mm diameter had a radial clearance of 80µm. With the exception of the joint of 36mm diameter, the simulator provided simplified loading and motion cycles whereas the joint of 36mm diameter was subjected to both the simplified loading cycle and the physiological loading cycle. In addition to the wear studies performed on the joints of 36mm diameter, using the simple resistivity technique, these workers also tested the surface separation within the simplified simulator and compared this with the physiological simulator. The proportion of surface separation per cycle was generally greater in the simplified simulator than in the physiological simulator. However, as far as the wear rates are concerned, simplified machines have been shown to produce very similar wear rates to the physiological machines. The joints tested were 28mm in diameter and articulated in 100 % bovine serum. Ten joints were tested, all with radial clearances of approximately 40µm. The mean volumetric wear was found to be approximately 0.13 mm³ per 10^6 cycles. A hip simulator study [Dowson *et al.*, 2006] tested joints of 36 mm diameter with low- and high-carbon components in wrought or cast forms. This work focused on the running in wear results. As with the previous study, the low-carbon cast material gave higher wear than the high-carbon cast or wrought materials. They reported no significant difference between the wear volumes of the high carbon wrought and cast materials. For the small range of radial clearances studied (52.5–73µm) the volumetric wear decreased as clearance decreased. However, this was only found to be significant at the extremes of clearance. In another study [Dowson *et al.*, 2006] using a hip joint simulator, the effects of different head diameters and clearances on the wear performance of high-carbon metal-on-metal joints of different diameters were investigated. Head diameters ranged from 28 to 54.5 mm and, as discussed previously the heads of 36 mm diameter were tested with diametral clearances of 52.5–73µm. All the joints were manufactured from high-carbon CoCrMo alloy; the joints of 28mm and 36mm diameters were manufactured from wrought material and the joints of 54 mm and 54.5mm diameters were cast materials. Again, as shown before, these joints exhibited an initial running-in wear phase that was higher than the steady state wear that developed after about $0.5X10^6$ cycles. Larger-diameter heads with small clearances gave a lower wear rate than the smaller- diameter heads. In conclusion it was noted that, for the best lubrication and wear performance, the head diameter should be as large as possible with a clearance as low as practicable. Eight high-carbon cast metal-on-metal joints of 40mm diameter were also tested [Dowson *et al.*, 2006], and although five joints showed a steady state wear rate of less than 1 mm³ per 10^6 cycles, the other joints showed considerably higher wear rates. This wide range of wear rates has been shown by other workers. This led to a mean wear rate of 6.3 mm³ per 10^6 cycles. This wear rate is higher than that recorded by most other groups. In a study [reported by Scholes *et al*., 2006] they tested joints of 38, 50, 54, and 56 mm diameter in 33% bovine serum and measured the linear wear on a coordinate-measuring machine. They found that all the joints exhibited a running-in wear period and, after that, little additional wear was measured. They reported an increase in running-in wear with an increase in clearance (see Table 2.6) [Scholes *et al*., 2006].

Table 2.6. Effect of bovine serum concentration on the wear of joints of 28 mm diameter and various diametral clearances [Scholes *et al*., 2006].

study	Bovine serum (%)	Mean radial clearance(μm)	Number tested	Volumetric wear (mm³ per 10^6 cycles)
1	100	44	10	0.13(s)
2	100	40	3	2.51(t)
3	100	42.5	2	0.4(t)
4	50	40	1	0.4 (t)
5	25	31.5	3	0.54(t)
6	25	56.3	4	0.45 (s)
7	25	30	3	1.6 (s)
8	25	40	2	0.25 (acetabular cup only) (s)
9	25	31	4	≈ 1(s)
10	25	30	Not stated	0.2 (s)

Clearance and diameter are of course, not directly related, and each has a separate effect on the tribological characteristics of a joint. Returning to the equation of film thickness, $(h_{\min})$, the overall dependence of $(h_{\min})$ on surface Roughness (*R)* can be seen, since this is a positive exponent, $(h_{\min})$ will increase as (*R)* increases if all other testing and material conditions remain constant. As previously shown, *R* is the product of the radii divided by the diametral clearance. Hence, larger radii and smaller radial clearances should induce a thicker film, while a combination of small radii and a larger clearance should induce a thinner film, assuming all other parameters affecting the film thickness remain the same.

There is of course a practical limit to the radial clearance, since if this is too small then the joint will not function. This is partly due to manufacturing tolerances, but also due to deformation of the cup at contact, which can cause a decrease in the diametral clearance. In extreme cases this may cause equatorial contact in the joint, a feature that was seen in early explanted joints leading to dislocation and implant failure. These factors (different diametral clearances and joint diameter) have been investigated experimentally and seem to tie in well with the theoretical predictions as shown in Table 2.7 [Vassiliou *et al*., 2007].

Table 2.7. Wear rates of hard bearing and conventional joints as reported in the literature [Vassiliou *et al*., 2007].

Implant type	Number of hips	Running-in per million cycles	Steady state per million cycles	Total wear per million cycles	Joint diameter	Study
CoCrMo on UHMWPE Zirconia on UHMWPE	5 5	n/a n/a	n/a n/a	40.8mm³ 33.3mm³	28mm	1
Alumina on polyethylene CoCrMo on CoCrMo Alumina on Alumina	3 8 9	n/a n/a n/a	n/a n/a n/a	51mm³ 0.04 mm³ 6.3 mm³	28mm 32mm 40mm	2
Alumina on Alumina	5	0.34mg ,0.24	n/a	0.3mg	28,32mm	3
Alumina on Alumina CoCrMo on CoCrMo		0.27 2.681	0.004 0.977	n/a n/a	28mm 28mm	4
Alumina on Alumina	5	n/a	n/a	<0.1mg	28mm	5
CoCrMo on CoCrMo	4	0.75 mm³	0.17 mm³	n/a	28mm	6
CoCrMo on CoCrMo CoCrMo on CoCrMo Low carbon, high carbon, mixed	2 2 2	1.38mm³ 0.32 mm³ 0.30mm³	0.322mm³ 0.023 mm³ 0.037 mm³	0.5mm³ 0.1 mm³ 0.1 mm³	28mm 28mm 28mm	7
CoCrMo on CoCrMo Low carbon wrought High carbon wrought	8 6 8	0.76 mm³ 0.24 mm³ 0.21 mm³	0.11 mm³ 0.067mm³ 0.063mm³	1.11 mm³ 0.42 mm³ 0.40 mm³	28mm 28mm 28mm	8
CoCrMo on CoCrMo	4 10 4	~4.2mm³ ~2.2 mm³ ~7 mm³	~1 mm³ ~0.5 mm³ ~0.5 mm		28mm 40mm 56mm	9

2.5 Friction studies

In the mid 1970s, a freely swinging pendulum machine was developed [Unsworth *et al*., 1978; Scholes *et al*., 2001] for the measurement of friction in which normal or artificial joints formed the pivot of the pendulum. The pendulum was later altered to enable friction to be measured under conditions that are more realistic. The mode of lubrication in metal on polymer hip joints was found to be columbic (boundary) in nature. The presence or of bovine synovial fluid had little consequence on the recorded friction. This was consistent with Charnley's analysis that boundary lubrication prevailed and it underlined the analysis following his preference of a relatively small-diameter femoral head for his low-friction arthroplasty. There was, however, an interesting suggestion from the use of high-viscosity silicone fluid that mixed or even fluid-film lubrication could be achieved with high-viscosity fluids. Various studies have established that friction factors for CoCrMo-on-CoCrMo joints were almost independent of the fluid dilution and typically lay in the range 0.2- 0.3. It is, therefore, known that such joints performed in the mixed lubrication regime, although the friction factors were high and the lambda rations were low. It appeared that the test on metal on metal joints exposed either boundary lubrication or severe mixed lubrication close to the boundary lubrication regime.

In this study we will show the progression from boundary to mixed and then fluid film lubrication as viscosity increases for the 50mm diameter metal-on-metal BHR hip resurfacing devices lubricated by 25 % bovine serum (BS as aqueous solutions of BS+ carboxymethyl cellulose to provide a range of viscosities).

Frictional measurements of all different kinds of joints are normally carried out on a Hip/Knee Function Friction Simulator. The loading cycles have maximum and minimum loads set at usually 2000-3000N and 100-300N, respectively. A simple harmonic oscillatory motion of amplitude 24° is usually applied to the femoral head in the flexion-extension plane (±12°). The period of motion is ~1.0 s. The simulator is described further in detail in the experimental procedure in chapter three. In these studies, friction factor (f) is defined in chapter one (equation 1.10). Friction factor is similar in magnitude to the coefficient of friction, but varies with the pressure distribution over the head. Table 2.9 gives typical friction factors for various material combinations [Scholes S and Unsworth A, 2000] and all joints were of 28mm diameter with 40 micron radial clearance.

Table 2.8 also shows the predicted lubrication modes and friction factors for each material pairing. Although the predicted minimum film thicknesses are usually similar for both the all ceramic and all metal couplings, an important difference is observed for the dimensionless parameter (λ). The metal-on-metal joints exhibit (λ) value of less than one, therefore suggesting a boundary lubrication regime whereas the ceramic-on-ceramic joints have (λ) value of greater than three suggesting a full fluid film lubricating regime. This difference is due to the much lower surface roughness of the ceramic components, see Table 2.8 the CoCrMo-on-UHMWPE joint exhibited (λ) value of less than one, suggesting, as expected, a boundary lubrication regime [Scholes S and Unsworth A, 2000].

Table 2.8. Predicted lubrication modes (η=0.01 Pas) [Scholes and Unsworth, 2000].

Femoral component	**Acetabular component**	**Femoral R1(µm)**	**Acetabular R2(µm)**	**Predicted Minimum film thickness(µm)**	**λ**	**Friction factors CMC/Bovine serum**
Co Cr Mo	Co Cr Mo	0.008(0.002)	0.08(0.00365)	0.05	<1	0.26/0.15
Alumina	Alumina	0.003(0.001)	0.01(0.0063)	0.06	>3	0.002/0.05
Co Cr Mo	UHMWPE	0.04(0.0060	1.29(0.086)	0.09	<1	0.017/0.032

Aqueous solutions of bovine serum (25%BS + 75% distilled water) with carboxy methylcellulose (CMC as gelling agent to give various viscosities) are normally used as the lubricants at viscosities of 0.001-0.2 Pas. BS+CMC fluids are used as the lubricants because of their similar rheological properties to synovial fluid. The joints may also be tested with 100% newborn calf serum with a viscosity of ~0.007 Pas. The joints are cleaned thoroughly between tests and Stribeck analysis are used to give an indication of the mode of lubrication, in which the friction factor is plotted against the Sommerfeld number, z, which is defined in equation (1.9).
As before, η is the viscosity of the lubricant, u is the entraining velocity of the bearing surfaces, l is the applied load and r is the head radius. The Sommerfeld number is varied by altering the viscosity of the lubricant. A decrease in friction factor with increase in Sommerfeld number is indicative of a mixed lubrication regime whereas a rising trend is indicative of a full fluid film regime. Friction testing, therefore, is a useful method to compare implants of various designs, materials and conditions. The measurement of friction may also be used as an indirect method to imply the lubrication of a bearing combination.
Also, typical friction factors associated with different lubrication regimens are given below in Table 2.9 [http/www.zimmer.co.uk].

Table 2.9. Typical friction factors for various artificial hip joints in the presence of bovine serum [http/www.zimmer.co.uk].

Lubrication regimes	**Friction factor**
Boundary lubrication	0.1–0.7
Mixed lubrication	0.01–0.1
Fluid-film lubrication	0.001–0.01

As mentioned earlier, a constant friction factor with increasing Sommerfeld number indicates boundary lubrication. A reducing friction factor with increasing Sommerfeld number is indicative of a mixed lubrication and increasing friction factor with increasing Sommerfeld number indicates fluid film lubrication. Typical friction factors in various hip joints are also summarised in Table 2.10 below.

Table 2.10. Typical friction factors for various artificial hip joints in the presence of bovine serum [http/www.zimmer.co.uk].

Bearings	**Friction factor**
UHMWPE-on-metal	0.06–0.08
UHMWPE-on-ceramic	0.06–0.08
Metal-on-metal	0.22–0.27
Ceramic-on-ceramic	0.002–0.07
Ceramic-on-metal	0.002–0.07

The femoral head made of a cobalt–chromium-molybdenum alloy has an elastic modulus of ~210 GPa and a Poisson's ratio of 0.3. Typical diameter of the femoral head (d) and the diametral clearance (C_d) are 28, 35, 50mm and 80-110μm, respectively. Typical load in the vertical direction and angular velocity representing the flexion-extension in the human hip joint can be chosen as 2500N and 1.5rad/s, respectively. A typical viscosity for peri-prosthetic synovial fluid is ~0.0025Pas. The equivalent radius, entraining velocity and equivalent elastic modulus can be calculated as ~20mm, 0.01m/s and 3.0GPa, respectively. The minimum film thickness can thus be determined as $h_{\min}$ = 0.06mm. Therefore, the calculated (λ) ratio is less than one and this indicates a boundary lubrication regimen. Lubrication regimens in other types of artificial hip joint can be analysed readily using the same procedure. Predictions for typical hip implants with metal-on-metal are shown in Table 2.11. It is clear from Table 2.11 that recently developed manufacturing techniques for metallic bearing surfaces are also capable of achieving a similar standard. The importance of design parameters, such as the femoral head diameter (d) and the diametral clearance (C_d), can be further explored for metal-on-metal bearings. It is clear from equation (6) that in order to promote fluid-film lubrication, it is necessary to increase the femoral head diameter and to reduce the diametral clearance so that the equivalent radius (R) is increased. The increase in the femoral head diameter also increases the entraining velocity. The importance of large diameter is manifest in the metal-on-metal hip resurfacing prosthesis. The estimated lubricant film thicknesses for a 28mm diameter total hip implant and a 50mm diameter hip resurfacing prosthesis, both using a metal-on-metal bearing is compared in Table 2.12. However, it should be pointed out that the diametral clearance also plays an equally important role in the large diameter metal-on-metal hip resurfacing prostheses. An increase in the diametral clearance can lead to a decrease in the equivalent radius and consequently the predicted lubricant film thickness is reduced.

Table 2.11. Calculation of (λ) ratio and determination of lubrication in a typical metal-on-metal hip implant [Jin *et al*., 2006].

Input parameters	
Femoral head diameter	28mm
Diametral clearance	0.06mm
Elastic modulus (Co–Cr)	210 GPa
Poisson's ratio (Co–Cr)	0.3
Load	2.5 kN

Angular velocity	1.5 rad/s
Viscosity	0.0025 Pas
Composite Ra	0.014μm
Calculation	
Equivalent radius	6.55M
Entraining velocity	0.0105 m/s
Equivalent elastic modulus	230 GPa
Minimum film thickness	0.024 mm
λ ratio	1.7
Lubrication regime	Mixed lubrication regimen

This is particularly important for large diameter bearings. If the reduction in the lubricant film thickness moves the lubrication regimen towards boundary lubrication, the adverse effect of increased sliding distance associated with the large femoral head diameter must be considered.

Table 2.12. Comparison of predicted lubricant film thickness between a total hip implant and a hip resurfacing prosthesis using a similar metal-on-metal bearing [Jin *et al*., 2006].

Parameters	Total hip implant	Hip resurfacing prosthesis
Diameter (mm)	28	50
Diamateral clearance (μm)	60	100
Load (N)	2500	2500
Angular vel. (rad/s)	1.5	1.5
Viscosity (Pas)	0.0025	0.0025
Equivalent diameter (mm)	13.1	25.1 (100%↑)
Entraining vel. (mm/s) Film thickness (mm)	10.5 0.024	18.75 (80%↑) 0.058 (142%↑)

Such a comparison is shown in Table 2.13. These simple theoretical analyses have recently been confirmed with the experimental simulator studies. However, it should also be pointed that metal-on-metal bearings depend on protection from the boundary layers and the effect of proteins can have a significant effect on the friction and wear [Jin *et al*., 2006].

Table 2.13. Effect of clearance on the predicted lubricating film thickness in metal-on-metal hip resurfacing prostheses [Jin *et al*, 2006].

Parameters	**Hip resurfacing prosthesis**	**Hip resurfacing prosthesis**
Diameter (mm)	**50**	**50**
Diameteral clearance (μm)	**100**	**300**
Load (N)	**2500**	**2500**
Angular vel. (rad/s)	**1.5**	**1.5**
Viscosity (Pas)	**0.0025**	**0.0025**
Equivalent diameter (m)	**25.1**	**8.38 (70%↓)**
Entraining vel. (mm/s)	**18.75**	**18.75 (0%)**
Film thickness (mm)	**0.058**	**0.025 (57 %↓)**

CHAPTER THREE

3.0 EXPERIMENTAL PROCEDURE

3.1 MATERIALS AND EQUIPMENT

Six as cast, high carbon Co-Cr-Mo Metal-on-Metal (MoM) 'Birmingham Hip Resurfacing (BHR) implants' (supplied by Smith and Nephew Orthopaedics Ltd, Coventry, UK) with a nominal diameter of 50 mm each and diametral clearances of 80, 135, 175, 200, 243 and 306 µm were used in this study. The initial surface roughnesses were measured by S&N Orthopaedics to be in the range 'R_a=10-30 nm' using a Form-Talysurf 50 (Taylor Hobson, Leicester, UK) which were similar to those of commercial MoM hip prostheses and within the accepted range.

Frictional measurements (and lubrication analyses) of all the BHR implants were carried out at University of Bradford-Medical Engineering Department, using a Prosim Hip Joint Friction Simulator (Simulation Solutions Ltd, Stockport, UK), Figure 3.1. The acetabular cup was positioned in a fixed low-friction carriage below and the femoral head in a moving-frame above as shown in Figure 3.2. The carriage sits on an externally pressurized hydrostatic bearings generating negligible friction compared to that generated between the articulating surfaces, also allowing for a self-centring mechanism. During the flexion-extension motion (see Figure 3.3), the friction generated between the BHR implants causes the pressurized carriage to move. This movement (or rotation) is restricted by a sensitive Kistler piezoelectric force transducer which is calibrated to measure torque directly. A pneumatic mechanism controlled by a microprocessor generates a dynamic loading cycle and the load is also measured by the same piezoelectric force transducer.

Figure 3.1. Picture of the Prosim Friction Hip Simulator used in this work for obtaining frictional torque and friction factor.

Figure 3.2.1. Friction hip simulator showing the fixed lower carriage with the cup holder and the moving carriage (rocker) with the femoral head.

Figure 3.2.2. Friction hip simulator in flexion (above) and extension (below) positions.

3.2 Friction factor and frictional torque measurements

Friction measurements (friction factor results given in chapter four) were made in the 'stable' part of the cycle at 2000N and to obtain accurate measurements for friction, the centre of rotation of the joint was aligned closely with the centre of rotation of the carriage. The loading cycle was set at maximum and minimum loads of 2000N and 100N, respectively. In the flexion/extension plane (see Figure 3.2.2), an oscillatory harmonic motion of amplitude ±24° was applied to the femoral head with a frequency of 1Hz in a period of 1.2s. The load was, therefore, applied to the femoral head with the artificial hip joint in an inverted position, i.e. femoral head on top of the acetabular component (see Figure 3.2.1), but with a 12° angle of loading between the two bearings as observed in human's body (12° medially to the vertical).

The angular displacement, frictional torque (T) and load (L) were recorded through each cycle (127 cycles for each friction test lasting 127x1.2=152.4 seconds=2.5 minutes). The frictional torque was then converted into friction factor (f) using the equation: $f = T/rL$, where r is the femoral head radius. An average of three independent runs (three friction tests) was taken for each friction factor.

3.3 Lubricants (and viscosities) used for friction testing

Initially, the test was conducted with non-clotted blood (whole blood with Lithium heparin to prevent clotting) and clotted blood as the lubricants for each joint. Viscosity of the non-clotted blood was found to be ~ 0.01 Pas and that of clotted blood was ~ 0.02 Pas. The test was then run with a combination of: (i) Aqueous solutions of bovine serum (BS, as new born calf serum via Harlan Sera-Lab with a total protein content of 61.27 mg/ml which had been sterile filtered to 0.1mm) with and without carboxymethyl cellulose (CMC), i.e. 25cc BS+75cc distilled water+CMC, to achieve viscosities of 0.0038, 0.0013, 0.0136, 0.0327, 0.105 and 0.19 Pas , and (ii) Bovine serum (BS) and hyaluronic acid (HA, Supartz ® supplied by Smith and Nephew Orthopaedics Ltd) with or without CMC to achieve viscosities of 0.00145, 0.0035, 0.01324, 0.037 and 0.138 Pas. Note that all the viscosities were measured at a shear rate of 3000 s^{-1} using the Anton Paar Physica MCR 301 Viscometer (see Figure 3.3); the content of the hyaluronic acid was equivalent to that contained in synovial fluid for a normal young adult (~3.1), and the bovine serum was diluted to 25% by volume, i.e. the BS concentration was kept at 25% with aqueous solutions of CMC (75% by volume of distilled water+CMC). The CMC was used as a gelling agent or viscosity enhancer. The CMC fluids are shown [Scholes, S. C et. al, 2000] to have similar rheological properties to synovial fluid, but it is possible that they may not produce the shear stresses created by the presence of macromolecules in the lubricant. Also, 0.2% sodium azide was added to the solutions (1g per litre of serum) as an anti-bacterial/antibiotic agent (biostatic) and 20 mMol of ethylenediaminetetra-aetic acid (EDTA) was also added to prevent calcium phosphate precipitation on the articulating surfaces of the implants.

Figure 3.3. Anton Paar Physica MCR 301 Viscometer used in this work.

3.4 Implant cleaning procedure

The joints with different clearances were cleaned thoroughly before each test using ultrasonic cleaning in water with liquid soap, followed by ultrasonic cleaning in methanol, and then ultrasonic cleaning in distilled water (10 minutes each time), and finally rinsed with methanol and dried off with soft tissue. Each joint was tested with each lubricant three times and the implants were cleaned with soft tissue and only distilled water after each 127 cycle for the same lubricant.

3.5 Stribeck analysis

To give an indication of the mode of lubrication, Stribeck analysis was performed by plotting friction factor against Sommerfeld number, z, which is defined as: $z=\eta ur/L$ where L is the load, r is the joint radius, η is the viscosity of the lubricant and u is the entraining velocity (=0.02 m/s) of the bearing surfaces. The Sommerfeld number is varied only by altering the viscosity of the lubricant since u, r and L remain constant. A decrease in friction factor with increase in Sommerfeld number is indicative of a mixed lubrication regime in which the load is carried in part by the contact between the asperities of the bearing surfaces and also by the pressure generated within the lubricant. A rising trend in friction factor with increase in Sommerfeld number is indicative of a full fluid film lubrication regime where the two surfaces are completely separated by the lubricant film and the frictional resistance is generated solely by the shear stress within the fluid.

3.6 ProSim friction simulator

The metal-on-metal friction tests were executed employing the ProSim Friction Simulator (ProSim Ltd, Stockport-Manchester) as mentioned earlier (see Figure 3.1). ProSim friction simulator is a compact single-station servo-hydraulic machine that consists of:

- A fixed frame which comprises of a friction measuring carriage that is placed on two externally pressurised hydrostatic bearings. The bearings allow negligible friction within the carriage, with respect to the friction generated between the articulating counterfaces of the joints.
- A loading frame in which the femoral head is attached through a motion arm (see Figure 3.4 for details).

As can be observed from Figure 3.4, a personal computer via a graphic user interface is employed in order to control the kinetics and kinematics of the machine.

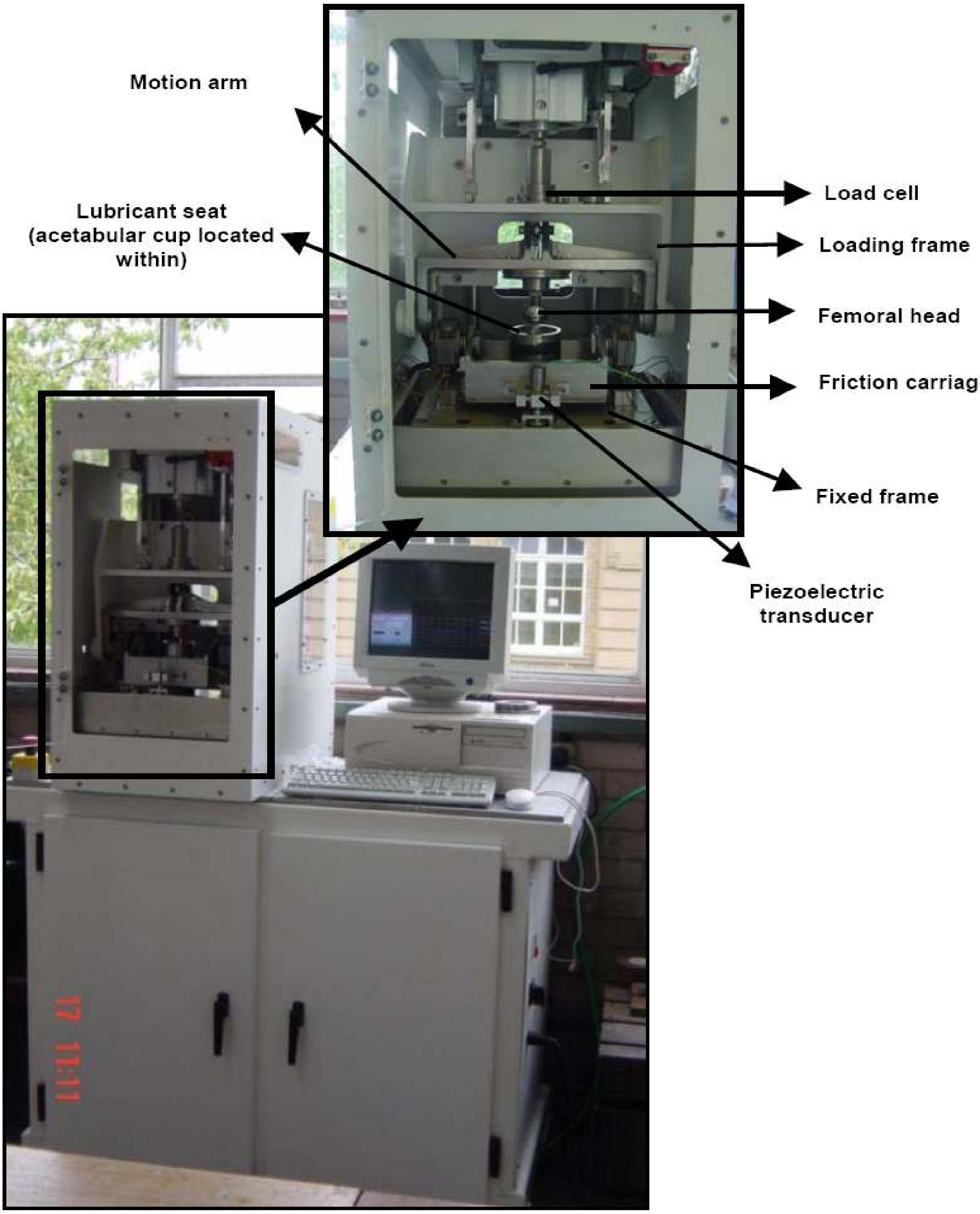


Figure 3.4. ProSim Friction Simulator with details.

A piezoelectric crystal transducer is attached to the friction carriage that prevents any undesired motion (as the femoral head flexes and extends). The piezoelectric transducer also determines the frictional torque within the system, by measuring the force transferred between the fixed frame and the carriage. An in-built charge amplifier is used in order to amplify the signals from the piezoelectric transducers (see Figure 3.5).

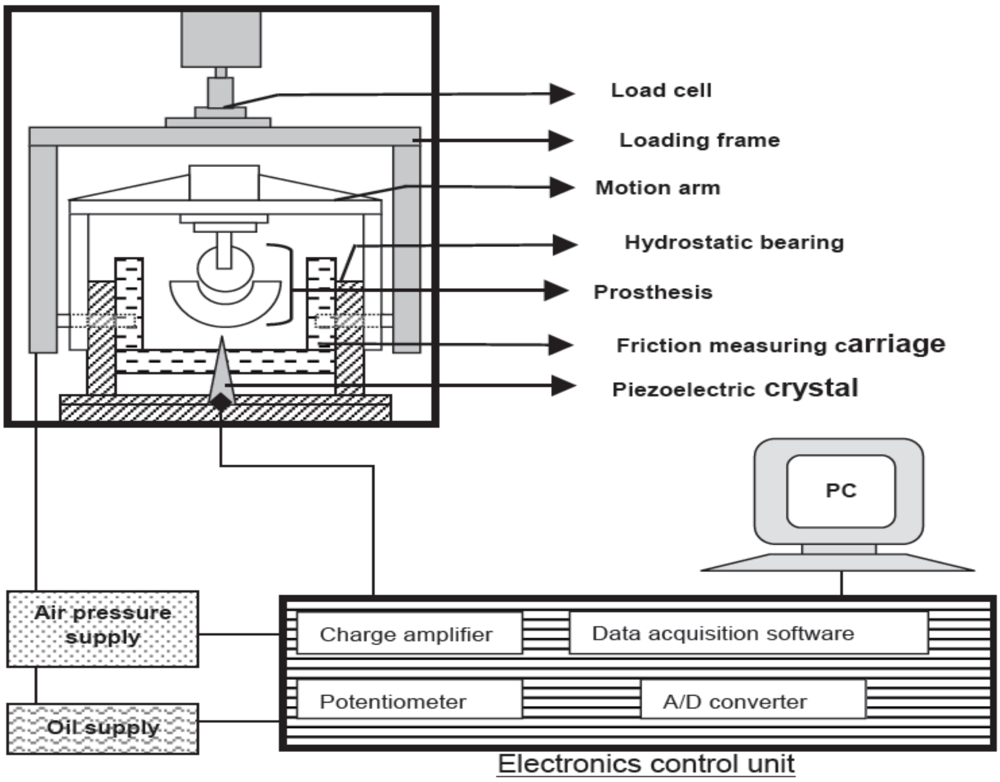


Figure 3.5. Schematic diagram of the ProSim friction simulator.

In order to achieve the true value of the frictional torque between the bearing surfaces for the duration of the experiment, correct alignment of centres of rotation of the head and cup within the friction carriage and the loading frame is necessary. The acetabular cup is placed in the lubricant seat within the friction carriage, such that the hip implant was inverted with respect to the *in vivo* condition (Figure 3.6).

Figure 3.6. The friction measuring carriage and loading frame of the ProSim friction hip simulator.

3.6.1 Alignment of the components

It should be noted that the alignment procedure must be carried out external to the machine. Alignment of the centre of rotation of the femoral component takes place by adjusting the femoral component using a stem holder (Figure 3.7). Furthermore, a specially designed rig was used in order to match the distance between the centre of the femoral head and the base of the stem holder, with the distance between the centre of rotation of the motion arm and the base of the stem holder. The femoral head height is then adjusted using slip gauges, to give a clearance between the top of the head and the roof of the rig. This clearance is determined by:

$(99.43 - 72.91 - R_1)$

Where R_1 is the radius of the femoral head (mm);

99.43 = the distance (mm) between the base and the foot of the stem holder jig;

72.91 = the distance (mm) between the centre of the femoral head to the base, which matches the centre of rotation of the motion arm and the base of the holder.

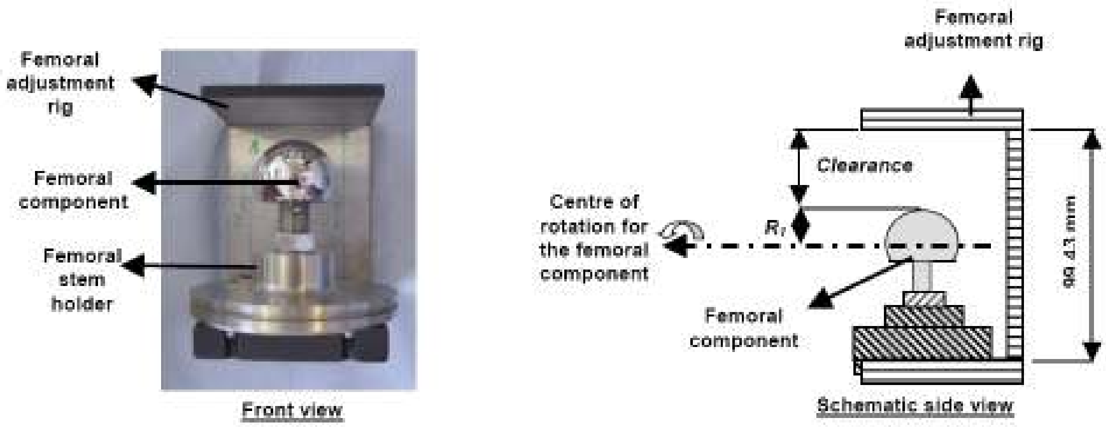


Figure 3.7. The rig used in adjusting the femoral component.

The position and height of the acetabular cup within the lubricant seat is adjusted by positioning a ceramic ball of a diameter less than the radius of the acetabular cup and also use of the adjustment screw in the base of the seat (Figure 3.8). The calculated value from: ($R_2 - 2R_{ball}$ + 14.92), is set on a depth gauge which can then be placed in the lubricant seat.

Where R_2 = the radius of the acetabular cup (mm)

R_{ball} = the radius of the ball bearing (mm)

14.91 = the distance (mm) from the centre of rotation of the friction measuring system to the top edge of the lubricant seat.

The acetabular cup is adjusted when the edge of the ceramic ball reaches the tip of the depth gauge.

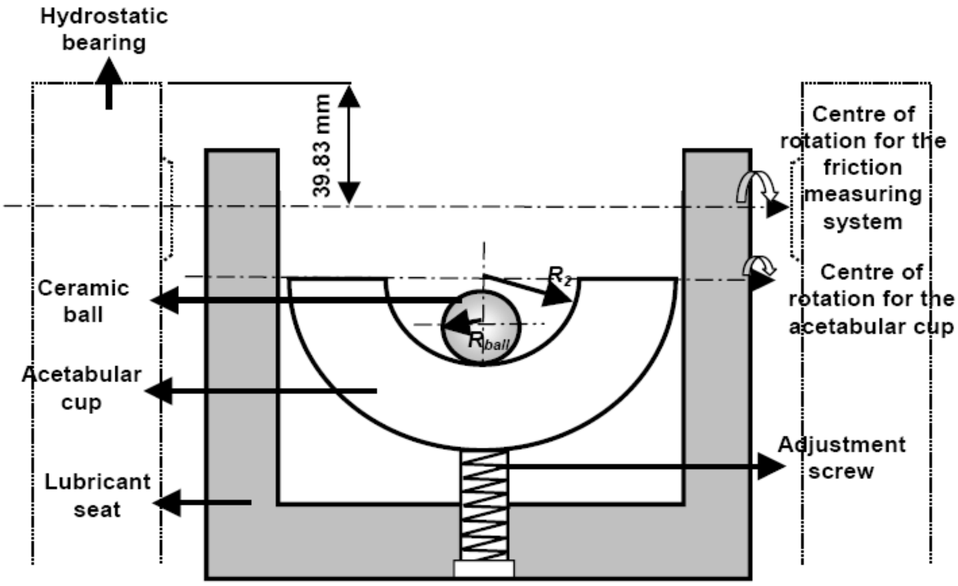


Figure 3.8. Schematic diagram of the lubricant seat showing the setup and alignment of the centre of rotation of the acetabular cup.

3.6.2 Kinetics and Kinematics

The friction simulator has two controlled axes of motion:

- rotation
- load

In order to simulate the dominant flexion/extension action of the natural hip joint in the friction hip simulator, the motion arm of the loading frame is used to flex and extend the femoral head in the range ±10° - ±30° (Figure 3.9). A hydraulic pressure system is controlling the loading cycle that has been applied vertically through the femoral head. A cam-follower mechanism applies the pressure to the hydraulic system as the femoral head undergoes flexion/extension motion. This will then pull the loading frame downwards and consequently will apply a load to the acetabular cup in the fixed frame. It should be pointed out that both kinetics and kinematics profiles are capable of being dynamic with fixed frequencies of 0.5, 1 and 2 Hz. The friction simulator can be programmed to generate maximum force of 3000N (Figure 3.9).

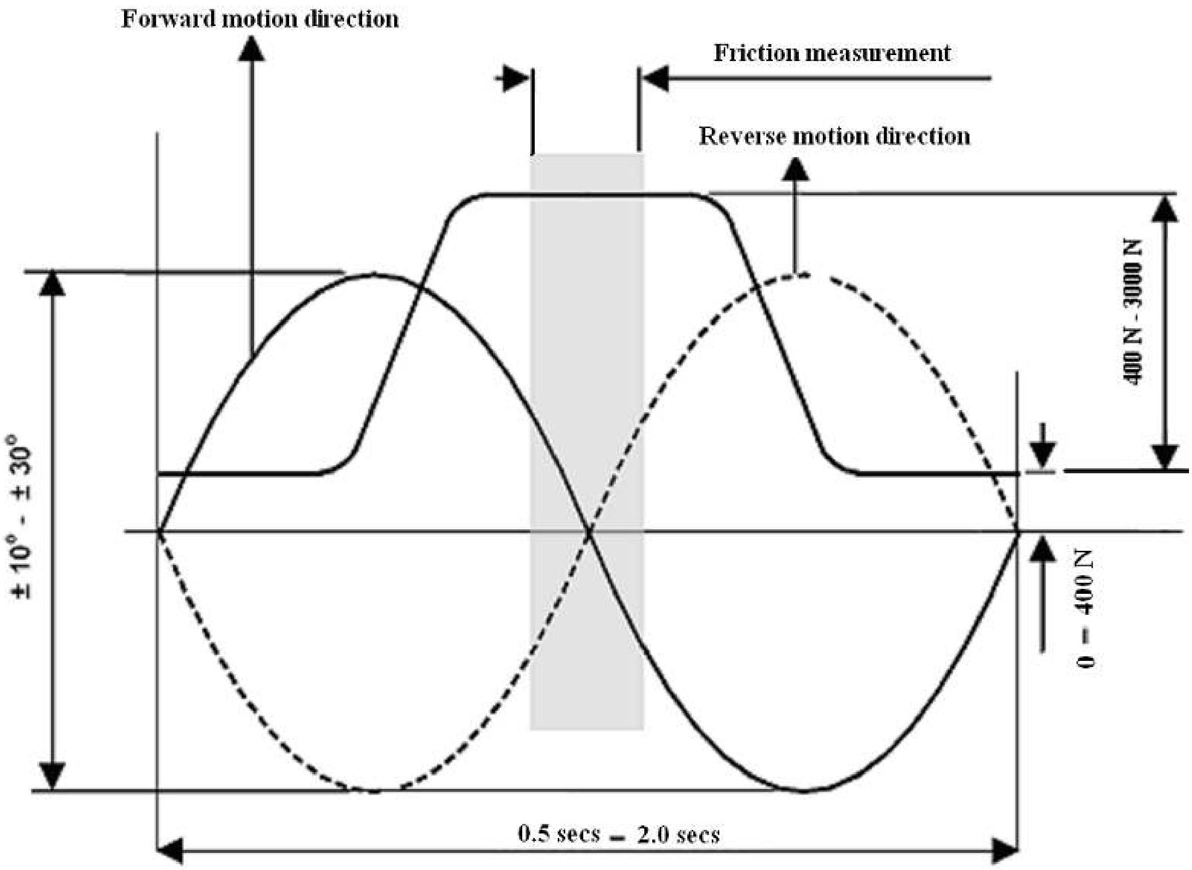


Figure 3.9 The dynamic loading cycle applied by the simulator indicating the forward and reverse motion directions and the friction measurement zone.

Prior to the start of each test the following checks were conducted:

- Alignment of the centres of rotation of the femoral head and acetabular cup of each hip joint with the simulator's centre of rotation. Furthermore, using a special alignment rod ensured the alignment of the loading frame to that of the friction measuring system.
- The simulator was allowed to run before each test for about 60-120 cycles to create steady state cyclic conditions.

Each lubricant was tested three times and it should also be pointed out that in order to minimise any further small misalignment within the simulator, each test was repeated in both forward and reverse direction. Kinetic and kinematic parameters as well as the frictional torque were recorded during each test, and the friction factor (*f*) was calculated from Equation 1.10.

Furthermore, data is logged at every 10 cycles, and the sample for each parameter is taken at 256 points per cycle. Data generated by 10 measurements were selected and the average of five points at high load and high velocity of the cycles were taken in order to calculate the friction factor. As already indicated, each test was repeated three times to eliminate any error or misalignment for the average friction calculation. All friction measurements obtained from the hip friction simulator did not show a great variation, thus, a negligible error (~0.0001) was not significant. In addition, lubrication mode was specified by Stribeck analysis, where friction factor was plotted against Sommerfeld number (z) which was calculated as shown by Equation 1.9.

3.6.2.1 Dynamic Load applied in Motion to the Human body

In order to study gait there are some basic definitions that need to be stated and understood. They are as follows:

Step – the act of lifting one foot and putting it down on a different part of the ground, such as when you walk or run.

Stride – the act of taking two steps thus returning to the original part of the walking cycle.

Step length – the distance travelled by taking one step.

Stride length – the distance travelled by taking one stride.

Velocity – the speed at which movement takes place. Calculated as stride distance /cycle time in m/s.

Cadence – how many steps are taken per minute.

Double support – both feet placed on the ground.

Float phase – neither foot is on the ground.

Using these basic definitions all aspects of gait can be observed and assessed.

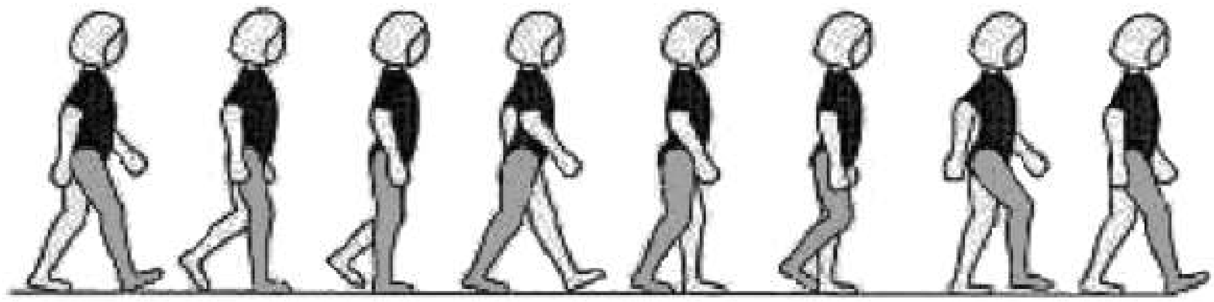


Figure 3.10. Phases of the gait cycle.

In the example above [Saleh *et al,* 1985] the right leg of the subject is highlighted so it can be studied through one entire stride. The left leg does the same actions but at different times to the right. For example it can be seen that while the right leg is in initial contact the left is in terminal stance.

The stance phase takes up 60% of the stride cycle time with 20% of this being double support and the swing phase takes up only 40%. By studying the right leg on the diagram below (Yellow) this can be seen to be true. It is also evident that while the right leg (Yellow) is in the stance phase the left leg (Green) is in the swing phase and visa versa, with the exception of the two periods of double support which are included in the stance phase.

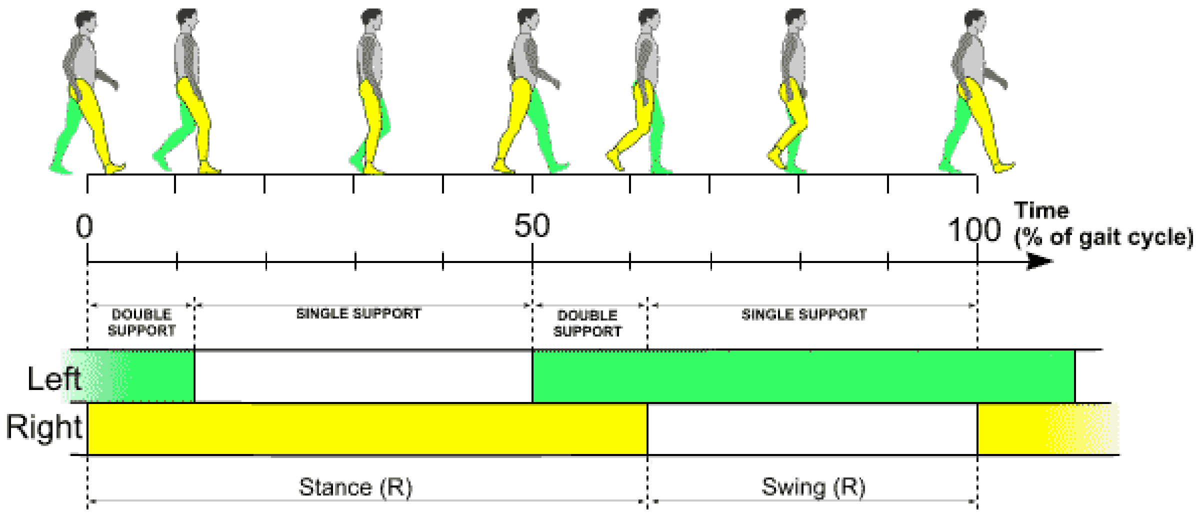


Figure 3.11. Gait cycle time analysis.

The vertical component of the ground reaction force can be split into four sections shown in Figure 3.12.

Heel Strike to 1st Peak (F1)

This is where the foot strikes the ground and the body decelerates downwards [Tanawongsuwan *et al,* 2003], and transfers the loading from the back foot to the front foot during initial double support. The 1st peak should be in the order of 1.2 times the person's body weight.

1st Peak (F1) to Trough (F2)

The trough should be in the order of 0.7 times the person's body weight.

Trough (F2) to 2nd Peak (F3)

The 2nd peak should be in the order of 1.2 times the person's body weight.

2nd Peak (F3) to Toe Off

The foot is unloaded as the load is transferred to the opposite foot.

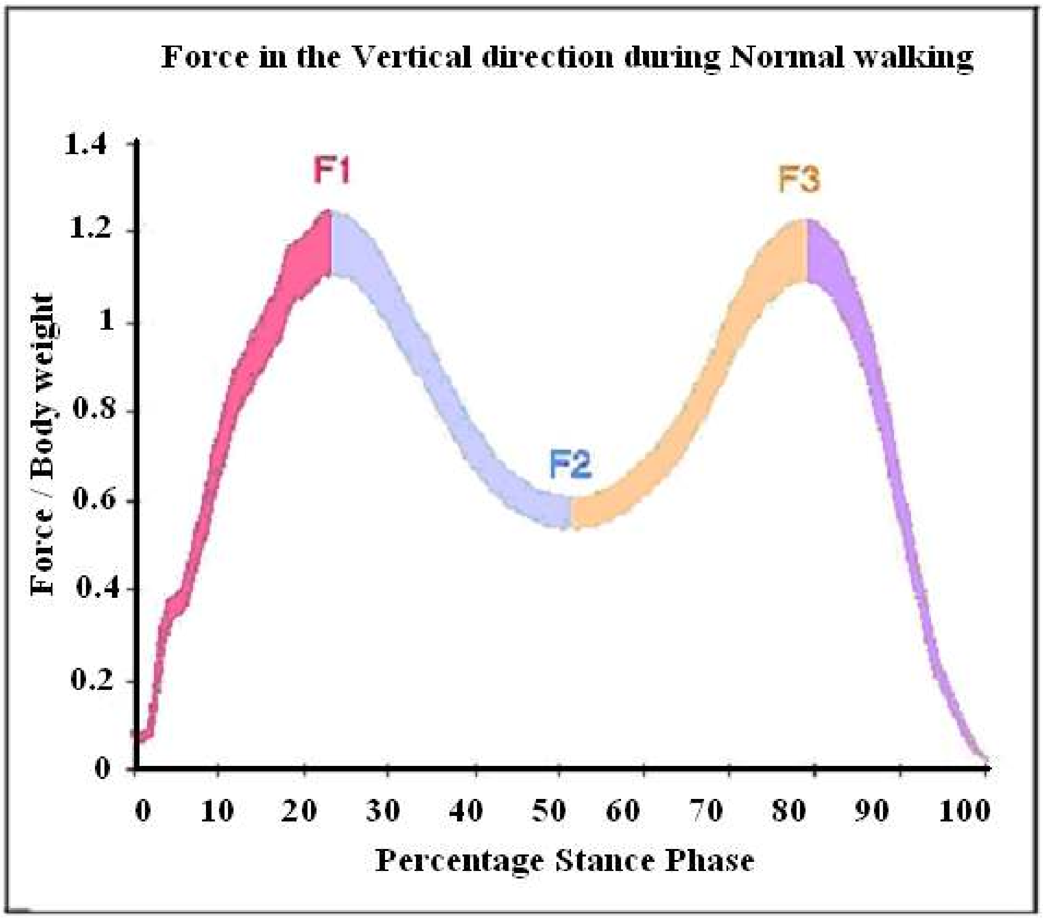


Figure 3.12. Force in the vertical direction during normal walking.

Ground reaction force (GRF), this is the force that is exerted on the body by the ground. From Newton's third law we know that "for every action there is an equal and opposite reaction" that is to say that if the ground is acting upwards on the body the body is acting downward in the same manner on the ground [http://www.upstate.edu/cdb/grossanat/limbs6.shtml]. These forces do not cancel each other out; they simply act against each other. The GRF is measured in 3 directions x, y and z and from the 3 a total force F can be calculated. The left diagram shows the planer co-ordinate system for calculating F.

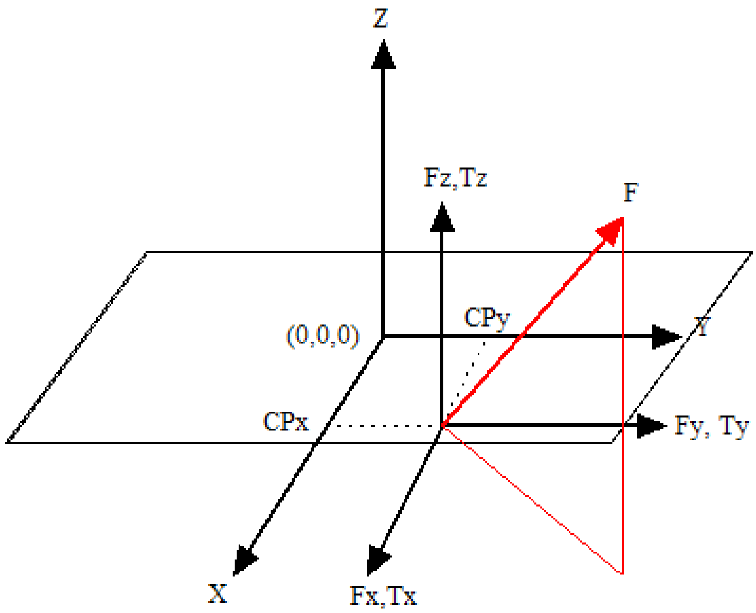


Figure 3.13. Ground reaction force measurement system.

The load and speed experienced in hip joints during walking are transient in nature, not only in magnitude but also in direction. However, the major load component is in the vertical direction, while the sliding and entraining speeds arise around a horizontal axis associated with flexion–extension, as schematically shown in Figure 3.13. Figure 3.14 shows the transient variation in load and speed during one walking/gait cycle [Dowson *et al*, 2005]. An average load of 1346 N for a complete cycle and 2500N in the stance phase (equivalent to about 3 times body weight of 750N) and an average resultant angular velocity of about 1.5 rad/s have been suggested for quasi steady state lubrication and normal gait analysis under *in vivo* conditions.

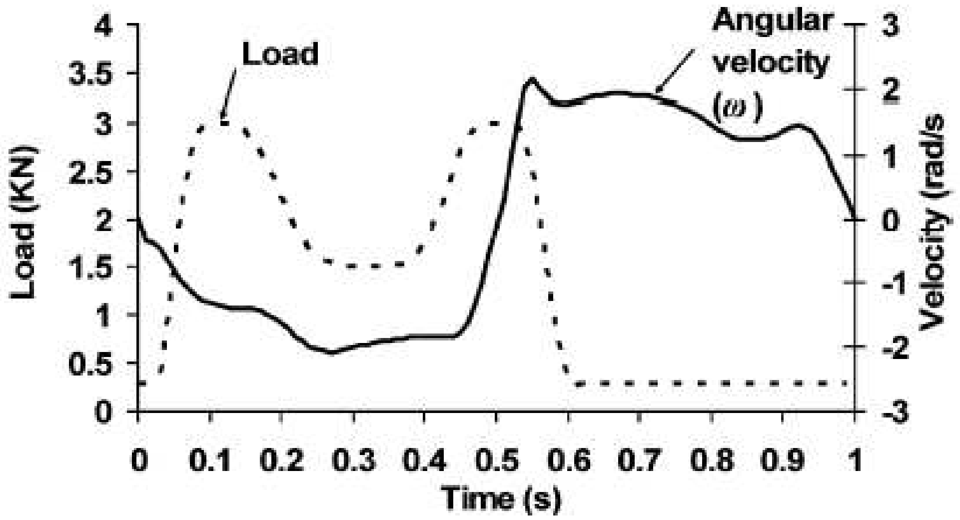


Figure 3.14. Typical variation in the transient load and angular velocity in hip joints during walking [Dowson *et al*, 2005].

3.6.3 Calibration process

The load cell mounted on the loading frame measures the load transmitted through the femoral head and acetabular cup (see Figure 3.4). A test load cell transducer was used to calibrate the load cell. An automatic load calibration mode in the ProSim Friction Simulator software is used to compute the calibration constants required to calculate the measured force against known forces of the test load cell. The air pressure valve is then opened at 5 positions from a zero value (close valve), to a maximum value (fully open valve). At each of these positions, axial force is applied to the simulator's load cell, and the actual force measured by the test load cell recorded in the graphic user interface (GUI) for each of the valve positions. Finally, in order to adjust the demand load applied by the pneumatic actuator of the friction rig, the calculated calibration constants should be corrected in the GUI.

An in-built automatic friction calibration facility is used to calibrate the piezoelectric crystal transducer. In order to do this, a loading arm of known length was used (Figure 3.15), on which various test weights were applied. As each of the test weights are applied on one side of the loading arm, the friction torque is measured and the corresponding torque calibration constants were calculated. Similar calibration process was applied on the other side of the loading arm. In order to ensure correct torque measurements, the calculated calibration constants were subsequently modified in the GUI.

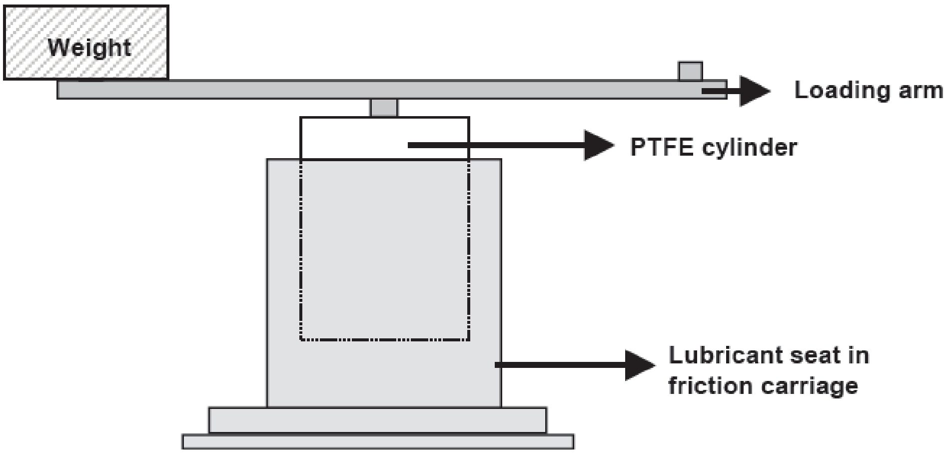


Figure 3.15. Schematic diagram of the friction torque loading arm.

3.6.4 PRE-TEST ALIGNMENT

As mentioned previously, the clearance between the centre of the femoral head and the base of the stem holder was determined by using the following equation:

$(99.43 - 72.91 - R_1)$

$(99.43 - 72.91 - 19) = 7.52$ mm

Where R_1 is the radius of the femoral head

99.43 = the distance (in mm) between the base and the foot of the stem holder jig

72.91 = the distance (in mm) between the centre of the femoral head to the base, which matches the centre of rotation of the motion arm and the base of the holder.

Furthermore, the femoral head height was then adjusted using various size slip gauges, to give a clearance between the top of the head and the roof of the rig.

The height of the acetabular cup was adjusted by placing a ceramic ball in the cup (Figure 3.16). The acetabular cup was then adjusted when the edge of the ceramic ball reached the tip of the depth gauge. The value for the depth gauge was determined by the following calculation:

$(R_2 - 2R_{ball} + 14.91)$

$(19 - 2(5) + 14.91 = 23.91$ mm

Where R_2 = the radius of the acetabular cup (mm)

R_{ball} = the radius of the ball bearing (mm)

14.91 = the distance (in mm) from the centre of rotation of the friction measuring system to the top edge of the lubricant seat.

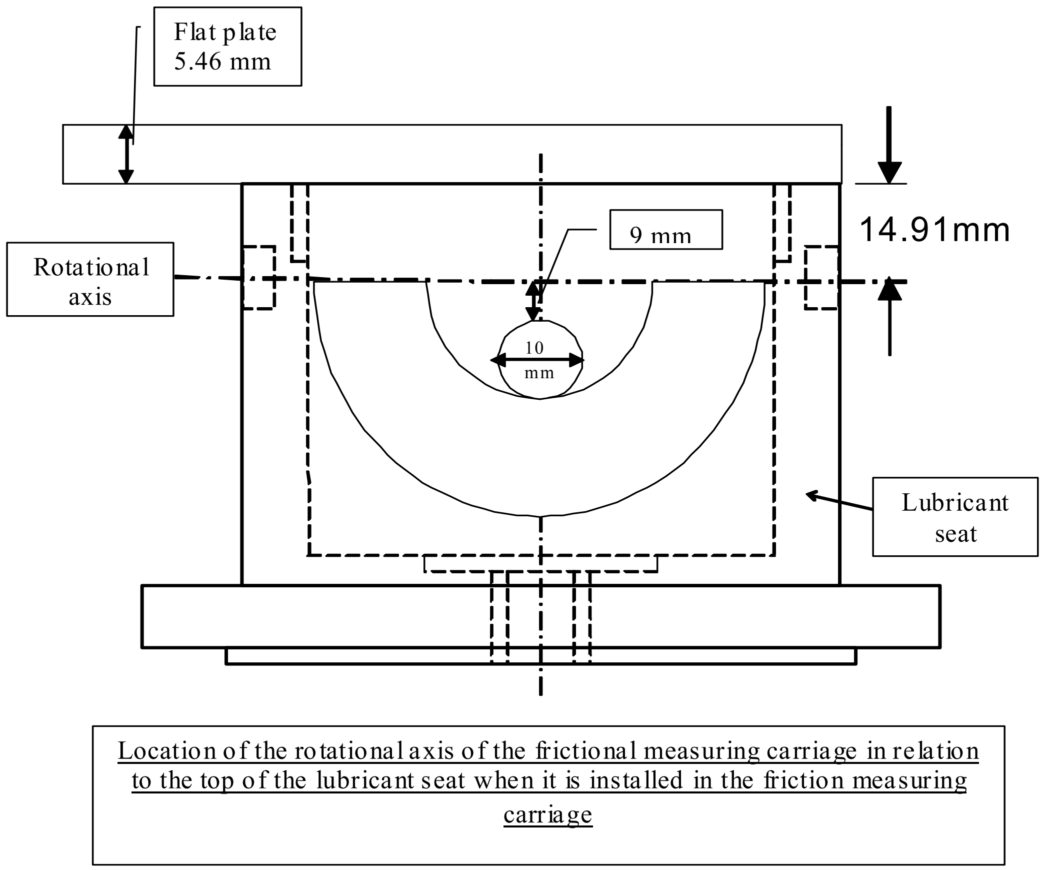


Figure 3.16. Schematic diagram of the friction measuring carriage.

3.7 PRE-TEST MEASUREMENTS

3.7.1 Surface roughness (R_a) measurements

Two dimensional measurements of the average surface roughness (R_a) were carried out at Smith & Nephew Orthopaedics Ltd. using a contacting Rank Taylor Hobson Talysurf profilometer with a Gaussian filter and a cut-off length of 0.25 mm. In this study the most commonly used surface roughness parameter, i.e. the average roughness (R_a), is therefore reported. R_a is defined as the arithmetic mean deviation of the surface height from the mean line through the profile. The average surface roughness for the 50 mm BHR devices are given in Table 3.1.

Table 3.1. Average surface roughness measurements of the 50 mm BHR devices.

Components	Average surface roughness (µm)
Cup	**0.011**
Cup	**0.010**
Head	**0.009**
Head	**0.009**

3.8 METAL-ON-METAL STRIBECK ANALYSIS AND CALCULATIONS

In order to generate Stribeck curves for the implants used in this study, Sommerfeld number calculated for each lubricant using Equation 1.9. The entraining velocity in the following calculations is taken from the average sliding speed at 2000 N.

Example 1:

$\eta = 0.0013$ Pa s

$u = 0.02$ m/s

$r = 25 \times 10^{-3}$ m

$L = 2000$ N

$$z = \frac{\eta u r}{L} = \frac{(0.0013)\times(0.02)\times(25\times(10^{(-3)}))}{2000} = 3.25\times10^{-10}$$

Example 2:

$\eta = 0.014$ Pa s

$u = 0.02$ m/s

$r = 25 \times 10^{-3}$ m

$L = 2000$ N

$$z = \frac{\eta u r}{L} = \frac{(0.014) \times (0.02) \times (25 \times (10^{(-3)}))}{2000} = 3.5 \times 10^{-9}$$

Example 3:

$\eta = 0.19$ Pa s

$u = 0.02$ m/s

$r = 25 \times 10^{-3}$ m

$L = 2000$ N

$$z = \frac{\eta u r}{L} = \frac{(0.193) \times (0.02) \times (25 \times (10^{(-3)}))}{2000} = 4.83 \times 10^{-8}$$

A Stribeck curve for each of the implants was generated from the above calculated Sommerfeld numbers against the experimental friction factors, as given in chapter four.

3.9 EXPERIMENTAL PROTOCOL

- **Cleaning regime for metal head and cup**

Before fixing the bearing at jig on the machine, all bearings were pre-washed and cleaned in three steps with different solutions to eliminate any friction error. First, cup and ball were bathed in a mixture of tap water and detergent in a clean beaker using an ultrasonic cleaner and ran for 10 minutes. After 10 minutes, the bearings were carefully rinsed with tap water. In the second step of cleaning, the bearings were submerged in the presence of methanol for 10 minutes using an ultrasonic cleaner and finally, the bearings were submerged in distilled water for the last 10 minutes in a beaker using an ultrasonic cleaner. Bearings were cleaned and dried by using soft wipes and ready to fix in the place of jig on the machine.

- **Component Alignment**

The alignment of the components explained in (section 3.6.4). The femoral head stem is stabilized on the femoral stem holder with the use of a screw. The acetabular cup must be placed in the cup holder (Figure 3.17) with the assistance of O-rings and an alignment screw at the base of the cup holder. The screw, as with the femoral head, allows the acetabular cup to be either lowered or raised within the cup holder.

Figure 3.17. Assembling Metal head and cup on the Prosim Hip Friction simulator machine.

After the acetabular cup and cup holder are secure, they are placed in the friction carriage at the base of the Prosim simulator. Also, the femoral head holder must be affixed to the superior pendulum arm (Figure 3.17). Lastly, alignment of the femoral head with the acetabular cup must be ensured. With the alignment of the head and cup there is alignment of the centres of rotation of the superior pendulum arm and the hydrostatic bearings. If these are aligned properly, a metal rod may be passed, with ease, through both sides of the bearings of the superior pendulum arm to the carriage friction state.

- **Machine set –up procedure**

Friction measurements were made in the 'stable' part of the cycle at 2000N and thus the loading cycle was set at maximum and minimum loads of 2000N and 400N, respectively. In the flexion/extension plane, an oscillatory harmonic motion of amplitude ±24°. was applied to the femoral head with a frequency of 1Hz in a period of 1.2s and all measurements are controlled via PC. By accessing the test option on the Prosim simulator program the edit option becomes available. Bearings being cleaned between each test and fresh lubricant being used for each test. Each test was completed after 127 cyclic loadings lasting 127 seconds. All data's generated by the Prosim Simulator supports the plotting of the Stribeck curve *z*, as well as the out- comes of the frictional coefficient (*f*).

- Lubricant viscosities measured by using Anton Paar Physica MCR 301 Viscometer.

CHAPTER FOUR

4.0 Results and Discussion

4.1 Friction factor results for the S&N BHR devices using BS+CMC with different viscosities

Figures 4.1–4.6 are the graphs of friction factor versus diametral clearance for all the six joints having different diametral clearance and using BS+CMC as lubricant with various viscosities. Standard error (SE) for all friction measurements obtained were negligible (~0.0001) and was not significant. Table 4.1 gives the actual friction factor values for all the joints and different viscosities.

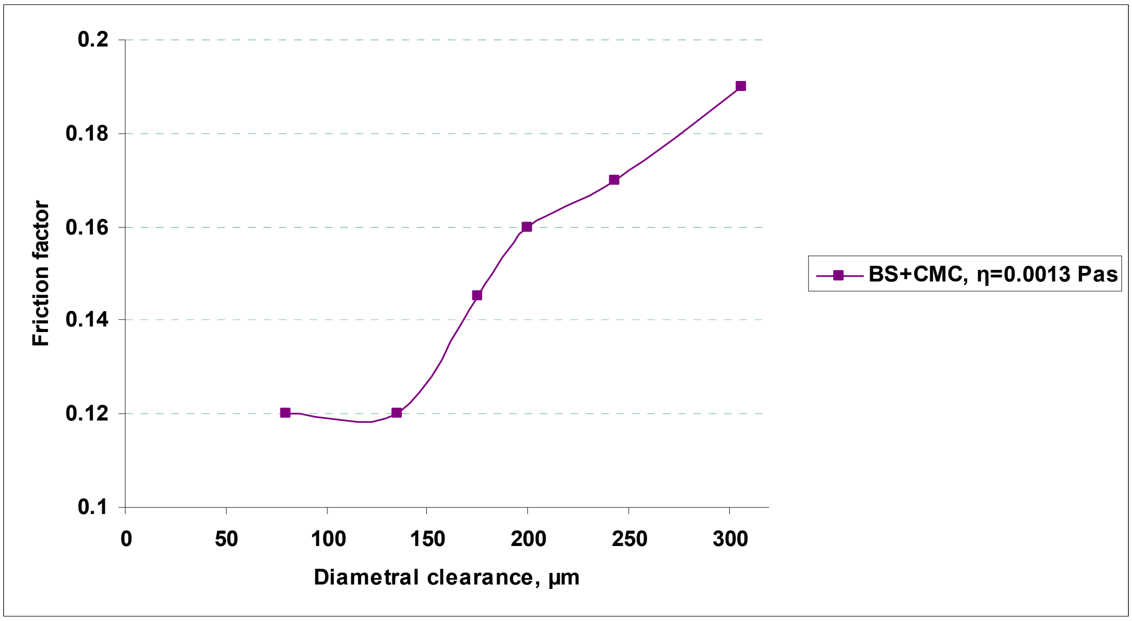


Figure 4-1. Friction factor versus diametral clearance for BS+CMC lubricant with η=0.0013 Pas before deflection.

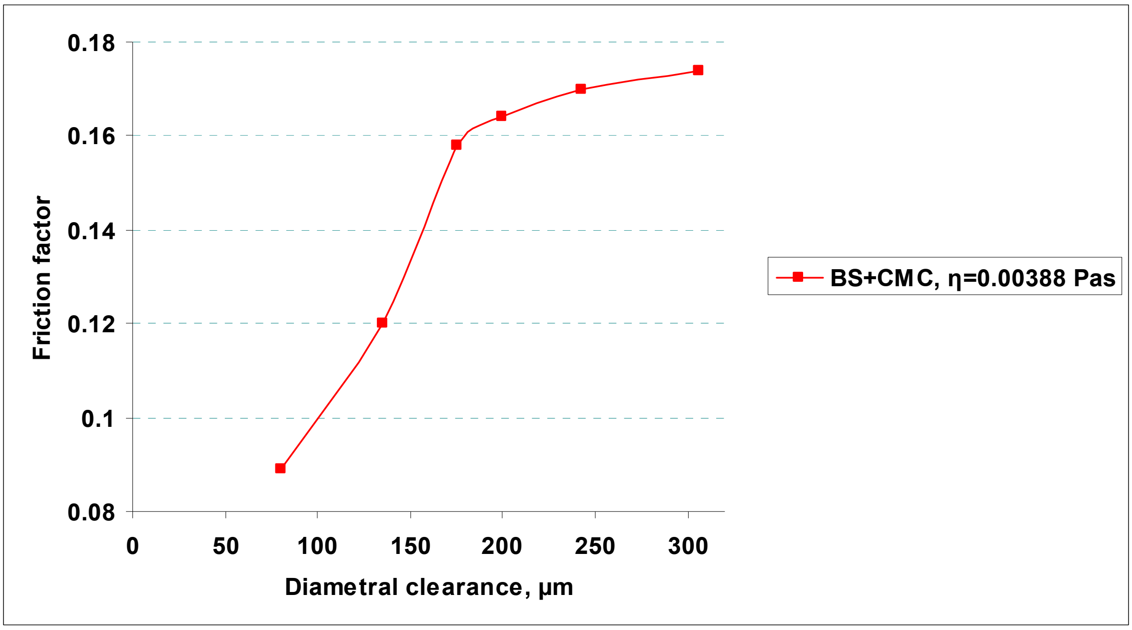


Figure 4-2. Friction factor versus diametral clearance for BS+CMC lubricant with η=0.00388 Pas before deflection.

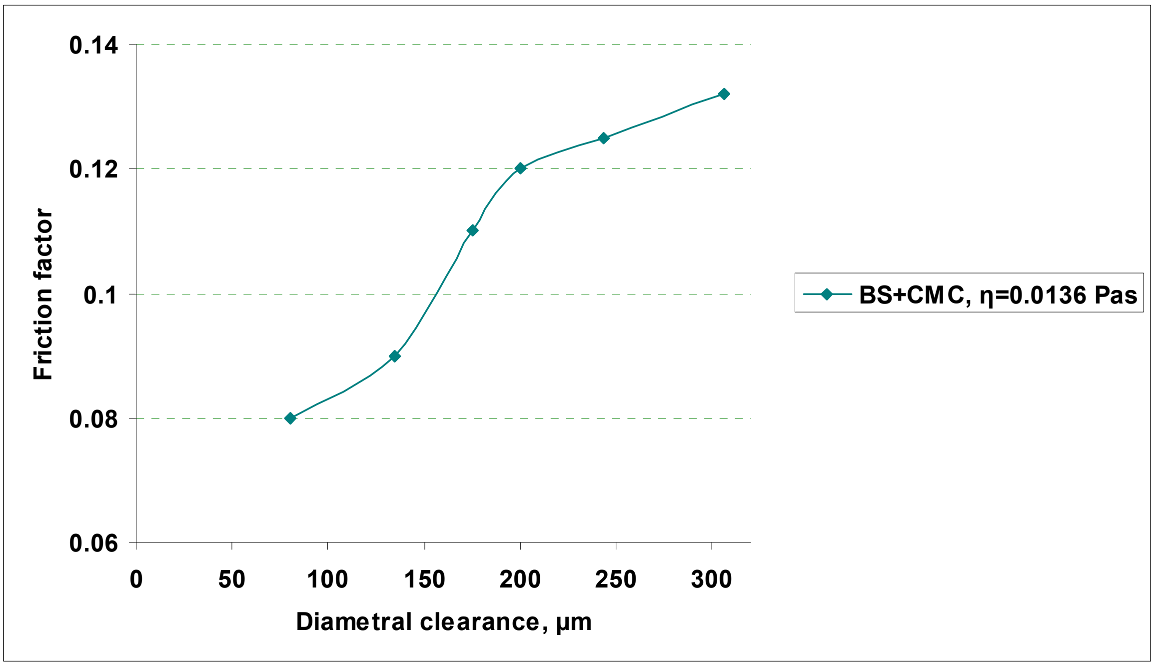


Figure 4-3. Friction factor versus diametral clearance for BS+CMC lubricant with η=0.0136 Pas before deflection.

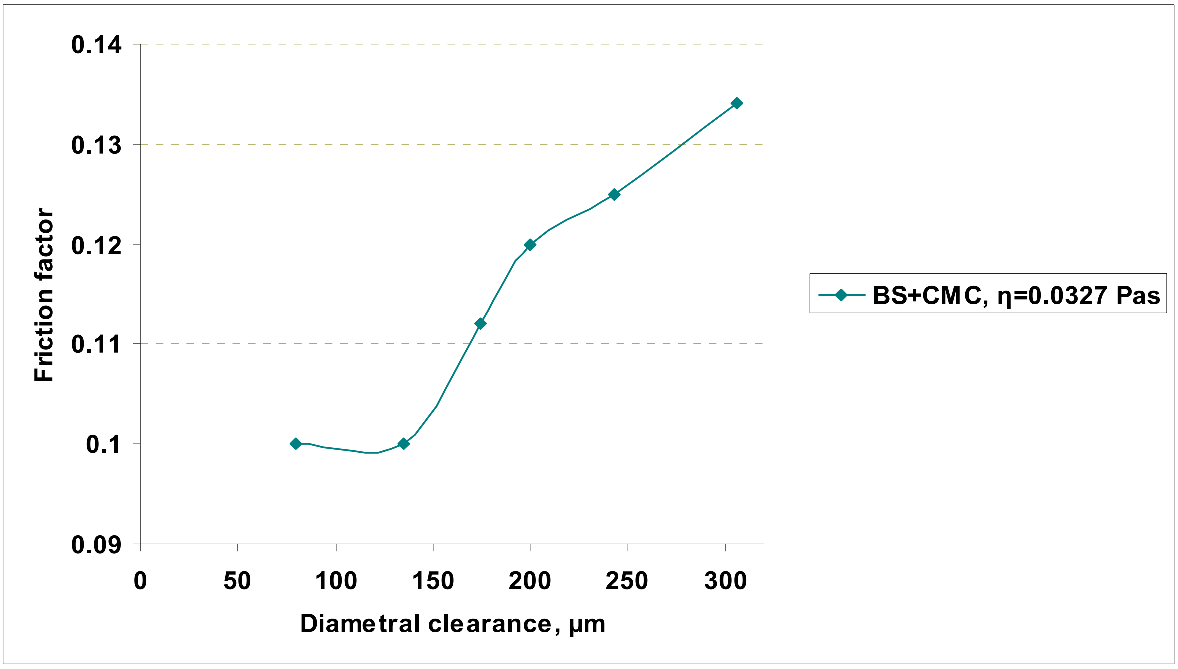


Figure 4-4. Friction factor versus diametral clearance for BS+CMC lubricant with η=0.0327 Pas before deflection.

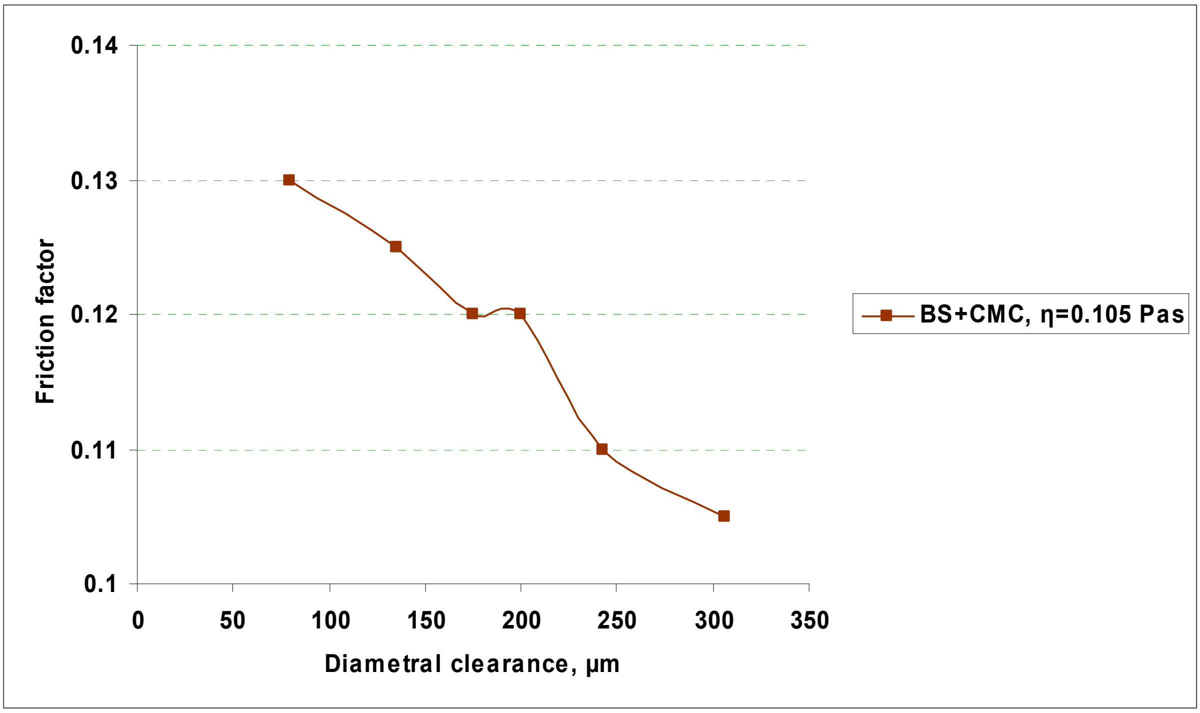


Figure 4-5. Friction factor versus diametral clearance for BS+CMC lubricant with η=0.105 Pas before deflection.

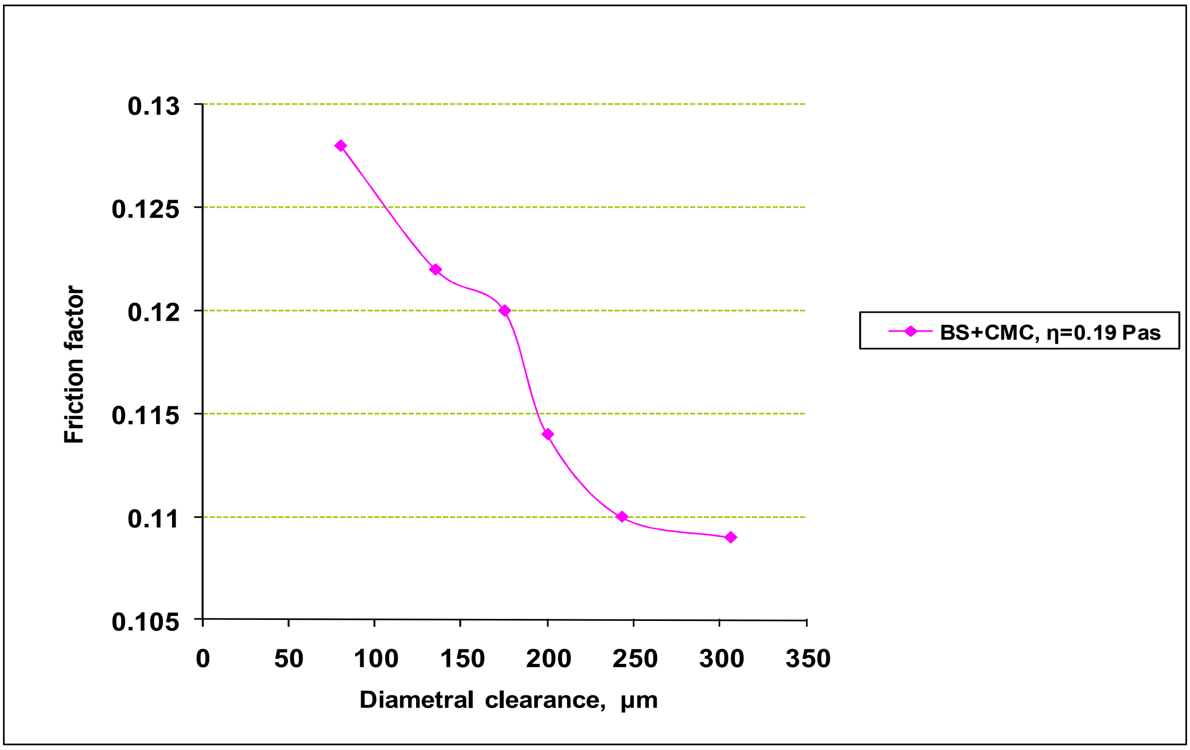


Figure 4-6. Friction factor versus diametral clearance for BS+CMC lubricant with η=0.19 Pas before deflection.

Table 4.1: Friction factors for different diametral clearance (80-306µm) using BS+CMC with various viscosities.

Reflect descending viscosity values	BS+CMC, η=0.00388 Pas	BS+CMC, η=0.0136 Pas	BS+CMC, η=0.19 Pas	BS+CMC, η=0.0013 Pas	BS+CMC, η=0.0327 Pas	BS+CMC, η=0.105 Pas
80	0.089	0.08	0.128	0.12	0.1	0.13
135	0.12	0.09	0.122	0.12	0.1	0.125
175	0.158	0.11	0.12	0.145	0.112	0.12
200	0.164	0.12	0.114	0.16	0.12	0.12
243	0.17	0.125	0.11	0.17	0.125	0.11
306	0.174	0.132	0.1094	0.19	0.134	0.105

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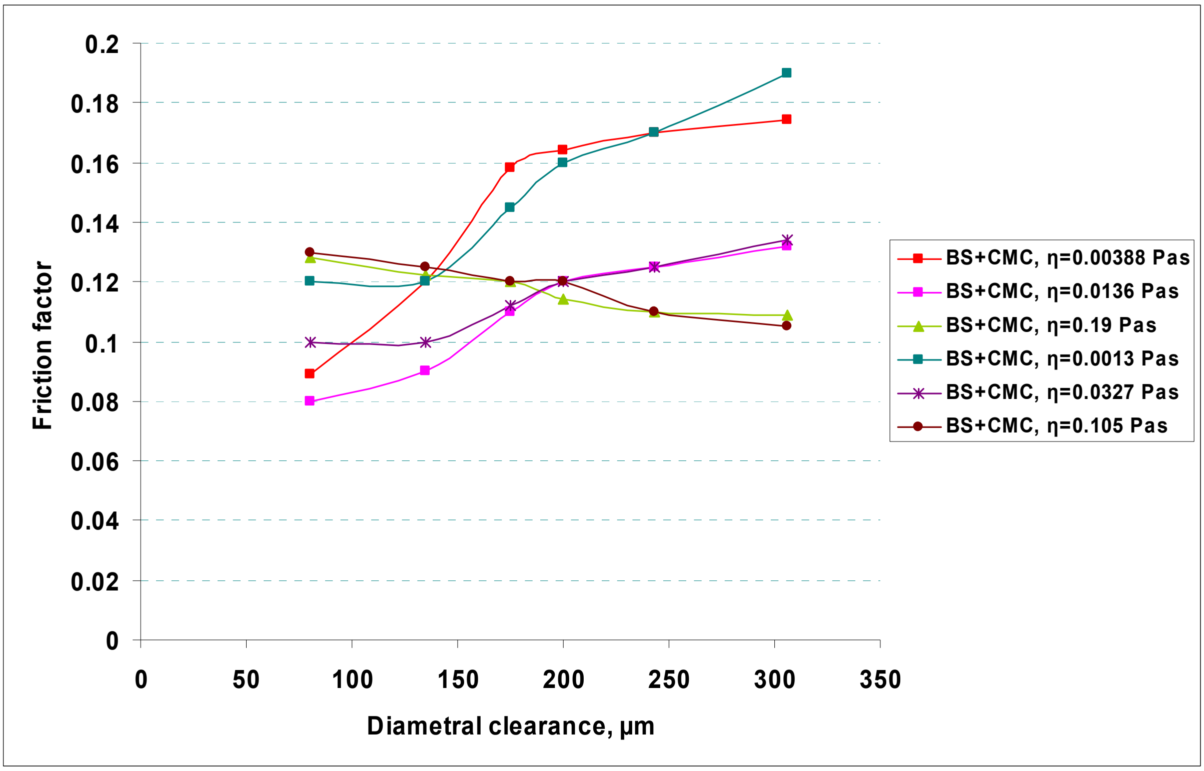

Figure 4-6a. Friction factors versus different diametral clearance (80-306μm) using BS+CMC with various viscosities.

Table 4.1 and Figures 4-1 to 4-6 give the average friction factors for different diametral clearances (80, 135, 175, 200, 243 and 306 μm) using aqueous solutions of bovine serum (25% BS +75% distilled water) with carboxymethyl cellulose (BS+CMC) as lubricants with various viscosities. The friction factors increased in the range 0.12-0.19, 0.08-0.175, 0.08-0.132 and 0.1-0.134 for viscosities 0.0013, 0.00388, 0.0136 and 0.0327 Pas, respectively, as given in Table 4.1, whereas friction factors decreased in the range 0.13-0.105 and 0.128-0.109 for viscosities of 0.105 and 0.19 Pas, respectively. This clearly suggests that higher clearances will cause less friction (and hence less wear) between the articulating surfaces of these large diameter S&N BHR MOM devices, for viscosities ≥0.1 Pas. On the other hand, BS+CMC lubricants with lower viscosities in the range η=0.0013 to η=0.0327 Pas showed the opposite effect, i.e. caused an increase in friction factor with increase in diametral clearance (from 80 to 306 μm). Also notable was that, the friction factors were generally higher for the BS+CMC lubricants with lower viscosities, e.g. for a viscosity of ~ 0.003 and 0.001 Pas the friction factor was in the range 0.19-0.08 as compared to that of 0.13-0.1 for a viscosity of 0.105 Pas (see Table 4.1).

According to Table 4.1 and related graphs (Figures 4.1-4.6), it can be seen clearly that friction factor in higher clearance bearings were found to be lower than those of the lower clearance bearings when a higher lubricant viscosity was used. Therefore, a significantly important finding is that the friction factors consistently decreased with increase in diametral clearance only for those lubricants with higher viscosities of 0.105 and 0.19 Pas (see Figures 4.1-4.6 and Table 4.1)

4.1.1 Stribeck Analysis

Table 4.2 gives the Sommerfeld numbers and friction factors for different diametral clearance and Figures 4-7 – 4-9 are the Stribeck plots, i.e. graphs of friction factor versus Sommerfeld number and the resulting Stribeck curves.

Table 4.2: Sommerfeld number and friction factors for various diametral clearances using BS+CMC as lubricants with different viscosities.

Sommerfeld Number, z ($x10^{-7}$)	80µm	135µm	175µm	200µm	243µm	306µm
0.00276	0.089	0.12	0.158	0.164	0.17	0.174
0.0076	0.12	0.12	0.145	0.16	0.17	0.19
0.0272	0.08	0.09	0.11	0.12	0.125	0.132
0.0654	0.1	0.1	0.112	0.12	0.125	0.134
0.21	0.13	0.125	0.12	0.12	0.11	0.105
0.38	0.128	0.122	0.12	0.114	0.11	0.109

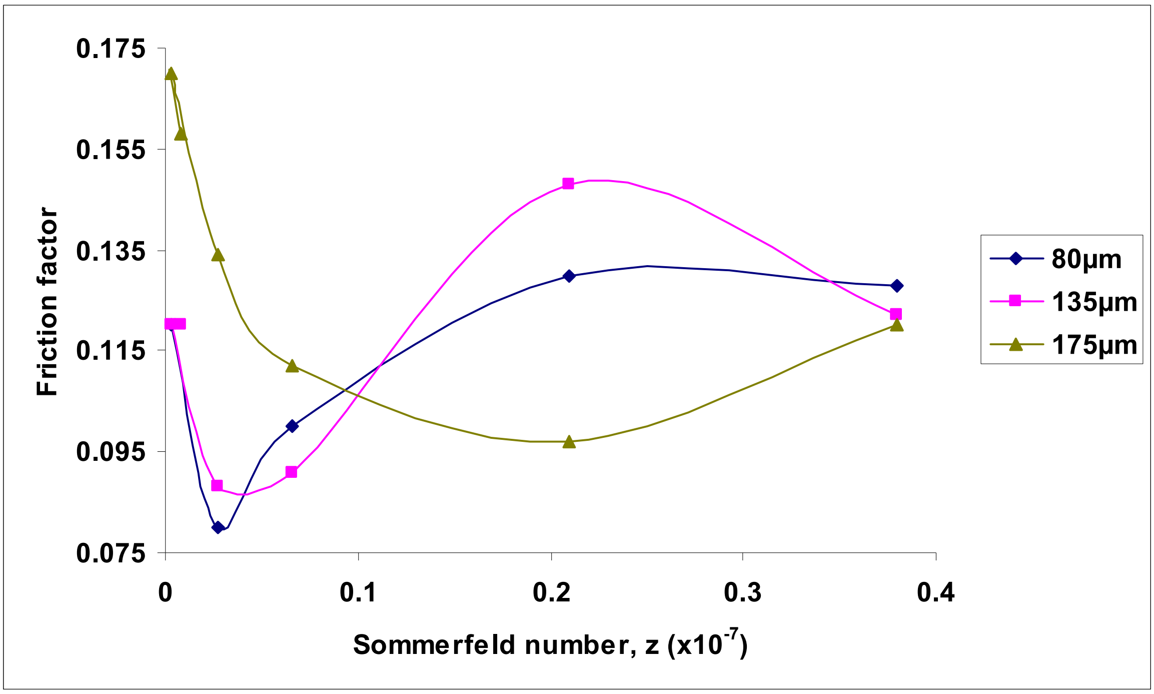

Figure 4-7. Friction factor versus Sommerfeld number for the 80, 135 and 175µm diametral clearance using BS+CMC as lubricant before deflection.

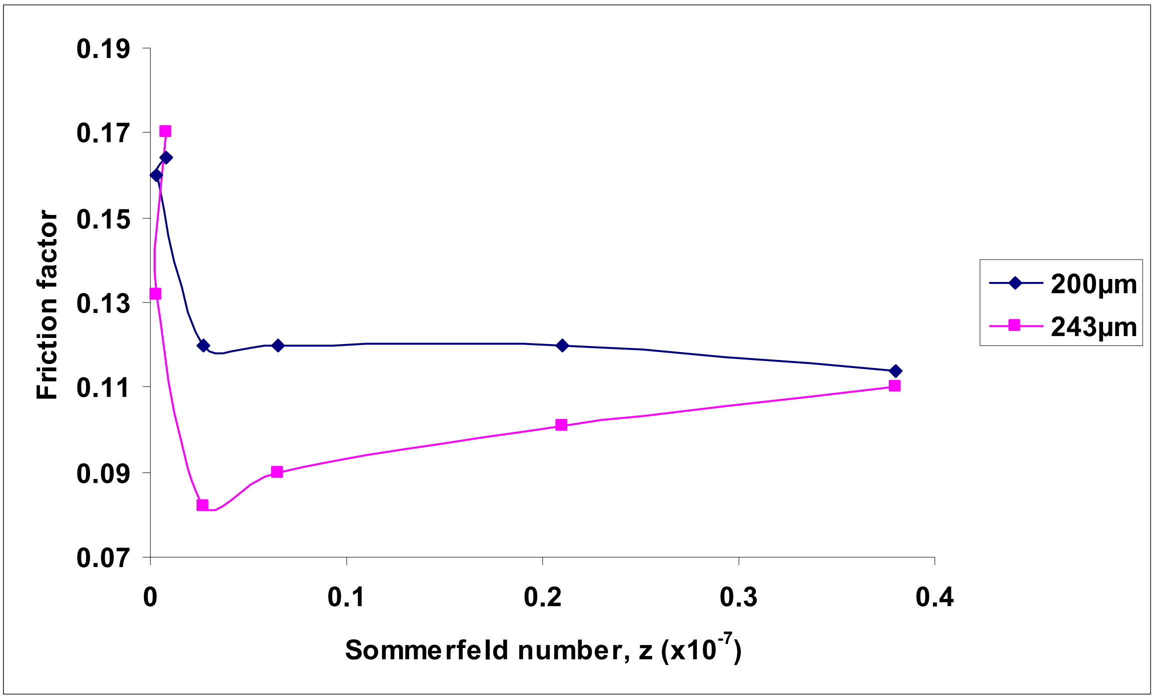

Figure 4-8. Friction factor versus Sommerfeld number for the 200 and 243µm diametral clearances using BS+CMC as lubricant before deflection.

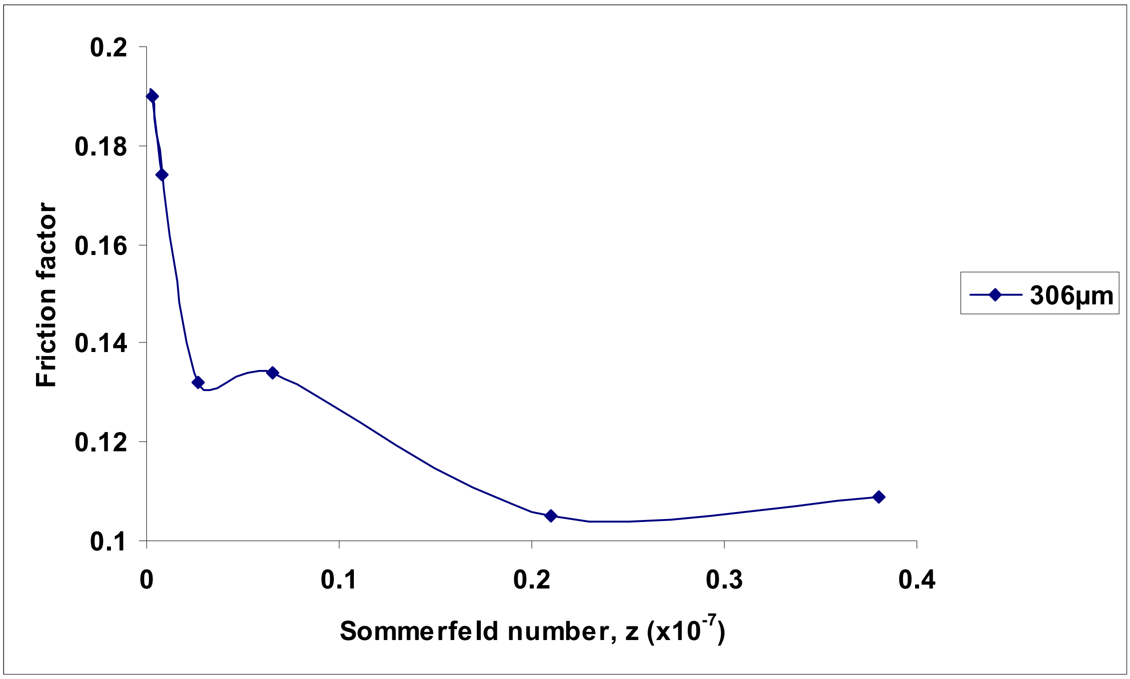

Figure 4-9. Friction factor versus Sommerfeld number for the 306μm diametral clearance using BS+CMC as lubricant before deflection.

The Stribeck curve in Figure 4-7 for the 80μm clearance shows an increasing friction factor from 0.08 to 0.128 as Sommerfeld number increases and then levels off, suggesting a transition from mixed to almost full fluid film lubrication during which the two bearing surfaces are completely separated by the lubricant film and that the frictional resistance is generated solely by the shear within the fluid.

The higher diametral clearances of 200, 243 and 306 μm did not show this transitional change and thus the mixed lubrication was the dominant mode (see Figures 4-8 and 4-9), during which the load is carried partly by the contact between the asperities of the bearing surfaces and also by the pressure generated within the lubricant. However, the friction factors were consistently lower for the higher clearances using the higher viscosity lubricants (η=0.105 and 0.19 Pas) indicating that these fluids with higher viscosities are effective in lowering the friction.

It has been reported [Scholes *et al*., 2000] that healthy synovial fluid would have a viscosity >0.03 Pas at a shear rate of 3000 s^{-1} and that rheumatoid fluid is likely to have a viscosity ≤ 0.005 Pas at the same shear rate. However, comparing the friction factors obtained in this work with those reported by others, e.g. 0.16-0.3 [Scholes *et al*., 2000] for the 28mm MOM bearings using similar lubricants (BS+CMC) with presence of proteins, of similar viscosities (0.001-0.154 Pas), it can be concluded clearly that the 50mm MOM S&N BHR prostheses have given lower friction factors (0.09-0.17 for η=0.00388 Pas, and 0.12-0.19 for η=0.0013 Pas which shows the advantages of having larger diameters over smaller diametral bearings.

4.2 Dynamic motion profiles for the S&N BHR devices using BS+CMC with different viscosities.

The dynamic loading cycles generated during the friction tests (for friction measurements) are plotted graphically in Figures 4.10 - 4.27. These are graphs of load, frictional torque and displacement (±24° oscillatory harmonic flexion- extension motion) versus the number of cycles (=127). It is to be noted that the friction factors were taken from the stable part of the cycle at 2000 N and thus the frictional torques were also from this part of the cycle which represents the normal loading cycle observed in human's body having a 12° angle of loading between the acetabular cup and the femoral head.

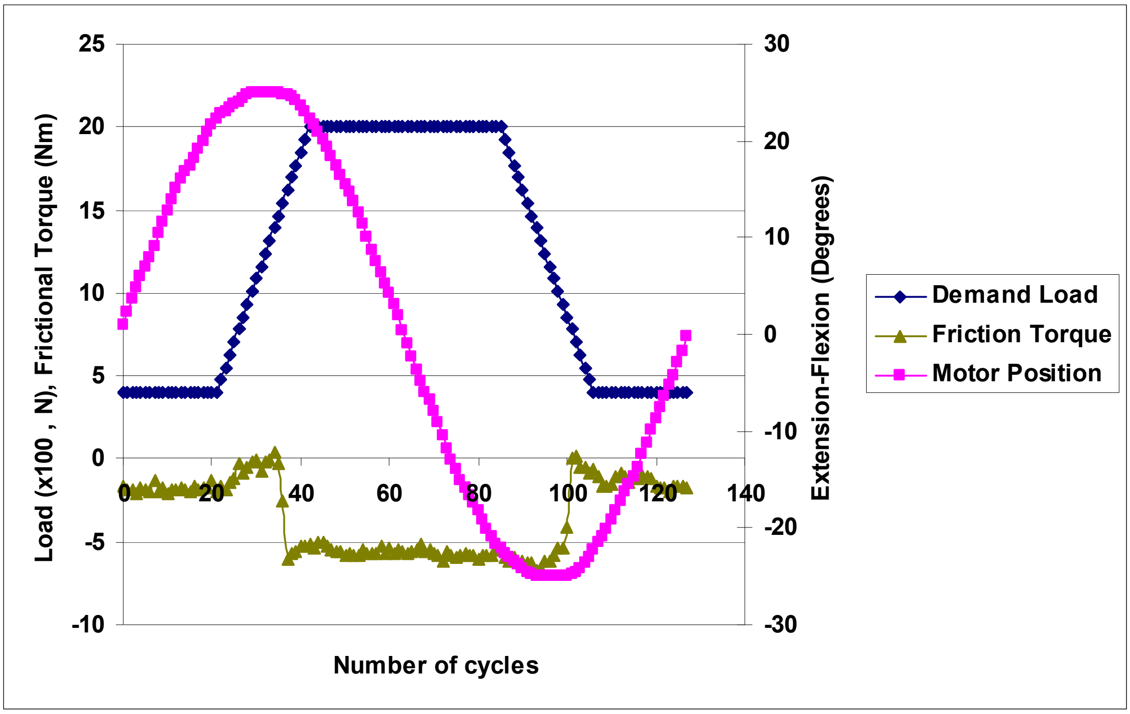

Figure 4-10. Friction Torque versus number of cycles for the 80 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0013 Pas) as lubricant before deflection.

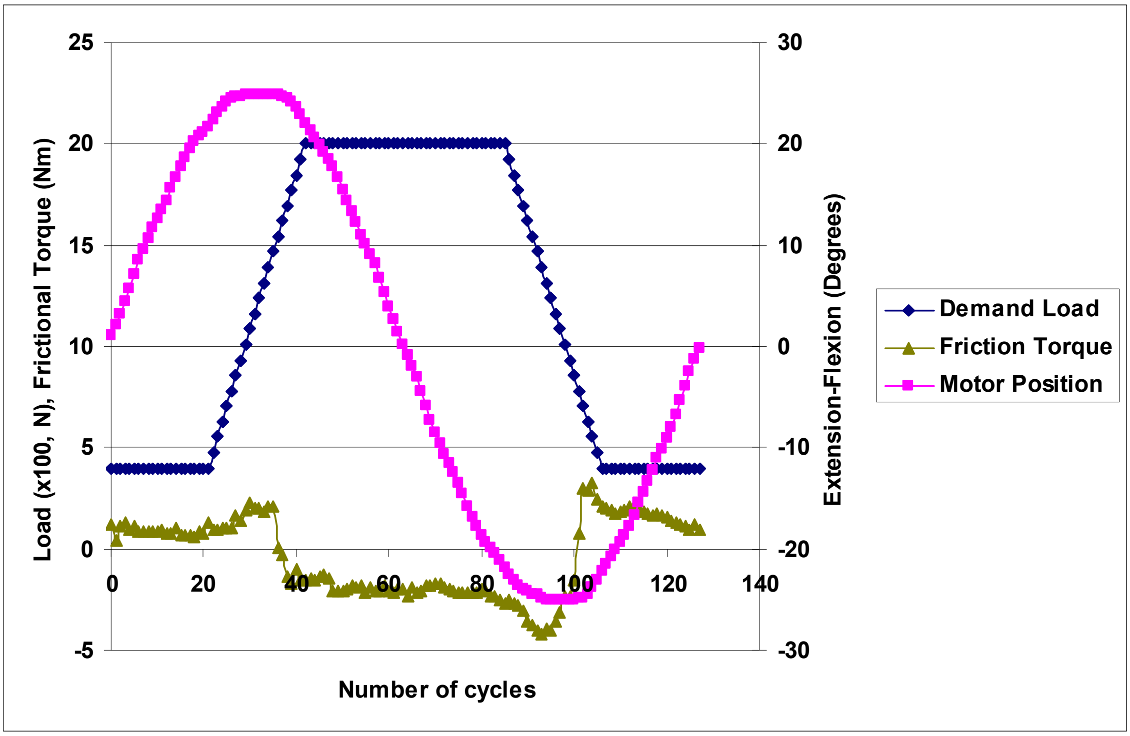

Figure 4-11. Friction Torque versus number of cycles for the 80 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.00388) Pas as lubricant before deflection.

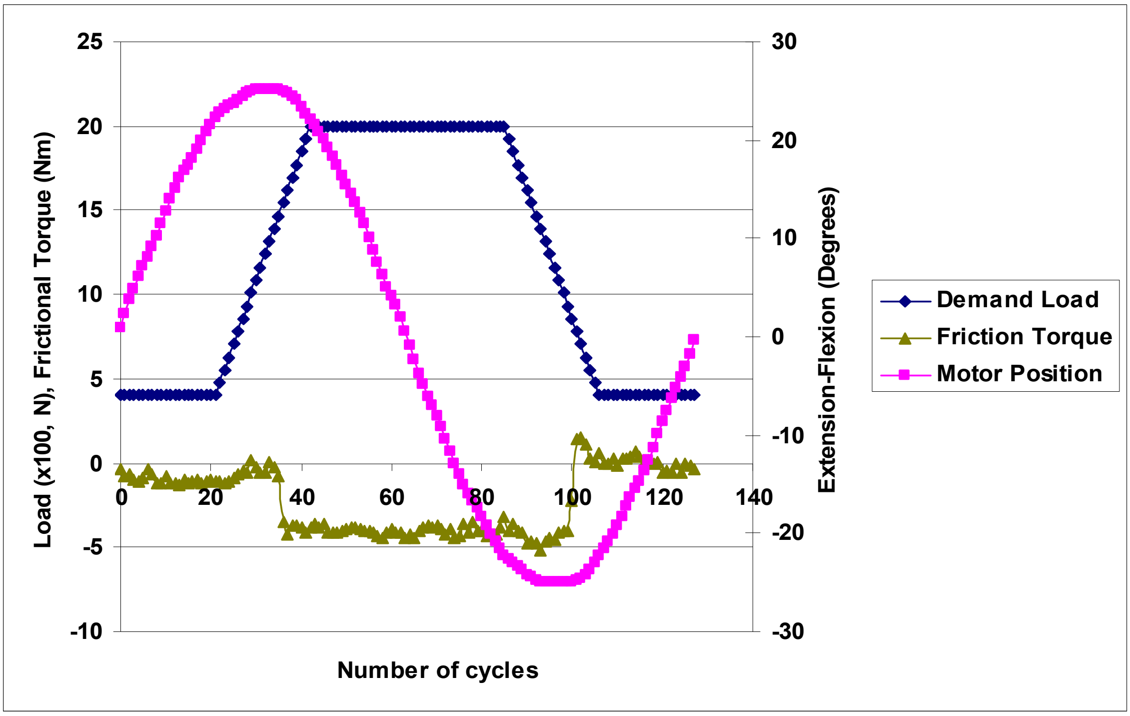

Figure 4-12. Friction Torque versus number of cycles for the 80 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0136 Pas) as lubricant before deflection.

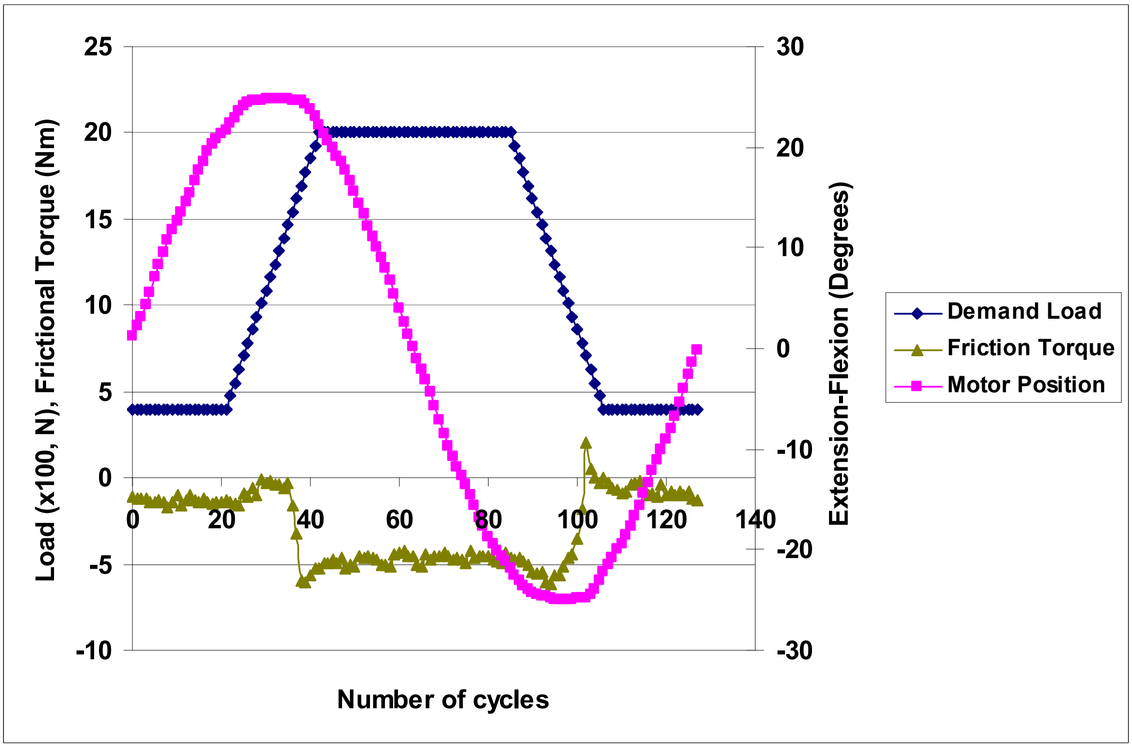

Figure 4-13. Friction Torque versus number of cycles for the 80 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0327 Pas) as lubricant before deflection.

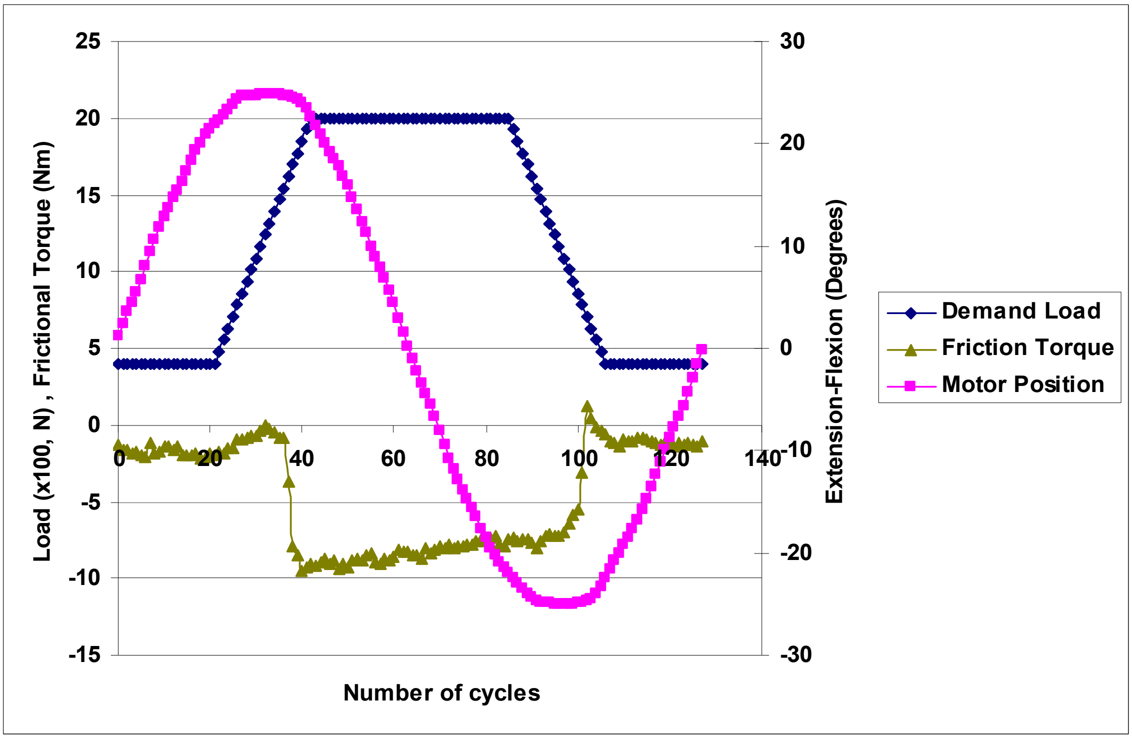

Figure 4-14. Friction Torque versus number of cycles for the 80 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.105 Pas) as lubricant before deflection.

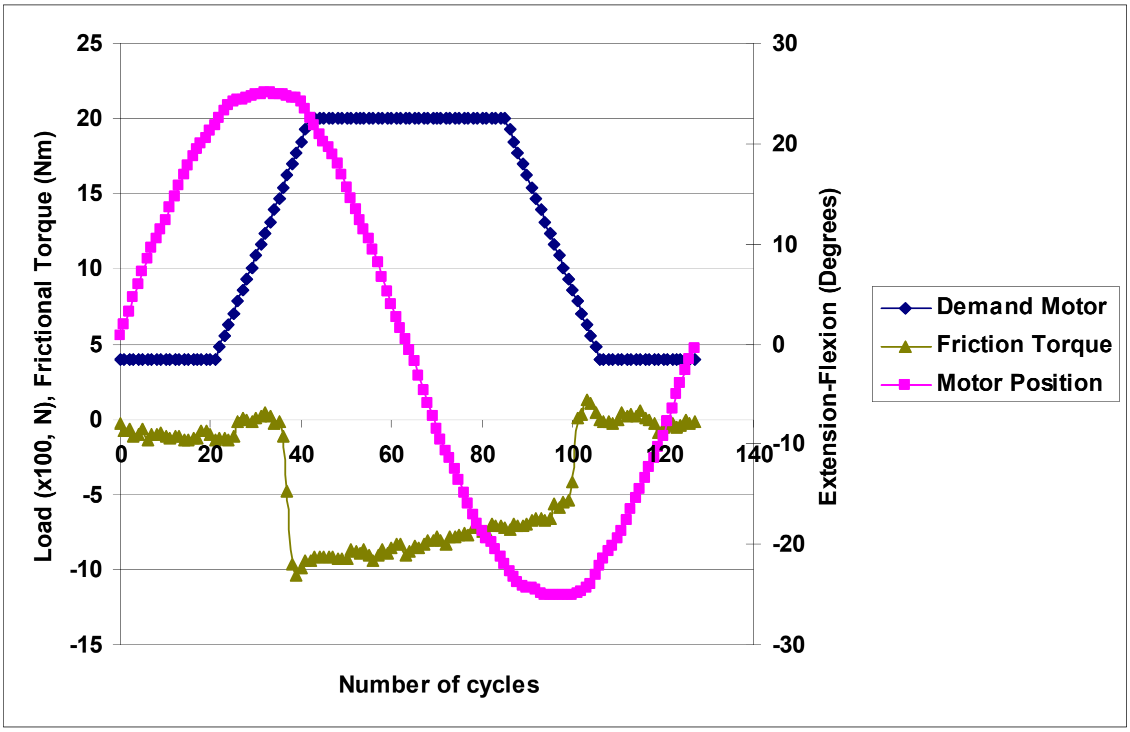

Figure 4-15. Friction Torque versus number of cycles for the 80 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.19 Pas) as lubricant before deflection.

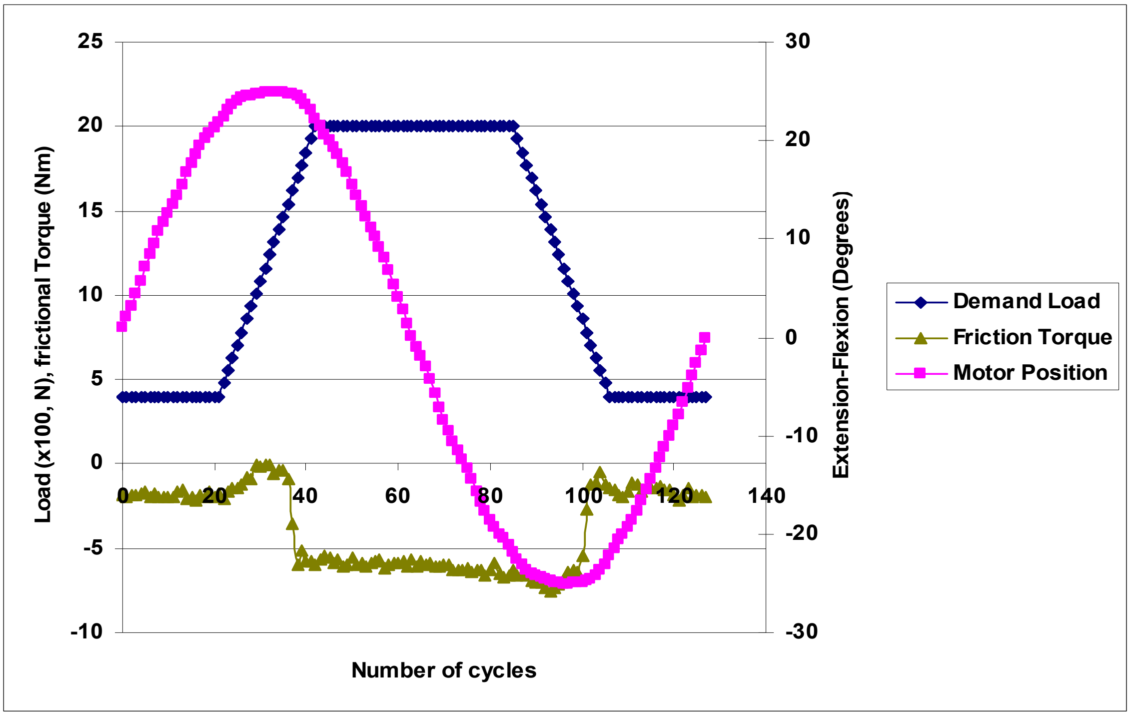

Figure 4-16. Friction Torque versus number of cycles for the 200 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0013 Pas) as lubricant before deflection.

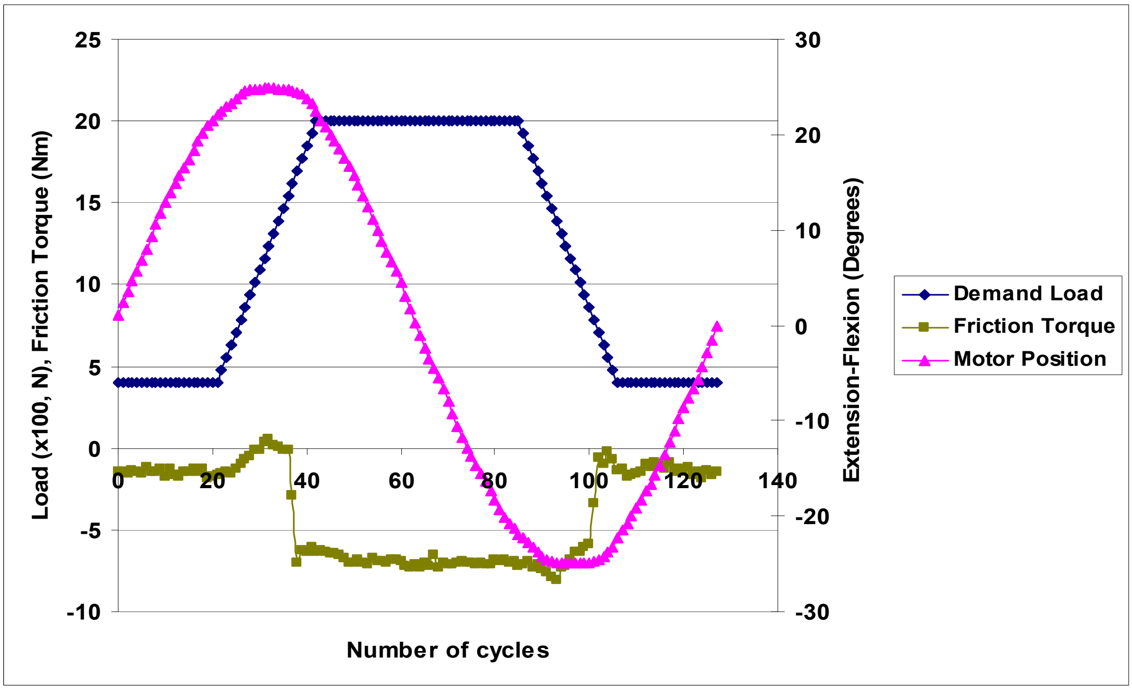

Figure 4-17. Friction Torque versus number of cycles for the 200 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.00388 Pas) as lubricant before deflection.

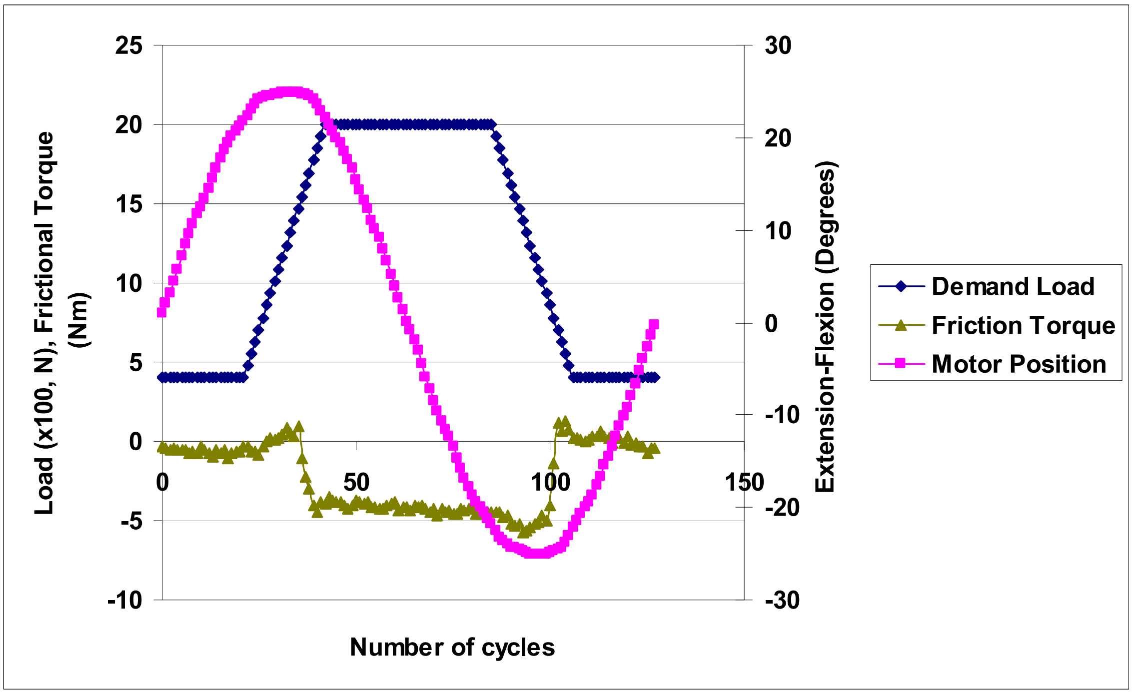

Figure 4-18. Friction Torque versus number of cycles for the 200 µm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0136 Pas) as lubricant before deflection.

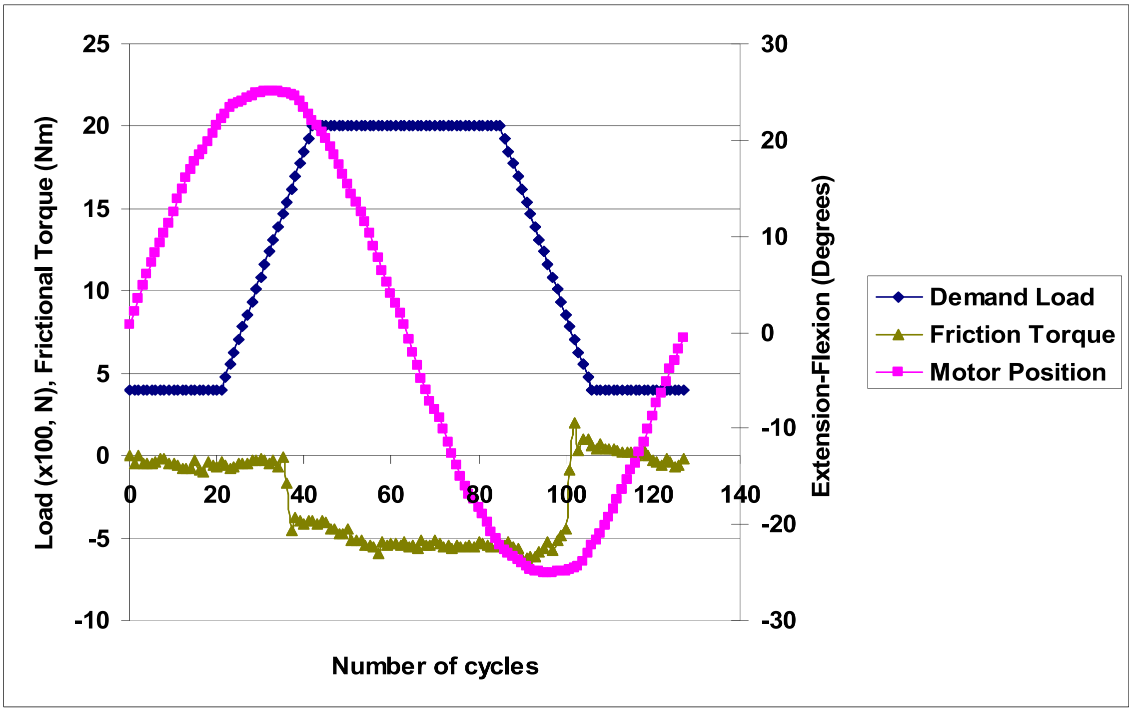

Figure 4-19. Friction Torque versus number of cycles for the 200 µm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0327 Pas) as lubricant before deflection.

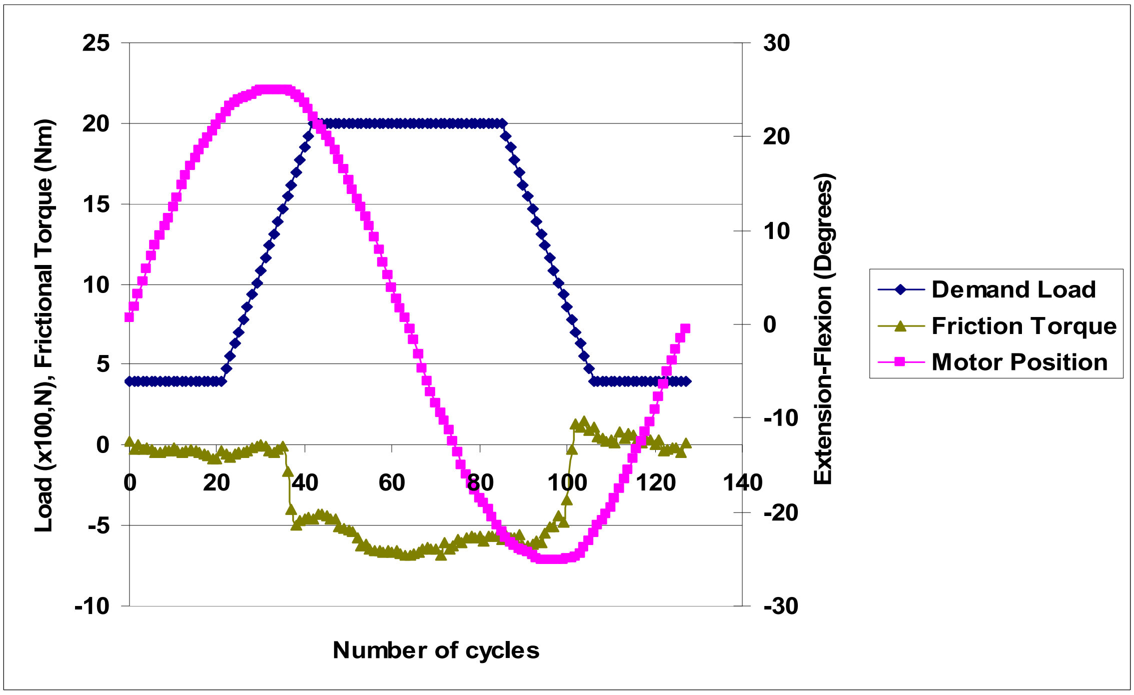

Figure 4-20. Friction Torque versus number of cycles for the 200 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.105 Pas) as lubricant before deflection.

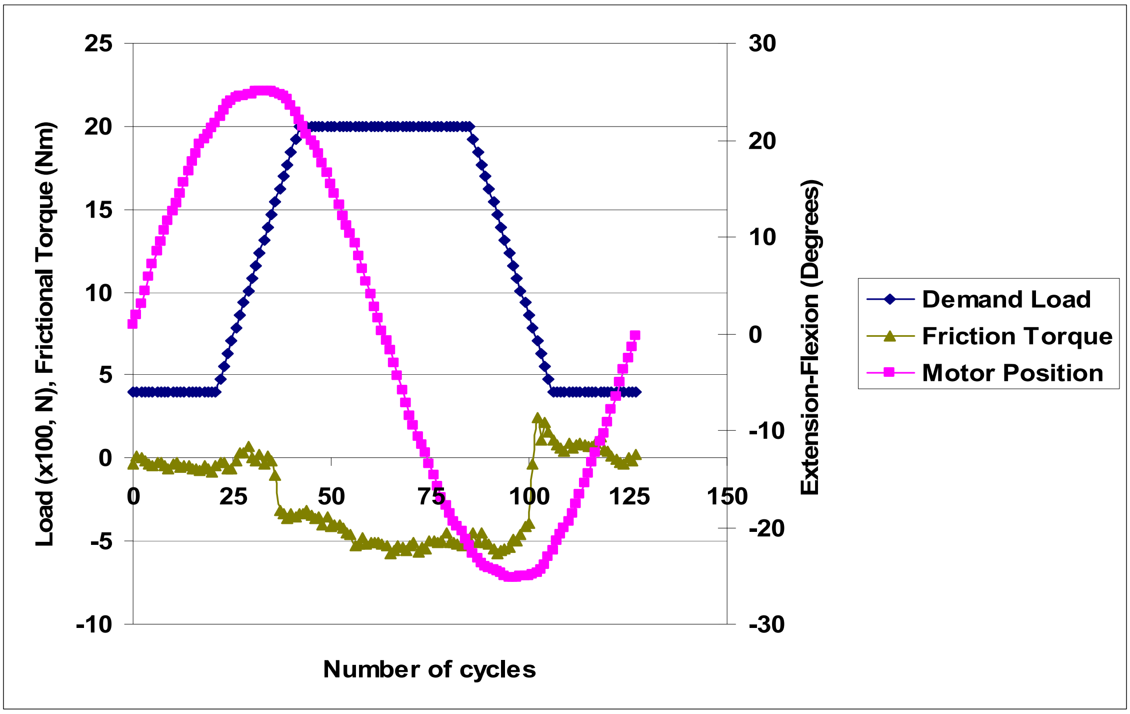

Figure 4-21. Friction Torque versus number of cycles for the 200 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.19 Pas) as lubricant before deflection

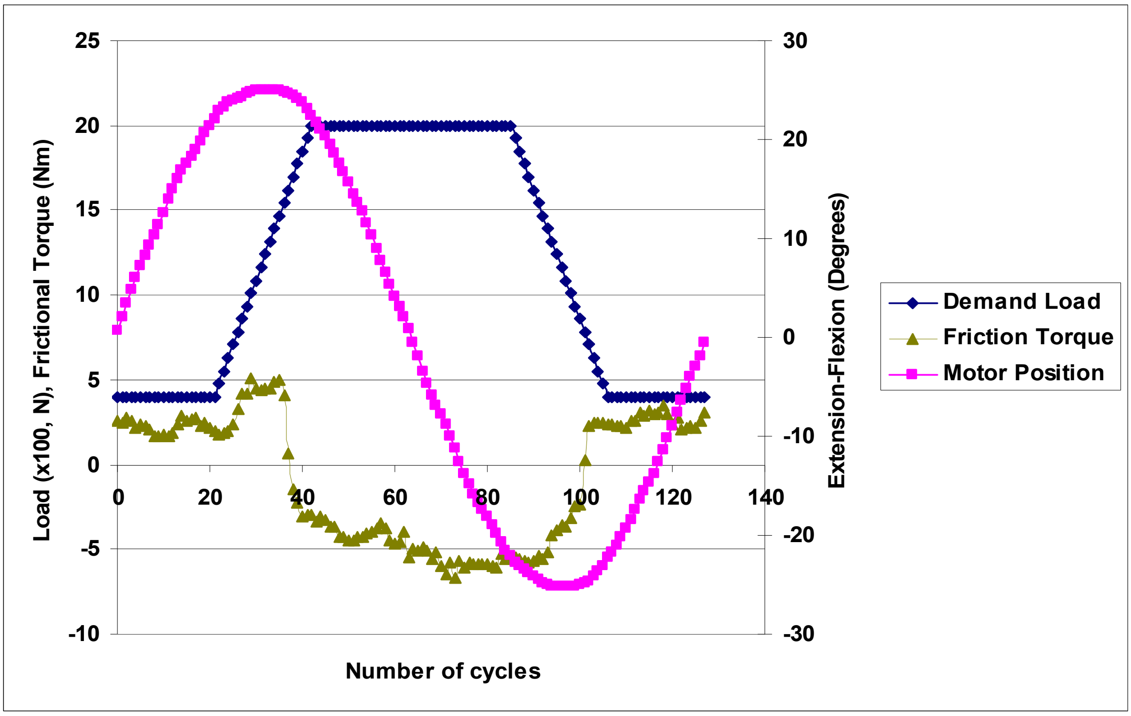

Figure 4–22. Friction Torque versus number of cycles for the 306 µm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0013 Pas) as lubricant before deflection.

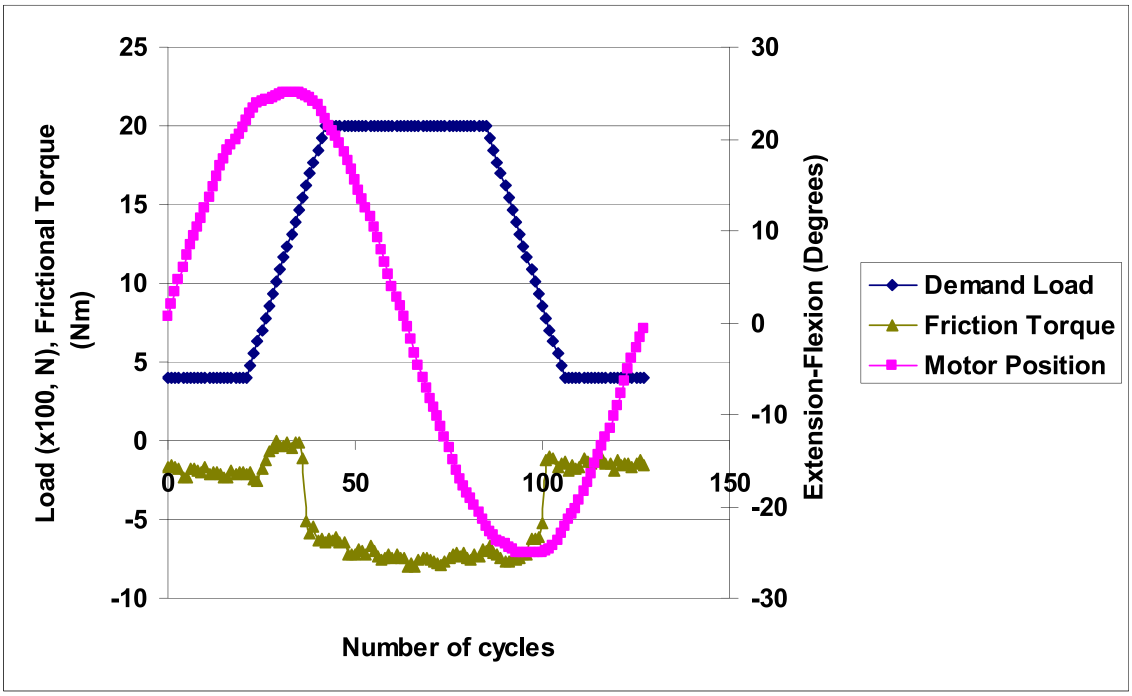

Figure 4-23. Friction Torque versus number of cycles for the 306 µm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.00388) Pas as lubricant before deflection.

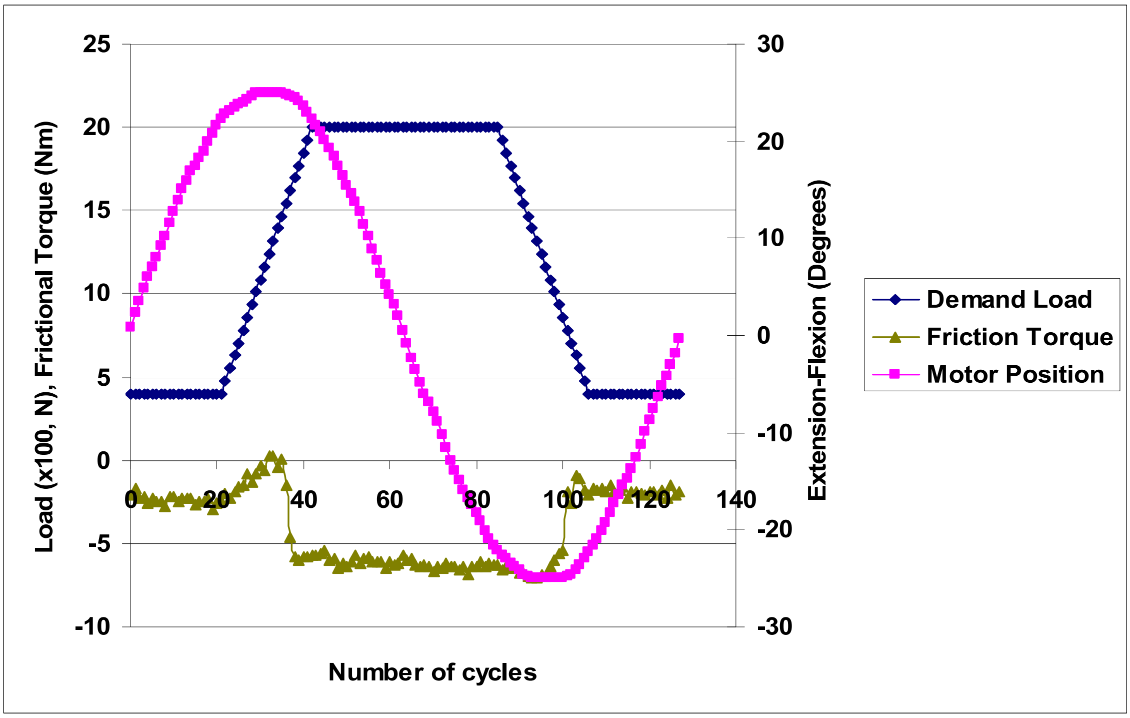

Figure 4-24. Friction Torque versus number of cycles for the 306 µm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0136 Pas) as lubricant before deflection.

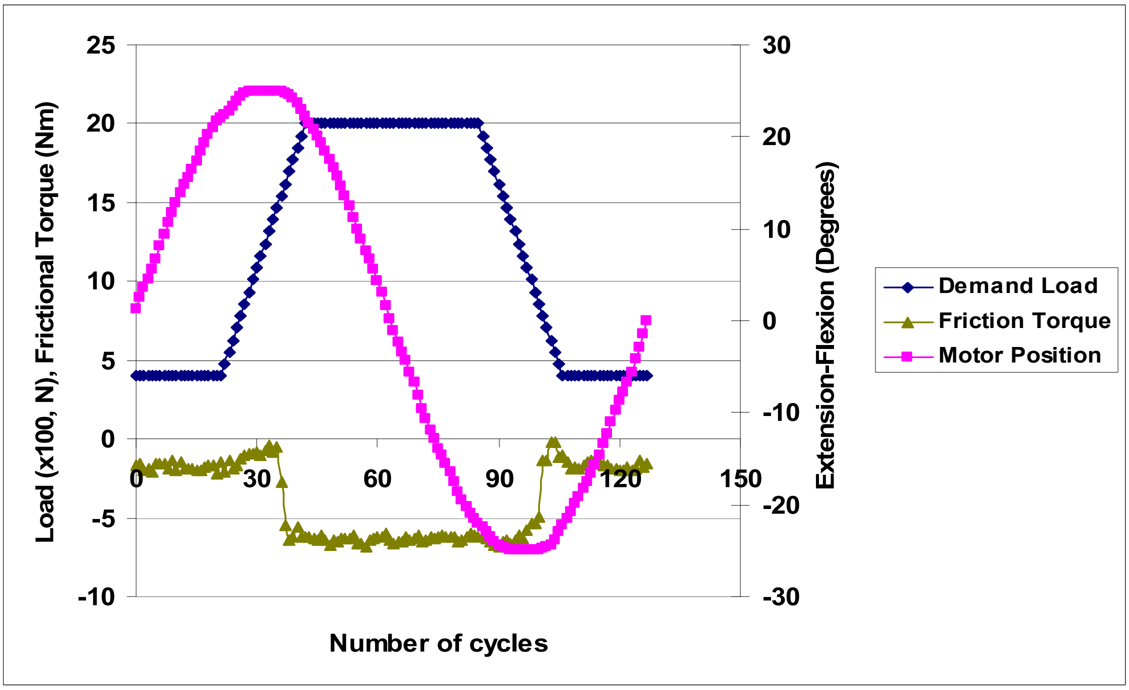

Figure 4-25. Friction Torque versus number of cycles for the 306 µm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.0327 Pas) as lubricant before deflection.

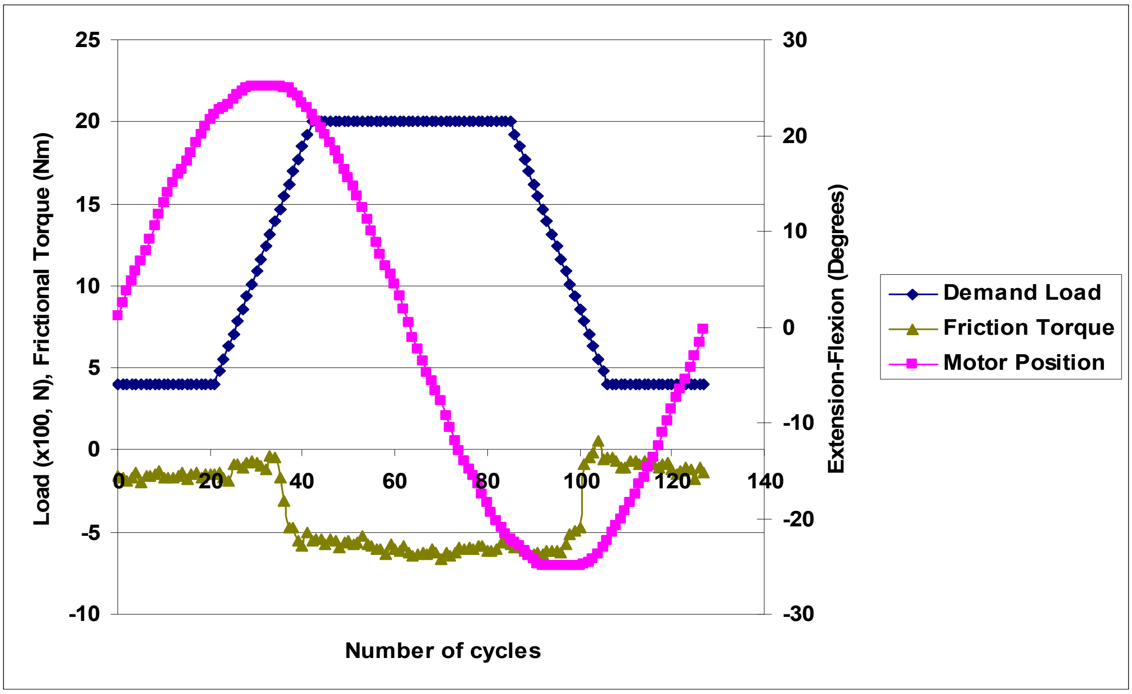

Figure 4-26. Friction Torque versus number of cycles for the 306 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.105 Pas) as lubricant before deflection.

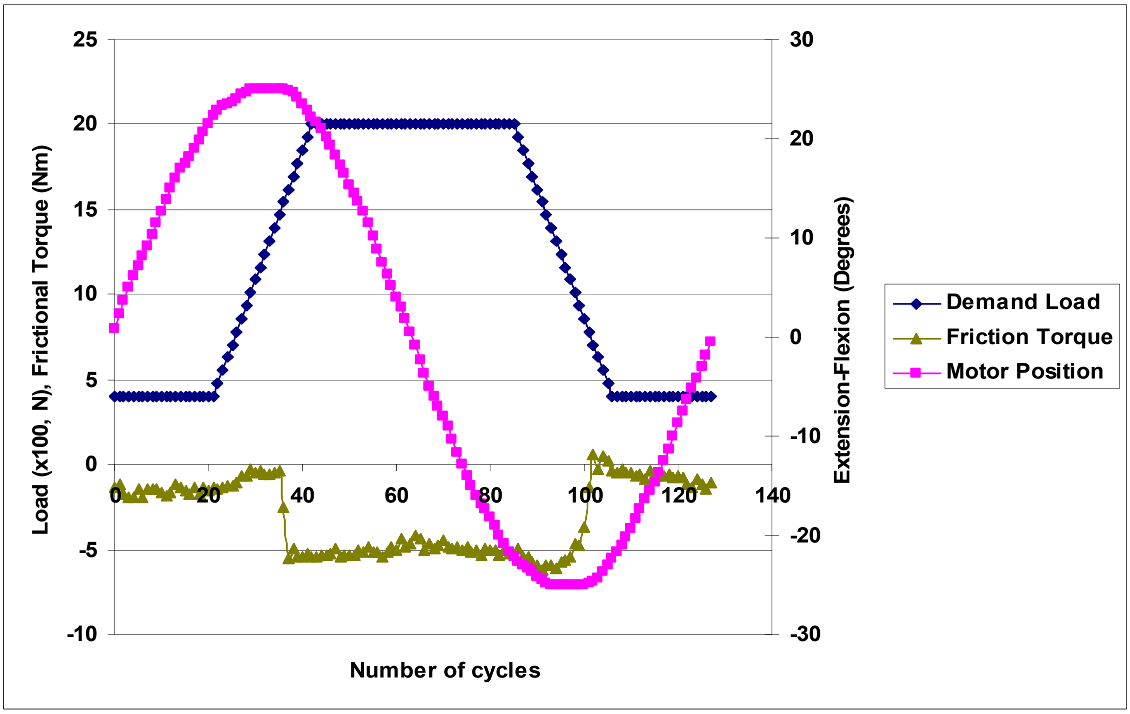

Figure 4-27. Friction Torque versus number of cycles for the 306 μm diametral clearance, 50mm BHR bearing using BS+CMC (η=0.19 Pas) as lubricant before deflection.

Table 4.3: Average frictional torque for diametral clearances of 80, 200 and 306 μm using BS+CMC lubricants with different viscosities.

BS+CMC Viscosity, η (Pas)	Friction Torque (Nm) for Diametral clearance of 80 μm	Friction Torque (Nm) for Diametral clearance of 200 μm	Friction Torque (Nm) for Diametral clearance of 306 μm
0.0013	**4.84**	**5.61**	**6.03**
0.00388	**1.97**	**6.93**	**7.22**
0.0136	**4.00**	**4.17**	**6.16**
0.0327	**4.72**	**5.18**	**6.33**
0.105	**8.34**	**5.92**	**5.92**
0.19	**8.37**	**4.75**	**5.02**

Table 4.3 shows the average friction torque produced during dynamic loading, i.e. friction tests. From Table 4.3 and Figures 4-10 to 4-27, it is clear that for viscosities of ≥0.1 Pas, friction torque increased from ~5.0 to ~8.3 Nm as diametral clearance decreased from 306 to 80μm likely due to higher contact between the bearing surfaces. The smaller torque in higher clearances and viscosities ≥0.1pas might also be because of bearing surfaces separated more efficiently by the more viscous lubricating film, and partly due to adsorbed protein from the bovine serum on the bearings causing lower friction torque (and lower friction factor) via protein rubbing against protein. However, these friction torques for all clearances were still within the reported safe range, i.e. no risk of dislocation or impaired fixation is expected for these torques. On the other hand, for viscosities <0.1Pas the friction torque decreased as diametral clearance decreased from 306 to 80μm depending on the viscosity of the lubricant used, i.e. lower viscosities resulted in lower torques especially for the diametral clearance of 80μm giving lower

frictional torques for all viscosities <0.1Pas. It is very interesting to note that similar trends were obtained for friction factors, i.e. depending on both viscosity and clearance, increasing as clearance increased for viscosities ≤0.1Pas, and decreasing as clearance increased for viscosities ≥0.1Pas, which are consistent with the friction torque results.

4.3 Friction factor results for the S&N BHR devices using BS+HA+ CMC with different viscosities

Figures 4.28–4.32 are the graphs of friction factor versus diametral clearance for all the five joints having different diametral clearance and using BS+HA+CMC as lubricant with various viscosities. Table 4.4 gives the actual friction factor values for all the joints and different viscosities.

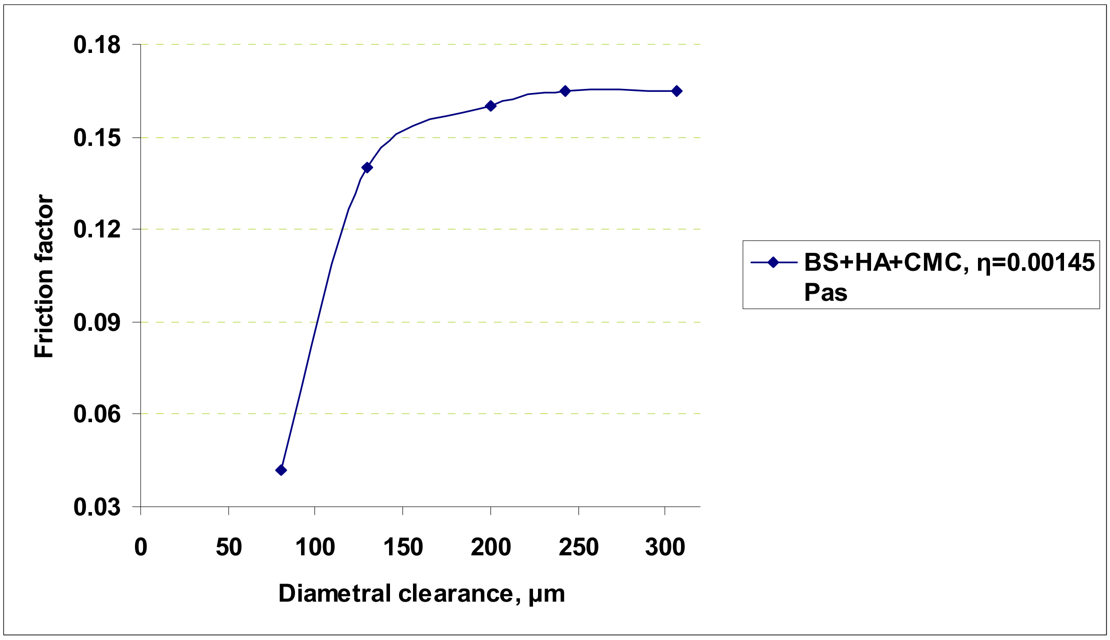

Figure 4-28. Friction factor versus diametral clearance for BS+HA+CMC lubricant with η=0.00145 Pas before deflection.

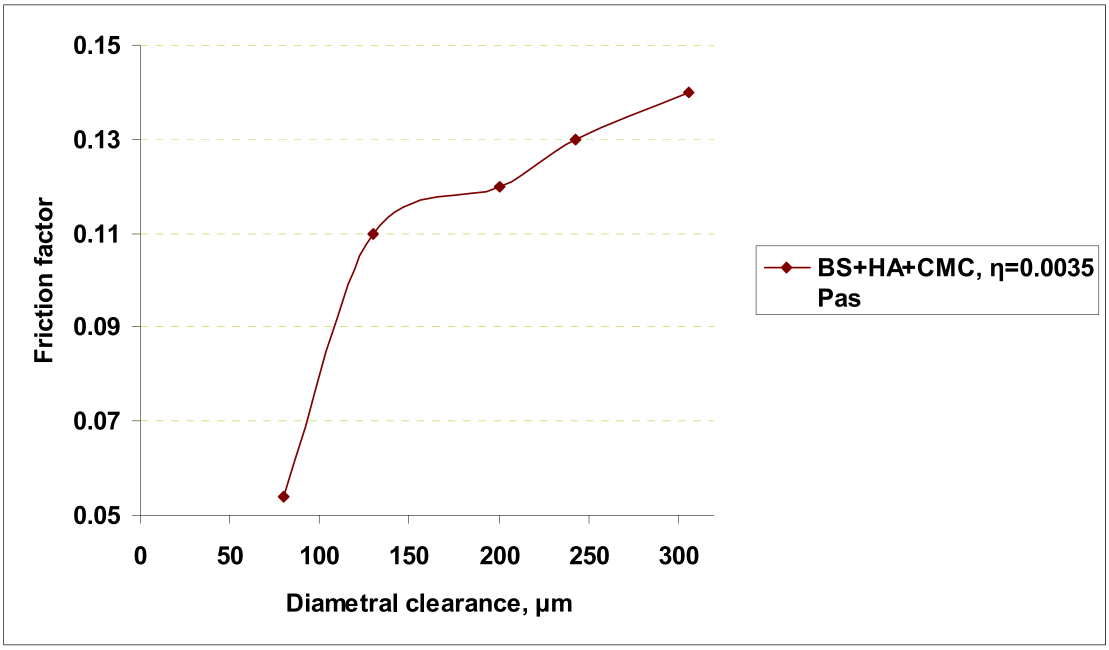

Figure 4-29. Friction factor versus diametral clearance for BS+HA+CMC lubricant with η=0.0035 Pas before deflection.

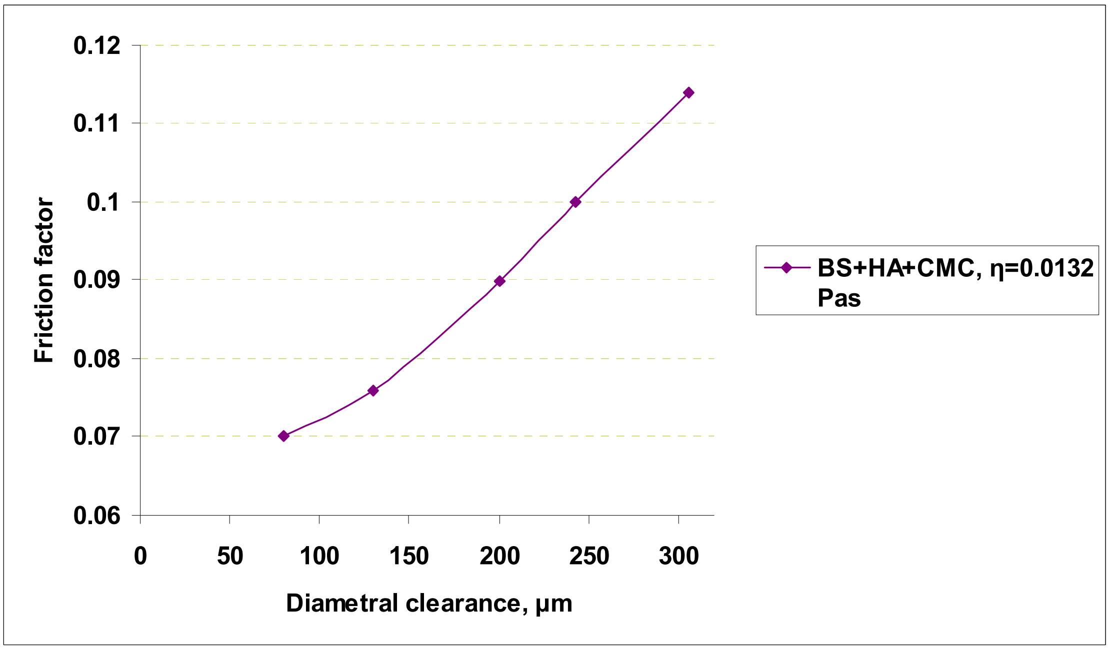

Figure 4-30. Friction factor versus diametral clearance for BS+HA+CMC lubricant with η=0.01324 Pas before deflection.

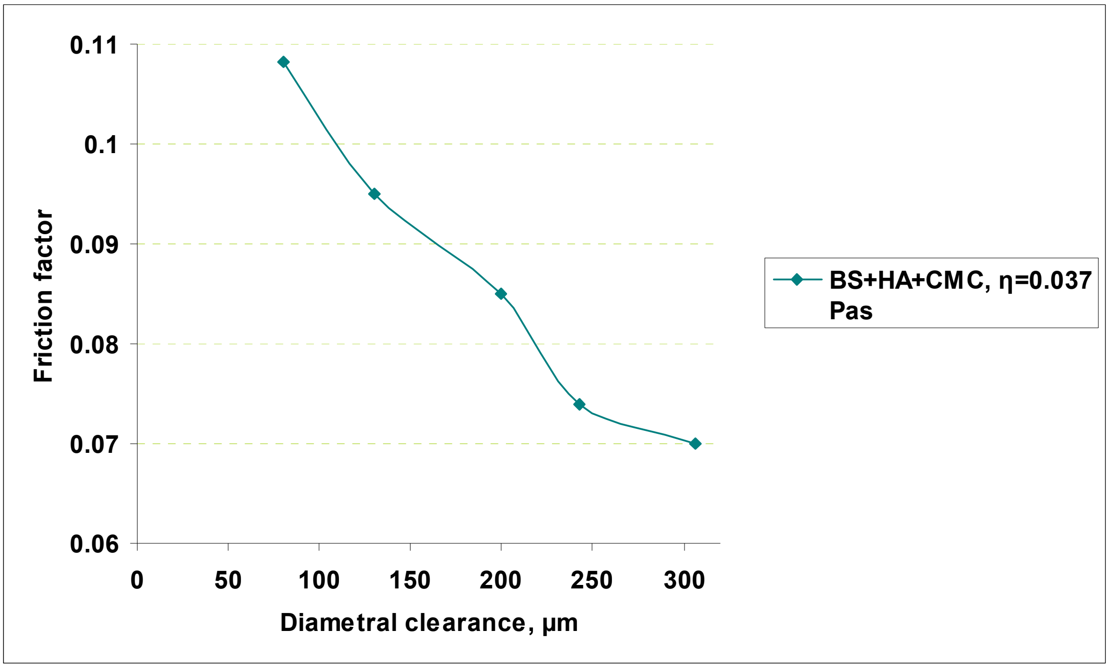

Figure 4-31. Friction factor versus diametral clearance for BS+HA+CMC lubricant with η=0.037 Pas before deflection.

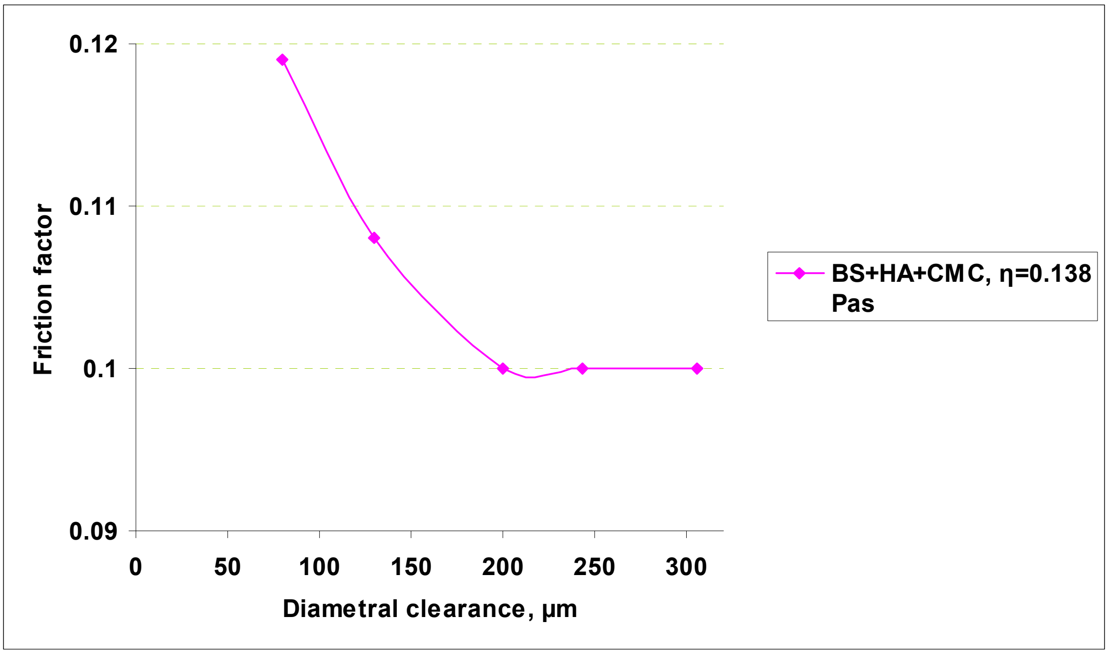

Figure 4-32. Friction factor versus diametral clearance for BS+HA+CMC lubricant with η=0.138 Pas before deflection.

Table 4.4: Average friction factors for different diametral clearances (80-306μm) using BS+HA+CMC with various viscosities.

Diametral Clearance, μm	η=0.00145 Pas	η=0.0035 Pas	η=0.0132 Pas	η=0.037 Pas	η=0.138 Pas
80	**0.042**	**0.054**	**0.07**	**0.1082**	**0.119**
130	**0.14**	**0.11**	**0.076**	**0.095**	**0.108**
200	**0.16**	**0.12**	**0.09**	**0.085**	**0.1**
243	**0.165**	**0.13**	**0.1**	**0.074**	**0.1**
306	**0.165**	**0.14**	**0.114**	**0.07**	**0.1**

Table 4.4 gives the friction factors for different diametral clearances using aqueous solutions of bovine serum with hyaluronic acid and carboxymethyl cellulose (BS+HA+CMC) as lubricant with various viscosities and Figure 4-28 to 4-32 are the graphs of these friction factors versus diametral clearances for only five joints. Again, it can be seen clearly and consistently that friction factors increase with increase in diametral clearance for viscosities of 0.00145, 0.0035 and 0.0132 Pas from ~ 0.04 to ~ 0.16, ~ 0.05 to ~ 0.14 and from ~ 0.07 to ~ 0.11, respectively. They imply that higher clearances do not give lower friction factors in this range of viscosities. On the other hand and opposite to this effect, the friction factors decreased consistently with increase in diametral clearance from ~ 0.108 to ~ 0.07 and from ~ 0.12 to ~ 0.1 for viscosities of 0.037 and 0.138 Pas, respectively. This suggests that the higher viscosity lubricants are effective in reducing the friction factors which was also the case with BS+CMC lubricants. The friction factors increased from ~ 0.04 to ~ 0.16 with increase in diametral clearances (80 to 306 μm) for

BS+HA+CMC lubricants having viscosities ≤0.00145 Pas, and decreased from ~ 0.11 to ~ 0.07 for viscosities ≥0.037 Pas, suggesting generally that higher friction factors are expected for lubricants with lower viscosities.

This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth resulting in impaired fixation with long-term implications for survival.

It is well recognized that the selection of optimum diametral clearance between the femoral head and the acetabular cup is a critical factor for the success of MOM bearings and thus an important consideration for the design/manufacturing of MOM hip prostheses. The current literature regarding the use of small clearances gives two different concluding remarks, i.e. for *in vitro* wear tests supported by theoretical studies it is claimed that smaller clearances reduce bedding-in wear and may improve lubrication conditions [Farrar *et al.,* 1997; Jin *et al.,* 2002]. So far, clinical studies, has not provided any evidence that larger clearances can cause reduction in the life of the MOM hip prostheses. In fact, we believe by observation that small clearances may increase the risk of equatorial or near equatorial contact causing the frictional torque to rise to high levels leading to loosening and eventual dislocation of the MOM hip prostheses which was a major reason for the earlier discontinuation of MOM bearings [Hall *et al.,* 1997; Scholes *et al.,* 2000]. We therefore believe that with small clearances, the bearing area can extend in the equatorial direction leading to higher contact stresses on the bearing surface near the equatorial area and thus causing higher frictional torque under the same loading condition.

Theoretical modelling has predicted that smaller diametral clearance may improve the lubrication by a thicker lubricating film in large diameter (50 mm) MOM hip resurfacing bearings [Jin *et al.,* 2006] and for UHMWPE on metal or ceramic femoral heads [Jalali *et al.,* 2001]. For example, an increase in head radius will enhance the film thickness, but it will also increase the sliding distance and hence wear in mixed or boundary lubrication conditions. Furthermore, it was pointed out that an increase in the predicted lubricant fi1m thickness is usually associated with an increase in the contact area, and this may cause lubricant starvation and stress concentration at the edge of the cup, and adversely affect the tribological performance of the implant.

The general trend in this study has been a mixed lubrication regime for clearances >80 µm and almost full fluid film lubrication for only the 80 µm clearance for both BS+CMC and BS+HA+CMC as lubricants with friction factors (BS+HA+CMC, 0.04-0.12 and 0.08-0.128, BS+CMC) outside the expected normal range for this regime (≤0.01).

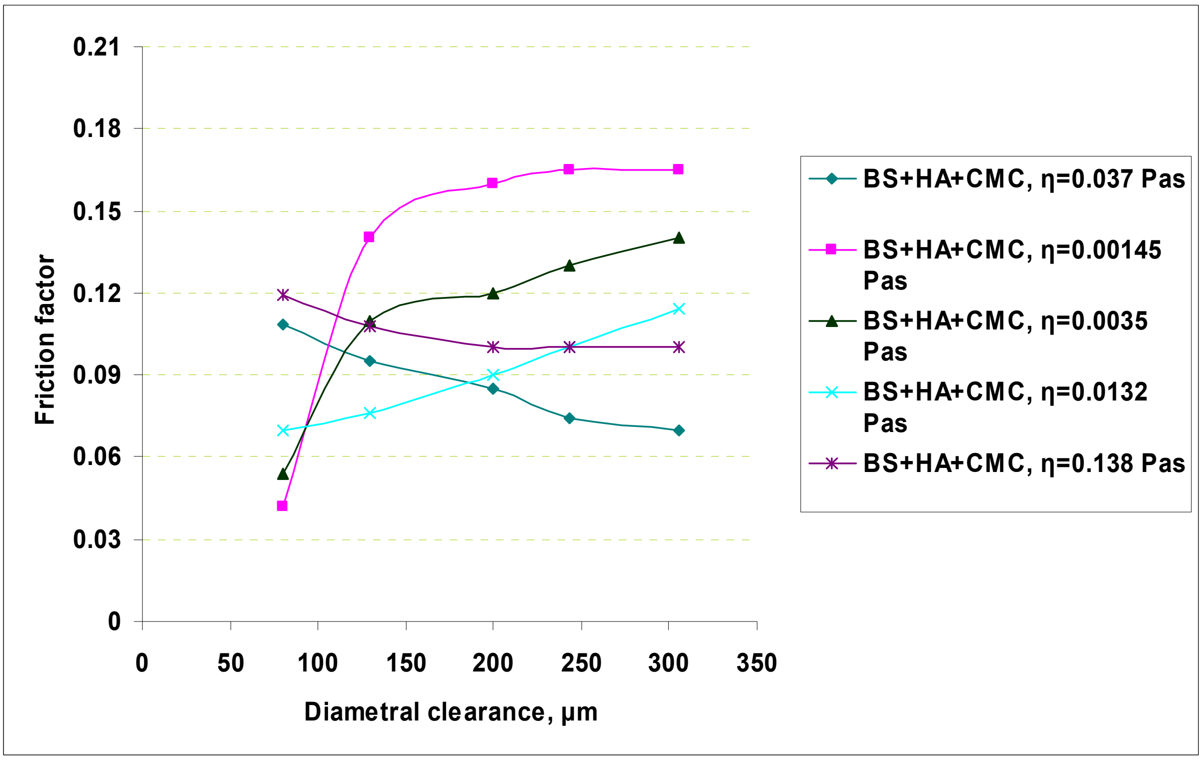

Figure 4-32 a. Friction factors versus different diametral clearance (80-306µm) using BS+CMC+HA with various viscosities.

4.3.1 Stribeck Analysis

Table 4.5 gives the calculated Sommerfeld number (z) and the related friction factors for all the five joints using BS+HA+CMC as lubricant with various viscosities and Figure 4-33 to 4-35 are the graphs of Stribeck curves using the results given in Table 4.5. The general trend is that of a decreasing friction factor with increase in Sommerfeld number (i.e. as viscosity increases) for the clearances ≥200 µm indicating a mixed lubrication regime.

Opposite to this effect is that of the 80 µm clearance for which friction factor increases with increase in Sommerfeld number from ~ 0.04 to ~ 0.12, implying a fluid film lubrication mode. These results also show clearly that the higher the diametral clearance the lower the friction factor for viscosities > 0.0132 Pas indicating that high viscosity fluids are effective in reducing friction as clearance increases, as observed previously for the BS+CMC lubricants (see section 4.1.1).

Table 4.5: Sommerfeld number versus friction factors for various diametral clearances using BS+HA +CMC as lubricants.

Sommerfeld Number, z ($x10^{-8}$)	**80µm**	**130µm**	**200µm**	**243µm**	**306µm**
0.025	**0.042**	**0.14**	**0.16**	**0.165**	**0.165**
0.07	**0.054**	**0.11**	**0.12**	**0.13**	**0.14**
0.265	**0.07**	**0.076**	**0.09**	**0.1**	**0.114**
0.74	**0.1082**	**0.095**	**0.085**	**0.074**	**0.07**
2.76	**0.119**	**0.108**	**0.1**	**0.07**	**0.1**

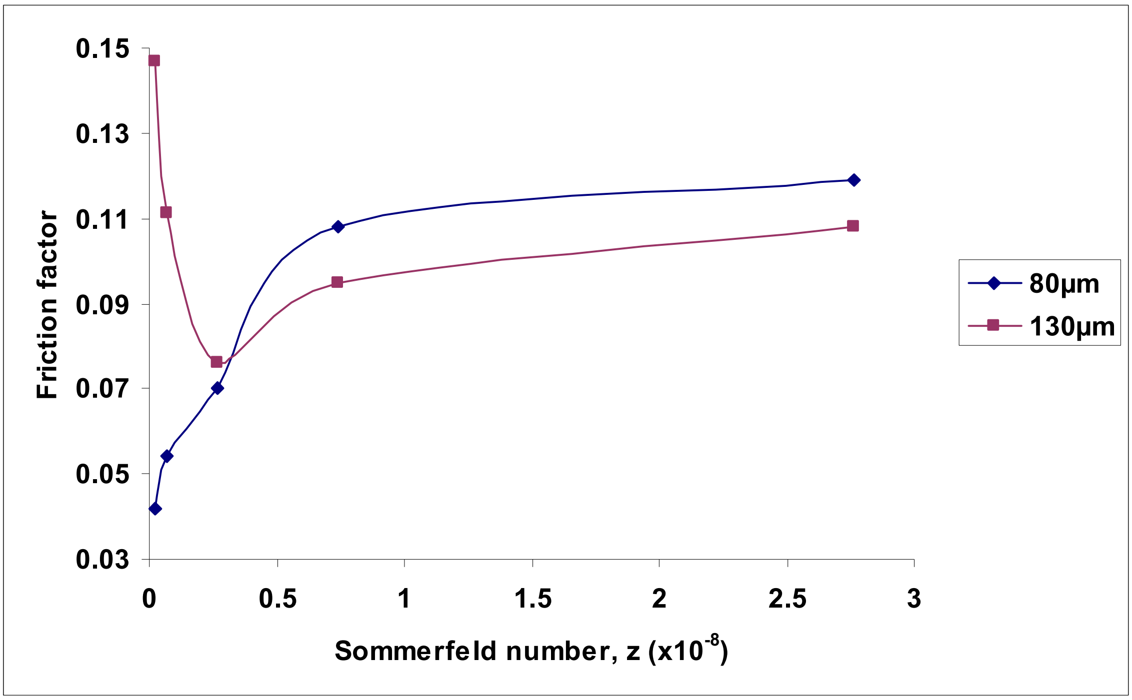

Figure 4-33. Friction factors versus Sommerfeld number for 80 and 135µm diametral clearance using BS+HA+CMC as lubricant before deflection.

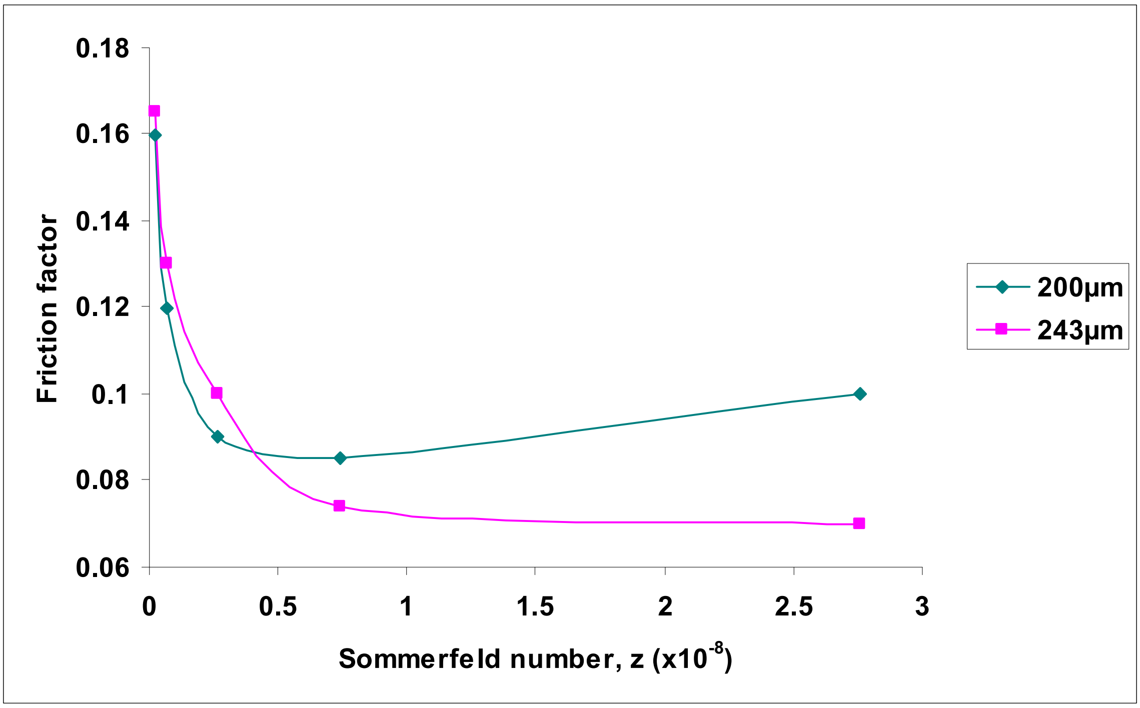

Figure 4-34. Friction factors versus Sommerfeld number for 200µm diametral clearance using BS+HA+CMC as lubricant before deflection.

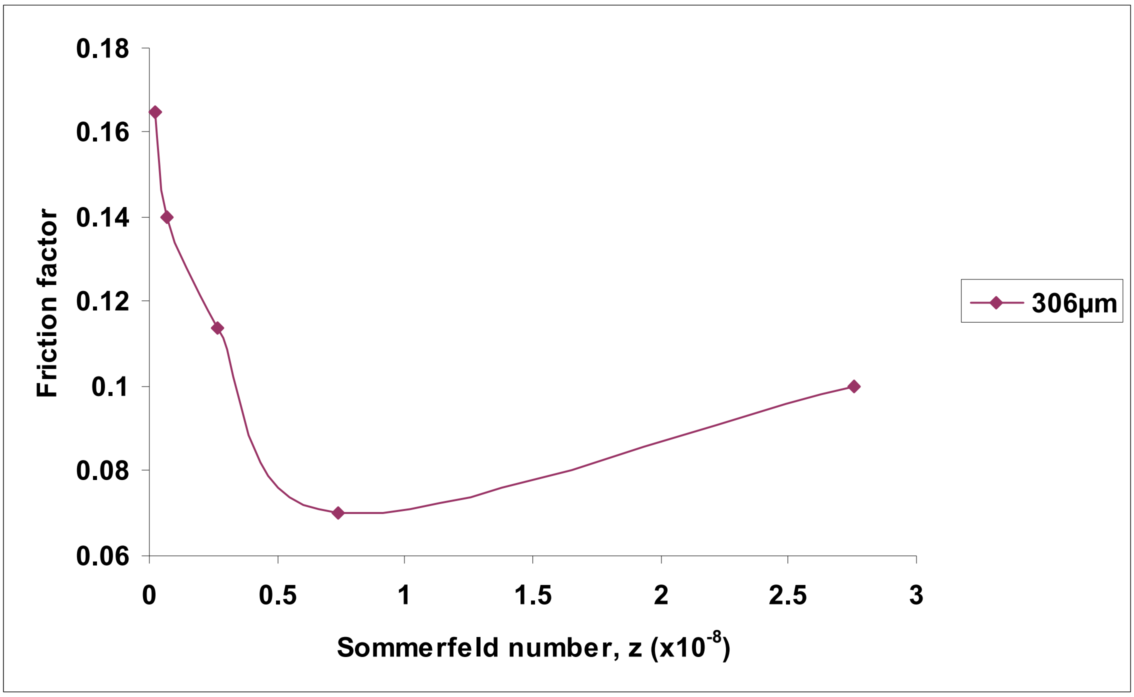

Figure 4-35. Friction factors versus Sommerfeld number for 306µm diametral clearance using BS+HA+CMC as lubricant before deflection.

4.4 Dynamic motion profiles for the S&N BHR devices using BS+HA+CMC with viscosities of 0.0035, 0.037, and 0.138 Pas and various clearances

Table 4.6 gives the average friction torque produced during dynamic friction tests for three different clearances using BS+HA+CMC with viscosities of 0.0035, 0.037, and 0.138 Pas. From Table 4.6 and Figures 4-36 to 4-38, it is clear that friction torque is dependent on both viscosity and clearance as also seen previously for the BS+CMC lubricants. However, for the 0.037 Pas lubricant there is only a negligible difference in frictional torque for diametral clearance of 200µm (2.97 Nm) and that of 306µm (3.19 Nm) with a similar frictional torque for clearances ≥175µm for the other viscosities. On the other hand, the 80µm clearance has caused slightly higher torque (4.72 Nm) which is very similar to that obtained for BS+CMC lubricant (see Table 4.3) of similar viscosity. It is to be noted, however, that the frictional torques generated in these tests for the MoM S&N BHR devices are significantly less than those reported by others [Wimmer *et al.,* 2003 and 2006] to cause instant loosening of the acetabular cup and depending upon the fixation and design ranged at 7-170 Nm. The same trend can also be seen here for the lowest and highest viscosities, i.e. friction torque increases as clearance increases and vice versa for the 0.0035 and 0.138 Pas viscosities, respectively, as also obtained for BS+CMC lubricants of similar viscosities.

Table 4.6: Average frictional torque for diametral clearances of 80, 200 and 306μm using BS+HA+CMC (η=0.037 Pas).

BS+HA+CMC Viscosity, η (Pas)	Friction Torque (Nm) for Diametral clearance of 80 μm	Friction Torque (Nm) for Diametral clearance of 200 μm	Friction Torque (Nm) for Diametral clearance of 306 μm
0.0035	**2.35**	**6.63**	**8.07**
0.037	**4.72**	**2.97**	**3.19**
0.138	**7.14**	**5.72**	**4.3**

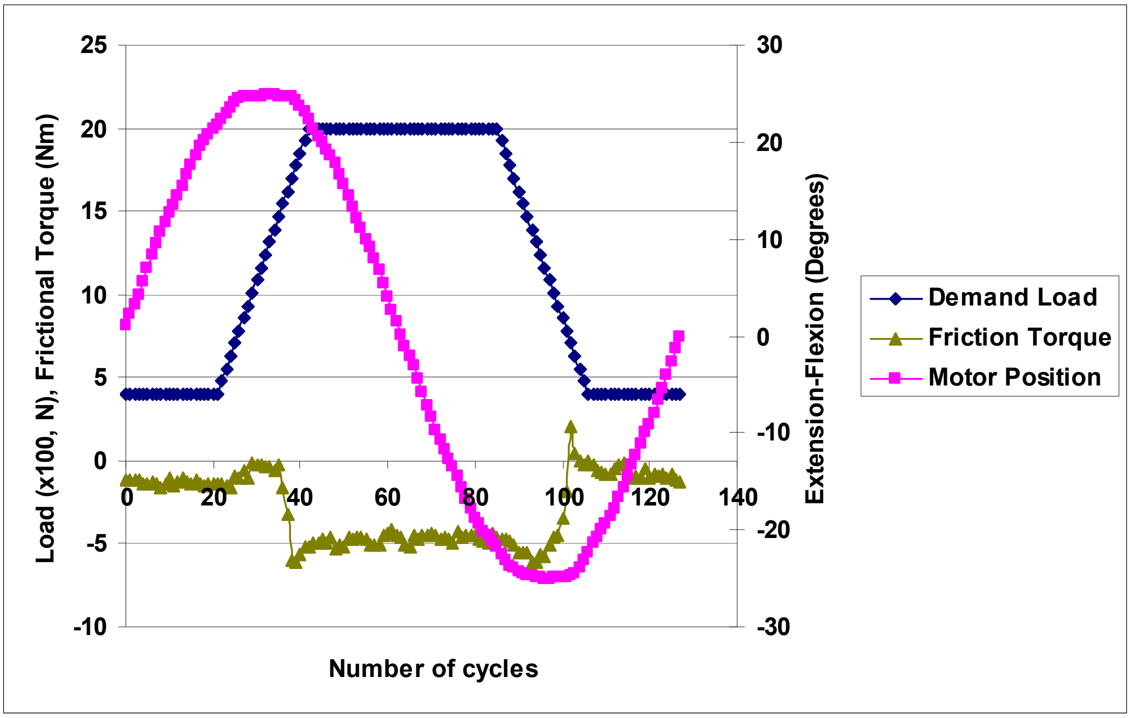

Figure 4-36. Friction Torque versus number of cycles for the 80µm diametral clearance, 50mm BHR bearing using BS+HA+CMC (η=0.037 Pas) as lubricant before deflection.

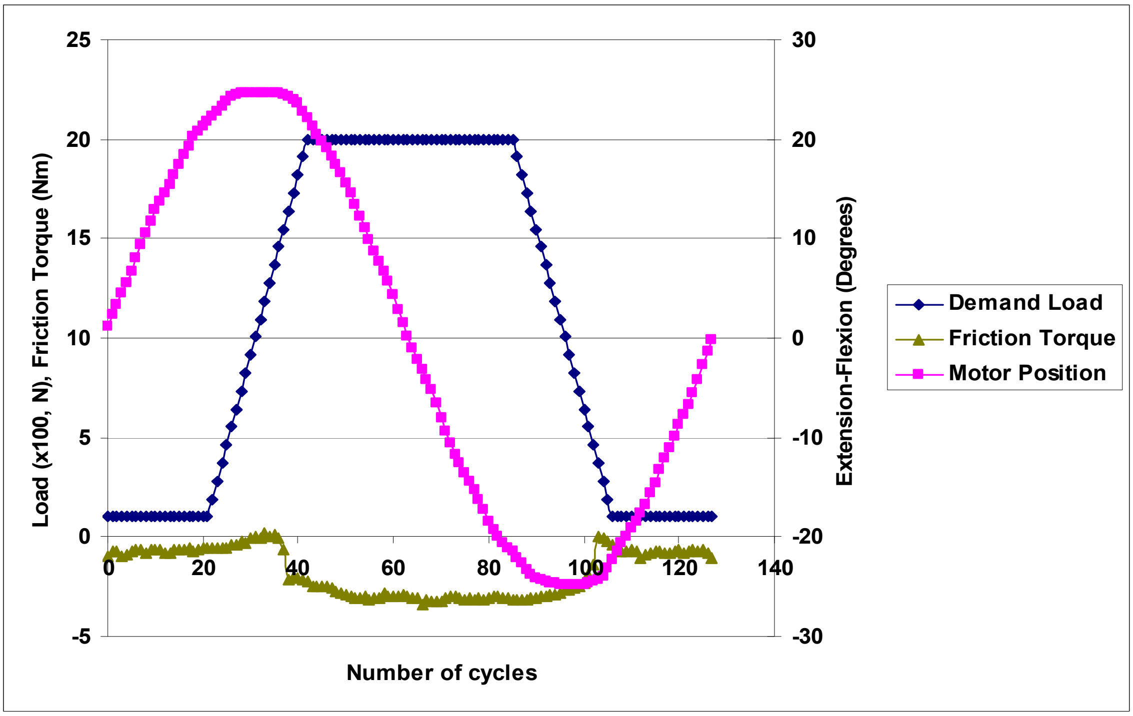

Figure 4-37. Friction Torque versus number of cycles for the 200µm diametral clearance, 50mm BHR bearing using BS+HA+CMC (η=0.037 Pas) as lubricant before deflection.

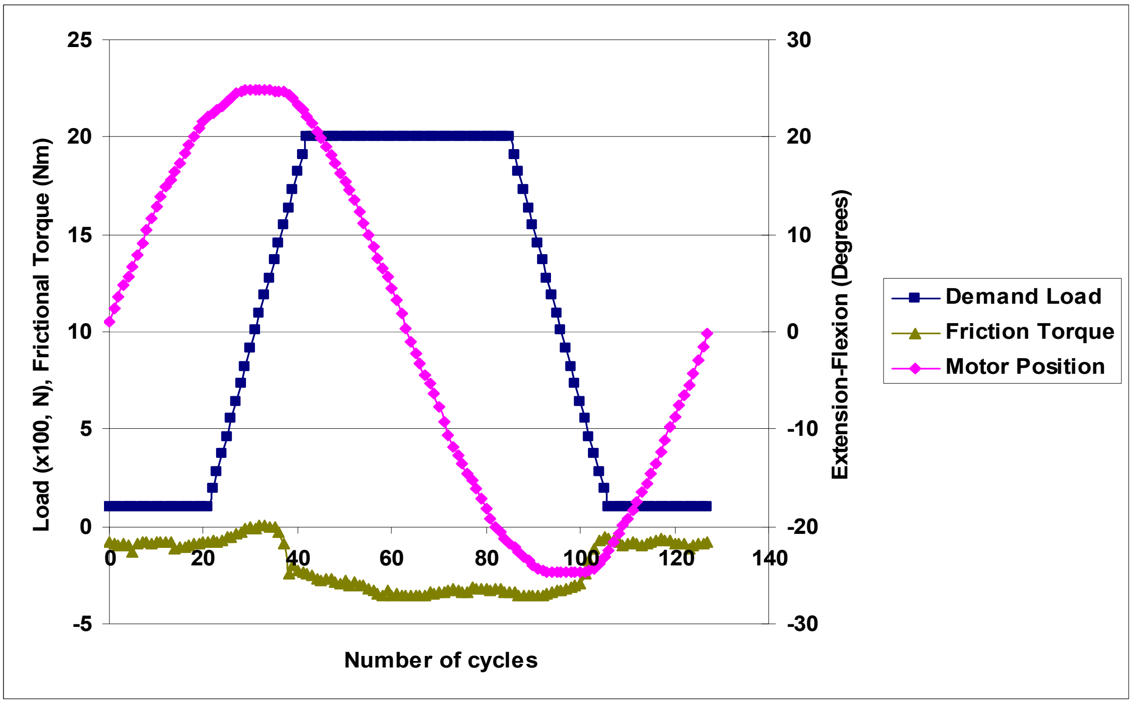

Figure 4-38. Friction Torque versus number of cycles for the 306μm diametral clearance, 50mm BHR bearing using BS+HA+CMC (η=0.037 Pas) as lubricant before deflection.

4.5 Friction factor and Viscosity results for the S&N BHR devices using Blood and Clotted blood as lubricants

As mentioned earlier, immediately after joint replacement, the artificial prosthesis is actually bathed in blood and clotted blood instead of synovial fluid. Blood contains large molecules and cells of size ~ 5 to 20 micron suspended in plasma and are considered to be a non-Newtonian fluid with density of 1060 Kg/m^3. The effect of these properties on friction is not fully understood and, so far, hardly any studies have been carried out regarding friction of metal-on-metal bearings with various clearances in the presence of lubricants such as blood or clotted blood. In this part of our work, therefore, we have investigated the frictional behaviour of a group of Smith and Nephew Birmingham Hip Resurfacing devices with a nominal diameter of

50mm and diametral clearances in the range ~ 80 to 300µm, in the presence of blood (clotted and whole blood).

4.5.1 Rheological properties of Clotted blood, Blood, Synovial fluid and Bovine serum

The procedure for assessing the flow behaviour was covered in the experimental methods (see section 3.3).

The viscosity curves for blood and clotted blood in Figures 4.1a and 4.1b, respectively, show a psuedoplastic (non-Newtonian) flow behaviour, i.e. a decrease in viscosity as shear rate increases, suggesting a shear thinning characteristic with the viscosity curve becoming asymptotic (levelling off) and remaining constant at high rates of shear >2000 s^{-1} implying that the lubricant becomes an incompressible isoviscous Newtonian fluid at these shear rates. This result, therefore, gives typical viscosities for blood and clotted blood expected between the articulating surfaces after implantation and allows some comparison with other biological lubricants such as synovial fluids and bovine serum. From Figure 4.1a, it can be seen that blood has a viscosity of ~ 0.01 Pas at a shear rate of 3000 s^{-1} as compared to ~ 0.02, ~ 0.04 and ~ 0.005 Pas for clotted blood (see Figure 4.1b), healthy synovial fluid, and rheumatoid fluid, respectively, at the same shear rates. This comparison clearly shows that blood has lower viscosity than both clotted blood and a healthy synovial fluid suggesting higher friction at the articulating surfaces is expected depending on the diametral clearance and when blood is the lubricating fluid. It is to be noted that the natural joint is surrounded by synovial fluid, a dialysate of blood plasma containing long-chain protein molecules such as human serum albumin (HSA) and glycoproteins, hyaluronic acid and phospholipids. The bovine serum with or without CMC also exhibited non-Newtonian shear thinning characteristics, i.e. psuedoplastic flow behaviour, as can be seen from Figures 4.1c and 4.1d, respectively. Noted is the viscosity values of ~0.002 and ~0.005 Pas at shear rates of 2000-3000 s^{-1} , for bovine serum (BS) and BS+CMC, respectively, which are almost 10 times less viscous than blood and clotted blood, indicating a different frictional behaviour will be expected depending on joint site and clearance.

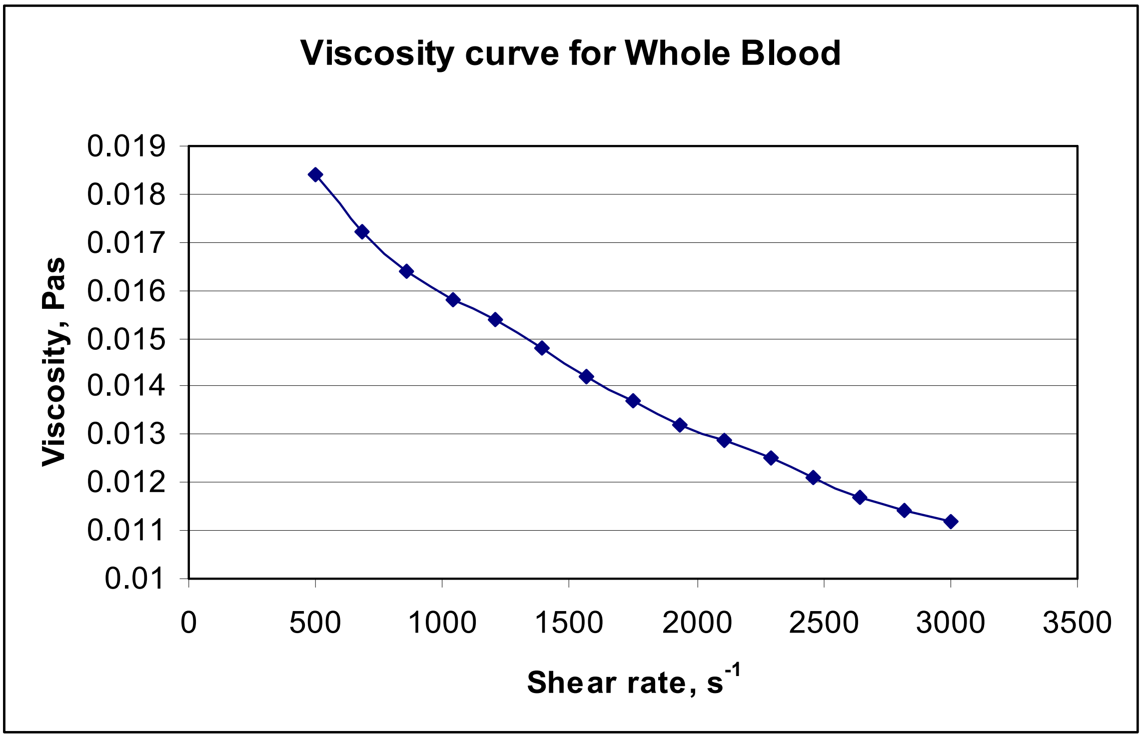

Figure 4.1a. Graph of viscosity versus shear rate for whole blood.

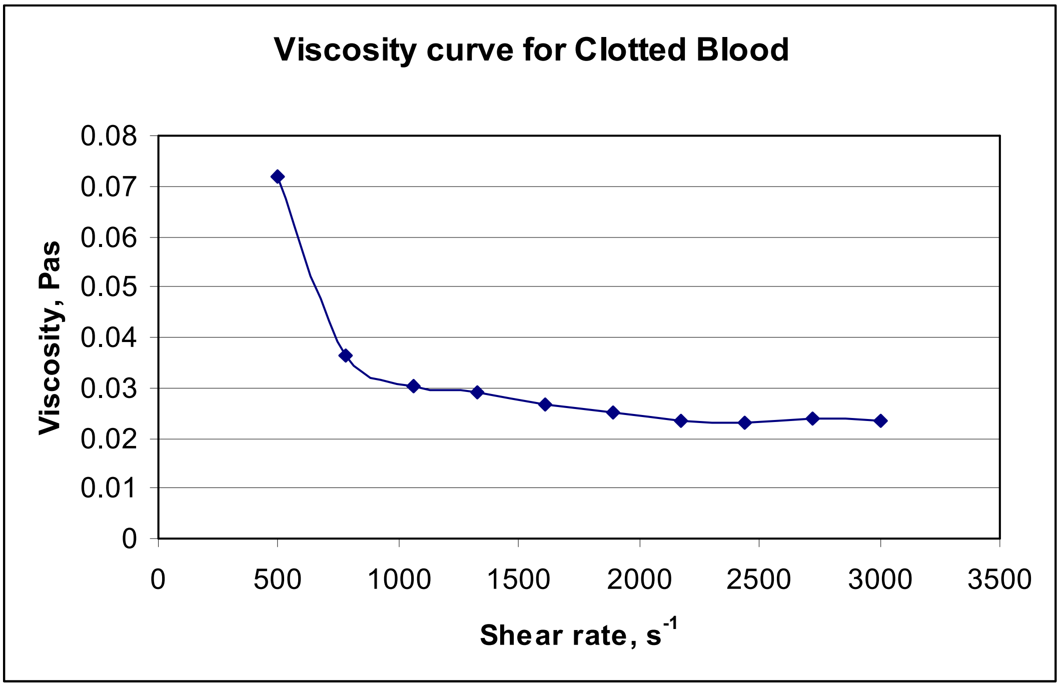

Figure 4.1b. Graph of viscosity versus shear rate for clotted blood.

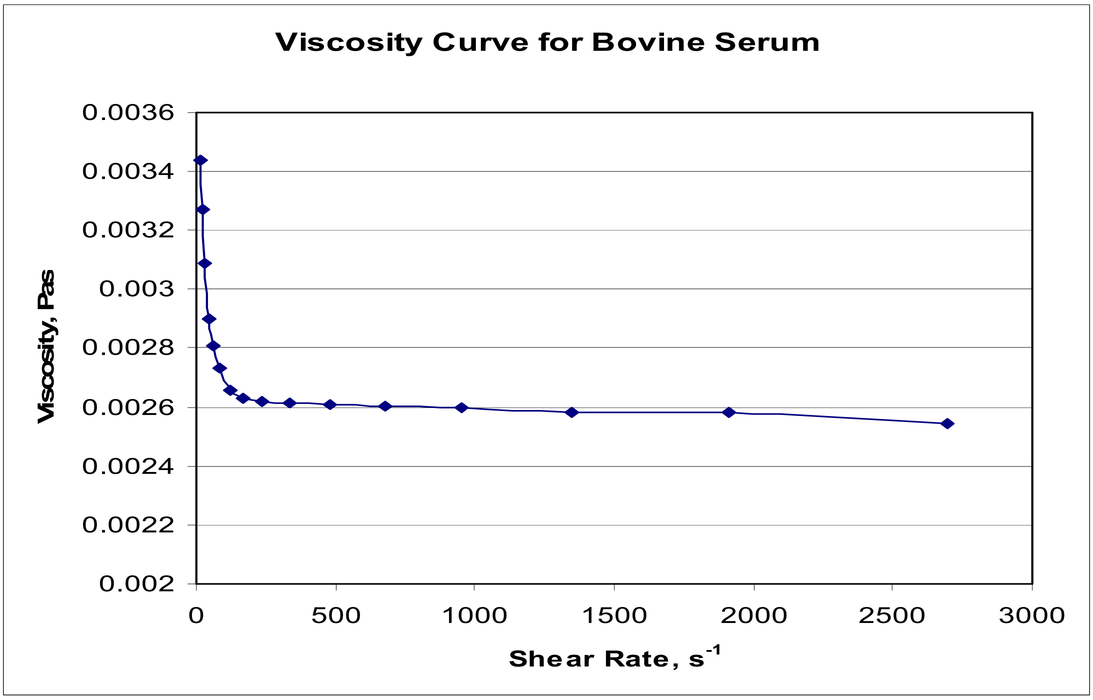

Figure 4.1c. Graph of viscosity versus shear rate for bovine serum.

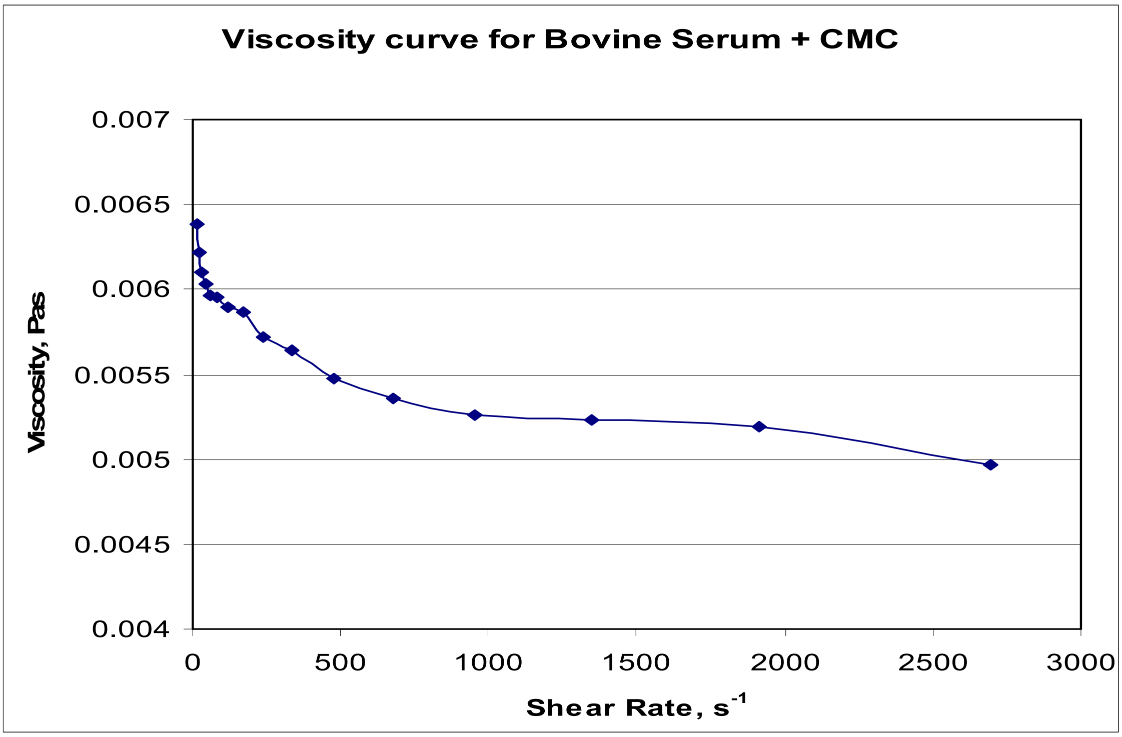

Figure 4.1d. Graph of viscosity versus shear rate for bovine serum with CMC.

4.6 Friction factor results for the S&N BHR devices using Blood and Clotted blood as lubricants at original diametral clearances

Table 4.7 and Figure 4.39 show a close comparison between friction factors for various diametral clearances of 80 to 306 µm using Blood and Clotted blood as lubricants. From Table 4.7 and Figure 4.39, it has become more obvious that both blood and clotted blood resulted in higher friction factors especially at lower clearances of 80 and 135 µm. This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth resulting in impaired fixation with long-term implications for survival. The friction factors in Table 4.7 have also shown that lower clearances do not necessarily reduce the friction factors to a level for the presence of full fluid film lubrication and that the friction factors decrease with increase in diametral clearance. This finding clearly suggests that lower clearances have higher potential for increasing the friction between the articulating joint surfaces and thus increase the risk of micromotion due to higher surface contacts, leading to higher risk of joint dislocation.

Table 4.7: Friction factors for the whole blood (η=0.0133 Pas) and clotted blood (η=0.02 Pas) for different diametral clearances.

Original Diametral Clearance (μm)	Average friction factor using Blood (η=0.013 Pas)	Average friction factor using Clotted blood (η=0.02 Pas)
80	0.19	0.17
135	0.19	0.165
200	0.18	0.16
243	0.143	0.15
306	0.14	0.14

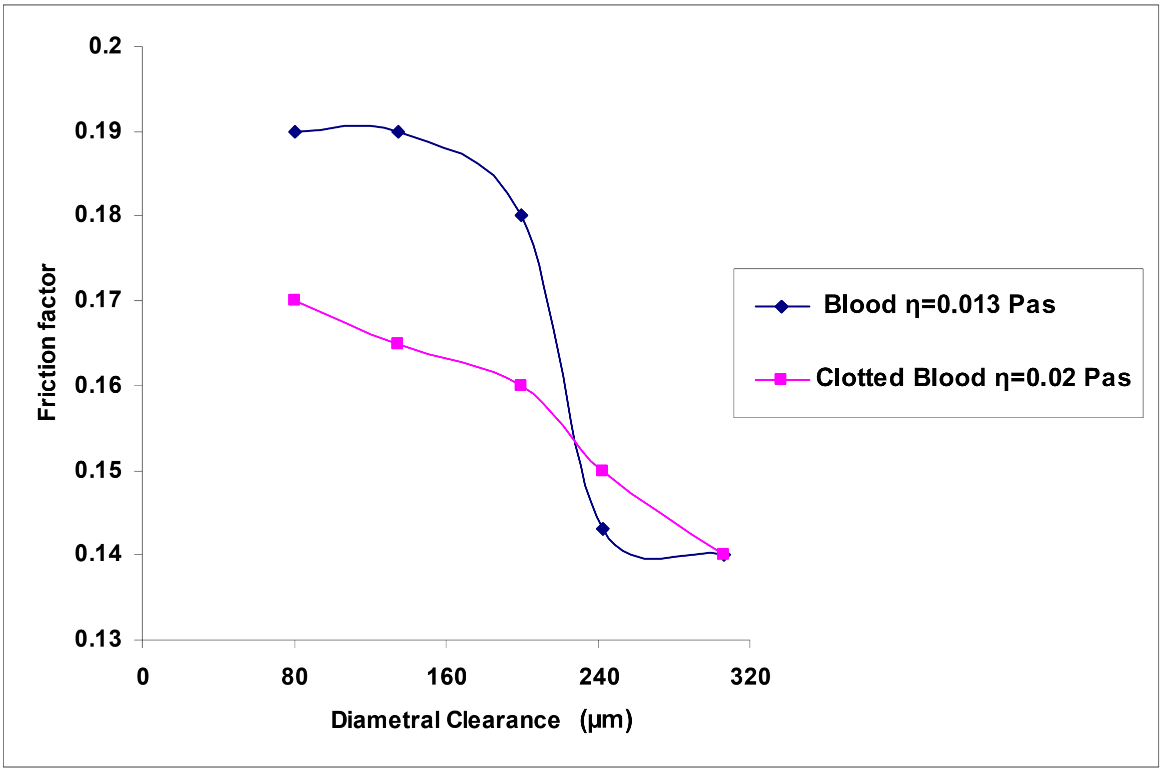

Figure 4.39. Graph of friction factor versus diametral clearance for the S&N BHR 50mm diameter devices using Blood and Clotted blood as lubricants.

4.7 Dynamic motion profiles for the S&N BHR devices using Clotted blood (η=0.02 Pas) and Blood (η=0.013Pas) as lubricants

Table 4.8 gives the average friction torque produced during dynamic friction tests for three different clearances using clotted blood and blood. From Table 4.8 and Figures 4-40 to 4-48, it is clear that there is a significant reduction in frictional torque when diametral clearance increases from 80 to 306 μm for both clotted blood and whole blood as lubricants. Friction torque decreased from ~7.15 to ~3.4 Nm and ~3.3 to ~1.7 Nm for blood and clotted blood, respectively, which indicate that using higher clearances a reduction in friction torque is expected.

Table 4.8: Average frictional torque for various diametral clearances of 80-306μm using blood and clotted blood as lubricants.

Diametral clearance, μm	Friction Torque (Nm) for blood (η=0.013 Pas)	Friction Torque (Nm) for clotted blood (η=0.02 Pas)
80	3.34	7.15
130	3.0	6.2
175	2.04	4.8
200	2.56	3.35
243	1.88	2.54
306	1.73	3.45

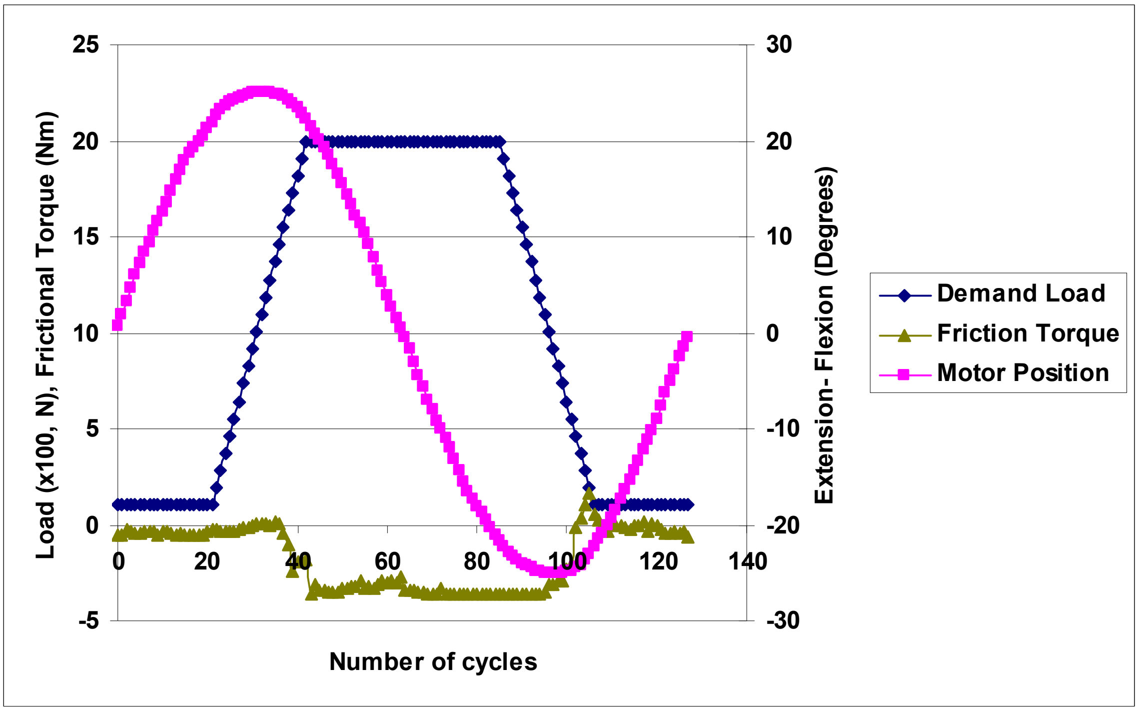

Figure 4-40. Friction Torque versus number of cycles for the 80 µm diametral clearance, 50mm BHR bearing using blood (η=0.013 Pas) as lubricant.

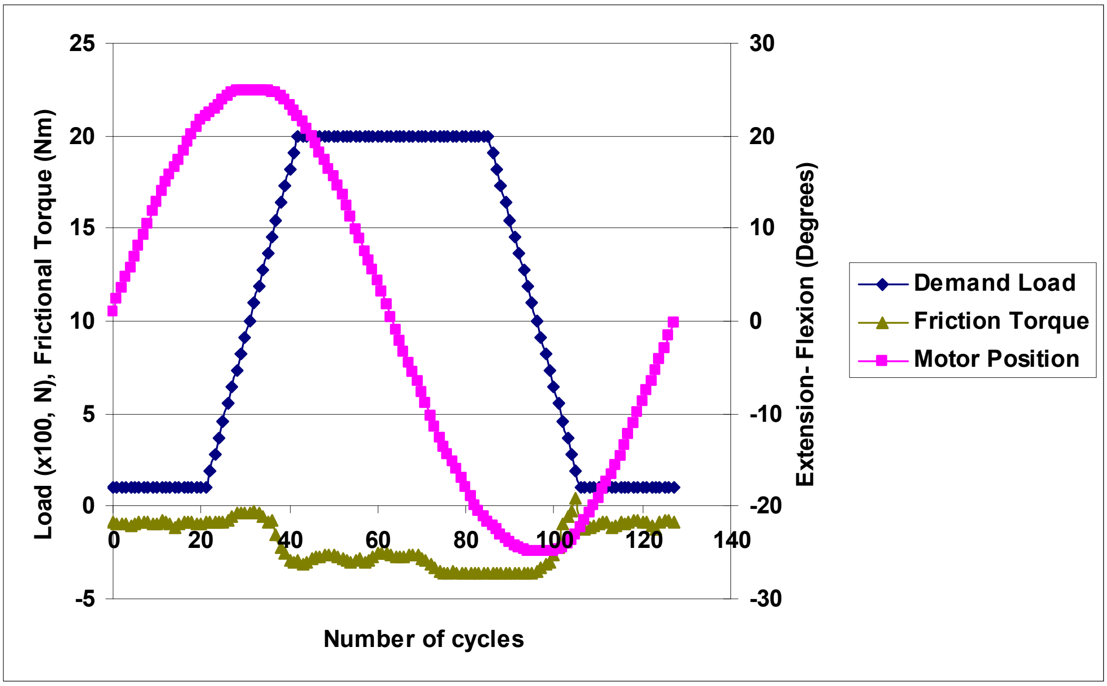

Figure 4-41. Friction Torque versus number of cycles for the 130 µm diametral clearance, 50mm BHR bearing using blood (η=0.013 Pas) as lubricant.

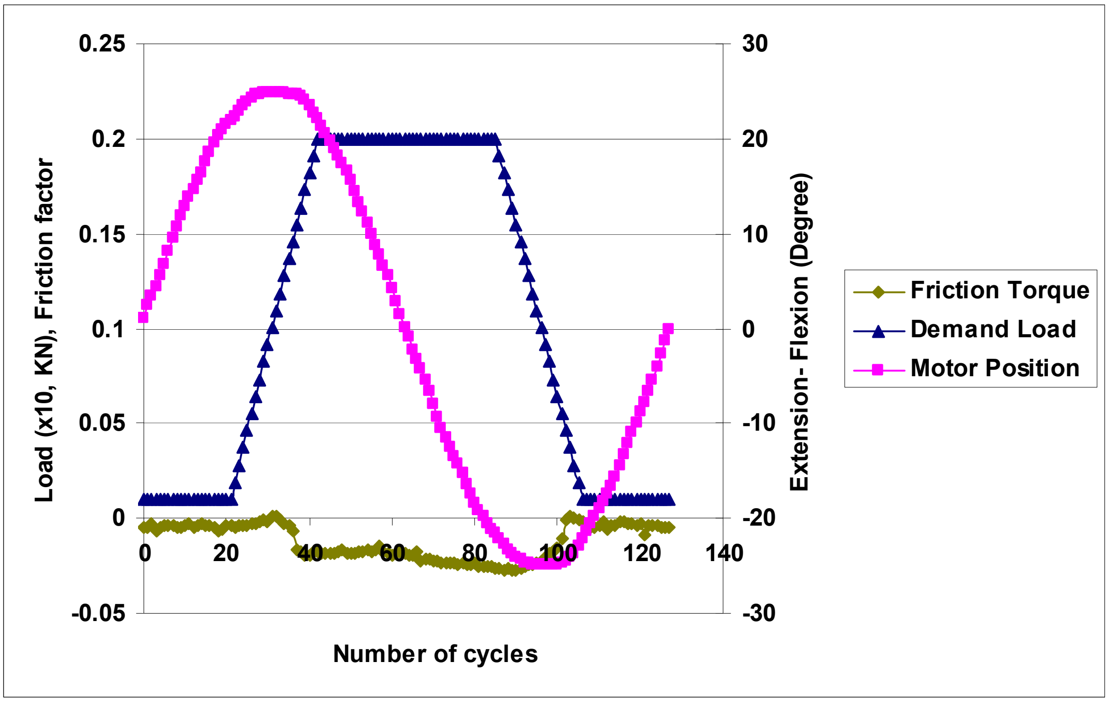

Figure 4-42. Friction Torque versus number of cycles for the 175 µm diametral clearance, 50mm BHR bearing using blood (η=0.013 Pas) as lubricant.

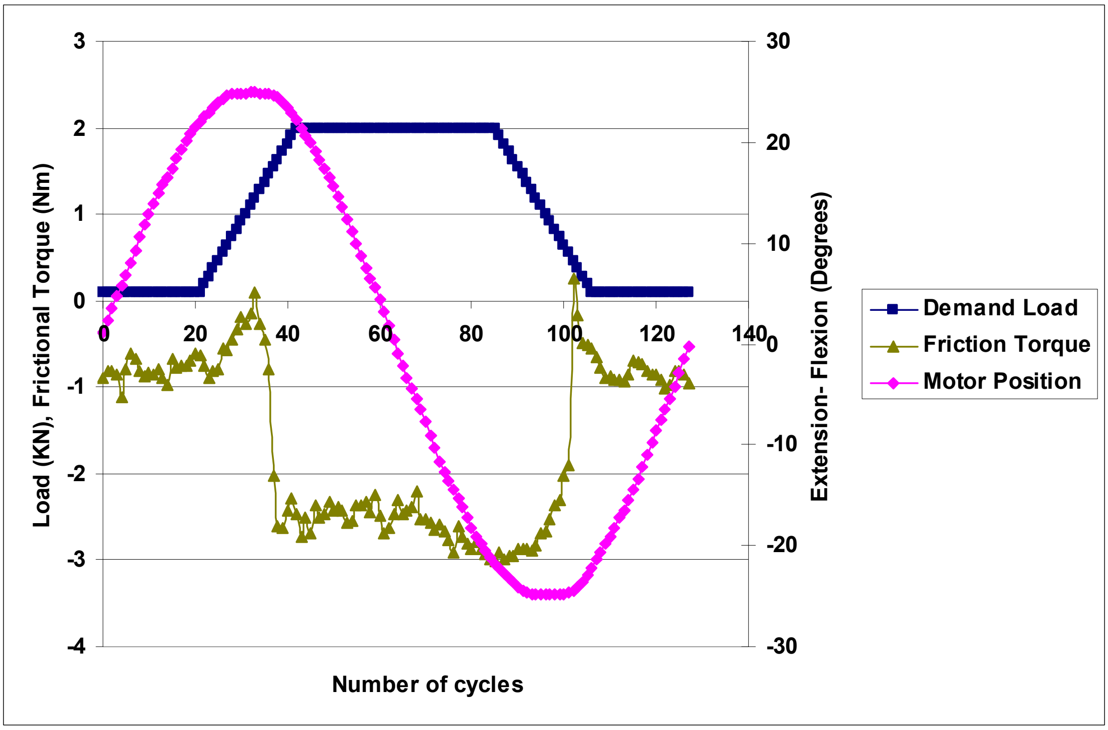

Figure 4-43. Friction Torque versus number of cycles for the 200 µm diametral clearance, 50mm BHR bearing using blood (η=0.013 Pas) as lubricant.

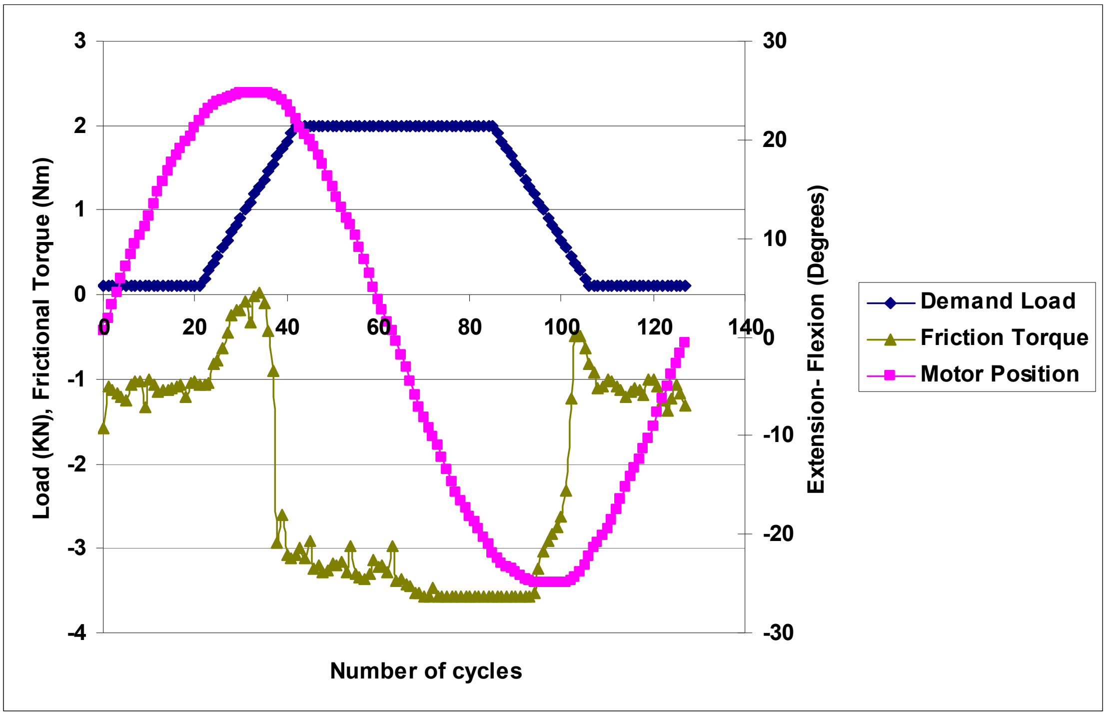

Figure 4-44. Friction Torque versus number of cycles for the 200 µm diametral clearance, 50mm BHR bearing using clotted blood (η=0.02 Pas) as lubricant.

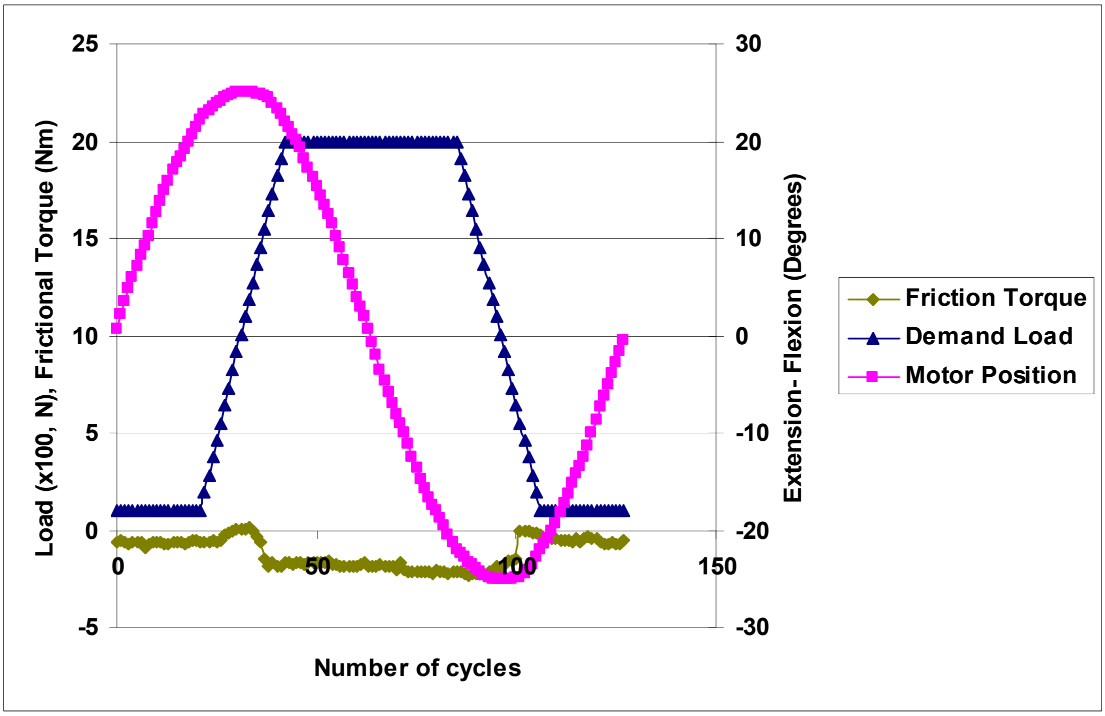

Figure 4-45. Friction Torque versus number of cycles for the 243 µm diametral clearance, 50mm BHR bearing using blood (η=0.013 Pas) as lubricant.

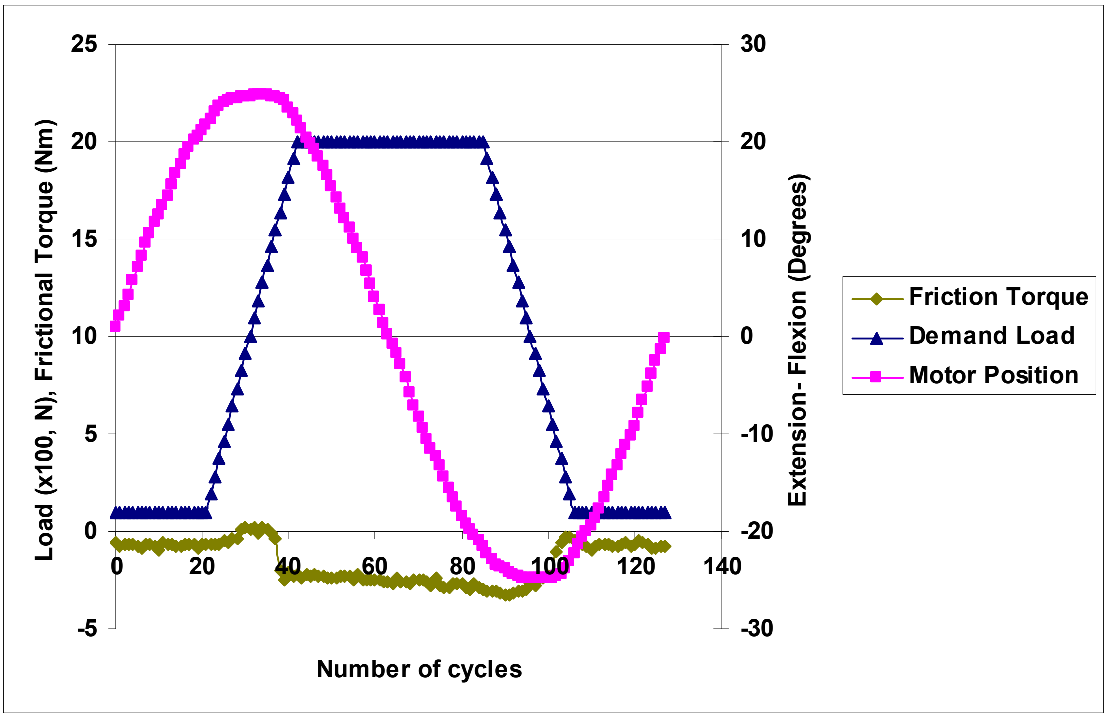

Figure 4-46. Friction Torque versus number of cycles for the 243 µm diametral clearance, 50mm BHR bearing using clotted blood (η=0.02 Pas) as lubricant.

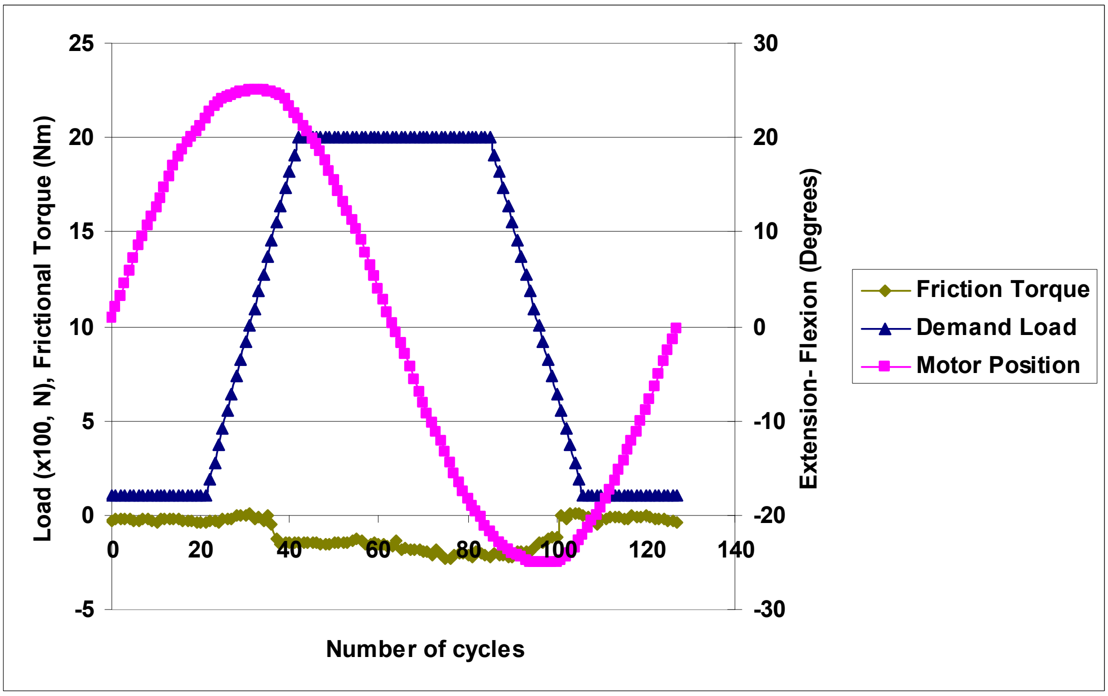

Figure 4-47. Friction Torque versus number of cycles for the 306 µm diametral clearance, 50mm BHR bearing using blood (η=0.013 Pas) as lubricant.

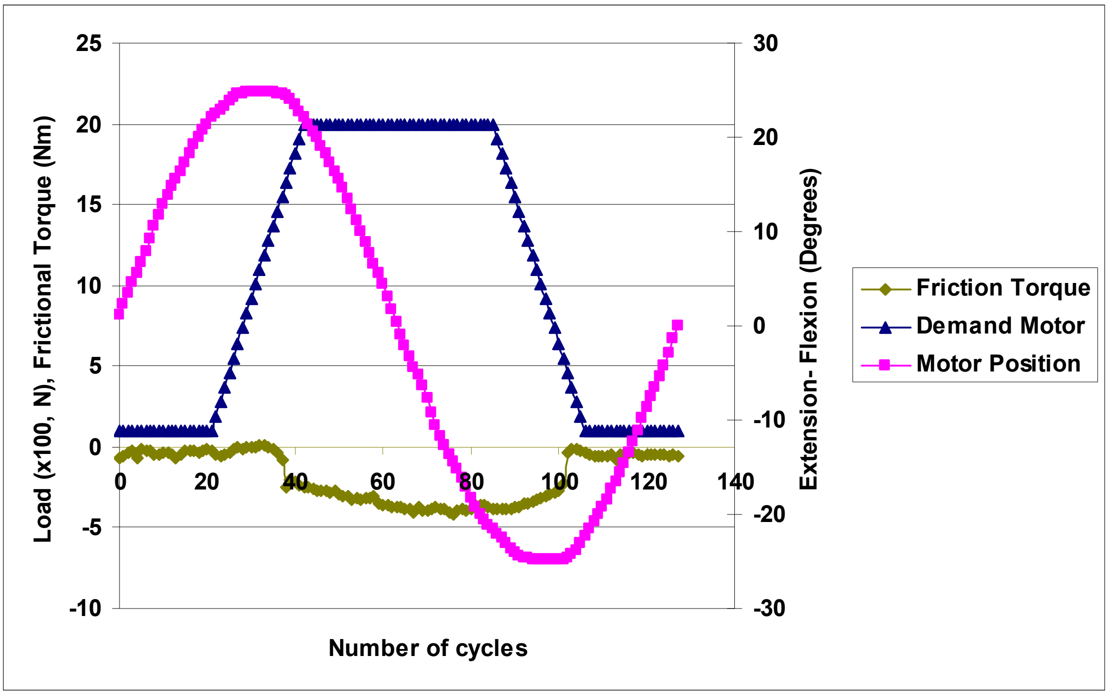

Figure 4-48. Friction Torque versus number of cycles for the 306 µm diametral clearance, 50mm BHR bearing using clotted blood (η=0.02 Pas) as lubricant.

4.8 Friction factor results for the deflected S&N BHR devices using Blood and Clotted blood as lubricants

Cementless cup designs for metal on metal hip resurfacing prostheses usually depend on a good primary press fit fixation which stabilizes the components in the early post-operative period. Press-fitting the cup into the acetabulum results in non-uniform compressive stresses on the cup and causes non-uniform cup deformation. This may result in equatorial contact and high frictional torque leading to femoral head seizure. It has been reported [Kamali *et al.,* 2006] that high frictional torque is likely to cause micromotion between the implant and its surrounding bone and thus adversely affecting the longevity of the implant.

The aim of this part of our work was to investigate the effect of cup deformation on friction between the articulating surfaces of the same six Birmingham Hip Resurfacing devices with various clearances but deformed initially by ~25-35 μm using two-point pinching action before friction tests, and finally deformed by ~60-70 μm (in total).

The friction test procedure was as before and covered in chapter three under experimental procedure (see section 3.2). However, the Birmingham Hip Resurfacing devices were tested in blood and clotted blood which is indeed the primary lubricants during the early weeks/months after implantation.

The average friction factors (average of 3 tests as before) for different clearances after initial and final deformation are given in Tables 4.9 and 4.10, respectively. Figures 4.49 and 4.50 are the graphs of friction factor versus diametral clearance after initial and final deformations, respectively.

Table 4.9: Average friction factors after initial (cup) deformation using blood and clotted blood as lubricants.

Original Diametral Clearance (μm)	Cup deflection (μm)	Average friction factor using Blood (η=0.0083 Pas)	Average friction factor using Clotted blood (η=0.0108 Pas)
80	30	0.18	0.19
130	35	0.201	0.2
175	25	0.194	0.2
200	24	0.147	0.18
243	26	0.13	0.134
306	26	0.15	0.16

Table 4.10: Average friction factors after final (cup) deformation using blood and clotted blood as lubricants.

Original Diametral Clearance (µm)	Cup deflection (µm)	Average friction factor using Blood (η=0.0112 Pas)	Average friction factor using Clotted blood (η=0.0234 Pas)
80	**67**	**0.173**	**0.203**
130	**63**	**0.193**	**0.201**
175	**69**	**0.18**	**0.185**
200	**69**	**0.171**	**0.167**
243	**61**	**0.097**	**0.1**
306	**59**	**0.14**	**0.136**

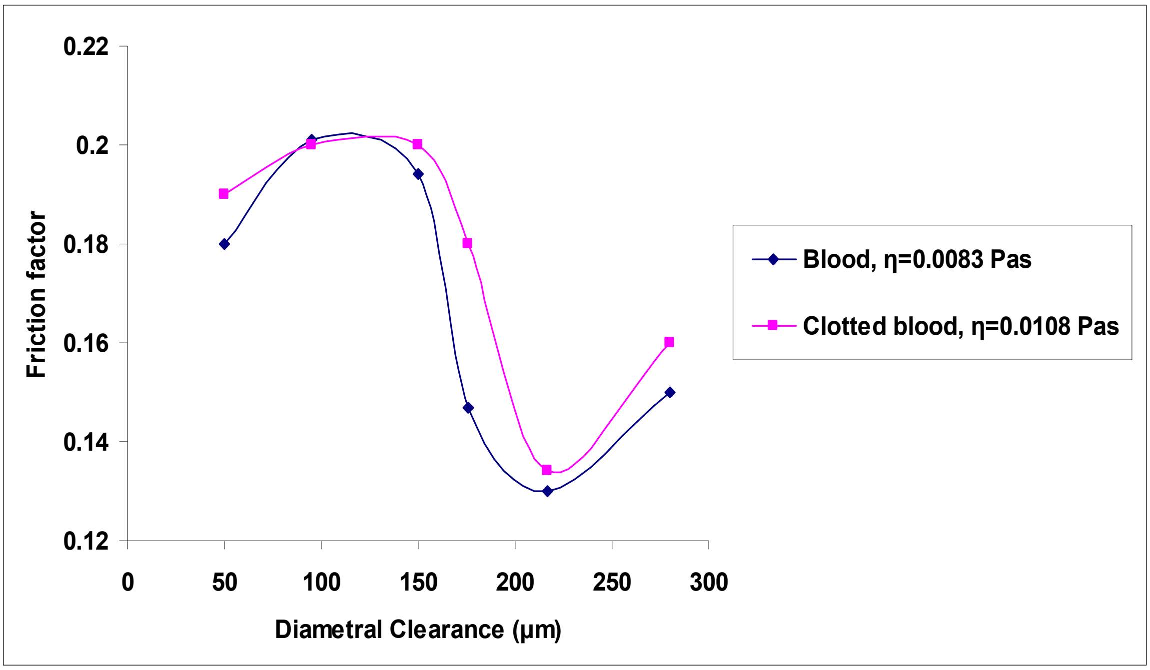

Figure 4.49. Graph of friction factor versus diametral clearance* after initial (cup) deformation.

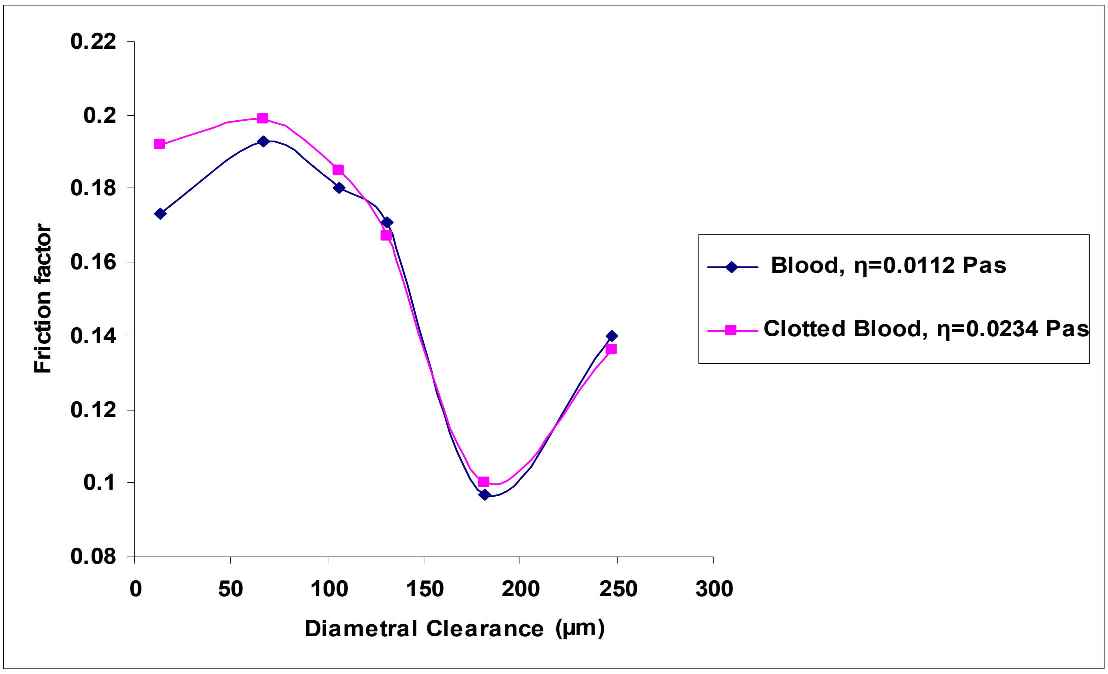

Figure 4.50. Graph of friction factor versus diametral clearance* after final (cup) deformation.

* Note that the actual clearances after cup deflections are:
Original diametral clearance - cup deflection, e.g. for the 306 µm - 59=247 µm= diametral clearance after final deflection. (see Tables 4.9 and 4.10)

From Tables 4.9 and 4.10 and Figures 4-49 and 4-50 it can be seen quite clearly that friction factor has decreased consistently as diametral clearance increases for both blood and clotted blood and also for both initial and final cup deflections. These results are in good agreement with those before cup deflection (covered in the previous section) when blood and clotted blood were also used on the original joints. This is an adequate (important) finding since the results obtained during this work clearly show that for reduced clearances friction increased clearly when the cups were deflected by ~30 µm and 60-70 µm. It is therefore clear that higher clearances can accommodate the amount of distortion introduced in the cups during this investigation.

The results of this study suggest therefore that reduced clearance bearings have the potential to generate high friction especially in the early weeks after implantation when blood is indeed the in vivo lubricant. This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth resulting in impaired fixation with long-term implications for survival.

4.8.1 Stribeck analysis

Tables 4.11 and 4.12 give the calculated Sommerfeld number (z) and the related friction factors for all the six joints using Blood and Clotted blood as lubricants and Figure 4-51 is the graph of Stribeck curves using the results given in Tables 4.11 and 4.12. The general trend is that of an increasing friction factor with increase in Sommerfeld number (i.e. as viscosity increases) for both initial and final deflections indicating possibility of fluid film lubrication. It is to be noted, however, that only two points could be obtained for the Stribeck analysis which may not be the true representation of the lubrication mode. The two points were for blood and clotted blood having different viscosities as the only possible variables in calculating the Sommerfeld number.

Table 4.11: Sommerfeld number versus friction factors for various diametral clearances using Blood (η=0.0083 Pas) and Clotted blood (η=0.0108 Pas) as lubricants after initial cup deflection.

Lubricant	Sommerfeld number, z ($x10^{-8}$)	50µm	95µm	150µm	176µm	217µm	280µm
Blood	0.205	0.18	0.2	0.194	0.147	0.13	0.150
Clotted blood	0.27	0.19	0.202	0.2	0.18	0.134	0.160

Table 4.12: Sommerfeld number versus friction factors for various diametral clearances using Blood (η=0.0112 Pas) and Clotted blood (η=0.0234 Pas) as lubricants after final cup deflection.

Lubricant	Sommerfeld number, z ($x10^{-8}$)	13 µm	67µm	131µm	182µm	247µm
Blood	0.28	0.178	0.192	0.178	0.096	0.132
Clotted blood	0.58	0.192	0.199	0.164	0.099	0.128

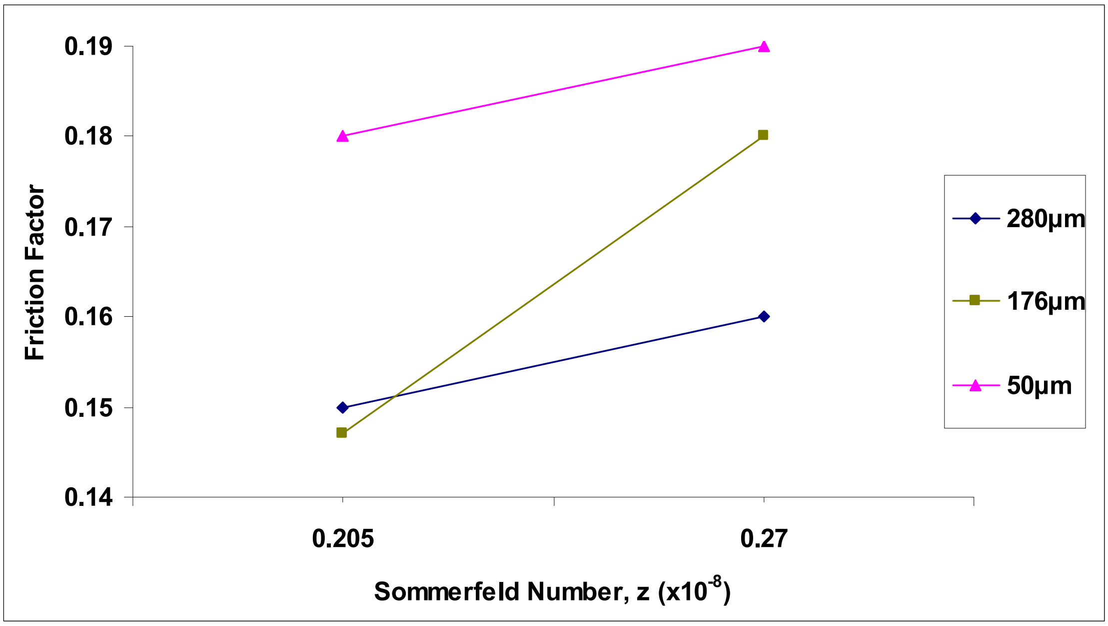

Figure 4-51. Friction factor versus Sommerfeld number for the 50, 176 and 280µm diametral clearance using blood and clotted blood as lubricant after initial deflection.

4.9 Dynamic motion profiles for the S&N BHR devices using Blood and Clotted blood as lubricants after initial cup deflection

The dynamic loading cycles generated during the friction tests for the deflected cups and original femoral heads are plotted graphically in Figures 4.52 - 4.57. Table 4.13 gives the average friction torque produced during dynamic friction tests.

Table 4.13 and Figures 4.52 - 4.57 show a general trend throughout the test, i.e. a falling frictional torque from ~8.8 to ~7.45 Nm with increasing diametral clearance when clotted blood was used as lubricant. Similar trend, but slightly lower frictional torques was obtained for blood, i.e. friction torque decreased from ~7.6 to ~6.5 Nm when blood was used as lubricant. This suggests that increasing the viscosity of a lubricant may also increase the frictional factor while maintaining the load applied in the joint. The affect is a rise in the frictional torque for higher viscosity fluid. It is exhibited that high friction torques (≥10-170 Nm) generated at the prosthetic interface could indeed be responsible to produce fatigue failure and result in loosening [Ma *et al.,* 1983].

It is also postulated that during acetabular fixation space limitation of replacement could be reduced due to cup deformation during insertion into the pelvis [Ma *et al.,*1983], thereby creating an imperfect bearing surface that could increase the frictional torque. It would be important to know at what viscosity a lower friction could be achieved and when the diametral clearance is small, the shear rate in the small clearance bearing could be 10 times higher than that of the larger clearance [Ma *et al.,* 1983] and thus a high friction factor and torque may result due to the internal friction of the lubricant, especially when the viscosity is high.

Table 4.13: Average friction torque for various diametral clearances of (50-280μm) using blood and clotted blood as lubricants after initial cup deflection.

Diametral clearance, μm	**Friction Torque (Nm), for blood of η=0.0083 Pas**	**Friction Torque (Nm), for clotted blood of η=0.0108 Pas**
50	**7.6**	**8.8**
176	**6.9**	**7.8**
280	**6.5**	**7.45**

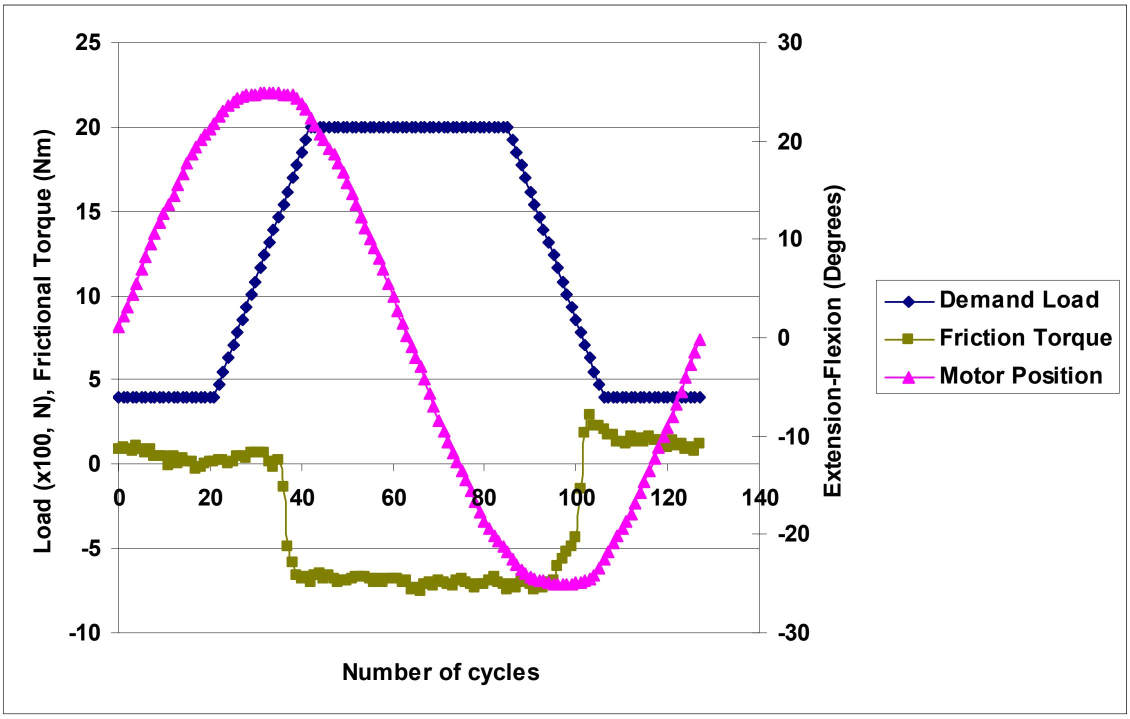

Figure 4-52. Friction Torque versus number of cycles for the 50 μm diametral clearance, 50mm BHR bearing using Blood (η=0.0083 Pas) as lubricant after initial deflection.

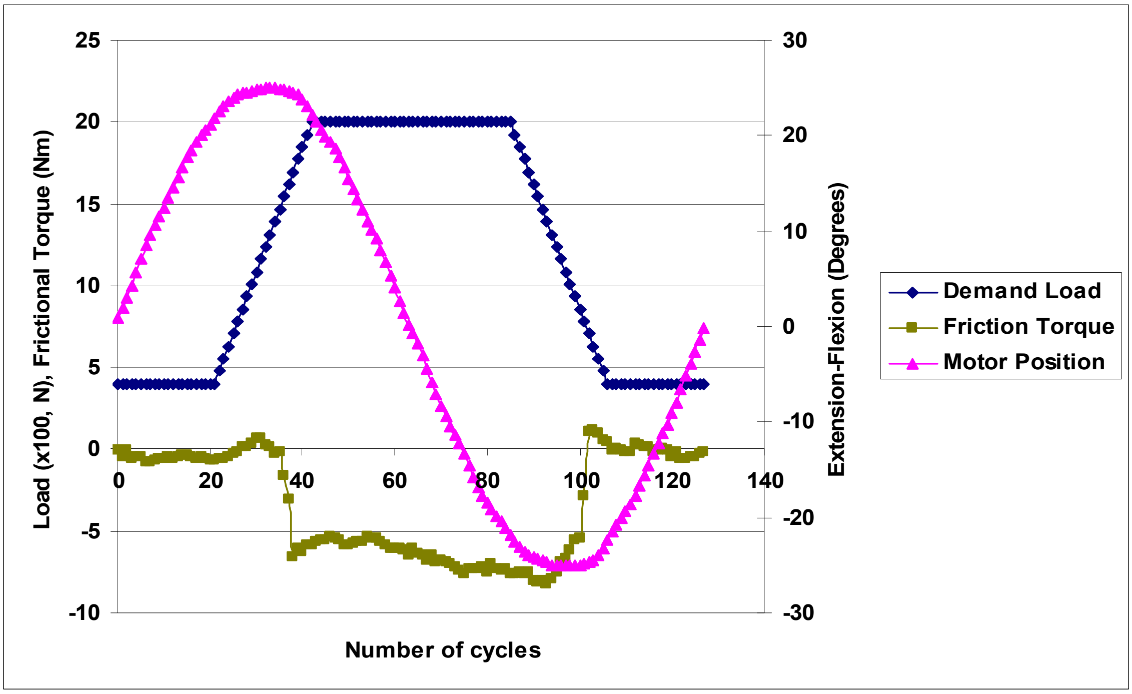

Figure 4-53. Friction Torque versus number of cycles for the 176 μm diametral clearance, 50mm BHR bearing using Blood (η=0.0083 Pas) as lubricant after initial deflection.

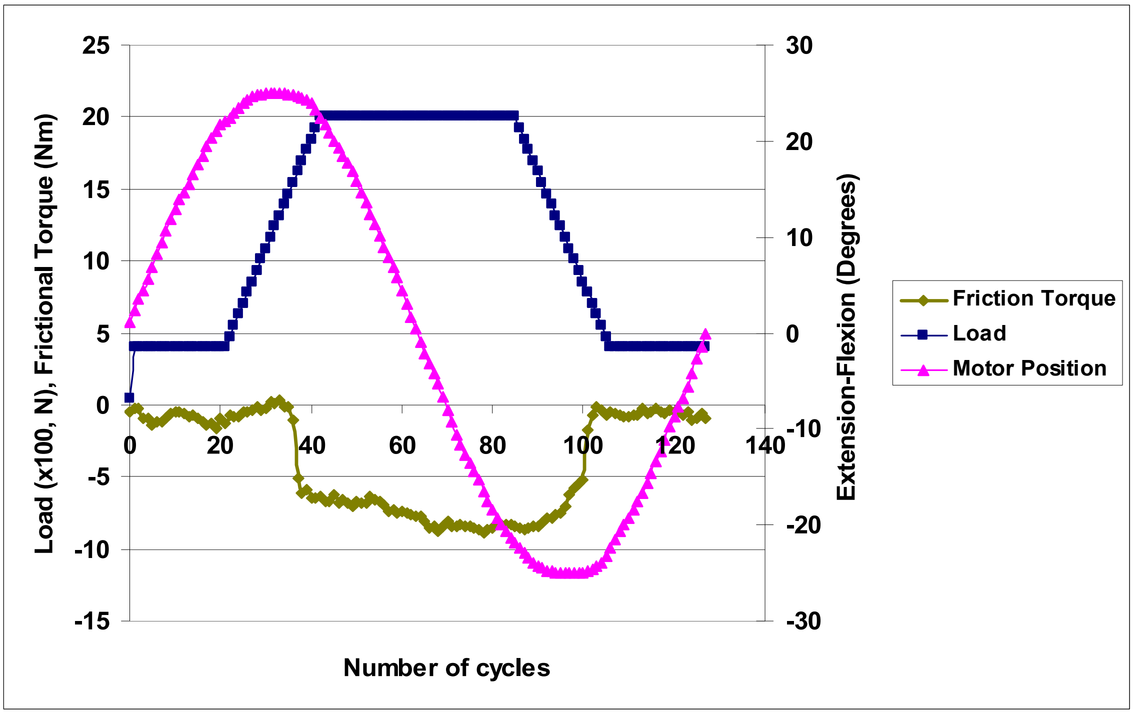

Figure 4-54. Friction Torque versus number of cycles for the 280 μm diametral clearance, 50mm BHR bearing using Blood (η=0.0083 Pas) as lubricant after initial deflection.

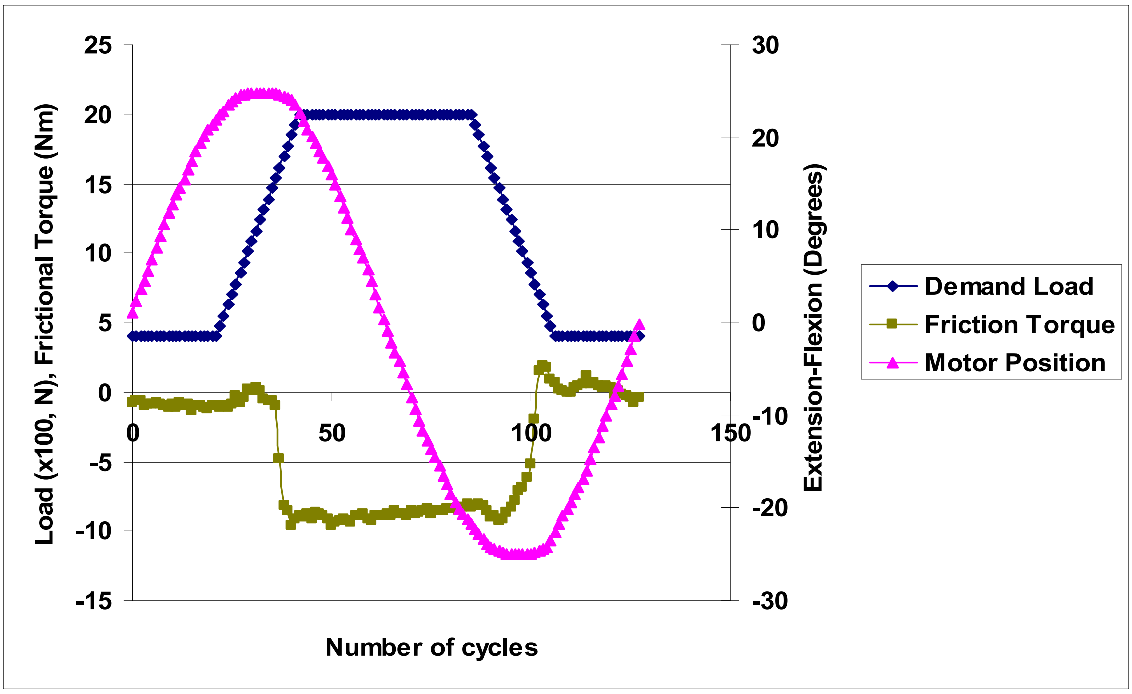

Figure 4-55. Friction Torque versus number of cycles for the 50 μm diametral clearance, 50mm BHR bearing using Clotted blood (η=0.0108 Pas) as lubricant after initial deflection.

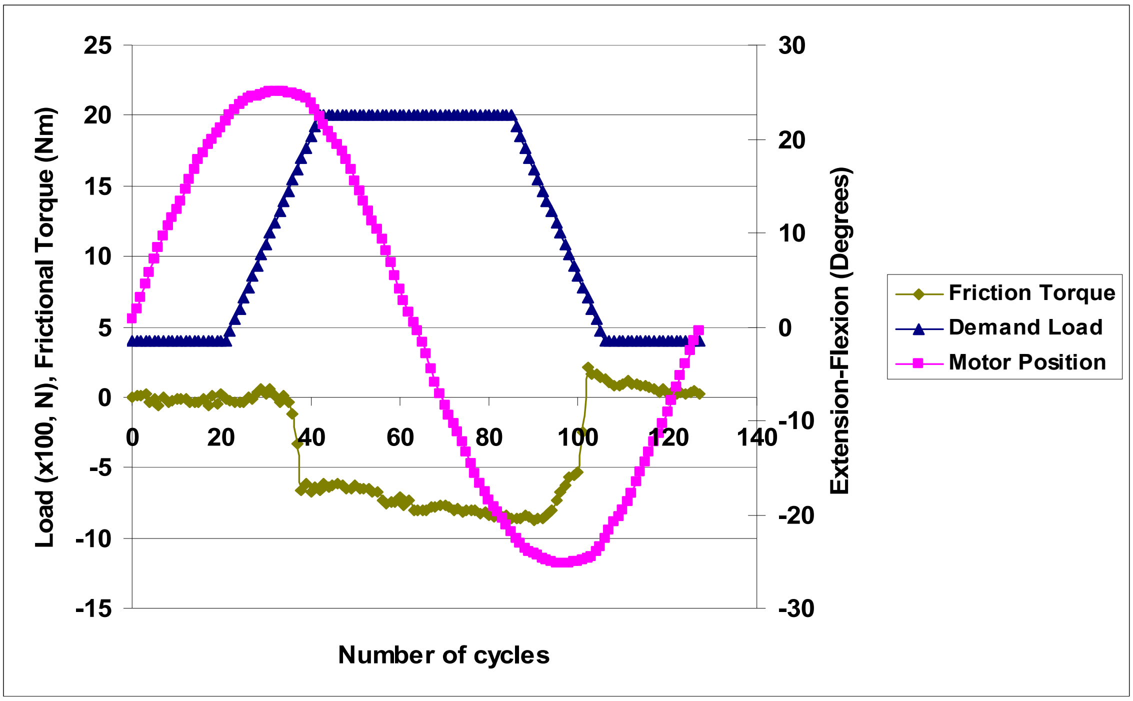

Figure 4-56. Friction Torque versus number of cycles for the 176 μm diametral clearance, 50mm BHR bearing using Clotted blood (η=0.0108 Pas) as lubricant after initial deflection.

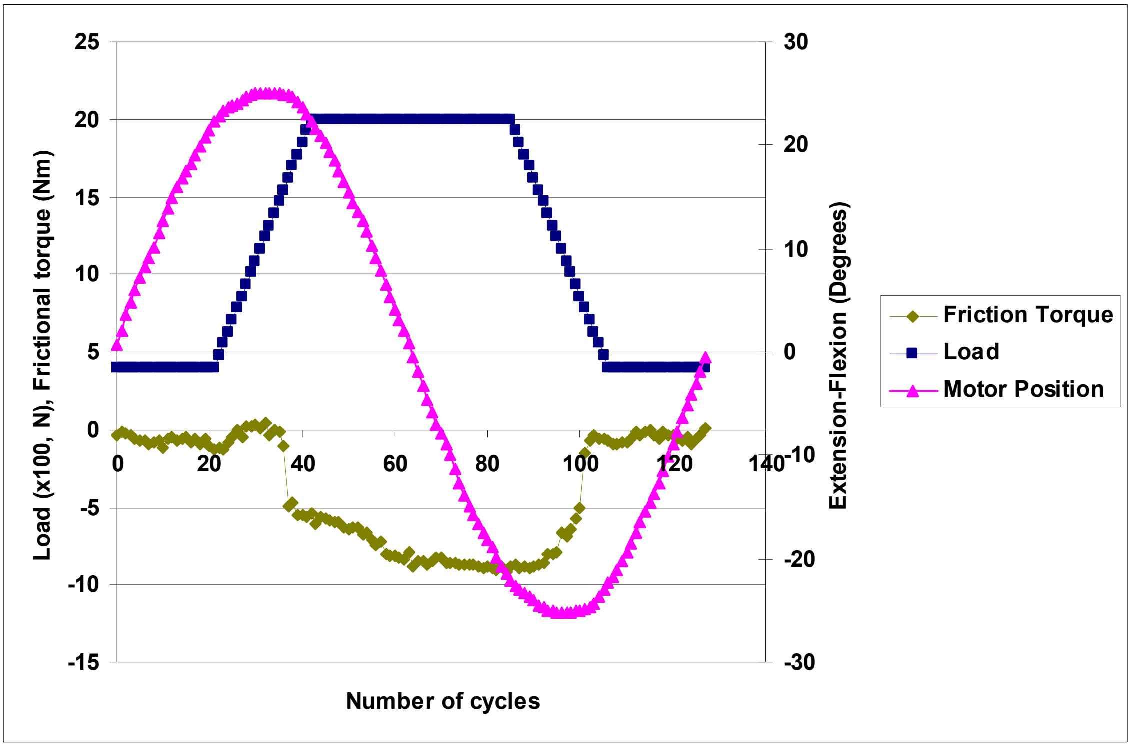

Figure 4-57. Friction Torque versus number of cycles for the 280 μm diametral clearance, 50mm BHR bearing using Clotted blood (η=0.0108 Pas) as lubricant after initial deflection.

It has been reported that smaller clearances can reduce bedding-in wear and hence reduced friction factors are expected [Farrar *et al.,* 1997; Jin *et al.,* 2002] and may improve lubrication conditions. From clinical experiences, however, no evidence has supported the theory that larger clearances can lead to a reduction in the life of MoM hip prostheses. Indeed, small clearances may increase the risk of equatorial or near equatorial contact raising the frictional torque. High friction torque was believed to be a major problem in the early generation of MoM hip prostheses [Unsworth *et al.,* 1988] and was a factor that led to their discontinued usage. Also, the role of frictional torque in loosening at the cement-bone interface have been evaluated [Mai *et al.,* 1996] for various hip resurfacing bearings of diameters 36, 39, 43, 47, 51 and 54 mm retrieved from 156 patients. It was reported that despite of high frictional torques due to the increased diameter of the bearing surface and the increased average load, the larger prostheses survived significantly longer than the smaller ones. Also, radiograph analysis of the retrieved specimens suggested that regardless of the size of the implant, the mechanism of loosening on both the acetabular and femoral side of the double-cup replacement was progressive resorption (migration due to lose of bone mass) of bone induced by polyethylene wear particles. It was therefore concluded that frictional torque was not the primary factor in the loosening of these prostheses with a large bearing surface and that high friction factor and friction torque can be tolerated if the range of worn debris is significantly reduced. These findings therefore strongly indicate clearly the importance of having an alternative to polyethylene bearings such as the large diameter MoM Birmingham Hip Resurfacing prostheses.

4.10 Dynamic Motion Profiles for the S&N BHR devices after final cup deflection using Clotted blood and Blood as lubricants

Table 4.14 gives the average friction torque produced during dynamic friction tests for three different clearances using clotted blood (η=0.0234 Pas) and blood (η=0.0112 Pas) as lubricants. From Table 4.14 and Figures 4-58 to 4-63, it is clear that there is a significant reduction in frictional torque from ~8.38 to ~6.12 Nm and ~8.85 to ~6.25 Nm for blood and clotted blood as diametral clearance increased from 13 to 247 µm.

Table 4.14: Average frictional torque for various diametral clearances of 13, 131 and 247µm using blood and clotted blood as lubricants.

Diametral clearance, µm	Friction Torque (Nm), for blood η=0.0112 Pas	Friction Torque (Nm), for clotted blood η=0.0234 Pas
13	8.38	8.85
131	8.10	7.9
247	6.12	6.25

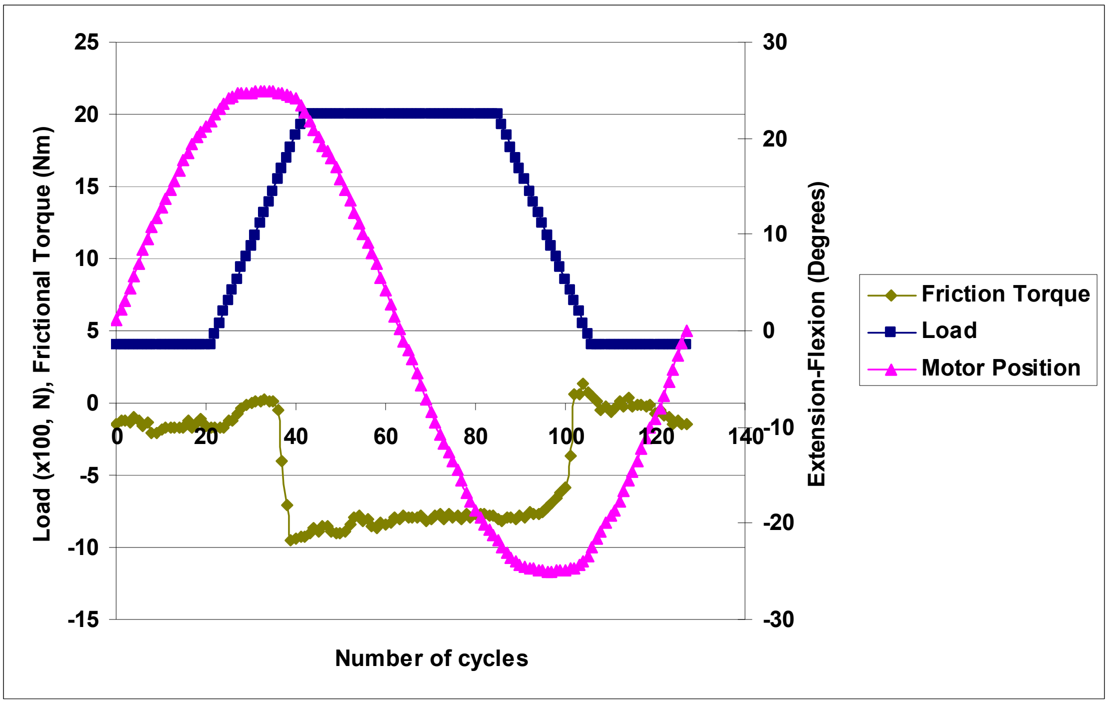

Figure 4-58. Friction Torque versus number of cycles for the 13 µm diametral clearance, 50mm BHR bearing using blood (η=0.0112 Pas) as lubricant after final deflection.

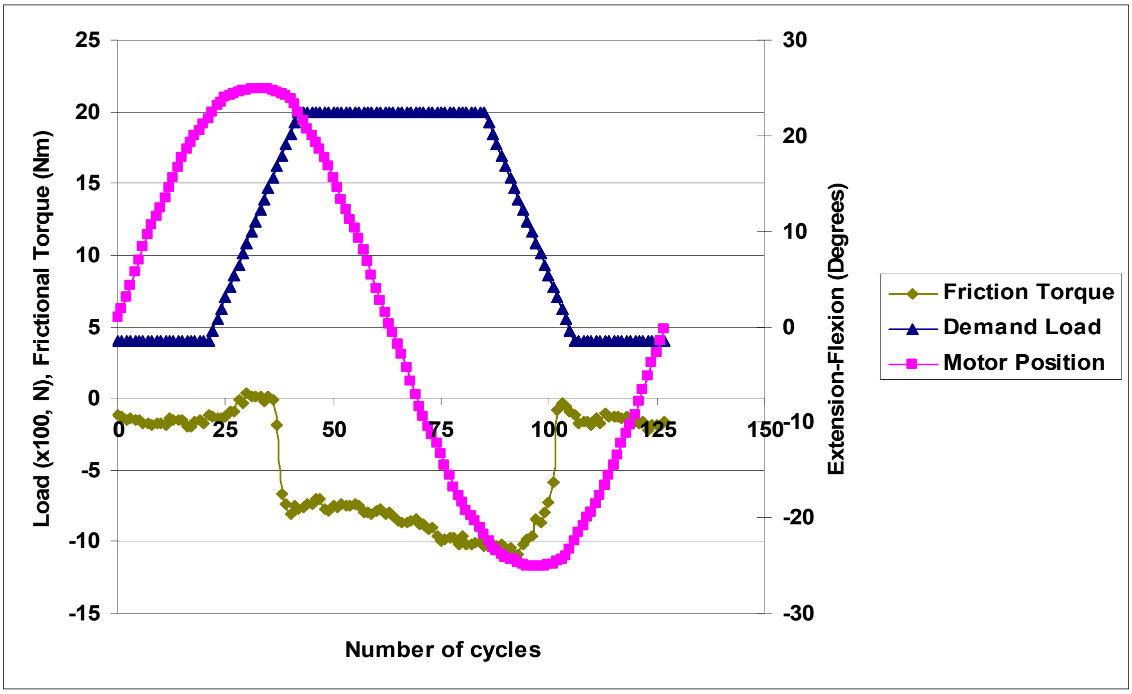

Figure 4-59. Friction Torque versus number of cycles for the 131 µm diametral clearance, 50mm BHR bearing using blood (η=0.0112 Pas) as lubricant after final deflection.

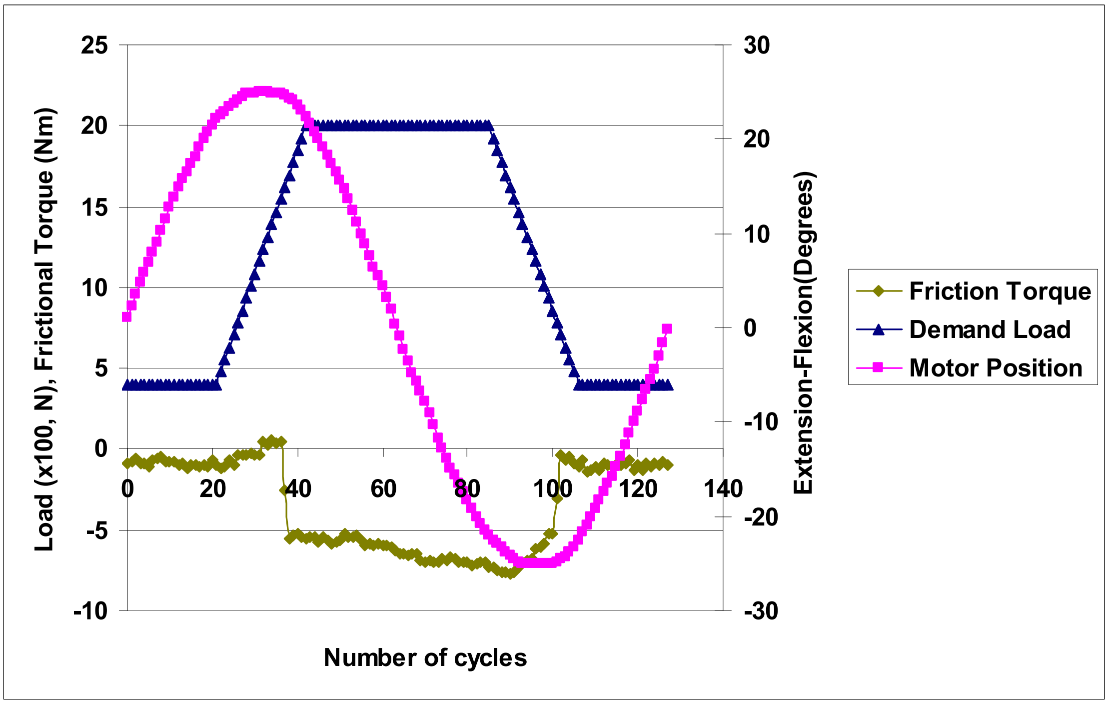

Figure 4-60. Friction Torque versus number of cycles for the 247 µm diametral clearance, 50mm BHR bearing using blood (η=0.0112 Pas) as lubricant after final deflection.

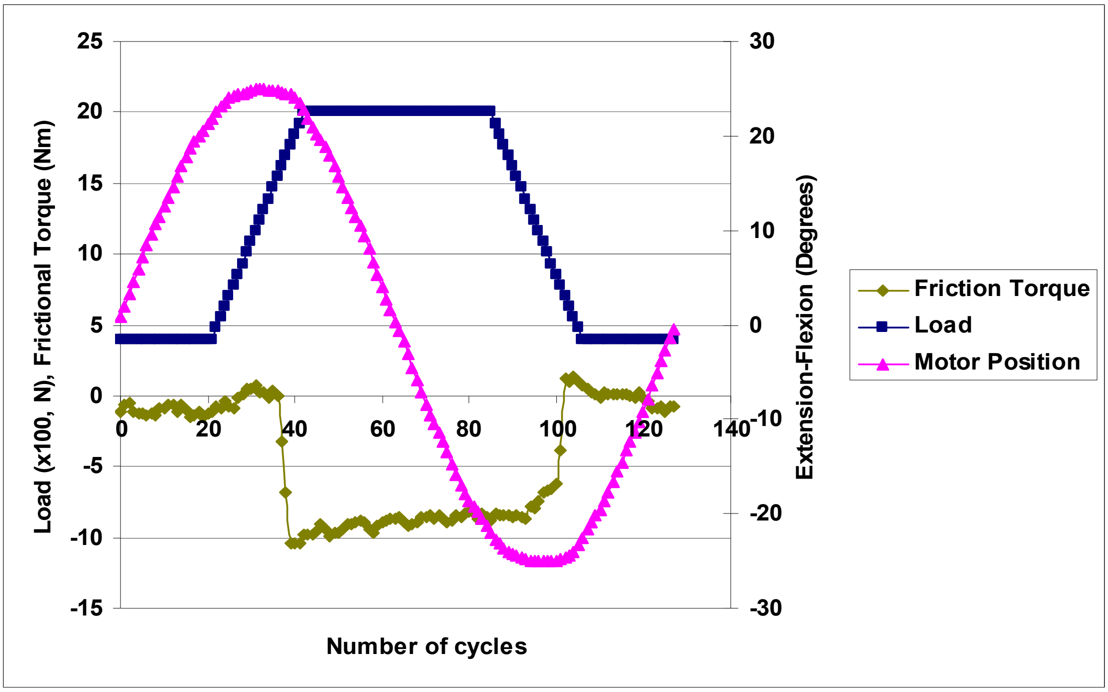

Figure 4-61. Friction Torque versus number of cycles for the 13 μm diametral clearance, 50mm BHR bearing using clotted blood (η=0.0234 Pas) as lubricant after final deflection.

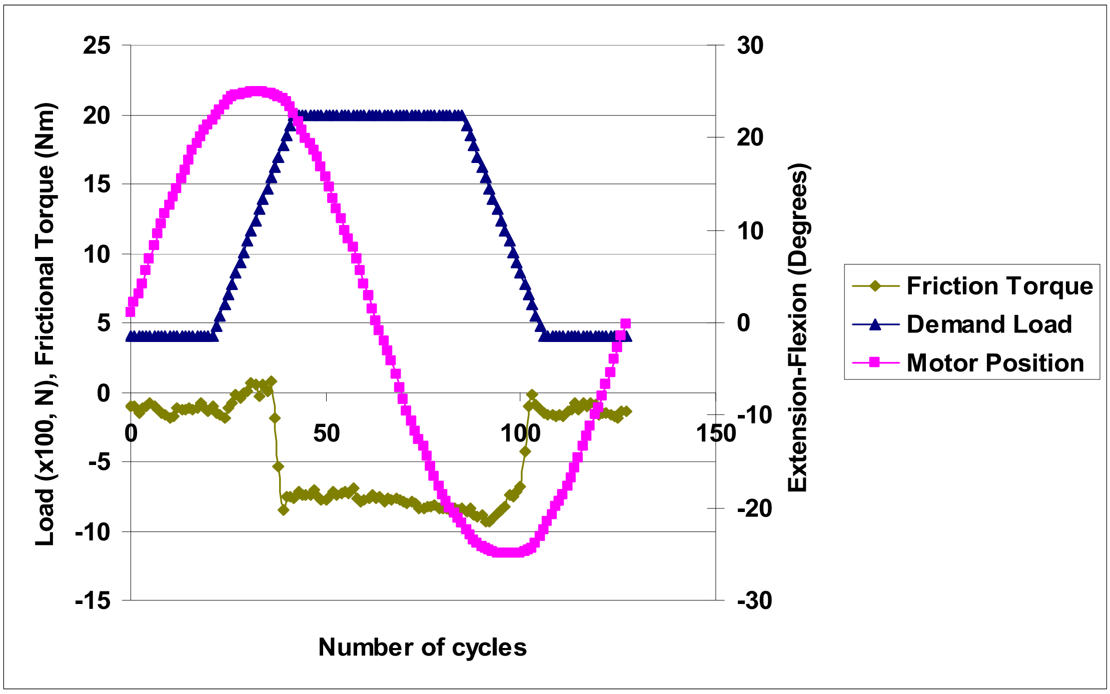

Figure 4-62. Friction Torque versus number of cycles for the 131 μm diametral clearance, 50mm BHR bearing using clotted blood (η=0.0234 Pas) as lubricant after final deflection.

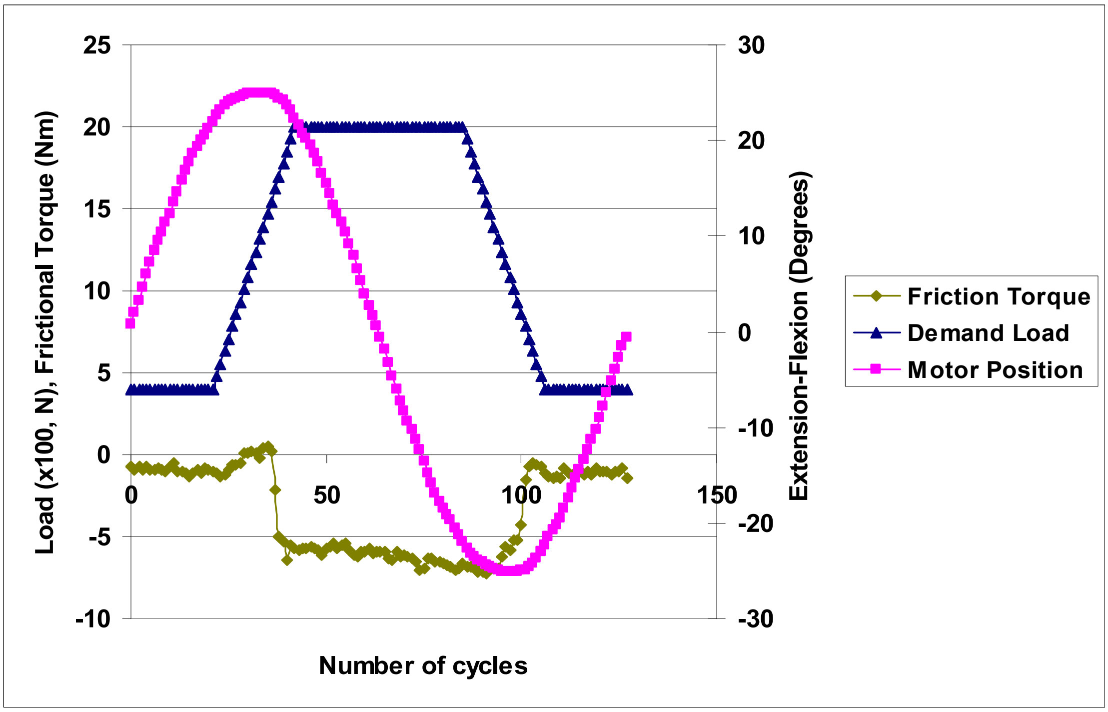

Figure 4-63. Friction Torque versus number of cycles for the 247 μm diametral clearance, 50mm BHR bearing using clotted blood (η=0.0234 Pas) as lubricant after final deflection.

Overall discussion

It has been shown via simulator studies [Jin 2002, Smith et al., 2004, 2001] that an increase in the femoral head diameter from 16 to 28mm led to an increase in wear as also predicted from the classical Lancaster equation, but a further increase from 28 to 36mm resulted in improved lubrication and formation of fluid film due to the elastohydrodynamic lubrication action. The range of diametral clearance for the entire family of various joint diameters is from ~90 to 200 microns, with each bearing size having an optimized gap (clearance) for maximum fluid film thickness. However, if the gap between the articulating components is too small or too large there will be a sharp increase in friction and wear rates.

It is to be noted that the introduction of the second-generation metal-on-metal (MOM) hip resurfacing prostheses has been based on extensive laboratory simulator testing and design optimization leading to optimization of the diametral clearance and hence lower friction and wear as a result of improved lubricity. Initially, however, a larger diametral clearance of ~ 300µm was mainly adopted in the first-generation of MOM hip resurfacing prostheses. This was optimized for the second-generation hip resurfacing bearings to smaller clearances, typically between 100 to 150µm [McMinn 2009] and in this work to be ≥150µm but <235µm with an optimum clearance of ~175µm as seen in this work giving the lowest friction factors in a range of physiological viscosities (0.001-0.2Pas).

It has become clear that there is a direct relationship between clearance and lubrication, and that metal-on-metal bearings are lubrication sensitive, and also clearance has a direct effect on wear. In this respect, it has been reported [Dowson and Jin, 2005] that for both 36 and 54 mm bearings as diametral clearance increased, bedding in wear of the metal-on-metal components increased significantly. For the resurfacing components, those couples with smaller diametral clearances (83–129µm) with a head diameter of 54mm exhibited running in wear rates that were four-fold lower and steady-state wear rates that were two-fold lower than those components with larger clearances (254–307µm) with the same head diameter. However, there appear to be an optimum band of clearance (126 µm), which produces favourable wear rates [Leslie et al, 2008].

Tribology theories and hip joint simulator studies have also predicted that friction, lubrication and wear within these bearing systems are affected by several factors including load applied, material hardness, surface roughness, bearing diameter, sliding speed, radial clearance and the viscosity of the lubricant [Dowson et al., 2004, Liu et al., 2006, 2005; Rieker et al., 2005; Smith et al., 2001a, b, c; Udofia et al., 2003].

As mentioned earlier in this thesis, immediately after joint implantation the artificial implant is actually soaked in blood (instead of synovial fluid) for couple of weeks or even months and that increased bearing friction in this postoperative period (due to the presence of blood) can lead to micromotion which has the potential to prevent effective bony ingrowth leading to fixation impairment and reduced longevity. The aim of present work was, therefore, to investigate the frictional and lubrication behaviour of a group of Birmingham Hip Resurfacing (BHR) prostheses with a nominal diameter of 50mm and different clearances in the range 80 to 306μm using lubricants such as blood and a combination of bovine serum with carboxymethyl cellulose (CMC), with or without hyaluronic acid (HA), adjusted to a range of physiological viscosities (0.001-0.2Pas). This was carried out using a friction hip simulator to obtain friction factors and then Stribeck analyses were carried out to assess the lubricating modes. The results of this study suggest, therefore, that reduced clearance MOM bearings have the potential to generate high friction especially in the early weeks after implantation when blood is indeed the in vivo lubricant. Friction factors in higher clearance bearings were much reduced in comparison. This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth resulting in impaired fixation with long-term implications for survival. The frictional studies in this work, therefore, have shown that lower clearances do not necessarily reduce the friction factors to a level for the presence of full fluid film lubrication and that the friction factors decrease with increase in diametral clearance for high viscosity (0.01-0.02 Pas) fluids. It is to be noted that friction between the bearing surfaces is the combination of direct contact between the bearing surfaces and the internal friction of the lubricant. For a small clearance, the shear rate of the lubricant will be higher than the larger clearance, e.g. the shear rate for the 80 μm clearance would be higher than that of the larger 306 μm clearance which suggests that a high friction factor may be caused due to the internal friction of the lubricant, especially when the viscosity is high as seen in this study. This means that the friction force will then be dominated by the internal friction of the lubricant and for a smaller clearance, the bearing area can easily extend in the equatorial direction, which can result in higher contact stresses on the bearing surface near the equatorial area and hence cause a higher friction torque under the same load.

It is only obvious that the engineering issues surrounding optimal metal-on-metal prostheses have been the centre of much debate and research in the past. Ongoing research into the in vitro friction, lubrication and wear performance of these bearings as a function of macrogeometry (bearing diameter, clearance, and component thickness) and microgeometry (roundness and surface finish) are carried out in hip/knee function simulators with lubricants that are believed to simulate the natural joint fluid in terms of viscosity. However, as discussed in this work and very clearly these lubricants have the limitation of being unable to simulate the friction effects of macromolecules, and thus, to our knowledge, factors such as cellular and macromolecular shear that can affect friction in these bearings, in vivo, have not been specifically investigated in vitro.

The results of this study suggest, therefore, that reduced clearance MOM bearings have the potential to generate high friction especially in the early weeks after implantation when blood is indeed the in vivo lubricant. This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth resulting in impaired fixation with long-term implications for survival. It became clear that the friction factors decreased consistently with increase in diametral clearance for both blood and clotted blood with opposite effect for BS+CMC and BS+HA+CMC of similar viscosities (~0.013 Pas). This therefore suggested that higher clearances will lower the friction for these large diameter S&N BHR devices depending on the type of lubricant and viscosity. The friction factors were higher for both blood and clotted blood especially at lower clearances as compared to the other lubricants indicating that lower diametral clearances may increase the risk of micromotion during the early weeks/months after hip implantation which in turn may adversely affect the longevity of the implant.

Finally, it is strongly believed that the selection of optimum diametral clearance between the femoral head and the acetabular cup is a critical factor for the success of MOM bearings and thus an important consideration for the design and manufacturing of MOM hip prostheses. So far, clinical studies, have not provided any evidence that larger clearances can cause reduction in the life of the MOM hip prostheses and, in fact, we believe by evidence from this work that small clearances may increase the risk of equatorial or near equatorial contact causing the frictional torque to rise to high levels leading to loosening and eventual dislocation of the MOM hip prostheses which was a major reason for the earlier discontinuation of MOM bearings [Scholes et al., 2006, 2001].

CHAPTER FIVE

5.1 Conclusions

- Various lubricants having different viscosities (0.001-0.2 Pas) were used to study the in vitro frictional and lubrication behaviour of six large diameter (50mm nominal) Smith & Nephew BHR prostheses with various diametral clearances (~ 80-300 μm). These lubricants included blood and clotted blood to understand and mimic the in vivo frictions generated at the articulating surfaces immediately after hip implantation. Other lubricants used were BS+CMC and BS+HA (+CMC) to compare (and understand the difference) with blood which is the actual in vivo lubricant for about couple of months after total hip joint replacement.
- It became clear that the friction factors decreased consistently with increase in diametral clearance for both blood and clotted blood and only for those lubricants with viscosities of 0.105 and 0.19 Pas for BS+CMC, and 0.037 and 0.138 Pas for BS+HA+CMC. This, therefore, suggested that higher clearances will lower the friction (and hence wear) for these large diameter S&N BHR devices depending on the type of lubricant and viscosity.
- The friction factors were higher for both blood and clotted blood especially at lower clearances as compared to the other lubricants indicating that lower diametral clearances may increase the risk of micromotion leading to dislocation of the bearings during the early weeks/months after hip implantation.
- The friction factors decreased in the range ~ 0.19-0.14 for blood, ~ 0.13-0.1 for BS+CMC and ~ 0.12-0.1 for BS+HA (+CMC) having viscosities of 0.01, 0.19 and 0.138 Pas, respectively, with increase in diametral clearance (from 80 to 306 μm).

- For the BS+CMC lubricants, the Stribeck analysis showed a decreasing friction factor with increase in Sommerfeld number (i.e. as viscosity increased) indicating a mixed lubrication regime up to a viscosity of 0.0136 Pas above which the friction factor increased slightly and then levelled off especially for the lower diametral clearances of 80, 135, 175 and 200 µm suggesting a transition from mixed to possibly fluid film lubrication regime. The higher diametral clearances of 243 and 306 µm did not show this transitional change and thus the mixed lubrication was the dominant mode.
- For the BS+HA+CMC lubricants, the Stribeck analysis showed a decreasing friction factor with increase in Sommerfeld number for clearances ≥200 µm indicating a mixed lubrication regime up to a viscosity of 0.037 Pas above which the friction factor increased slightly or levelled off suggesting the possibility of a fluid film lubrication regime. Opposite to this effect was that of the 80 (and almost 130) µm clearance for which friction factor increased with increase in Sommerfeld number implying a fluid film lubrication mode.
- The friction factors obtained in this work for the 50mm (nominal diameter) MOM S&N BHR prostheses were lower than those for the 28mm (nominal diameter) MOM THR bearings (reported by others [Scholes, S. et al, 2000]) using similar lubricants (BS+CMC) of similar viscosities.
- Another six large diameter (50mm nominal) BHR deflected prostheses with various clearances (~ 50-280µm after cup deflection) were also friction tested in vitro in the presence of blood and clotted blood to study the effect of cup deflection on friction. It was found that the biological lubricants caused higher friction factors at the lower diametral clearances for blood and clotted blood as clearance decreased from 280µm to

50μm (after cup deflection). It is postulated that if the cup is deflected by press fitting, this may result in increased contact at bearing surfaces around the equatorial rib of the cup and result in higher frictional torque which can increase the risk of dislocation and hamper fixation. This has been the case for some early loosening of the implants after few weeks of implantation. This work, therefore, showed clearly that higher clearances will lower the friction for large diameter BHR bearings, which, in turn, may accommodate for the amount of deflection that occurs in the cups during press-fit arthroplasty.

- Finally, it is believed strongly that the optimum clearance for the tested 50 mm diameter BHR implants is about ~200 μm using the above mentioned lubricants [Afshinjavid and Youseffi, 2010] in order to obtain low friction with good lubrication (being mixed mode dominantly).

5.2 Further future work

Further friction tests and lubrication analyses using similar lubricants with similar viscosities for different sizes of metal-on-metal hip resurfacing implants, i.e. 40 to 60 mm diameter, and various clearances (100-300 μm) should be carried out in order to establish the optimal clearance for each size, and hence be able to compare their tribological properties with those of the 50 mm BHR implants obtained in this work. This will allow the orthopaedic manufacturers to have the necessary data for their production lines and the surgeons for choosing the correct size (depending on the size of the patient's hip) and clearance for longer lasting implants and thus for improving patients life and avoiding revision surgery.

It is also necessary to carry out the above mentioned tests for any change in design of the current hip resurfacing implants *in vitro* and *in vivo* to avoid rejection by the patient due to high frictions leading to massive amount of wear and causing the occurrence of pseudo-tumours as reported for only one type of hip resurfacing prosthesis via a major orthopaedic company.

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List of Publications

1. S. Afshinjavid and M. Youseffi, 2010, Effect of cup deflection on friction of hip resurfacing prostheses with various clearances using blood and clotted blood as lubricants, The World Congress on Engineering, WCE 2010, London, U.K., Volume I, ISBN: 978-988-17012-9-9 and ISNN: 2078-0958, full paper, pages 598-600.

2. S. Afshinjavid and M. Youseffi, 2009, The effect of clearance upon friction of large diameter hip resurfacing prostheses using blood, clotted blood and bovine serum as lubricants, World Congress on Medical Physics and Biomedical Engineering, Munich-Germany, 7-12 September 2009, International Federation for Medical and Biological Engineering 2009 (IFMBE) Proceedings 25/IX, Abstract/Poster; www.springerlink.com

3. S. Afshinjavid and M. Youseffi, 2009. The effect of clearance upon friction of large diameter hip resurfacing prostheses using blood, clotted blood and bovine serum as lubricants. World Congress on Medical Physics and Biomedical Engineering, Munich-Germany, 7-12 September 2009, International Federation for Medical and Biological Engineering (IFMBE) Proceedings, Volume 25/IX, ISBN: 978-3-642-03888-4, full paper, pages 418-420; www.springerlink.com

4. S. Afshinjavid and M. Youseffi, 2009, Effect of Cup Deflection on Friction of Hip Resurfacing Prostheses with various Clearances using Blood and Clotted blood as Lubricants (Abstract), 15th International Biomedical Science and Technology Symposium, Middle East Technical University- Northern Cyprus Campus, Guzelyut, TRNC (Biomed 16-19 August 2009).

5. S. Afshinjavid; M. Youseffi; A. Kamali; J.T. Daniel; T. Band; R. Ashton; A. Hussain; C.X. Li; J. Daniel; D.J.W. McMinn, 2007, The effects of cup deflection on friction in metal-on-metal bearings, Journal of Bone and Joint Surgery - British Volume, Vol 90-B, Issue SUPP_III, 552.

Appendix A

Appendix 1:

Friction Test Results (Data) obtained from the Friction Hip Simulator for the 50 mm Diameter Smith & Nephew Birmingham Hip Resurfacing Prosthesis with 80 μm Diametral Clearance and using Whole Blood (of viscosity ~ 0.01 Pas) as Lubricant.

Whole blood, 50mm diameter BHR, 80µm diametral clearance

Index	Demand Load(N)	Demand Motor(○)	Load(N)	Friction Torque(Nm)	Motor Position(○)	Friction Coefficient
0	100	0	236.36	-0.59	0.7	-0.0995
1	100	1.23	271.82	-0.5	1.93	-0.0742
2	100	2.45	286.59	-0.22	3.25	-0.0309
3	100	3.67	310.23	-0.3	4.57	-0.0382
4	100	4.88	333.86	-0.48	5.98	-0.057
5	100	6.08	333.86	-0.43	7.21	-0.0512
6	100	7.26	325	-0.3	8.44	-0.0365
7	100	8.42	345.68	-0.36	9.49	-0.0414
8	100	9.57	313.18	-0.3	10.55	-0.0383
9	100	10.69	268.86	-0.51	11.51	-0.0755
10	100	11.79	224.55	-0.36	12.48	-0.0646
11	100	12.85	192.05	-0.36	13.62	-0.0759
12	100	13.89	180.23	-0.42	14.77	-0.0941
13	100	14.89	174.32	-0.54	15.91	-0.1249
14	100	15.86	162.5	-0.48	16.96	-0.1193
15	100	16.79	165.45	-0.55	17.84	-0.1328
16	100	17.68	197.95	-0.52	18.63	-0.1043
17	100	18.52	218.64	-0.51	19.34	-0.0932
18	100	19.32	257.05	-0.57	19.95	-0.0893
19	100	20.08	277.73	-0.54	20.57	-0.0779
20	100	20.79	277.73	-0.36	21.27	-0.0515
21	100	21.44	277.73	-0.27	21.97	-0.0382
22	190.48	22.05	277.73	-0.25	22.59	-0.0367
23	280.95	22.6	277.73	-0.31	23.2	-0.0442
24	371.43	23.1	277.73	-0.37	23.64	-0.054
25	461.9	23.54	280.68	-0.33	23.99	-0.0475
26	552.38	23.92	372.27	-0.31	24.26	-0.0332
27	642.86	24.25	466.82	-0.26	24.52	-0.0221
28	733.33	24.52	573.18	-0.19	24.7	-0.0135
29	823.81	24.73	694.32	-0.17	24.78	-0.0097
30	914.29	24.88	803.64	-0.02	24.96	-0.0011
31	1004.76	24.97	871.59	0.01	24.96	0.0006
32	1095.24	25	945.45	0.04	24.96	0.0018
33	1185.71	24.97	1010.45	0.07	24.96	0.0028
34	1276.19	24.88	1090.23	0	24.96	0

35	1366.67	24.73	1187.73	0.12	24.87	0.0041
36	1457.14	24.52	1252.73	0.06	24.78	0.002
37	1547.62	24.25	1285.23	-0.41	24.7	-0.0128
38	1638.1	23.92	1382.73	-1.08	24.34	-0.0313
39	1728.57	23.54	1459.54	-2.42	23.91	-0.0663
40	1819.05	23.1	1509.77	-1.93	23.47	-0.0511
41	1909.52	22.6	1619.09	-1.91	22.94	-0.0471
42	2000	22.05	1690	-1.85	22.24	-0.0438
43	2000	21.44	1763.86	-3.57	21.53	-0.081
44	2000	20.79	1828.86	-3.08	20.74	-0.0674
45	2000	20.08	1787.5	-3.42	20.13	-0.0765
46	2000	19.32	1769.77	-3.42	19.34	-0.0774
47	2000	18.52	1752.04	-3.47	18.54	-0.0793
48	2000	17.68	1755	-3.56	17.58	-0.0812
49	2000	16.79	1734.32	-3.54	16.52	-0.0817
50	2000	15.86	1728.41	-3.32	15.47	-0.0769
51	2000	14.89	1710.68	-3.29	14.41	-0.077
52	2000	13.89	1707.73	-3.19	13.36	-0.0748
53	2000	12.85	1707.73	-3.24	12.22	-0.0758
54	2000	11.79	1775.68	-2.95	11.34	-0.0666
55	2000	10.69	1808.18	-3.3	10.37	-0.073
56	2000	9.57	1802.27	-3.21	9.23	-0.0711
57	2000	8.42	1793.41	-3.29	7.91	-0.0734
58	2000	7.26	1781.59	-3.12	6.5	-0.07
59	2000	6.08	1772.73	-2.92	5.19	-0.0659
60	2000	4.88	1784.54	-3.04	3.95	-0.0681
61	2000	3.67	1808.18	-2.97	2.72	-0.0657
62	2000	2.45	1820	-3.05	1.58	-0.067
63	2000	1.23	1834.77	-2.68	0.53	-0.0584
64	2000	0	1831.82	-3.39	-0.53	-0.0739
65	2000	-1.23	1817.04	-3.4	-1.67	-0.0749
66	2000	-2.45	1817.04	-3.39	-2.99	-0.0747
67	2000	-3.67	1805.23	-3.49	-4.39	-0.0772
68	2000	-4.88	1811.14	-3.52	-5.71	-0.0777
69	2000	-6.08	1843.64	-3.57	-6.94	-0.0775
70	2000	-7.26	1864.32	-3.57	-8.09	-0.0766
71	2000	-8.42	1879.09	-3.57	-9.14	-0.076
72	2000	-9.57	1893.86	-3.36	-10.02	-0.0709

73	2000	-10.69	1902.73	-3.57	-10.99	-0.0751
74	2000	-11.79	1917.5	-3.57	-12.04	-0.0745
75	2000	-12.85	1905.68	-3.57	-13.18	-0.075
76	2000	-13.89	1837.73	-3.57	-14.41	-0.0777
77	2000	-14.89	1825.91	-3.57	-15.56	-0.0782
78	2000	-15.86	1817.04	-3.57	-16.52	-0.0786
79	2000	-16.79	1822.95	-3.57	-17.4	-0.0784
80	2000	-17.68	1870.23	-3.57	-18.19	-0.0764
81	2000	-18.52	1890.91	-3.57	-18.81	-0.0756
82	2000	-19.32	1929.32	-3.57	-19.51	-0.0741
83	2000	-20.08	1917.5	-3.57	-20.21	-0.0745
84	2000	-20.79	1917.5	-3.57	-21.01	-0.0745
85	2000	-21.44	1920.45	-3.57	-21.71	-0.0744
86	1909.52	-22.05	1926.36	-3.57	-22.32	-0.0742
87	1819.05	-22.6	1926.36	-3.57	-22.85	-0.0742
88	1728.57	-23.1	1822.95	-3.57	-23.29	-0.0784
89	1638.1	-23.54	1692.95	-3.57	-23.64	-0.0844
90	1547.62	-23.92	1627.95	-3.57	-23.99	-0.0878
91	1457.14	-24.25	1509.77	-3.57	-24.26	-0.0946
92	1366.67	-24.52	1453.64	-3.57	-24.52	-0.0983
93	1276.19	-24.73	1373.86	-3.57	-24.78	-0.104
94	1185.71	-24.88	1311.82	-3.57	-24.87	-0.1089
95	1095.24	-24.97	1243.86	-3.53	-24.96	-0.1135
96	1004.76	-25	1178.86	-3.1	-24.96	-0.1052
97	914.29	-24.97	1110.91	-3.07	-24.96	-0.1105
98	823.81	-24.88	1040	-2.88	-24.96	-0.1108
99	733.33	-24.73	957.27	-2.91	-24.96	-0.1216
100	642.86	-24.52	892.27	-2.31	-24.87	-0.1037
101	552.38	-24.25	806.59	-2	-24.78	-0.0992
102	461.9	-23.92	732.73	-0.1	-24.52	-0.0055
103	371.43	-23.54	673.64	0.32	-24.26	0.0188
104	280.95	-23.1	617.5	1.01	-23.73	0.0656
105	190.48	-22.6	555.45	1.64	-23.11	0.1181
106	100	-22.05	455	0.52	-22.32	0.0454
107	100	-21.44	387.05	0.22	-21.53	0.0227
108	100	-20.79	310.23	-0.01	-20.83	-0.0009
109	100	-20.08	265.91	-0.37	-20.13	-0.0554
110	100	-19.32	271.82	0.15	-19.34	0.0218

111	100	-18.52	260	-0.01	-18.46	-0.0011
112	100	-17.68	265.91	-0.07	-17.4	-0.0102
113	100	-16.79	345.68	-0.14	-16.35	-0.0161
114	100	-15.86	381.14	-0.2	-15.29	-0.0205
115	100	-14.89	401.82	-0.01	-14.33	-0.001
116	100	-13.89	390	-0.06	-13.27	-0.0066
117	100	-12.85	395.91	0.13	-12.22	0.0136
118	100	-11.79	413.64	-0.35	-11.25	-0.0341
119	100	-10.69	378.18	0.02	-10.2	0.0022
120	100	-9.57	333.86	-0.03	-8.96	-0.0036
121	100	-8.42	301.36	-0.2	-7.65	-0.0271
122	100	-7.26	233.41	-0.46	-6.33	-0.0795
123	100	-6.08	221.59	-0.49	-5.01	-0.0888
124	100	-4.88	221.59	-0.38	-3.78	-0.0693
125	100	-3.67	221.59	-0.43	-2.55	-0.0774
126	100	-2.45	227.5	-0.33	-1.49	-0.0576
127	100	-1.23	224.55	-0.64	-0.44	-0.114

Appendix 2:

Friction Test Results (Data) obtained from the Friction Hip Simulator for the 50 mm Diameter Smith & Nephew Birmingham Hip Resurfacing Prosthesis with 80 µm Diametral Clearance and using Bovine Serum with CMC (of viscosity ~ 0.105 Pas) as Lubricant.

BS+CMC of η=0.105 Pas, 50mm diameter BHR, 80μm diametral clearance

Index	Demand Load(N)	Demand Motor(○)	Load(N)	Friction Torque(Nm)	Motor Position(○)	Friction Coefficient
0	400	0	542.34	-1.25	1.14	-0.0921
1	400	1.23	510.8	-1.62	2.37	-0.1272
2	400	2.45	459.85	-1.62	3.52	-0.1413
3	400	3.67	413.76	-1.83	4.48	-0.1769
4	400	4.88	404.05	-1.73	5.54	-0.171
5	400	6.08	411.33	-2	6.77	-0.1946
6	400	7.26	455	-2.04	8.09	-0.179
7	400	8.42	481.69	-1.21	9.4	-0.1008
8	400	9.57	525.36	-1.8	10.63	-0.1368
9	400	10.69	532.63	-1.76	11.78	-0.1323
10	400	11.79	530.21	-1.42	12.83	-0.1071
11	400	12.85	542.34	-1.39	13.8	-0.1022
12	400	13.89	518.08	-1.62	14.68	-0.1255
13	400	14.89	474.41	-1.45	15.47	-0.1226
14	400	15.86	421.04	-2	16.35	-0.1902
15	400	16.79	408.9	-1.97	17.4	-0.1924
16	400	17.68	423.46	-1.93	18.37	-0.1826
17	400	18.52	425.89	-1.9	19.34	-0.1783
18	400	19.32	447.72	-2.17	20.13	-0.1941
19	400	20.08	493.82	-2.1	20.92	-0.1704
20	400	20.79	501.09	-1.86	21.44	-0.1488
21	400	21.44	513.22	-2	21.97	-0.156
22	476.19	22.05	513.22	-1.73	22.32	-0.1346
23	552.38	22.6	508.37	-1.8	22.76	-0.1413
24	628.57	23.1	505.95	-1.49	23.29	-0.1176
25	704.76	23.54	612.69	-1.49	23.82	-0.0971
26	780.95	23.92	736.42	-0.97	24.26	-0.0529
27	857.14	24.25	845.59	-0.91	24.61	-0.0428
28	933.33	24.52	942.64	-0.84	24.7	-0.0355
29	1009.52	24.73	969.32	-0.77	24.7	-0.0317
30	1085.71	24.88	1039.68	-0.77	24.7	-0.0296
31	1161.9	24.97	1100.33	-0.32	24.78	-0.0118
32	1238.1	25	1148.85	-0.05	24.78	-0.0017
33	1314.29	24.97	1214.35	-0.26	24.78	-0.0084

34	1390.48	24.88	1323.52	-0.46	24.78	-0.0139
35	1466.67	24.73	1372.05	-0.84	24.78	-0.0244
36	1542.86	24.52	1410.86	-0.8	24.7	-0.0228
37	1619.05	24.25	1503.05	-3.65	24.61	-0.097
38	1695.24	23.92	1575.83	-7.86	24.52	-0.1994
39	1771.43	23.54	1641.34	-8.51	24.34	-0.2073
40	1847.62	23.1	1692.28	-9.5	24.08	-0.2246
41	1923.81	22.6	1791.75	-9.29	23.55	-0.2075
42	2000	22.05	1891.22	-9.02	22.68	-0.1908
43	2000	21.44	1910.63	-9.12	21.71	-0.191
44	2000	20.79	1930.04	-8.99	20.83	-0.1862
45	2000	20.08	1944.59	-8.68	20.04	-0.1785
46	2000	19.32	1934.89	-9.05	19.25	-0.1872
47	2000	18.52	1879.09	-8.88	18.54	-0.1891
48	2000	17.68	1820.86	-9.43	17.75	-0.2072
49	2000	16.79	1808.73	-9.02	16.87	-0.1995
50	2000	15.86	1842.7	-9.29	15.91	-0.2018
51	2000	14.89	1881.52	-8.88	14.77	-0.1889
52	2000	13.89	1947.02	-8.75	13.53	-0.1797
53	2000	12.85	1954.3	-8.88	12.39	-0.1818
54	2000	11.79	1954.3	-8.54	11.25	-0.1748
55	2000	10.69	1939.74	-8.4	10.02	-0.1733
56	2000	9.57	1837.85	-8.95	8.88	-0.1948
57	2000	8.42	1796.6	-9.09	7.82	-0.2024
58	2000	7.26	1801.46	-8.68	6.86	-0.1927
59	2000	6.08	1845.13	-8.85	5.71	-0.1918
60	2000	4.88	1917.91	-8.64	4.39	-0.1803
61	2000	3.67	1934.89	-8.13	2.99	-0.1681
62	2000	2.45	1942.17	-8.3	1.58	-0.171
63	2000	1.23	1942.17	-8.23	0.18	-0.1696
64	2000	0	1913.05	-8.44	-1.05	-0.1764
65	2000	-1.23	1852.4	-8.51	-2.29	-0.1837
66	2000	-2.45	1837.85	-8.71	-3.43	-0.1896
67	2000	-3.67	1835.42	-8.03	-4.39	-0.175
68	2000	-4.88	1888.79	-8.34	-5.54	-0.1765
69	2000	-6.08	1961.58	-8.13	-6.77	-0.1658
70	2000	-7.26	2002.82	-7.93	-8.09	-0.1583
71	2000	-8.42	1993.11	-8.03	-9.49	-0.1611

72	2000	-9.57	1988.26	-7.79	-10.72	-0.1567
73	2000	-10.69	1915.48	-8.03	-11.86	-0.1676
74	2000	-11.79	1862.11	-7.96	-12.92	-0.171
75	2000	-12.85	1840.27	-7.86	-13.89	-0.1708
76	2000	-13.89	1857.26	-7.75	-14.68	-0.167
77	2000	-14.89	1893.65	-7.75	-15.56	-0.1638
78	2000	-15.86	1978.56	-7.55	-16.52	-0.1526
79	2000	-16.79	1990.69	-7.51	-17.58	-0.151
80	2000	-17.68	1985.84	-7.55	-18.54	-0.152
81	2000	-18.52	1973.71	-7.34	-19.51	-0.1488
82	2000	-19.32	1905.78	-7.21	-20.21	-0.1512
83	2000	-20.08	1864.53	-7.79	-20.92	-0.1671
84	2000	-20.79	1852.4	-7.86	-21.44	-0.1697
85	2000	-21.44	1859.68	-7.41	-21.88	-0.1594
86	1923.81	-22.05	1908.2	-7.34	-22.41	-0.1539
87	1847.62	-22.6	1915.48	-7.58	-22.94	-0.1583
88	1771.43	-23.1	1908.2	-7.51	-23.47	-0.1575
89	1695.24	-23.54	1888.79	-7.48	-23.99	-0.1584
90	1619.05	-23.92	1799.03	-7.65	-24.34	-0.1701
91	1542.86	-24.25	1651.04	-8.03	-24.61	-0.1945
92	1466.67	-24.52	1473.94	-7.62	-24.78	-0.2067
93	1390.48	-24.73	1372.05	-7.24	-24.87	-0.2111
94	1314.29	-24.88	1287.13	-7.17	-24.87	-0.2229
95	1238.1	-24.97	1253.17	-7.27	-24.96	-0.2322
96	1161.9	-25	1233.76	-7.27	-24.96	-0.2358
97	1085.71	-24.97	1199.8	-6.97	-24.96	-0.2322
98	1009.52	-24.88	1151.28	-6.38	-24.96	-0.2218
99	933.33	-24.73	1037.25	-5.8	-24.96	-0.2238
100	857.14	-24.52	930.51	-5.56	-24.87	-0.2391
101	780.95	-24.25	865	-3.1	-24.78	-0.1432
102	704.76	-23.92	801.92	1.22	-24.7	0.0607
103	628.57	-23.54	746.13	0.43	-24.52	0.023
104	552.38	-23.1	690.33	-0.15	-24.08	-0.0088
105	476.19	-22.6	607.84	-0.36	-23.38	-0.0236
106	400	-22.05	544.76	-0.56	-22.5	-0.0414
107	400	-21.44	479.26	-1.01	-21.53	-0.0842
108	400	-20.79	433.17	-1.15	-20.74	-0.1058
109	400	-20.08	435.59	-1.39	-20.04	-0.1272

110	400	-19.32	455	-0.91	-19.25	-0.0796
111	400	-18.52	525.36	-1.11	-18.46	-0.0846
112	400	-17.68	552.04	-1.11	-17.67	-0.0805
113	400	-16.79	561.75	-0.8	-16.79	-0.0572
114	400	-15.86	566.6	-0.8	-15.82	-0.0567
115	400	-14.89	578.73	-0.91	-14.68	-0.0626
116	400	-13.89	573.88	-1.08	-13.53	-0.0751
117	400	-12.85	542.34	-1.15	-12.3	-0.0845
118	400	-11.79	484.11	-1.25	-11.07	-0.1031
119	400	-10.69	438.02	-1.21	-9.93	-0.1109
120	400	-9.57	450.15	-1.11	-8.79	-0.0987
121	400	-8.42	433.17	-1.35	-7.73	-0.1248
122	400	-7.26	467.13	-1.21	-6.77	-0.104
123	400	-6.08	510.8	-1.25	-5.62	-0.0977
124	400	-4.88	549.62	-1.18	-4.31	-0.0859
125	400	-3.67	544.76	-1.25	-2.9	-0.0917
126	400	-2.45	554.47	-1.35	-1.49	-0.0975
127	400	-1.23	556.89	-1.08	-0.09	-0.0774

Appendix 3:

Friction Test Results (Data) obtained from the Friction Hip Simulator for the 50 mm Diameter Smith & Nephew Birmingham Hip Resurfacing Prosthesis with 80 µm Diametral Clearance and using Bovine Serum with CMC (of viscosity ~ 0.0013 Pas) as Lubricant.

BS+CMC of η=0.0013 Pas, 50mm diameter BHR, 80μm diametral clearance

Index	Demand Load(N)	Demand Motor(○)	Load(N)	Friction Torque(Nm)	Motor Position(○)	Friction Coefficient
0	400	0	501.09	-1.66	0.97	-0.1324
1	400	1.23	493.82	-1.9	2.37	-0.1538
2	400	2.45	493.82	-1.86	3.69	-0.151
3	400	3.67	484.11	-2.07	4.92	-0.171
4	400	4.88	493.82	-1.8	6.06	-0.1455
5	400	6.08	491.39	-1.9	7.03	-0.1546
6	400	7.26	491.39	-1.97	8	-0.1601
7	400	8.42	493.82	-1.32	9.14	-0.1067
8	400	9.57	488.96	-1.97	10.37	-0.1609
9	400	10.69	486.54	-1.76	11.6	-0.1448
10	400	11.79	486.54	-2.1	12.83	-0.173
11	400	12.85	484.11	-1.86	13.97	-0.1541
12	400	13.89	474.41	-1.86	15.12	-0.1572
13	400	14.89	474.41	-1.8	16.08	-0.1514
14	400	15.86	469.56	-1.9	16.79	-0.1618
15	400	16.79	467.13	-2.04	17.4	-0.1743
16	400	17.68	462.28	-1.69	18.19	-0.1465
17	400	18.52	464.7	-1.86	19.07	-0.1605
18	400	19.32	467.13	-1.73	19.95	-0.1479
19	400	20.08	462.28	-1.8	20.83	-0.1554
20	400	20.79	469.56	-1.35	21.62	-0.1151
21	400	21.44	467.13	-2.04	22.32	-0.1743
22	476.19	22.05	457.43	-1.66	22.85	-0.1451
23	552.38	22.6	476.83	-1.83	23.11	-0.1535
24	628.57	23.1	498.67	-1.39	23.38	-0.1111
25	704.76	23.54	607.84	-1.15	23.73	-0.0754
26	780.95	23.92	709.73	-0.32	24.08	-0.0182
27	857.14	24.25	857.72	-0.8	24.43	-0.0375
28	933.33	24.52	920.8	-0.46	24.78	-0.02
29	1009.52	24.73	1000.86	-0.22	24.96	-0.0088
30	1085.71	24.88	1039.68	-0.12	25.05	-0.0046
31	1161.9	24.97	1061.51	-0.7	25.05	-0.0264
32	1238.1	25	1129.44	-0.19	25.05	-0.0066
33	1314.29	24.97	1214.35	-0.08	25.05	-0.0028

34	1390.48	24.88	1279.86	0.36	24.96	0.0113
35	1466.67	24.73	1367.19	-0.26	24.96	-0.0075
36	1542.86	24.52	1454.53	-2.58	24.87	-0.071
37	1619.05	24.25	1478.79	-6.01	24.78	-0.1625
38	1695.24	23.92	1527.31	-5.73	24.61	-0.1502
39	1771.43	23.54	1585.54	-5.63	24.17	-0.1421
40	1847.62	23.1	1697.14	-5.29	23.55	-0.1246
41	1923.81	22.6	1769.92	-5.29	22.94	-0.1195
42	2000	22.05	1849.98	-5.12	22.24	-0.1106
43	2000	21.44	1925.18	-5.39	21.62	-0.112
44	2000	20.79	1930.04	-5.05	20.92	-0.1046
45	2000	20.08	1932.46	-5.01	20.13	-0.1038
46	2000	19.32	1934.89	-5.29	19.34	-0.1093
47	2000	18.52	1900.92	-5.53	18.46	-0.1163
48	2000	17.68	1883.94	-5.63	17.49	-0.1196
49	2000	16.79	1864.53	-5.6	16.44	-0.1201
50	2000	15.86	1876.66	-5.8	15.47	-0.1237
51	2000	14.89	1876.66	-5.7	14.68	-0.1215
52	2000	13.89	1876.66	-5.77	13.71	-0.1229
53	2000	12.85	1874.24	-5.77	12.57	-0.1231
54	2000	11.79	1881.52	-5.46	11.43	-0.1161
55	2000	10.69	1883.94	-5.7	10.11	-0.121
56	2000	9.57	1888.79	-5.73	8.79	-0.1214
57	2000	8.42	1883.94	-5.73	7.56	-0.1217
58	2000	7.26	1886.37	-5.25	6.33	-0.1114
59	2000	6.08	1886.37	-5.67	5.27	-0.1201
60	2000	4.88	1886.37	-5.39	4.31	-0.1143
61	2000	3.67	1886.37	-5.7	3.16	-0.1209
62	2000	2.45	1886.37	-5.49	1.93	-0.1165
63	2000	1.23	1888.79	-5.56	0.44	-0.1178
64	2000	0	1883.94	-5.73	-0.97	-0.1217
65	2000	-1.23	1881.52	-5.56	-2.37	-0.1183
66	2000	-2.45	1883.94	-5.46	-3.69	-0.1159
67	2000	-3.67	1893.65	-5.15	-4.92	-0.1088
68	2000	-4.88	1888.79	-5.63	-5.98	-0.1192
69	2000	-6.08	1891.22	-5.43	-6.86	-0.1148
70	2000	-7.26	1891.22	-5.67	-7.91	-0.1198
71	2000	-8.42	1891.22	-5.84	-9.14	-0.1234

72	2000	-9.57	1896.07	-6.18	-10.46	-0.1303
73	2000	-10.69	1900.92	-5.63	-11.78	-0.1185
74	2000	-11.79	1893.65	-5.87	-12.92	-0.124
75	2000	-12.85	1896.07	-5.9	-13.97	-0.1246
76	2000	-13.89	1896.07	-5.84	-15.03	-0.1231
77	2000	-14.89	1898.5	-5.67	-15.91	-0.1194
78	2000	-15.86	1900.92	-5.84	-16.61	-0.1228
79	2000	-16.79	1922.76	-5.8	-17.4	-0.1207
80	2000	-17.68	1917.91	-6.04	-18.28	-0.126
81	2000	-18.52	1910.63	-5.87	-19.16	-0.1229
82	2000	-19.32	1898.5	-5.87	-20.13	-0.1237
83	2000	-20.08	1900.92	-5.87	-20.92	-0.1235
84	2000	-20.79	1905.78	-5.43	-21.62	-0.1139
85	2000	-21.44	1900.92	-5.53	-22.15	-0.1163
86	1923.81	-22.05	1898.5	-5.94	-22.59	-0.1251
87	1847.62	-22.6	1900.92	-6.11	-22.94	-0.1286
88	1771.43	-23.1	1896.07	-5.94	-23.38	-0.1253
89	1695.24	-23.54	1876.66	-6.25	-23.82	-0.1332
90	1619.05	-23.92	1832.99	-6.18	-24.26	-0.1348
91	1542.86	-24.25	1757.79	-6.32	-24.61	-0.1437
92	1466.67	-24.52	1626.78	-6.28	-24.78	-0.1545
93	1390.48	-24.73	1469.09	-6.73	-24.96	-0.1832
94	1314.29	-24.88	1323.52	-6.56	-25.05	-0.1981
95	1238.1	-24.97	1270.15	-6.11	-25.05	-0.1924
96	1161.9	-25	1228.91	-6.11	-25.05	-0.1989
97	1085.71	-24.97	1187.67	-5.8	-25.05	-0.1954
98	1009.52	-24.88	1117.31	-5.36	-25.05	-0.1918
99	933.33	-24.73	1044.53	-5.32	-25.05	-0.2038
100	857.14	-24.52	971.75	-4.16	-24.96	-0.1712
101	780.95	-24.25	898.97	0.05	-24.78	0.0024
102	704.76	-23.92	835.89	0.19	-24.61	0.0091
103	628.57	-23.54	775.24	-0.46	-24.26	-0.0238
104	552.38	-23.1	726.72	-0.56	-23.64	-0.031
105	476.19	-22.6	649.08	-0.8	-22.94	-0.0495
106	400	-22.05	581.15	-0.67	-22.24	-0.0459
107	400	-21.44	498.67	-1.04	-21.53	-0.0836
108	400	-20.79	450.15	-1.62	-20.83	-0.1444
109	400	-20.08	447.72	-1.66	-20.04	-0.1482

110	400	-19.32	462.28	-1.49	-19.16	-0.1287
111	400	-18.52	479.26	-1.04	-18.28	-0.087
112	400	-17.68	518.08	-0.91	-17.31	-0.0699
113	400	-16.79	527.78	-1.11	-16.35	-0.0842
114	400	-15.86	535.06	-1.42	-15.56	-0.1061
115	400	-14.89	539.91	-0.87	-14.68	-0.0646
116	400	-13.89	549.62	-1.18	-13.71	-0.0859
117	400	-12.85	544.76	-1.01	-12.48	-0.0741
118	400	-11.79	542.34	-1.08	-11.16	-0.0794
119	400	-10.69	525.36	-1.18	-9.93	-0.0898
120	400	-9.57	505.95	-1.59	-8.7	-0.1258
121	400	-8.42	493.82	-1.73	-7.47	-0.1399
122	400	-7.26	479.26	-1.8	-6.33	-0.1499
123	400	-6.08	479.26	-2	-5.27	-0.1671
124	400	-4.88	484.11	-1.62	-4.22	-0.1343
125	400	-3.67	484.11	-1.59	-2.99	-0.1314
126	400	-2.45	491.39	-1.66	-1.67	-0.1351
127	400	-1.23	488.96	-1.8	-0.26	-0.1469

Appendix 4:

Friction Test Results (Data) obtained from the Friction Hip Simulator for the 50 mm Diameter Smith & Nephew Birmingham Hip Resurfacing Prosthesis with 80 μm Diametral Clearance and using Bovine Serum with CMC (of viscosity ~ 0.19 Pas) as Lubricant.

BS+CMC of η=0.19 Pas, 50mm diameter BHR, 80μm diametral clearance

Index	Demand Load(N)	Demand Motor(○)	Load(N)	Friction Torque(Nm)	Motor Position(○)	Friction Coefficient
0	400	0	532.63	-0.29	0.79	-0.0217
1	400	1.23	471.98	-0.73	1.85	-0.0623
2	400	2.45	406.48	-0.6	3.16	-0.0588
3	400	3.67	404.05	-1.18	4.57	-0.1168
4	400	4.88	408.9	-0.97	5.98	-0.0953
5	400	6.08	486.54	-0.7	7.29	-0.0576
6	400	7.26	530.21	-1.42	8.44	-0.1071
7	400	8.42	532.63	-0.97	9.58	-0.0732
8	400	9.57	527.78	-1.01	10.55	-0.0764
9	400	10.69	542.34	-0.84	11.43	-0.0618
10	400	11.79	522.93	-1.18	12.39	-0.0902
11	400	12.85	491.39	-1.21	13.53	-0.0988
12	400	13.89	430.74	-1.08	14.68	-0.1
13	400	14.89	411.33	-1.18	15.82	-0.1147
14	400	15.86	404.05	-1.42	16.87	-0.1405
15	400	16.79	413.76	-1.35	17.75	-0.1306
16	400	17.68	486.54	-1.32	18.63	-0.1083
17	400	18.52	527.78	-1.28	19.34	-0.0972
18	400	19.32	535.06	-0.77	19.95	-0.0575
19	400	20.08	554.47	-0.77	20.57	-0.0555
20	400	20.79	539.91	-1.04	21.18	-0.0773
21	400	21.44	522.93	-1.35	21.88	-0.1033
22	476.19	22.05	488.96	-1.21	22.59	-0.0993
23	552.38	22.6	462.28	-1.21	23.2	-0.105
24	628.57	23.1	464.7	-1.39	23.73	-0.1192
25	704.76	23.54	629.68	-1.08	24.08	-0.0684
26	780.95	23.92	753.4	-0.22	24.34	-0.0117
27	857.14	24.25	894.11	0.09	24.43	0.0039
28	933.33	24.52	957.19	-0.05	24.52	-0.0021
29	1009.52	24.73	986.3	-0.12	24.7	-0.0048
30	1085.71	24.88	988.73	0.02	24.87	0.0008
31	1161.9	24.97	1061.51	0.19	24.96	0.0072
32	1238.1	25	1163.41	0.43	25.05	0.0148
33	1314.29	24.97	1216.78	0.16	25.05	0.0051
34	1390.48	24.88	1313.82	-0.29	24.96	-0.0088

35	1466.67	24.73	1381.75	-0.12	24.96	-0.0034
36	1542.86	24.52	1410.86	-1.08	24.87	-0.0305
37	1619.05	24.25	1478.79	-4.81	24.78	-0.1301
38	1695.24	23.92	1573.41	-9.71	24.61	-0.2467
39	1771.43	23.54	1663.17	-10.32	24.52	-0.2482
40	1847.62	23.1	1704.41	-9.88	24.08	-0.2318
41	1923.81	22.6	1791.75	-9.4	23.47	-0.2098
42	2000	22.05	1881.52	-9.43	22.59	-0.2005
43	2000	21.44	1908.2	-9.16	21.71	-0.192
44	2000	20.79	1942.17	-9.16	20.83	-0.1886
45	2000	20.08	1934.89	-9.16	20.21	-0.1893
46	2000	19.32	1922.76	-9.12	19.6	-0.1898
47	2000	18.52	1852.4	-9.19	18.9	-0.1985
48	2000	17.68	1828.14	-9.33	17.93	-0.2041
49	2000	16.79	1820.86	-9.29	16.87	-0.2042
50	2000	15.86	1874.24	-9.33	15.64	-0.1991
51	2000	14.89	1925.18	-8.64	14.5	-0.1796
52	2000	13.89	1932.46	-8.78	13.45	-0.1818
53	2000	12.85	1934.89	-8.92	12.39	-0.1844
54	2000	11.79	1925.18	-8.64	11.43	-0.1796
55	2000	10.69	1842.7	-8.99	10.46	-0.1951
56	2000	9.57	1801.46	-9.36	9.4	-0.2079
57	2000	8.42	1808.73	-9.09	8.09	-0.201
58	2000	7.26	1905.78	-8.71	6.68	-0.1829
59	2000	6.08	1951.87	-8.88	5.27	-0.1821
60	2000	4.88	1971.28	-8.61	3.95	-0.1747
61	2000	3.67	1978.56	-8.3	2.72	-0.1678
62	2000	2.45	1951.87	-8.3	1.49	-0.1701
63	2000	1.23	1852.4	-8.99	0.44	-0.1941
64	2000	0	1823.29	-8.78	-0.62	-0.1926
65	2000	-1.23	1818.44	-8.47	-1.76	-0.1864
66	2000	-2.45	1900.92	-8.51	-3.16	-0.179
67	2000	-3.67	1925.18	-8.3	-4.57	-0.1725
68	2000	-4.88	1939.74	-8.13	-5.98	-0.1677
69	2000	-6.08	1925.18	-8.06	-7.29	-0.1675
70	2000	-7.26	1920.33	-7.82	-8.53	-0.1629
71	2000	-8.42	1891.22	-8.1	-9.58	-0.1712
72	2000	-9.57	1854.83	-8.27	-10.63	-0.1783

73	2000	-10.69	1845.13	-7.86	-11.43	-0.1703
74	2000	-11.79	1922.76	-7.86	-12.48	-0.1634
75	2000	-12.85	1956.72	-7.75	-13.62	-0.1585
76	2000	-13.89	1949.45	-7.55	-14.85	-0.1549
77	2000	-14.89	1934.89	-7.75	-16	-0.1603
78	2000	-15.86	1920.33	-7.27	-17.05	-0.1515
79	2000	-16.79	1876.66	-7.17	-17.93	-0.1529
80	2000	-17.68	1852.4	-7.48	-18.72	-0.1615
81	2000	-18.52	1840.27	-7.41	-19.34	-0.1611
82	2000	-19.32	1947.02	-6.97	-19.86	-0.1431
83	2000	-20.08	1980.98	-7.14	-20.57	-0.1441
84	2000	-20.79	1973.71	-7.07	-21.27	-0.1433
85	2000	-21.44	1964	-7.21	-22.06	-0.1468
86	1923.81	-22.05	1927.61	-7.38	-22.76	-0.1531
87	1847.62	-22.6	1828.14	-7	-23.29	-0.1532
88	1771.43	-23.1	1735.95	-7.14	-23.73	-0.1645
89	1695.24	-23.54	1636.49	-7.07	-24.08	-0.1728
90	1619.05	-23.92	1600.09	-7.03	-24.26	-0.1759
91	1542.86	-24.25	1549.15	-6.73	-24.43	-0.1737
92	1466.67	-24.52	1478.79	-6.59	-24.61	-0.1782
93	1390.48	-24.73	1401.16	-6.56	-24.87	-0.1871
94	1314.29	-24.88	1345.36	-6.69	-25.05	-0.199
95	1238.1	-24.97	1287.13	-6.56	-25.14	-0.2037
96	1161.9	-25	1228.91	-5.63	-25.14	-0.1833
97	1085.71	-24.97	1151.28	-5.9	-25.14	-0.2052
98	1009.52	-24.88	1071.22	-5.46	-25.14	-0.2039
99	933.33	-24.73	1000.86	-5.39	-25.05	-0.2155
100	857.14	-24.52	940.21	-4.16	-25.05	-0.1769
101	780.95	-24.25	898.97	0.02	-24.96	0.0008
102	704.76	-23.92	840.74	0.36	-24.78	0.0172
103	628.57	-23.54	750.98	1.35	-24.43	0.0721
104	552.38	-23.1	683.05	1.08	-23.91	0.0633
105	476.19	-22.6	617.54	0.46	-23.11	0.03
106	400	-22.05	556.89	-0.05	-22.24	-0.0036
107	400	-21.44	491.39	-0.15	-21.44	-0.0124
108	400	-20.79	452.57	-0.15	-20.74	-0.0135
109	400	-20.08	459.85	-0.26	-20.13	-0.0222
110	400	-19.32	469.56	-0.08	-19.51	-0.0072

111	400	-18.52	547.19	0.4	-18.63	0.0289
112	400	-17.68	569.02	0.22	-17.58	0.0158
113	400	-16.79	588.43	0.36	-16.52	0.0245
114	400	-15.86	610.27	0.16	-15.47	0.0102
115	400	-14.89	619.97	0.53	-14.41	0.0343
116	400	-13.89	607.84	0.02	-13.36	0.0012
117	400	-12.85	556.89	-0.02	-12.3	-0.0011
118	400	-11.79	491.39	-0.26	-11.43	-0.0208
119	400	-10.69	433.17	-0.94	-10.28	-0.0868
120	400	-9.57	430.74	-0.56	-9.14	-0.0523
121	400	-8.42	447.72	-0.39	-7.73	-0.035
122	400	-7.26	539.91	-0.32	-6.42	-0.024
123	400	-6.08	547.19	-0.49	-5.1	-0.0362
124	400	-4.88	564.17	-0.43	-3.87	-0.0302
125	400	-3.67	588.43	-0.02	-2.72	-0.0011
126	400	-2.45	569.02	-0.29	-1.49	-0.0204
127	400	-1.23	561.75	-0.22	-0.44	-0.0157

Appendix 5:

Friction Test Results (Data) obtained from the Friction Hip Simulator for the 50 mm Diameter Smith & Nephew Birmingham Hip Resurfacing Prosthesis with 200 µm Diametral Clearance and using Whole Blood (of viscosity ~ 0.01 Pas) as Lubricant.

Whole blood of η=0.01 Pas, 50mm diameter BHR, 200μm diametral clearance

Index	Demand Load(N)	Demand Motor(○)	Load(N)	Friction Torque(Nm)	Motor Position(○)	Friction Coefficient
0	100	0	271.82	-0.9	1.05	-0.1329
1	100	1.23	295.45	-0.81	2.37	-0.11
2	100	2.45	316.14	-0.81	3.6	-0.1024
3	100	3.67	325	-0.85	4.75	-0.1047
4	100	4.88	354.55	-1.12	5.8	-0.1267
5	100	6.08	336.82	-0.79	6.86	-0.0942
6	100	7.26	307.27	-0.62	8	-0.0808
7	100	8.42	271.82	-0.67	9.23	-0.0983
8	100	9.57	215.68	-0.81	10.55	-0.1501
9	100	10.69	180.23	-0.87	11.78	-0.1931
10	100	11.79	189.09	-0.84	12.92	-0.1775
11	100	12.85	171.36	-0.85	13.97	-0.1995
12	100	13.89	183.18	-0.8	14.94	-0.174
13	100	14.89	180.23	-0.9	15.82	-0.1997
14	100	15.86	236.36	-0.97	16.52	-0.1641
15	100	16.79	271.82	-0.68	17.4	-0.1003
16	100	17.68	271.82	-0.77	18.37	-0.1139
17	100	18.52	277.73	-0.76	19.25	-0.1098
18	100	19.32	280.68	-0.76	20.13	-0.1086
19	100	20.08	289.55	-0.69	20.92	-0.0952
20	100	20.79	283.64	-0.62	21.53	-0.0871
21	100	21.44	254.09	-0.64	22.06	-0.1013
22	190.48	22.05	212.73	-0.75	22.5	-0.1417
23	280.95	22.6	177.27	-0.9	22.94	-0.2023
24	371.43	23.1	183.18	-0.82	23.47	-0.179
25	461.9	23.54	274.77	-0.8	23.91	-0.1168
26	552.38	23.92	363.41	-0.55	24.34	-0.0607
27	642.86	24.25	505.23	-0.57	24.7	-0.0454
28	733.33	24.52	593.86	-0.44	24.78	-0.0295
29	823.81	24.73	717.95	-0.32	24.87	-0.0177
30	914.29	24.88	791.82	-0.19	24.87	-0.0094
31	1004.76	24.97	880.45	-0.26	24.87	-0.0118
32	1095.24	25	969.09	-0.15	24.96	-0.0063
33	1185.71	24.97	1045.91	0.1	24.96	0.0039
34	1276.19	24.88	1140.45	-0.26	24.87	-0.0092

35	1366.67	24.73	1193.64	-0.44	24.87	-0.0146
36	1457.14	24.52	1237.95	-0.79	24.78	-0.0255
37	1547.62	24.25	1297.04	-2.02	24.7	-0.0623
38	1638.1	23.92	1338.41	-2.6	24.43	-0.0778
39	1728.57	23.54	1424.09	-2.63	23.99	-0.0739
40	1819.05	23.1	1533.41	-2.42	23.38	-0.063
41	1909.52	22.6	1571.82	-2.28	22.85	-0.0581
42	2000	22.05	1660.45	-2.47	22.15	-0.0595
43	2000	21.44	1769.77	-2.72	21.44	-0.0614
44	2000	20.79	1793.41	-2.5	20.65	-0.0558
45	2000	20.08	1775.68	-2.68	19.95	-0.0603
46	2000	19.32	1757.95	-2.36	19.07	-0.0538
47	2000	18.52	1728.41	-2.5	18.19	-0.0579
48	2000	17.68	1734.32	-2.46	17.31	-0.0567
49	2000	16.79	1719.54	-2.32	16.44	-0.0541
50	2000	15.86	1701.82	-2.43	15.64	-0.0572
51	2000	14.89	1692.95	-2.39	14.68	-0.0565
52	2000	13.89	1701.82	-2.43	13.53	-0.0572
53	2000	12.85	1778.64	-2.56	12.3	-0.0575
54	2000	11.79	1790.45	-2.54	11.16	-0.0568
55	2000	10.69	1778.64	-2.37	9.93	-0.0533
56	2000	9.57	1772.73	-2.36	8.79	-0.0533
57	2000	8.42	1757.95	-2.33	7.56	-0.053
58	2000	7.26	1746.14	-2.45	6.5	-0.056
59	2000	6.08	1772.73	-2.24	5.54	-0.0506
60	2000	4.88	1805.23	-2.49	4.39	-0.0552
61	2000	3.67	1796.36	-2.68	3.16	-0.0596
62	2000	2.45	1817.04	-2.63	1.76	-0.0578
63	2000	1.23	1805.23	-2.47	0.35	-0.0548
64	2000	0	1787.5	-2.31	-0.97	-0.0517
65	2000	-1.23	1802.27	-2.47	-2.2	-0.0548
66	2000	-2.45	1790.45	-2.42	-3.43	-0.0542
67	2000	-3.67	1814.09	-2.39	-4.48	-0.0527
68	2000	-4.88	1825.91	-2.21	-5.45	-0.0483
69	2000	-6.08	1837.73	-2.53	-6.5	-0.055
70	2000	-7.26	1876.14	-2.53	-7.73	-0.0538
71	2000	-8.42	1908.64	-2.56	-9.05	-0.0536
72	2000	-9.57	1944.09	-2.64	-10.37	-0.0543
73	2000	-10.69	1970.68	-2.58	-11.6	-0.0524
74	2000	-11.79	1923.41	-2.66	-12.66	-0.0553
75	2000	-12.85	1846.59	-2.77	-13.62	-0.06
76	2000	-13.89	1825.91	-2.92	-14.5	-0.064
77	2000	-14.89	1834.77	-2.6	-15.29	-0.0567

78	2000	-15.86	1837.73	-2.73	-16.17	-0.0595
79	2000	-16.79	1837.73	-2.81	-17.23	-0.0611
80	2000	-17.68	1923.41	-2.88	-18.19	-0.06
81	2000	-18.52	1947.04	-2.79	-19.07	-0.0573
82	2000	-19.32	1944.09	-2.87	-19.86	-0.059
83	2000	-20.08	1950	-2.93	-20.57	-0.0602
84	2000	-20.79	1950	-2.99	-21.18	-0.0614
85	2000	-21.44	1950	-3.02	-21.8	-0.062
86	1909.52	-22.05	1885	-2.91	-22.15	-0.0618
87	1819.05	-22.6	1840.68	-2.99	-22.68	-0.0649
88	1728.57	-23.1	1775.68	-2.96	-23.2	-0.0666
89	1638.1	-23.54	1684.09	-2.96	-23.64	-0.0703
90	1547.62	-23.92	1610.23	-2.88	-24.08	-0.0714
91	1457.14	-24.25	1536.36	-2.88	-24.43	-0.0749
92	1366.67	-24.52	1456.59	-2.87	-24.7	-0.0789
93	1276.19	-24.73	1388.64	-2.9	-24.78	-0.0834
94	1185.71	-24.88	1323.64	-2.83	-24.87	-0.0854
95	1095.24	-24.97	1258.64	-2.69	-24.87	-0.0856
96	1004.76	-25	1181.82	-2.66	-24.87	-0.0899
97	914.29	-24.97	1113.86	-2.53	-24.87	-0.0908
98	823.81	-24.88	1042.95	-2.37	-24.87	-0.0908
99	733.33	-24.73	972.04	-2.31	-24.87	-0.0949
100	642.86	-24.52	889.32	-2.03	-24.78	-0.0915
101	552.38	-24.25	821.36	-1.9	-24.7	-0.0923
102	461.9	-23.92	744.55	0.25	-24.43	0.0133
103	371.43	-23.54	682.5	-0.17	-24.08	-0.01
104	280.95	-23.1	608.64	-0.49	-23.55	-0.0323
105	190.48	-22.6	531.82	-0.5	-22.94	-0.0378
106	100	-22.05	452.05	-0.56	-22.24	-0.0498
107	100	-21.44	375.23	-0.66	-21.44	-0.0705
108	100	-20.79	301.36	-0.78	-20.65	-0.1042
109	100	-20.08	254.09	-0.9	-19.86	-0.1411
110	100	-19.32	257.05	-0.87	-19.07	-0.1349
111	100	-18.52	248.18	-0.91	-18.19	-0.1464
112	100	-17.68	251.14	-0.92	-17.23	-0.1472
113	100	-16.79	369.32	-0.94	-16.44	-0.102
114	100	-15.86	375.23	-0.85	-15.56	-0.0904
115	100	-14.89	395.91	-0.7	-14.5	-0.0703
116	100	-13.89	387.05	-0.71	-13.45	-0.0732
117	100	-12.85	390	-0.74	-12.22	-0.0758
118	100	-11.79	398.86	-0.81	-10.99	-0.0815
119	100	-10.69	351.59	-0.85	-9.76	-0.097
120	100	-9.57	292.5	-0.85	-8.61	-0.1159

121	100	-8.42	248.18	-0.92	-7.47	-0.1487
122	100	-7.26	239.32	-1.02	-6.5	-0.1705
123	100	-6.08	248.18	-0.97	-5.45	-0.156
124	100	-4.88	254.09	-0.82	-4.22	-0.1285
125	100	-3.67	257.05	-0.81	-2.9	-0.1267
126	100	-2.45	260	-0.86	-1.49	-0.1325
127	100	-1.23	271.82	-0.96	-0.18	-0.1414

Appendix 6:

Friction Test Results (Data) obtained from the Friction Hip Simulator for the 50 mm Diameter Smith & Nephew Birmingham Hip Resurfacing Prosthesis with 200 μm Diametral Clearance and using Bovine Serum with CMC (of viscosity ~ 0.0013 Pas) as Lubricant.

BS+CMC of η=0.0013 Pas, 50mm diameter BHR, 200μm diametral clearance

Index	Demand Load(N)	Demand Motor(○)	Load(N)	Friction Torque(Nm)	Motor Position(○)	Friction Coefficient
0	400	0	571.45	-1.9	1.05	-0.1329
1	400	1.23	535.06	-1.93	2.11	-0.1445
2	400	2.45	484.11	-1.86	3.16	-0.1541
3	400	3.67	421.04	-1.86	4.39	-0.1771
4	400	4.88	416.18	-1.8	5.71	-0.1726
5	400	6.08	421.04	-1.66	7.12	-0.1576
6	400	7.26	459.85	-1.97	8.44	-0.1711
7	400	8.42	484.11	-1.73	9.58	-0.1427
8	400	9.57	539.91	-1.93	10.72	-0.1432
9	400	10.69	552.04	-1.97	11.78	-0.1425
10	400	11.79	552.04	-2	12.66	-0.145
11	400	12.85	554.47	-1.97	13.53	-0.1419
12	400	13.89	539.91	-1.69	14.5	-0.1255
13	400	14.89	496.24	-1.56	15.56	-0.1255
14	400	15.86	450.15	-2.04	16.7	-0.1809
15	400	16.79	433.17	-1.97	17.75	-0.1817
16	400	17.68	425.89	-2.17	18.63	-0.2041
17	400	18.52	425.89	-1.9	19.51	-0.1783
18	400	19.32	445.3	-2	20.21	-0.1798
19	400	20.08	474.41	-1.69	20.74	-0.1428
20	400	20.79	488.96	-2	21.27	-0.1637
21	400	21.44	491.39	-1.73	21.8	-0.1406
22	476.19	22.05	488.96	-2.07	22.41	-0.1693
23	552.38	22.6	491.39	-1.66	23.03	-0.1351
24	628.57	23.1	539.91	-1.39	23.64	-0.1026
25	704.76	23.54	639.38	-1.39	24.08	-0.0867
26	780.95	23.92	784.94	-1.25	24.43	-0.0636
27	857.14	24.25	874.71	-0.84	24.61	-0.0383
28	933.33	24.52	974.17	-0.94	24.61	-0.0386
29	1009.52	24.73	988.73	-0.08	24.7	-0.0034
30	1085.71	24.88	993.58	-0.19	24.78	-0.0075
31	1161.9	24.97	1059.09	-0.08	24.87	-0.0032
32	1238.1	25	1151.28	-0.05	24.96	-0.0017
33	1314.29	24.97	1221.63	-0.56	24.96	-0.0184
34	1390.48	24.88	1321.1	-0.43	24.96	-0.0129
35	1466.67	24.73	1401.16	-0.43	24.87	-0.0122

36	1542.86	24.52	1430.27	-0.94	24.78	-0.0263
37	1619.05	24.25	1544.3	-3.54	24.7	-0.0918
38	1695.24	23.92	1595.24	-6.01	24.52	-0.1506
39	1771.43	23.54	1626.78	-5.15	24.26	-0.1267
40	1847.62	23.1	1704.41	-5.73	23.73	-0.1346
41	1923.81	22.6	1784.47	-5.8	23.03	-0.1301
42	2000	22.05	1832.99	-5.94	22.24	-0.1296
43	2000	21.44	1903.35	-5.7	21.44	-0.1198
44	2000	20.79	1922.76	-5.49	20.65	-0.1143
45	2000	20.08	1922.76	-5.6	19.95	-0.1164
46	2000	19.32	1908.2	-5.84	19.25	-0.1223
47	2000	18.52	1883.94	-5.63	18.63	-0.1196
48	2000	17.68	1830.57	-6.08	17.75	-0.1328
49	2000	16.79	1828.14	-5.97	16.7	-0.1307
50	2000	15.86	1828.14	-5.6	15.56	-0.1225
51	2000	14.89	1893.65	-5.97	14.41	-0.1262
52	2000	13.89	1925.18	-5.87	13.36	-0.122
53	2000	12.85	1915.48	-6.04	12.22	-0.1262
54	2000	11.79	1920.33	-5.87	11.16	-0.1223
55	2000	10.69	1908.2	-5.77	10.2	-0.1209
56	2000	9.57	1854.83	-5.7	9.23	-0.1229
57	2000	8.42	1840.27	-6.18	8.09	-0.1343
58	2000	7.26	1842.7	-5.94	6.77	-0.1289
59	2000	6.08	1893.65	-5.9	5.36	-0.1247
60	2000	4.88	1905.78	-5.9	4.04	-0.1239
61	2000	3.67	1910.63	-5.77	2.72	-0.1208
62	2000	2.45	1908.2	-6.04	1.41	-0.1266
63	2000	1.23	1913.05	-5.7	0.18	-0.1192
64	2000	0	1905.78	-6.08	-0.97	-0.1275
65	2000	-1.23	1898.5	-5.77	-1.93	-0.1215
66	2000	-2.45	1896.07	-5.97	-2.99	-0.126
67	2000	-3.67	1913.05	-5.9	-4.31	-0.1235
68	2000	-4.88	1905.78	-6.08	-5.8	-0.1275
69	2000	-6.08	1864.53	-6.04	-7.12	-0.1296
70	2000	-7.26	1876.66	-6.01	-8.44	-0.128
71	2000	-8.42	1939.74	-5.94	-9.58	-0.1225
72	2000	-9.57	1927.61	-6.32	-10.63	-0.1311
73	2000	-10.69	1927.61	-6.25	-11.6	-0.1296
74	2000	-11.79	1927.61	-6.25	-12.39	-0.1296
75	2000	-12.85	1915.48	-6.21	-13.36	-0.1297
76	2000	-13.89	1903.35	-6.42	-14.5	-0.1349
77	2000	-14.89	1881.52	-6.35	-15.73	-0.135
78	2000	-15.86	1883.94	-6.25	-16.79	-0.1326

79	2000	-16.79	1883.94	-6.62	-17.75	-0.1406
80	2000	-17.68	1881.52	-6.35	-18.63	-0.135
81	2000	-18.52	1959.15	-5.9	-19.34	-0.1206
82	2000	-19.32	1964	-6.52	-19.95	-0.1328
83	2000	-20.08	1956.72	-6.69	-20.48	-0.1368
84	2000	-20.79	1949.45	-6.62	-21.09	-0.1359
85	2000	-21.44	1905.78	-6.35	-21.8	-0.1333
86	1923.81	-22.05	1871.81	-6.62	-22.5	-0.1416
87	1847.62	-22.6	1825.72	-6.56	-23.11	-0.1436
88	1771.43	-23.1	1767.49	-6.59	-23.64	-0.1491
89	1695.24	-23.54	1677.73	-6.97	-23.99	-0.1661
90	1619.05	-23.92	1609.8	-7	-24.26	-0.1739
91	1542.86	-24.25	1524.89	-7.03	-24.43	-0.1845
92	1466.67	-24.52	1461.81	-7.31	-24.61	-0.2
93	1390.48	-24.73	1425.42	-7.58	-24.78	-0.2128
94	1314.29	-24.88	1386.6	-7.34	-24.87	-0.2118
95	1238.1	-24.97	1299.26	-7.14	-24.96	-0.2197
96	1161.9	-25	1221.63	-6.83	-25.05	-0.2236
97	1085.71	-24.97	1158.55	-6.42	-25.05	-0.2216
98	1009.52	-24.88	1073.64	-6.28	-24.96	-0.234
99	933.33	-24.73	1008.14	-6.38	-24.96	-0.2533
100	857.14	-24.52	954.77	-5.43	-24.87	-0.2273
101	780.95	-24.25	891.69	-2.69	-24.78	-0.1205
102	704.76	-23.92	826.19	-1.18	-24.61	-0.0571
103	628.57	-23.54	758.26	-1.15	-24.26	-0.0604
104	552.38	-23.1	687.9	-0.53	-23.73	-0.0308
105	476.19	-22.6	624.82	-1.18	-23.03	-0.0755
106	400	-22.05	559.32	-1.39	-22.24	-0.0991
107	400	-21.44	498.67	-1.52	-21.44	-0.1221
108	400	-20.79	450.15	-1.83	-20.65	-0.1626
109	400	-20.08	455	-1.97	-19.95	-0.1729
110	400	-19.32	467.13	-1.62	-19.34	-0.1391
111	400	-18.52	566.6	-1.11	-18.54	-0.0785
112	400	-17.68	586.01	-1.25	-17.67	-0.0852
113	400	-16.79	598.14	-1.69	-16.61	-0.1132
114	400	-15.86	605.41	-1.39	-15.47	-0.0915
115	400	-14.89	622.4	-1.45	-14.41	-0.0934
116	400	-13.89	607.84	-1.45	-13.27	-0.0957
117	400	-12.85	571.45	-1.28	-12.22	-0.0898
118	400	-11.79	518.08	-1.56	-11.07	-0.1202
119	400	-10.69	455	-1.49	-10.11	-0.1308
120	400	-9.57	428.31	-1.83	-9.05	-0.1709
121	400	-8.42	438.02	-2.14	-7.91	-0.1953

122	400	-7.26	481.69	-1.86	-6.59	-0.1548
123	400	-6.08	537.49	-1.45	-5.19	-0.1082
124	400	-4.88	561.75	-1.93	-3.87	-0.1376
125	400	-3.67	571.45	-1.86	-2.55	-0.1305
126	400	-2.45	566.6	-1.86	-1.32	-0.1316
127	400	-1.23	571.45	-1.93	-0.09	-0.1353

<u>Appendix 7:</u>

Friction Test Results (Data) obtained from the Friction Hip Simulator for the 50 mm Diameter Smith & Nephew Birmingham Hip Resurfacing Prosthesis with 200 µm Diametral Clearance and using Bovine Serum with CMC (of viscosity ~ 0.19 Pas) as Lubricant.

BS+CMC of η=0.19 Pas, 50mm diameter BHR, 200µm diametral clearance

Index	Demand Load(N)	Demand Motor(○)	Load(N)	Friction Torque(Nm)	Motor Position(○)	Friction Coefficient
0	400	0	590.86	-0.32	0.88	-0.0219
1	400	1.23	566.6	0.16	1.93	0.011
2	400	2.45	520.5	-0.02	3.08	-0.0012
3	400	3.67	462.28	-0.19	4.39	-0.0162
4	400	4.88	394.35	-0.36	5.8	-0.0363
5	400	6.08	377.37	-0.46	7.21	-0.0488
6	400	7.26	387.07	-0.26	8.44	-0.0264
7	400	8.42	450.15	-0.26	9.58	-0.0227
8	400	9.57	508.37	-0.46	10.72	-0.0362
9	400	10.69	535.06	-0.63	11.69	-0.0472
10	400	11.79	552.04	-0.39	12.57	-0.0284
11	400	12.85	549.62	-0.29	13.45	-0.0211
12	400	13.89	544.76	-0.56	14.5	-0.0414
13	400	14.89	525.36	-0.46	15.64	-0.0351
14	400	15.86	491.39	-0.46	16.79	-0.0375
15	400	16.79	438.02	-0.63	17.75	-0.0577
16	400	17.68	406.48	-0.6	18.63	-0.0588
17	400	18.52	394.35	-0.7	19.51	-0.071
18	400	19.32	399.2	-0.49	20.13	-0.0496
19	400	20.08	440.44	-0.63	20.65	-0.0574
20	400	20.79	469.56	-0.84	21.18	-0.0713
21	400	21.44	498.67	-0.46	21.8	-0.037
22	476.19	22.05	503.52	-0.29	22.41	-0.023
23	552.38	22.6	503.52	-0.29	23.11	-0.023
24	628.57	23.1	537.49	-0.67	23.64	-0.0496
25	704.76	23.54	653.94	-0.63	24.08	-0.0387
26	780.95	23.92	792.22	-0.19	24.43	-0.0094
27	857.14	24.25	860.15	0.33	24.52	0.0152
28	933.33	24.52	959.62	0.29	24.61	0.0122
29	1009.52	24.73	954.77	0.74	24.7	0.0309
30	1085.71	24.88	981.45	-0.02	24.78	-0.0006
31	1161.9	24.97	1066.36	-0.19	24.96	-0.007
32	1238.1	25	1158.55	0.19	25.05	0.0066
33	1314.29	24.97	1231.34	-0.32	24.96	-0.0105
34	1390.48	24.88	1316.25	0.09	24.96	0.0026
35	1466.67	24.73	1406.01	-0.15	24.96	-0.0043

36	1542.86	24.52	1437.55	-1.04	24.87	-0.029
37	1619.05	24.25	1481.22	-3.17	24.78	-0.0855
38	1695.24	23.92	1549.15	-3.3	24.61	-0.0853
39	1771.43	23.54	1655.89	-3.65	24.17	-0.088
40	1847.62	23.1	1723.82	-3.3	23.55	-0.0766
41	1923.81	22.6	1777.2	-3.58	22.85	-0.0805
42	2000	22.05	1869.39	-3.37	22.06	-0.0721
43	2000	21.44	1925.18	-3.3	21.36	-0.0686
44	2000	20.79	1954.3	-3.17	20.65	-0.0648
45	2000	20.08	1954.3	-3.41	19.95	-0.0697
46	2000	19.32	1922.76	-3.65	19.34	-0.0758
47	2000	18.52	1842.7	-3.54	18.54	-0.0769
48	2000	17.68	1803.88	-3.99	17.67	-0.0884
49	2000	16.79	1794.18	-3.58	16.61	-0.0797
50	2000	15.86	1854.83	-4.16	15.47	-0.0897
51	2000	14.89	1886.37	-4.02	14.41	-0.0853
52	2000	13.89	1932.46	-3.99	13.27	-0.0825
53	2000	12.85	1930.04	-4.19	12.22	-0.0869
54	2000	11.79	1930.04	-4.54	11.16	-0.094
55	2000	10.69	1879.09	-4.6	10.28	-0.098
56	2000	9.57	1820.86	-5.29	9.32	-0.1162
57	2000	8.42	1823.29	-5.19	8.09	-0.1138
58	2000	7.26	1830.57	-4.81	6.77	-0.1051
59	2000	6.08	1891.22	-5.15	5.36	-0.109
60	2000	4.88	1951.87	-5.08	4.04	-0.1042
61	2000	3.67	1949.45	-5.08	2.72	-0.1043
62	2000	2.45	1944.59	-5.12	1.49	-0.1053
63	2000	1.23	1913.05	-5.15	0.26	-0.1077
64	2000	0	1869.39	-5.29	-0.7	-0.1132
65	2000	-1.23	1845.13	-5.73	-1.76	-0.1243
66	2000	-2.45	1837.85	-5.53	-2.99	-0.1203
67	2000	-3.67	1903.35	-5.32	-4.39	-0.1119
68	2000	-4.88	1927.61	-5.39	-5.8	-0.1119
69	2000	-6.08	1927.61	-5.53	-7.21	-0.1147
70	2000	-7.26	1934.89	-5.29	-8.44	-0.1093
71	2000	-8.42	1934.89	-5.05	-9.58	-0.1044
72	2000	-9.57	1891.22	-5.67	-10.63	-0.1198
73	2000	-10.69	1857.26	-5.36	-11.43	-0.1154
74	2000	-11.79	1852.4	-5.49	-12.39	-0.1186
75	2000	-12.85	1876.66	-4.98	-13.45	-0.1062
76	2000	-13.89	1939.74	-4.95	-14.68	-0.102
77	2000	-14.89	1947.02	-5.12	-15.82	-0.1051
78	2000	-15.86	1944.59	-4.95	-16.87	-0.1017

79	2000	-16.79	1954.3	-4.54	-17.84	-0.0928
80	2000	-17.68	1891.22	-5.05	-18.63	-0.1068
81	2000	-18.52	1862.11	-5.08	-19.34	-0.1092
82	2000	-19.32	1862.11	-5.22	-19.86	-0.1121
83	2000	-20.08	1871.81	-5.29	-20.48	-0.113
84	2000	-20.79	1942.17	-5.05	-21.18	-0.104
85	2000	-21.44	1951.87	-4.88	-21.97	-0.1
86	1923.81	-22.05	1966.43	-4.5	-22.68	-0.0916
87	1847.62	-22.6	1951.87	-5.05	-23.2	-0.1035
88	1771.43	-23.1	1915.48	-4.54	-23.64	-0.0947
89	1695.24	-23.54	1816.01	-5.12	-23.99	-0.1127
90	1619.05	-23.92	1675.3	-5.15	-24.26	-0.123
91	1542.86	-24.25	1527.31	-5.49	-24.43	-0.1439
92	1466.67	-24.52	1418.14	-5.73	-24.61	-0.1617
93	1390.48	-24.73	1357.49	-5.6	-24.78	-0.1649
94	1314.29	-24.88	1340.51	-5.49	-24.96	-0.1639
95	1238.1	-24.97	1313.82	-5.39	-25.14	-0.1641
96	1161.9	-25	1279.86	-4.91	-25.14	-0.1535
97	1085.71	-24.97	1190.09	-4.98	-25.14	-0.1674
98	1009.52	-24.88	1105.18	-4.64	-25.05	-0.1679
99	933.33	-24.73	988.73	-4.16	-25.05	-0.1682
100	857.14	-24.52	911.1	-3.88	-24.96	-0.1706
101	780.95	-24.25	874.71	-0.39	-24.87	-0.0179
102	704.76	-23.92	821.33	2.45	-24.7	0.1193
103	628.57	-23.54	765.53	1.11	-24.43	0.0582
104	552.38	-23.1	697.6	2.14	-23.91	0.1228
105	476.19	-22.6	622.4	1.56	-23.11	0.1002
106	400	-22.05	547.19	1.05	-22.32	0.0765
107	400	-21.44	486.54	0.81	-21.44	0.0663
108	400	-20.79	435.59	0.63	-20.74	0.0583
109	400	-20.08	433.17	0.4	-20.04	0.0365
110	400	-19.32	455	0.87	-19.42	0.0769
111	400	-18.52	508.37	0.57	-18.63	0.0446
112	400	-17.68	573.88	0.81	-17.67	0.0562
113	400	-16.79	629.68	0.87	-16.61	0.0556
114	400	-15.86	644.23	0.84	-15.47	0.0522
115	400	-14.89	651.51	0.7	-14.41	0.0432
116	400	-13.89	649.08	0.74	-13.36	0.0455
117	400	-12.85	629.68	0.91	-12.3	0.0577
118	400	-11.79	590.86	1.29	-11.16	0.087
119	400	-10.69	532.63	0.5	-10.28	0.0374
120	400	-9.57	452.57	0.4	-9.14	0.0349
121	400	-8.42	401.63	0.12	-7.91	0.0121

122	400	-7.26	413.76	-0.05	-6.5	-0.0048
123	400	-6.08	428.31	-0.22	-5.1	-0.0206
124	400	-4.88	493.82	-0.36	-3.78	-0.029
125	400	-3.67	581.15	0.02	-2.55	0.0013
126	400	-2.45	583.58	-0.15	-1.41	-0.0105
127	400	-1.23	593.28	0.19	-0.26	0.0128

Appendix 8:

Friction Test Results (Data) obtained from the Friction Hip Simulator for the 50 mm Diameter Smith & Nephew Birmingham Hip Resurfacing Prosthesis with 200 μm Diametral Clearance and using Bovine Serum with CMC (of viscosity ~ 0.105 Pas) as Lubricant.

BS+CMC of η=0.105 Pas, 50mm diameter BHR, 200µm diametral clearance

Index	Demand Load(N)	Demand Motor(○)	Load(N)	Friction Torque(Nm)	Motor Position(○)	Friction Coefficient
0	400	0	547.19	0.19	0.7	0.0139
1	400	1.23	549.62	-0.26	1.85	-0.0186
2	400	2.45	537.49	0.05	3.25	0.0039
3	400	3.67	510.8	-0.22	4.57	-0.0173
4	400	4.88	476.83	-0.15	5.98	-0.0128
5	400	6.08	457.43	-0.22	7.21	-0.0193
6	400	7.26	433.17	-0.46	8.44	-0.0425
7	400	8.42	433.17	-0.49	9.49	-0.0457
8	400	9.57	430.74	-0.36	10.37	-0.0332
9	400	10.69	462.28	-0.39	11.34	-0.0339
10	400	11.79	467.13	-0.19	12.39	-0.016
11	400	12.85	493.82	-0.32	13.53	-0.0262
12	400	13.89	498.67	-0.43	14.77	-0.0342
13	400	14.89	505.95	-0.39	15.82	-0.031
14	400	15.86	501.09	-0.22	16.79	-0.0176
15	400	16.79	503.52	-0.39	17.75	-0.0312
16	400	17.68	488.96	-0.46	18.54	-0.0377
17	400	18.52	493.82	-0.6	19.16	-0.0484
18	400	19.32	479.26	-0.63	19.86	-0.0527
19	400	20.08	447.72	-0.84	20.48	-0.0748
20	400	20.79	447.72	-0.87	21.18	-0.0779
21	400	21.44	438.02	-0.39	21.97	-0.0358
22	476.19	22.05	435.59	-0.6	22.59	-0.0549
23	552.38	22.6	469.56	-0.77	23.2	-0.0655
24	628.57	23.1	535.06	-0.6	23.64	-0.0447
25	704.76	23.54	632.1	-0.46	23.99	-0.0292
26	780.95	23.92	753.4	-0.46	24.17	-0.0245
27	857.14	24.25	865	-0.39	24.34	-0.0181
28	933.33	24.52	913.52	-0.12	24.52	-0.0052
29	1009.52	24.73	1010.56	-0.08	24.7	-0.0033
30	1085.71	24.88	1037.25	0.05	24.87	0.002
31	1161.9	24.97	1066.36	-0.05	24.96	-0.0019
32	1238.1	25	1148.85	-0.36	24.96	-0.0125
33	1314.29	24.97	1224.06	-0.43	24.96	-0.0139
34	1390.48	24.88	1284.71	-0.26	24.96	-0.0079

35	1466.67	24.73	1357.49	-0.08	24.96	-0.0025
36	1542.86	24.52	1476.37	-1.62	24.87	-0.044
37	1619.05	24.25	1517.61	-4.02	24.78	-0.106
38	1695.24	23.92	1590.39	-4.95	24.52	-0.1244
39	1771.43	23.54	1651.04	-4.74	24.17	-0.1149
40	1847.62	23.1	1701.99	-4.64	23.55	-0.109
41	1923.81	22.6	1760.21	-4.5	22.85	-0.1023
42	2000	22.05	1871.81	-4.57	22.06	-0.0976
43	2000	21.44	1925.18	-4.33	21.27	-0.09
44	2000	20.79	1966.43	-4.33	20.65	-0.0881
45	2000	20.08	1968.85	-4.4	20.13	-0.0894
46	2000	19.32	1976.13	-4.6	19.42	-0.0932
47	2000	18.52	1896.07	-4.64	18.54	-0.0978
48	2000	17.68	1864.53	-5.08	17.58	-0.109
49	2000	16.79	1840.27	-5.19	16.52	-0.1127
50	2000	15.86	1835.42	-5.29	15.38	-0.1153
51	2000	14.89	1837.85	-5.39	14.33	-0.1173
52	2000	13.89	1859.68	-5.73	13.27	-0.1233
53	2000	12.85	1903.35	-6.28	12.39	-0.132
54	2000	11.79	1903.35	-6.14	11.51	-0.1291
55	2000	10.69	1900.92	-6.45	10.46	-0.1358
56	2000	9.57	1900.92	-6.52	9.23	-0.1372
57	2000	8.42	1886.37	-6.56	7.91	-0.139
58	2000	7.26	1874.24	-6.66	6.5	-0.1421
59	2000	6.08	1866.96	-6.52	5.19	-0.1397
60	2000	4.88	1866.96	-6.66	3.95	-0.1427
61	2000	3.67	1864.53	-6.56	2.72	-0.1406
62	2000	2.45	1874.24	-6.73	1.67	-0.1436
63	2000	1.23	1891.22	-6.83	0.7	-0.1444
64	2000	0	1893.65	-6.9	-0.44	-0.1457
65	2000	-1.23	1893.65	-6.8	-1.76	-0.1435
66	2000	-2.45	1898.5	-6.62	-3.25	-0.1396
67	2000	-3.67	1891.22	-6.49	-4.66	-0.1372
68	2000	-4.88	1896.07	-6.35	-6.06	-0.134
69	2000	-6.08	1903.35	-6.42	-7.29	-0.1349
70	2000	-7.26	1896.07	-6.42	-8.35	-0.1354
71	2000	-8.42	1891.22	-6.86	-9.4	-0.1452
72	2000	-9.57	1913.05	-6.11	-10.28	-0.1278
73	2000	-10.69	1915.48	-6.42	-11.25	-0.134
74	2000	-11.79	1881.52	-6.28	-12.39	-0.1335
75	2000	-12.85	1896.07	-5.84	-13.62	-0.1231
76	2000	-13.89	1900.92	-6.08	-14.94	-0.1279

77	2000	-14.89	1913.05	-5.77	-16	-0.1206
78	2000	-15.86	1913.05	-5.67	-16.96	-0.1185
79	2000	-16.79	1915.48	-5.77	-17.75	-0.1204
80	2000	-17.68	1910.63	-5.63	-18.54	-0.1179
81	2000	-18.52	1913.05	-5.94	-19.07	-0.1242
82	2000	-19.32	1910.63	-5.7	-19.77	-0.1193
83	2000	-20.08	1910.63	-5.7	-20.57	-0.1193
84	2000	-20.79	1908.2	-5.67	-21.36	-0.1188
85	2000	-21.44	1910.63	-5.84	-22.15	-0.1222
86	1923.81	-22.05	1908.2	-5.56	-22.76	-0.1166
87	1847.62	-22.6	1900.92	-5.77	-23.2	-0.1214
88	1771.43	-23.1	1871.81	-5.73	-23.55	-0.1225
89	1695.24	-23.54	1803.88	-5.56	-23.91	-0.1233
90	1619.05	-23.92	1699.56	-6.28	-24.08	-0.1478
91	1542.86	-24.25	1551.57	-6.25	-24.34	-0.1611
92	1466.67	-24.52	1471.51	-6.18	-24.61	-0.168
93	1390.48	-24.73	1391.45	-5.94	-24.87	-0.1707
94	1314.29	-24.88	1379.32	-6.04	-25.05	-0.1752
95	1238.1	-24.97	1340.51	-5.43	-25.14	-0.1619
96	1161.9	-25	1275	-5.08	-25.14	-0.1595
97	1085.71	-24.97	1190.09	-5.12	-25.14	-0.172
98	1009.52	-24.88	1095.48	-4.4	-25.05	-0.1606
99	933.33	-24.73	993.58	-4.77	-25.05	-0.1922
100	857.14	-24.52	928.08	-3.37	-24.96	-0.1453
101	780.95	-24.25	879.56	-0.26	-24.87	-0.0116
102	704.76	-23.92	838.32	1.32	-24.7	0.063
103	628.57	-23.54	777.66	1.15	-24.43	0.0591
104	552.38	-23.1	707.31	1.53	-23.82	0.0863
105	476.19	-22.6	634.53	0.87	-23.11	0.0551
106	400	-22.05	573.88	1.08	-22.24	0.0753
107	400	-21.44	491.39	0.53	-21.44	0.0433
108	400	-20.79	445.3	0.46	-20.83	0.0417
109	400	-20.08	442.87	0.22	-20.21	0.0202
110	400	-19.32	450.15	0.36	-19.51	0.0321
111	400	-18.52	501.09	0.12	-18.54	0.0097
112	400	-17.68	569.02	0.81	-17.49	0.0567
113	400	-16.79	593.28	0.43	-16.44	0.029
114	400	-15.86	612.69	0.74	-15.47	0.0482
115	400	-14.89	612.69	0.63	-14.41	0.0415
116	400	-13.89	627.25	0.26	-13.36	0.0165
117	400	-12.85	588.43	0.16	-12.39	0.0106
118	400	-11.79	561.75	0.43	-11.43	0.0306

119	400	-10.69	518.08	0.33	-10.37	0.0252
120	400	-9.57	459.85	-0.02	-9.05	-0.0014
121	400	-8.42	418.61	0.36	-7.73	0.0345
122	400	-7.26	416.18	-0.39	-6.33	-0.0377
123	400	-6.08	457.43	-0.29	-5.1	-0.0253
124	400	-4.88	484.11	-0.15	-3.87	-0.0126
125	400	-3.67	537.49	-0.15	-2.64	-0.0114
126	400	-2.45	561.75	-0.46	-1.58	-0.0328
127	400	-1.23	549.62	0.16	-0.53	0.0113

Appendix B: Publications

THE EFFECT OF CUP DEFLECTION ON FRICTION IN METAL-ON-METAL BEARINGS

*A Kamali; *JT Daniel; **SA Javid; **M Youseffi; *T Band; *R Ashton; *A Hussain; *CX Li; ***J Daniel; ***McMinn DJW
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Introduction: Cementless cup designs in metal-on-metal (MoM) hip resurfacing devices generally depend on a good primary press-fit fixation which stabilises the components in the early post-operative period. Press-fitting the cup into the acetabulum generates non-uniform compressive stresses on the cup and consequently causes non-uniform cup deformation. That in turn may result in equatorial contact, high frictional torque and femoral head seizure. It has been reported that high frictional torque has the potential to generate micromotion between the implant and its surrounding bone and as a result adversely affect the longevity of the implant. The aim of this study was to investigate the effects of cup deformation on friction between the articulating surfaces in MoM bearings with various clearances.

Materials and methods: Six Birmingham Hip Resurfacing (BHR) devices with various clearances (80 to 306 µm) were tested in a hip friction simulator to determine the friction between the bearing surfaces. The components were tested in clotted blood which is the primary lubricant during the early post-operative period. The joints were friction tested initially in their pristine conditions and subsequently the cups were deflected by 25-35 µm using two points pinching action before further friction tests were carried out.

Results and Discussions: It has been reported that reduced clearance results in reduced friction. However, none of the previous studies have taken cup deflection into consideration nor have they used physiologically relevant lubricant. The results presented in this study show that for the reduced clearance components, friction was significantly increased when the cups were deflected by only 30 µm. However, for the components with higher clearance the friction did not change before and after deflection. It is postulated that the larger clearances can accommodate for the amount of distortion introduced to the cups in this study.

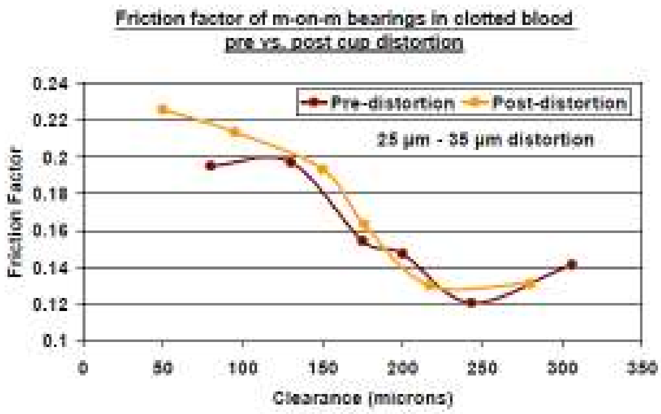

Figure 1. Friction vs. clearance in pre and post distorted cups for MoM joints

The effect of clearance upon friction of large diameter hip resurfacing prostheses using blood, clotted blood and bovine serum as lubricants

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Abstract: **Total hip joint implantation is an effective solution for reducing pain and ailing induced by arthritis or other diseases at the hip joint. Hence, a conventional metal on polyethylene (PE) bearing device has been introduced since late 1950's for implantation. However, due to significant release of PE wear debris causing swelling at joints and osteolysis leading to implant loosening and failure in fixation, attempts are made to optimize implant design, manufacturing and surgical procedures for a relatively new metal on metal hip resurfacing prostheses of larger diameters to have lower friction and wear, better fixation and reduced risk of dislocation.**

The aim of the present study was to investigate the role of diametral clearance on friction using a large diameter metal on metal hip resurfacing prosthesis and various lubricants including blood, clotted blood and combinations of bovine serum with aqueous solutions of carboxymethyl cellulose (CMC) and hyaluronic acid (HA).

Keywords: **Friction, 50mm diameter metal-on-metal hip resurfacing prostheses, Diametral clearances, Blood and Clotted Blood, BS+CMC and BS+HA (+CMC).**

I. INTRODUCTION

The major objectives in the design of joint prostheses are the development of stable articulations, low friction and wear, solid fixation into the bone, and normal range of motion. New synthetic replacement materials are being designed by biomedical/medical engineers to accomplish these objectives. Total Hip Joint Replacements (THRs) are usually composed of cobalt chrome molybdenum (Co-Cr-Mo) alloys in combination with modern plastic, Ultra-High Molecular Weight Polyethylene (UHMWPE). In recent years, the proportions of younger and more active patients undergoing arthroplasty have increased mainly because of advanced osteoarthritic joints in ~10% of young adults. Due to their higher activity levels compared to the inactive population, the elderly, and the limited survivorship of the conventional replacement, there are still concerns about the use of metal on polyethylene prosthesis for the younger generation. Highly active patients could potentially generate extremely high wear rates, which would in turn result in premature failure of the implant. The most commonly quoted survivorship rates followed by revision for a particular implant or treatment is 10 – 15 years. Hence, it would be advantageous to develop a replacement that could either survive the patients' lifetime, or use a replacement that would conserve bone stock for the eventual revision. One of the hip replacement procedures in which the head of the femur is retained resulting in minimum bone removal is called hip resurfacing. Instead of removing the head completely, it is shaped to accept an anatomically sized metal sphere. There is no large stem to go down the central part of the femur (or femoral shaft) and the surface of the acetabulum is also replaced with a metal implant, which is wedged directly into the bone.

The modern resurfacing components are made of Co-Cr-Mo metal alloys, which are finely polished to produce a very smooth surface finish giving low friction and wear. There are many other advantages of using hip resurfacing arthroplasty including bone conservation, improved function due to retention of the femoral head and neck and hence better biomechanical restoration, decreased morbidity at the time of revision arthroplasty, reduced dislocation rates and stress-shielding, less infection, and reduced occurrences of thromboembolic phenomena (less blood clotting due to not using any tools/stems in the femur). Typical examples of such devices include the Birmingham Hip Resurfacing prostheses from Smith and Nephew Orthopaedic Ltd. (see Figure 1 [1]), ASR from DePuy International, and DUROM from Zimmer. Following their promising short to medium term clinical results, second-generation metal-on-metal hip resurfacing prostheses have, therefore, been extensively introduced since the 1990's by almost all major orthopaedic companies.

It has been illustrated via both simulator studies and clinical trials that correct manufacturing of the prosthesis will lead to excellent sphericity, tolerances, and an optimum radial clearance which are the main reason for their success. Use of larger diameter bearings (>35-50mm diameter) and hip resurfacing prostheses have the advantages of increased range of motion and decreased incidence of dislocation for younger and more active patients.

The clearance between the articulating components is size-dependent, i.e. the larger the diameter the higher the gap/clearance between the components. The range for the entire family of various diameters is from 90 to 200 microns of diametral clearance, with each bearing size having an optimized gap for maximum fluid film thickness. The diametral clearances between articulation components play a major role in their generation of wear debris which is probably the most influential factor in wear behaviour. Proper clearance is essential for entrapping the synovial fluid between the articulating surfaces. This fluid is largely responsible for separating the surfaces while the joint is in motion and, thereby, reducing wear. If the gap between components is too small or too large there will be a sharp increase in wear rates.

O. Dössel and W.C. Schlegel (Eds.): WC 2009, IFMBE Proceedings 25/IX, pp. 418–420, 2009.
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The aim of this work was therefore to investigate the effect of various clearances on friction and lubrication of as-cast Co-Cr-Mo hip resurfacing components with 50mm diameter via hip friction simulator studies which have strongly indicated that clearance plays a major role in generation of high friction and wear, and that wear appears to be relatively insensitive to changes in materials that have similar chemical compositions but different microstructures [2].

II. Materials and Methods

Immediately after joint replacement, the artificial prosthesis is bathed in blood and clotted blood instead of synovial fluid. Blood contains large molecules and cells of size ~ 5 to 20 micron suspended in plasma and is considered to be a non-Newtonian fluid with density of 1060 Kg/m^3. The effect of these properties on friction is not fully understood and, so far, hardly any studies have been carried out regarding friction of metal-on-metal bearings with various clearances in the presence of lubricants such as blood or clotted blood. In this work, therefore, we have investigated the frictional behavior of a group of Smith and Nephew Birmingham Hip Resurfacing devices with a nominal diameter of 50mm and diametral clearances in the range ~ 80 to 300µm, in the presence of blood (clotted and whole blood).

Five as cast high carbon Co-Cr-Mo MOM 'Birmingham Hip Resurfacing (BHR) devices' (supplied by Smith and Nephew Orthopaedics Ltd, Coventry, UK) with a nominal diameter of 50 mm each and diametral clearances of 80, 135, 200, 243 and 306 µm were used in this study. The initial surface roughnesses were in the range 'R_a=10-30 nm' similar to those of commercial MOM hip prostheses.

Frictional measurements of all the joints were carried out using a Prosim Hip Joint Friction Simulator (Simulation Solutions Ltd, Stockport, UK). Friction measurements were made in the 'stable' part of the cycle at 2000N and thus the loading cycle was set at maximum and minimum loads of 2000N and 100N, respectively. In the flexion/extension plane, an oscillatory harmonic motion of amplitude ±24° was applied to the femoral head with a frequency of 1Hz in a period of 1.2s. The load was therefore applied to the femoral head with the artificial hip joint in an inverted position, i.e. femoral head on top of the acetabular component, but with a 12° angle of loading between the two bearings as observed in human's body (12° medially to the vertical). The angular displacement, frictional torque (T) and load (L) were recorded through each cycle. The frictional torque was then converted into friction factor (f) using the equation: $f = T/rL$, where r is the femoral head radius. An average of three independent runs (tests) was taken.

Initially, the test was conducted with non-clotted blood (whole blood with Lithium heparin to prevent clotting) and then clotted blood as the lubricants for each joint. Viscosity of the non-clotted blood was found to be ~ 0.01 Pas and that of clotted blood was ~ 0.02 Pas. For comparison, combinations of bovine serum (BS, 25%) with aqueous solutions of carboxymethyl cellulose (CMC+75% distilled water) and Hyaluronic acid (HA) as lubricants were also used with viscosities of 0.0136 Pas (BS+CMC) and 0.0132 Pas (BS+HA+CMC). Note that CMC was used as the gelling agent to obtain the required viscosities.

III. Results and Discussion

Table 1 and Figure 2 show a close comparison between friction factors for various diametral clearances of 80 to 306 µm using Blood and Clotted blood as lubricants. It became more obvious that both blood and clotted blood resulted in higher friction factors especially at lower clearances of 80 and 135 µm. This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth immediately after implantation resulting in impaired fixation with long-term implications for survival. The friction factors in Table 1 have also shown that lower clearances do not necessarily reduce the friction factors to a level for the presence of full fluid film lubrication and that the friction factors decrease with increase in diametral clearance. This finding clearly suggests that lower clearances have higher potential for increasing the friction between the articulating joint surfaces using blood and clotted blood as lubricants and thus increase the risk of micromotion due to higher surface contacts, leading to higher risk of joint dislocation. On the other hand, the BS+CMC and BS+HA+CMC lubricants with similar viscosities showed the opposite effect, i.e. caused an increase in friction factor with increase in diametral clearance (see Table 1 and Figure 2). Also notable was that, the friction factors were consistently higher for blood and clotted blood as compared to those of lubricants based on bovine serum.

Table 1: Friction factors for the whole blood, clotted blood, BS+CMC and BS+HA+CMC for different diametral clearances.

Diametral Clearance, µm	Whole Blood, η=0.013 Pas	Clotted Blood, η=0.02 Pas	BS+CMC, η=0.0136 Pas	BS+HA(+CMC), η=0.0132 Pas
80	0.19	0.17	0.08	0.07
135	0.19	0.165	0.09	0.076
200	0.18	0.16	0.12	0.095
243	0.143	0.15	0.125	0.1
306	0.14	0.14	0.132	0.114

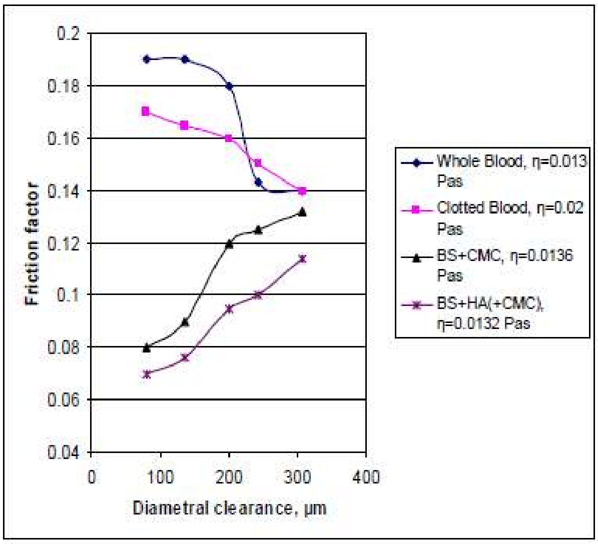

Figure 2. Graph of friction factor versus diametral clearance for the S&N BHR 50mm diameter devices using Blood, Clotted blood and bovine serum as lubricants.

Figure 1. The Birmingham Hip Resurfacing (BHR) device consists of two parts: an acetabular component (or cup), and a femoral resurfacing component (or head).

IV. CONCLUSIONS

- The in vitro frictional behaviour of five large diameter (50mm nominal) S & N BHR prostheses with various diametral clearances (~ 80-300 µm) has been investigated using blood, clotted blood and combinations of bovine serum with aqueous solutions of CMC and Hyaluronic acid as lubricants to understand and mimic the in vivo frictions generated at the articulating surfaces immediately after hip implantation.
- It became clear that the friction factors decreased consistently with increase in diametral clearance for both blood and clotted blood. This therefore suggested that higher clearances will lower the friction (and hence wear) for these large diameter S&N BHR devices depending on the type of lubricant and viscosity.
- On the other hand, the bovine serum based lubricants with similar viscosities showed the opposite effect, i.e. caused an increase in friction factor with increase in diametral clearance and that the friction factors were consistently higher for blood and clotted blood as compared to those of lubricants based on bovine serum.

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* Oral

Effect of Cup Deflection on Friction of Hip Resurfacing Prostheses with various Clearances using Blood and Clotted blood as Lubricants

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Abstract. Clinical studies have indicated that deflection of the acetabular cup may influence the generation of high friction in metal-on-metal total hip arthroplasty (THA). This may result in micromotion leading to dislocation of hip prostheses requiring revision surgery which is clinically a complicated operation. A successful total hip joint implantation can be achieved when the correct surgical method is used to eliminate cup deflection during implantation [1].

Keywords: Friction, Cup deflection, 50mm diameter metal-on-metal hip resurfacing prosthesis, diametral clearances, Blood, Clotted Blood.

MATERIALS AND METHODS

We investigated the frictional behavior of a group of Smith& Nephew (BHR) devices with a nominal diameter of 50mm and diametral clearances in the range ~ 80 to 300μm, in the presence of blood and clotted blood after cup deflection. Frictional measurements were carried out using a Prosim Hip Joint Friction Simulator (Stockport, UK). The angular displacement, frictional torque (T) and load (L) were recorded through each cycle. The frictional torque was then converted into friction factor (f) using the equation: $f = T/rL$, where r is the femoral head radius. An average of three independent tests was taken. Viscosity of the blood was found to be ~ 0.0083 Pas and that of clotted blood was ~ 0.0108 Pas using an Anton Paar Rheometer.

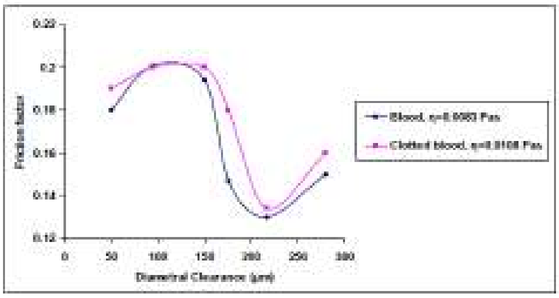

FIGURE 1. Graph of friction factor versus diametral clearance after ~25-35μm cup deflection.

RESULTS AND DISCUSSION

The average friction factors of (3 tests) for different clearances after ~25-35μm deformation is given in Table 1 and Figure 1 which is a graph of friction factor versus diametral clearance after cup deflection. From Table 1 and Figure 1 it can be seen quite clearly that friction factor has decreased consistently as diametral clearance increases for both blood and clotted blood. This is a significant finding since the results obtained during this work clearly show that for lower clearances friction increased significantly when the cups were deflected by ~30 μm. It is therefore illustrated that higher clearances can accommodate the amount of distortion or deflection that occurs in the cups during press-fit arthroplasty. The results of this study suggest therefore that reduced clearance bearings have the potential to generate high equatorial friction especially in the early weeks after implantation when blood is indeed the in vivo lubricant. This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth resulting in impaired fixation with long-term implications for survival.

TABLE 1. Average friction factors after ~25-35μm cup deformation using blood and clotted blood as lubricants.

Diametral Clearance μm	Cup deflection μm	Blood η=0.0083 Pas	Clotted Blood η=0.0108 Pas
80	30	0.18	0.19
200	24	0.147	0.18
306	26	0.15	0.16

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Effect of Cup Deflection on Friction of Hip Resurfacing Prosthesis with Various Clearances Using Blood and Clotted Blood as Lubricants

S. Afshinjavid and M. Youseffi

Abstract— **Clinical studies have indicated that deflection of the acetabular cup may influence the generation of high friction and wear in metal-on-metal total hip arthroplasty (THA). This may result in micromotion leading to dislocation of hip prostheses requiring revision surgery which is clinically a complicated operation. A successful total hip joint implantation can be achieved when the correct surgical method is used to eliminate cup deflection during implantation. Clearance plays a unique role in squeezing lubricants between contact surfaces allowing the formation of a fluid film, and thus any deflection of the cup during surgery may result in negative action during articulation. This work investigated the effect of cup deflection (~25-30μm) on friction of large diameter (50mm) metal on metal Birmingham Hip Resurfacing (BHR) prosthesis with various clearances (80-306μm) using blood and clotted blood as lubricants. It was found that the physiological lubricants caused higher friction factors at lower diametral clearances which is opposite to the serum-based lubricants causing higher friction at higher clearances.**

Keywords— **Blood and Clotted Blood; Diametral clearances; Friction and Cup deflection; 50mm diameter metal-on-metal hip resurfacing prosthesis.**

I. Introduction

Hip resurfacing arthroplasty has become increasingly popular for young and more active patients, who may require a second procedure in their lifetime. Metal-on-Metal (MoM) hip resurfacing arthroplasty is currently one of the most common implantation for patients with advanced arthritis disorders. Metal-on-metal hip resurfacing systems were introduced by the majority of implant manufacturers by the end of 2004. The goal of hip resurfacing arthroplasty is to remove the two damaged and worn parts of the hip joint, i.e. the arthritic acetabulum and femoral surfaces, and replace them with artificial implants to reproduce the form and function of the natural joint, relief pain, restore function and correct deformity.

The first metal-on-metal resurfacing prostheses were established by McMinn and Wagner [1, 2]. These prostheses were made from Co-Cr-Mo alloys and were initially implanted cementless. To improve the stability and osteointegration, the McMinn prosthesis has been modified to a cemented femoral component and with a hydroxyapatite (HA) coated cup [3] for which a survival rate of 99.8 % at four years has been reported with short rehabilitation periods allowing patients to return to their preoperative levels of activity. It has been acknowledged that the lower friction characteristics of metal-on-metal implants are related to the generation of full or partial lubricating films throughout the walking cycle.

Historically, there have been very few incidences of mechanical failures with metal-on-metal total hip replacements causing dislocation. While the optimal clearance to achieve elastohydrodynamic lubrication and avoid equatorial seizing is still being studied and debated, tribologists recommend that the diametral clearance be as small as possible in large-diameter bearings [4, 5]. This requirement must be balanced against practical limitations of manufacturing tolerances and also must take into account the possibility that deformation of the acetabular cup may occur when it is implanted into the acetabulum with a press-fit of 1 to 2 mm. Initial stability can be influenced by the method of fixation (press-fit), the surgical technique, the quantity and quality of the bone structure, bearing geometry and applied loading conditions. Press-fit fixation involves inserting an acetabular cup into an under reamed acetabulum, where the primary stability is gained through the frictional compressive forces generated about the acetabular periphery. The press-fit procedures have moderate influence on the contact mechanics at the bearing surfaces, but produce remarkable deformation of the acetabular cup. Further deformation of the acetabular cup, and subsequent reduction of the effective clearance, may also occur with physiological loading. The effect of cup deflection on clearance has been studied experimentally in cadaver pelvis and with the use of finite-element modeling [6]. The wall thickness of the cup showed to be the most important factor influencing deformation of the acetabular cup in both studies, although diametral clearance and bearing diameter were also important. Therefore, design and manufacturing parameters such as diametral clearance, femoral head diameter, surface finish or roughness, have shown to significantly influence the contact mechanics and tribology at the bearing surfaces of hip resurfacing arthroplasty. It is therefore important for the orthopaedic manufacturers to ensure that deformation of the component does not adversely affect clearance since this would lead to increased friction and hence joint dislocation [7].

The aim of this work was to investigate the effect of cup deformation on friction between the articulating surfaces of six Birmingham Hip Resurfacing devices with various clearances using blood and clotted blood as lubricants.

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II. Materials and Methods

Immediately after joint replacement, the artificial prosthesis is actually bathed in blood and clotted blood instead of synovial fluid. Blood contains large molecules and cells of size ~ 5 to 20 micron suspended in plasma and is considered to be a non-Newtonian fluid with density of 1060 Kg/m^3. The effect of these properties on friction is not fully understood and, so far, hardly any studies have been carried out regarding friction of metal-on-metal bearings with various clearances in the presence of lubricants such as blood or clotted blood. Therefore, we investigated the frictional behaviour of a group of Birmingham Hip Resurfacing devices (supplied by Smith & Nephew Orthopaedics Ltd, UK) with a nominal diameter of 50mm and original diametral clearances in the range ~ 80 to 300μm, in the presence of blood (clotted and whole blood) after cup deflection (See Table 1 for details) using two-point pinching action before friction tests.

Frictional measurements of all the joints were carried out using a Prosim Hip Joint Friction Simulator (Simulation Solutions Ltd, Stockport, UK). The acetabular cup was positioned in a fixed low-friction carriage below and the femoral head in a moving-frame above. The carriage sits on an externally pressurized hydrostatic bearings generating negligible friction compared to that generated between the articulating surfaces, also allowing for a self-centring mechanism. During the flexion-extension motion, the friction generated between the BHR devices causes the pressurized carriage to move. This movement (or rotation) is restricted by a sensitive Kistler piezoelectric force transducer which is calibrated to measure torque directly. A pneumatic mechanism controlled by a microprocessor generates a dynamic loading cycle and the load is also measured by the same piezoelectric force transducer. Friction measurements (friction factor results given in Table 1) were made in the 'stable' part of the cycle at 2000N and to obtain accurate measurements for friction, the centre of rotation of the joint was aligned closely with the centre of rotation of the carriage. The loading cycle was set at maximum and minimum loads of 2000N and 100N to represent the stance and the swing phase of the walking cycle, respectively.

In the flexion/extension plane, an oscillatory harmonic motion of amplitude ±24° was applied to the femoral head with a frequency of 1Hz in a period of 1.2s. The load was, therefore, applied to the femoral head with the artificial hip joint in an inverted position, i.e. femoral head on top of the acetabular component, but with a 12° angle of loading between the two bearings as observed in human's body (12° medially to the vertical).

The angular displacement, frictional torque (T) and load (L) were recorded through each cycle. The frictional torque was then converted into friction factor (f) using the equation: $f = T/rL$, where r is the femoral head radius. An average of three independent runs (tests) was taken.

Initially, the test was conducted with non-clotted blood (whole blood with Lithium heparin to prevent clotting) and clotted blood as the lubricants for each joint. Viscosity of the non-clotted blood was found to be ~ 0.0083 Pas and that of clotted blood was ~ 0.0108 Pas using an Anton Paar Rheometer.

III. Results and Discussion

The dynamic loading cycles generated during the friction tests are plotted graphically in Figures 1 and 2 as graphs of load, frictional torque, friction factor and flexion-extension (±24° oscillatory harmonic motion) versus the number of cycles (=127). It is to be noted that the friction factors were taken from the stable part of the cycle at 2000N and thus the frictional torques were also from this part of the cycle which represents the normal loading cycle observed in human's body having ~ 12° angle of loading between the acetabulum and the femoral head. The friction torques (~4.8-7.2 Nm) obtained during friction tests were within the safe range causing no risk of dislocation.

The average friction factors (average of 3 tests) for different clearances after ~25-35μm deformation is given in Table 1 and Figure 3 which is a graph of friction factor versus diametral clearance after cup deflection.

From Table 1 and Figure 3 it can be seen quite clearly that friction factor has decreased consistently as diametral clearance increases for both blood and clotted blood. These results are in agreement with those before cup deflection, when blood and clotted blood were also used on the original joints [8]. This is a significant finding since the results obtained during this work clearly show that for lower clearances friction increased significantly when the cups were deflected by ~30 μm. It is therefore illustrated that higher clearances can accommodate the amount of distortion introduced in the cups during this investigation or indeed during implantation. The results of this study suggest therefore that reduced clearance bearings have the potential to generate high equatorial friction especially in the early weeks after implantation when blood is indeed the in vivo lubricant. This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth resulting in impaired fixation with long-term implications for survival.

Table1. Average friction factors after ~25-35μm cup deformation using blood and clotted blood as lubricants.

Original Diametral Clearance (μm)	Cup deflection (μm)	Diametral Clearance after deflection (μm)	Blood η=0.0083 (Pas)	Clotted blood η=0.0108 (Pas)
80	30	50	0.18	0.19
130	35	95	0.201	0.2
175	25	150	0.194	0.2
200	24	176	0.147	0.18
243	26	217	0.13	0.134
306	26	280	0.15	0.16

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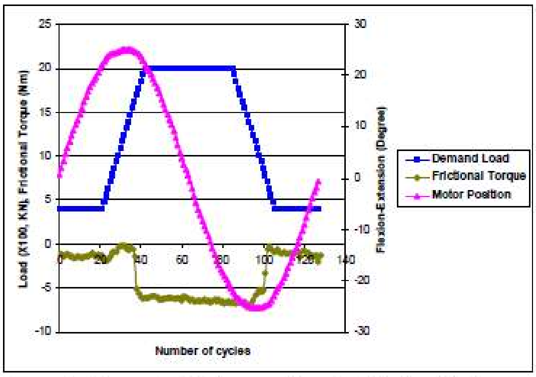

Figure 1. Dynamic-Motion profile plotted after friction test showing variation in frictional torque with the applied load in ±24° extension-flexion versus number of cycles for the 50μm diametral clearance after initial deflection, 50mm BHR bearing using Blood as lubricant.

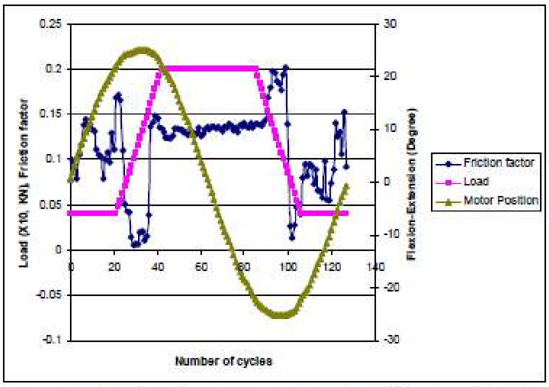

Figure 2. Friction factor versus number of cycles for the 50mm BHR prosthesis with a diametral clearance of 217μm after cup deflection using blood as lubricant.

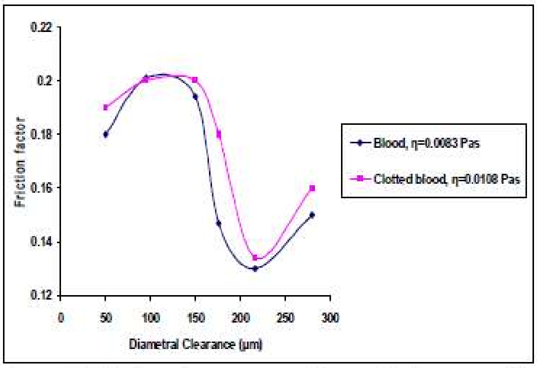

Figure 3. Friction factor versus diametral clearance after ~25-35μm cup deflection.

IV. Conclusions

Six large diameter (50mm nominal) BHR deflected prostheses with various clearances (~ 50-280μm) were friction tested in vitro in the presence of blood and clotted blood to study the effect of cup deflection on friction. It was found that the biological lubricants caused higher friction factors at the lower diametral clearances for blood and clotted blood as clearance decreased from 280μm to 50μm (after deflection). It is postulated that if the cup is deflected by press fitting, this may result in increased contact at bearing surfaces around the equatorial rib of the cup and result in higher frictional torque which can increase the risk of dislocation and hamper fixation. This has been the case for some early loosening of the implants after few weeks of implantation. This work therefore showed clearly that higher clearances will lower the friction for large diameter BHR bearings, which, in turn, may accommodate for the amount of deflection that occurs in the cups during press-fit arthroplasty.

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The effect of clearance upon friction of large diameter hip resurfacing prostheses using blood, clotted blood and bovine serum as lubricants

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Abstract: **Total hip joint implantation is an effective solution for reducing pain and ailing induced by arthritis or other diseases at the hip joint. Hence, a conventional metal on polyethylene (PE) bearing device has been introduced since late 1950's for implantation. However, due to significant release of PE worn debris causing swelling at joints and osteolysis leading to implant loosening and failure in fixation, attempts are made to optimize implant design, manufacturing and surgical procedures for a relatively new metal on metal hip resurfacing prostheses of larger diameters to have lower friction and wear, better fixation and reduced risk of dislocation.**

The aim of the present study was to investigate the effect of diametral clearance on friction using a large diameter metal on metal hip resurfacing prosthesis and various lubricants including blood, clotted blood and bovine serum with aqueous solutions of carboxymethyl cellulose (CMC) and hyaluronic acid (HA).

INTRODUCTION

One of the hip replacement procedures in which the head of the femur is retained resulting in minimum bone removal is called hip resurfacing. Instead of removing the head completely, it is shaped to accept an anatomically sized metal sphere. There is no large stem to go down the central part of the femur (or femoral shaft) and the surface of the acetabulum is also replaced with a metal implant, which is wedged directly into the bone (see Figure 1). The modern resurfacing components are made of Co-Cr-Mo metal alloys, which are finely polished to produce a very smooth surface finish giving low friction and wear. There are many other advantages of using hip resurfacing arthroplasty including bone conservation, improved function due to retention of the femoral head and neck and hence better biomechanical restoration, decreased morbidity at the time of revision arthroplasty, reduced dislocation rates and stress-shielding, less infection, and reduced occurrences of thromboembolic phenomena (less blood clotting due to not using any tools/stems in the femur). It has been illustrated via both simulator studies and clinical trials that correct manufacturing of the prosthesis will lead to excellent sphericity, tolerances, and an optimum radial clearance which are the main reason for their success. Use of larger diameter bearings (>35-50mm diameter) and hip resurfacing prostheses have the advantages of increased range of motion and decreased incidence of dislocation for younger and more active patients.

The clearance between the articulating components is size-dependent, i.e. the larger the diameter the higher the gap/clearance between the components. The range for the entire family of various diameters is from ~90 to 200 microns of diametral clearance, with each bearing size having an optimized gap for maximum fluid film thickness [1-2]. The diametral clearances between articulation components play a major role in their generation of wear debris which is probably the most influential factor in wear behaviour.

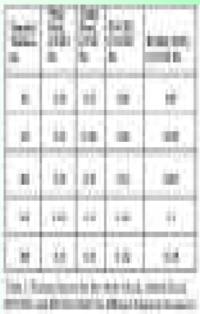
MATERIALS AND METHODS

Five as cast high carbon Co-Cr-Mo 'Birmingham Hip Resurfacing (BHR) devices' (supplied by Smith and Nephew Orthopaedics Ltd, Coventry, UK) with a nominal diameter of 50mm each and diametral clearances of 80, 135, 200, 243 and 306 µm were used in this study. Frictional measurements of all the joints were carried out using a Prosim Hip Joint Friction Simulator (Simulation Solutions Ltd, Stockport, UK). Friction measurements were made in the 'stable' part of the cycle at 2000N and thus the loading cycle was set at maximum and minimum loads of 2000N and 100N, respectively. In the flexion/extension plane, an oscillatory harmonic motion of amplitude ±24° was applied to the femoral head with a frequency of 1Hz in a period of 1.2s. The angular displacement, frictional torque (T) and load (L) were recorded through each cycle. The frictional torque was then converted into friction factor (*f*) using the equation: $f = T/rL$, where r is the femoral head radius. An average of three independent runs (tests) was taken.

Initially, the test was conducted with non-clotted blood (whole blood with Lithium heparin to prevent clotting) and then clotted blood as the lubricants for each joint. Viscosity of the non-clotted blood was found to be ~ 0.01 Pas and that of clotted blood was ~ 0.02 Pas. For comparison, combinations of bovine serum (BS, 25%) with aqueous solutions of carboxymethyl cellulose (CMC+75% distilled water) and Hyaluronic acid (HA) as lubricants were also used with viscosities of 0.0136 Pas (BS+CMC) and 0.0132 Pas (BS+HA+CMC). Note that CMC was used as the gelling agent to obtain the required viscosities.

RESULTS AND DISCUSSION

Table 1 and Figure 2 show a close comparison between friction factors for various diametral clearances of 80 to 306 µm using Blood and Clotted blood as lubricants. It became more obvious that both blood and clotted blood resulted in higher friction factors especially at lower clearances of 80 and 135 µm. This higher friction in the low clearance bearings may produce micromotion and hamper bony ingrowth immediately after implantation resulting in impaired fixation with long-term implications for survival. The friction factors in Table 1 have also shown that lower clearances do not necessarily reduce the friction factors to a level for the presence of full fluid film lubrication and that the friction factors decrease with increase in diametral clearance. This finding clearly suggests that lower clearances have higher potential for increasing the friction between the articulating joint surfaces using blood and clotted blood as lubricants and thus increase the risk of micromotion due to higher surface contacts, leading to higher risk of joint dislocation. On the other hand, the BS+CMC and BS+HA+CMC lubricants with similar viscosities showed the opposite effect, i.e. caused an increase in friction factor with increase in diametral clearance (see Table 1 and Figure 2). Also notable was that, the friction factors were consistently higher for blood and clotted blood as compared to those of lubricants based on bovine serum.



CONCLUSIONS

- **The in vitro frictional behaviour of five large diameter (50mm nominal) S & N BHR prostheses with various diametral clearances (~ 80-300 µm) has been investigated using blood, clotted blood and combinations of bovine serum with aqueous solutions of CMC and Hyaluronic acid as lubricants to understand and mimic the in vivo frictions generated at the articulating surfaces immediately after hip implantation.**
- **It became clear that the friction factors decreased consistently with increase in diametral clearance for both blood and clotted blood. This therefore suggested that higher clearances will lower the friction (and hence wear) for these large diameter S&N BHR devices depending on the type of lubricant and viscosity.**
- **On the other hand, the bovine serum based lubricants with similar viscosities showed the opposite effect, i.e. caused an increase in friction factor with increase in diametral clearance and that the friction factors were consistently higher for blood and clotted blood as compared to those of lubricants based on bovine serum.**

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