An Investigation into the Effects of Mechanical Loading Patterns from Everyday Activities on Long Term Stability after Cementless Hip Replacement Surgery

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DECLARATION

I hereby declare that this thesis is entirely my own work, and has not been submitted for any other awards at this or any other academic establishment. Where use has been made of the work of other people it has been fully acknowledged and referenced.

______________  ______________
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Abstract

The number of Total Hip Replacement (THR) surgeries carried out per year is increasing year on year as people live longer and partake in a more active lifestyle. Other factors such as obesity and weight problems are also increasing and adding to the number of surgeries carried out. Aseptic loosening is the most frequent complication for THR for both cemented and cementless prostheses. However, the long term stability of the cementless prostheses is not well documented. In vitro studies of cementless prostheses have all shown primary stability of the prosthesis whereas the long term stability of the prosthesis and the corresponding effects of mechanical loading have not been investigated. Of the published literature that focused on numerical modelling only Jung et al. (2004) was found to evaluate long term stability and the loosening of cementless prostheses with most other studies concentrating on either primary or secondary stability.

Motion analysis was carried out on 23 subjects consisting of 12 THR subjects, 6 Birmingham replacement subjects and 5 healthy subjects. Walking, sitting, standing, up steps and down steps motions were carried out by each subject during the testing. Using musculo skeletal modelling and motion data, average hip joint loading forces for THR subjects were calculated for each motion. These loads were then used as the hip joint forces for the experimental and numerical modelling throughout the study. Statistical analysis was carried out on the forces which highlighted statistical significance between a number of forces when comparing the following groups, male vs. female, large prosthesis head size vs. standard prosthesis head size, anterior surgical approach vs. posterior surgical approach, healthy hip joints vs. replaced hip joints in replacement subjects and THR subjects vs. Birmingham subjects vs. healthy subjects. Four of these significant differences were investigated further with numerical models to determine if any difference in stresses were found at the interface.

There are limitations associated with loading patterns in the existing published experimental simulators as most previous research has used a dynamic ramp load which does not take into account realistic forces created at the hip joint such as walking gait often having 2 peaks during one cycle. An experimental rig was designed, fabricated and commissioned. Three pneumatic pistons were used to create the hip joint forces and also change the angle throughout a cycle. Three experimental models were conducted with varying loading conditions. The results demonstrated that both elastic and plastic displacements occurred at the interface due to the mechanical loading. The displacements recorded indicate that the mechanical loading could lead to the loosening of the prosthesis. However, it is important to note that none of the three models became unstable even after 5 million cycles. However, these results do show that the mechanical loading can change the position of the implanted prosthesis over a long period of time.

Finally, the effect of the loading directions was investigated using numerical modelling and the average and maximum peak joint forces for each motion were calculated from the musculo skeletal modelling results. The loading direction was also varied in each model to investigate the effects of each loading direction. The Y direction (anterior/posterior) was identified as the main force associated with high stresses that may lead to instability and aseptic loosening while the X direction (medial/lateral) force was seen to play a role in reducing the stresses on the interface surface especially in the neck region where high stresses occurred. The Y direction was also found to be a greater indicator of possible loosening rather than the load magnitude which is used in literature to date.
Dedication

I dedicate this thesis to my parents Stephanie and Richard and my grandparents Ellen, Frank, Anna and Especially Martin.
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<th>Symbol</th>
<th>Description</th>
<th>Units</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>Age</td>
<td>years</td>
</tr>
<tr>
<td>Aᵢ</td>
<td>Area</td>
<td>m²</td>
</tr>
<tr>
<td>B</td>
<td>Thickness</td>
<td>m</td>
</tr>
<tr>
<td>C</td>
<td>Constraint term</td>
<td>-</td>
</tr>
<tr>
<td>E</td>
<td>Young’s Modulus</td>
<td>Pa</td>
</tr>
<tr>
<td>F</td>
<td>Force</td>
<td>N</td>
</tr>
<tr>
<td>K</td>
<td>Stiffness matrix</td>
<td>-</td>
</tr>
<tr>
<td>Kₑ</td>
<td>Fracture toughness</td>
<td>Pam²</td>
</tr>
<tr>
<td>L</td>
<td>Coriolis term</td>
<td>-</td>
</tr>
<tr>
<td>M</td>
<td>Mass matrix</td>
<td>Kg</td>
</tr>
<tr>
<td>N</td>
<td>Number of Steps per Year</td>
<td>-</td>
</tr>
<tr>
<td>P</td>
<td>Pressure</td>
<td>Pa</td>
</tr>
<tr>
<td>Q</td>
<td>Generalized forces</td>
<td>N</td>
</tr>
<tr>
<td>R</td>
<td>Generalized forces</td>
<td>N</td>
</tr>
<tr>
<td>S</td>
<td>Number of Steps per Day</td>
<td>-</td>
</tr>
<tr>
<td>U</td>
<td>Nodal displacements</td>
<td>m</td>
</tr>
<tr>
<td>W</td>
<td>Width</td>
<td>m</td>
</tr>
<tr>
<td>d</td>
<td>Diameter</td>
<td>m</td>
</tr>
<tr>
<td>f</td>
<td>Nodal force</td>
<td>N</td>
</tr>
</tbody>
</table>
Generalized states

Time (s)

Centroid (m)

Greek Symbols

α Dundurs’ Parameter

β Dundurs’ Parameter

λ Lagrange multiplier

ρ Density (Kgm⁻³)

ψ Correction Factor

Latin Symbols

Ü Nodal acceleration (ms⁻²)

Co-ordinate Symbols

Fx Force in the X direction (Cartesian co-ordinate) (N)

Fy Force in the Y direction (Cartesian co-ordinate) (N)

Fz Force in the Z direction (Cartesian co-ordinate) (N)

Fxₙ Normalised force in the X direction

Fyₙ Normalised force in the Y direction

Fzₙ Normalised force in the Z direction

θ Hoop direction (Cylindrical co-ordinate)

R Radial direction (Cylindrical co-ordinate)

L Longitudinal direction (Cylindrical co-ordinate)

Mathematical symbols

° Degrees (Deg)

∑ Sum of

xxiv
## Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>AAOS</td>
<td>American Academy of Orthopaedic Surgeons</td>
</tr>
<tr>
<td>ANOVA</td>
<td>Analysis of Variance</td>
</tr>
<tr>
<td>BEL</td>
<td>Biomechanics European Laboratories</td>
</tr>
<tr>
<td>BIRM</td>
<td>Birmingham Hip Replacement</td>
</tr>
<tr>
<td>BW</td>
<td>Body Weight</td>
</tr>
<tr>
<td>CDM</td>
<td>Continuum Damage Mechanics</td>
</tr>
<tr>
<td>COF</td>
<td>Coefficient OF Friction</td>
</tr>
<tr>
<td>CT</td>
<td>Computed Tomography</td>
</tr>
<tr>
<td>DAE</td>
<td>Differential and Algebraic Equations</td>
</tr>
<tr>
<td>DEXA</td>
<td>Dual-Energy X-ray Absorptiometry</td>
</tr>
<tr>
<td>DVRT</td>
<td>Differential Variable Reluctance Transducer</td>
</tr>
<tr>
<td>FEM</td>
<td>Finite Element Modelling</td>
</tr>
<tr>
<td>GZ</td>
<td>Gruen Zones</td>
</tr>
<tr>
<td>LVDT</td>
<td>Linear Voltage Displacement Transducer</td>
</tr>
<tr>
<td>NL</td>
<td>Neck Length</td>
</tr>
<tr>
<td>NSA</td>
<td>Neck Shaft Angle</td>
</tr>
<tr>
<td>PCSA</td>
<td>Physiological Cross Sectional Area</td>
</tr>
<tr>
<td>PESS</td>
<td>Physical Education and Sports Science department</td>
</tr>
<tr>
<td>THR</td>
<td>Total Hip Replacement</td>
</tr>
<tr>
<td>TKR</td>
<td>Total Knee Replacement</td>
</tr>
<tr>
<td>VI</td>
<td>Virtual Instrument</td>
</tr>
</tbody>
</table>
CHAPTER 1

Introduction
Chapter 1  Introduction

1.1 Introduction
Total hip replacement (THR) surgery was one of the major surgical advances of the twentieth century, with an estimated occurrence of between 500,000 and 1 million operations per year (Charnley, 1979, Malchau et al., 1993, Dowson and Wright, 1981, Bennett et al., 2008a, Heisel et al., 2003, Konttinen et al., 1997). It is also predicted that in the next 10 years the demand for total hip and knee replacements is set to grow exponentially as people live longer and partake in a more active lifestyle. On the other side of the spectrum obesity and weight problems are also increasing and becoming a major issue (Watkins-Castillo, 2006). Although the surgery is a successful operation with good success rates reported over 90%, complications do occur and as the number of patients undergoing THR surgery increases the number of failed primary hip surgeries increases as well. There are several complications ranging from short term to long term. The most frequent complication for THR is long term loosening unrelated to infection and usually called “aseptic” or “mechanical” loosening. Aseptic loosening is a gradual process whereby the mechanical integrity of the implant-bone interface is lost, and a fibrous tissue is formed between the two surfaces (Eftekar et al., 1985, Goldring et al., 1983, Jasty et al., 1992, Radin et al., 1982, Schmalzried et al., 1992, Mow and Huiskes, 2005). In the beginning THR was used for elderly or older patients and therefore long term stability was not a substantial complication as many of these patients did not live for 15 or 20 years post surgery. But in today’s environment the age of patients is dropping and patients are living longer and therefore the long term stability has become a greater issue.

Figure 1-1 A diagram of before and after Total Hip Replacement surgery (www.adam.com)
Chapter 2 presents an in depth literature review of the total hip replacement field and the studies associated with this. Previous work on gait analysis and musculoskeletal modelling is investigated along with the survival rates of the different total hip replacement techniques. This chapter also looks at previous numerical modelling concepts that have been used to identify survival rates and possible loosening and identifies the experimental setups that are presently used for analysing hip prosthesis stability. The goals for this study are also presented at the end of the literature review.

The progression from motion data, recorded from motion analysis studies, to the calculation of muscle and joint forces is presented in chapter 3. The testing procedure used during the motion analysis and the parameters used during the musculoskeletal modelling in LifeMod are also discussed.

Chapter four discusses the numerical modelling approach used in this study including boundary conditions. Chapter 5 presents comparisons between the numerical modelling approach and published experimental data that used strain gauge measurements to examine the stresses at the stem-cement interface of a cemented hip prosthesis.

Chapter 6 describes the development of an experimental rig to investigate if the forces found in chapter 3 would lead to aseptic loosening and this can be seen in figure 1-2.

Using the parameters discussed in chapters 4 and 5, chapter 7 investigates the numerical modelling results of a reconstructed prosthesis implanted into a femur. Different loading conditions are investigated using the gait data collected in the motion analysis section of the study.

The final chapter of this study discusses the findings from the work carried out, concludes on the main objectives achieved and recommends future work that would lead on from this study.
Figure 1-2 The experimental rig designed, fabricated and commissioned to investigate different loading patterns from everyday activities.
CHAPTER 2

Literature Review
Chapter 2  Literature Review

2.1  Hip Joint Anatomy

2.1.1  Bones of Hip Joint

The hip joint is a ball and socket joint which joins the lower limbs to the trunk of the body. The hip joint is made up of two parts, the femur bone and the pelvis. The femur is a long bone consisting of the upper extremity, the body or shaft and the lower extremity. The upper extremity consists of the head, neck, greater trochanter and the lesser trochanter. These can be seen in Figure 2-1 below. The lower extremity is part of the knee joint and consists of two condyles, the lateral and medial condyles. The body or shaft of the bone is a long cylindrical part joining the upper extremity and the lower extremity.

![Figure 2-1 The upper and lower extremities of the femoral bone (Gray, 1967)](image)

The pelvis consists of four bones, the sacrum, coccyx and two hip bones. The hip bone consists of three sections bonded together on each side in the adult skeleton. These are the ilium, ischium and the pubis bones. The acetabulum cup is a cup shaped depression in the pelvis. It is where the head of the femur and the pelvis meet. This is surrounded by ligaments called the articular capsule. The articular capsule surrounds the joint and is made up of several ligaments including the iliofemoral ligament, the pubocapsular and the ischiocapsular ligaments.
2.1.2 Muscles of the hip joint

There are 26 muscles involved in movement of the hip joint. The muscles are listed below in Table 2-1 along with their origin and insertion area.
Table 2-1 Origins and insertions of each muscle associated with motion at the hip joint

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Origin</th>
<th>Insertion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Psoas Major</td>
<td>lumbar vertebrae (L1-5)</td>
<td>lesser trochanter</td>
</tr>
<tr>
<td>Psoas Minor</td>
<td>lumbar vertebrae (L1-5)</td>
<td>iliac fascia</td>
</tr>
<tr>
<td>Iliacus</td>
<td>inner surface of the Ilium</td>
<td>lesser trochanter</td>
</tr>
<tr>
<td>Gluteus Minimus</td>
<td>outer surface of the Ilium</td>
<td>anterior surface of the greater trochanter</td>
</tr>
<tr>
<td>Gluteus Medius</td>
<td>outer surface of the Ilium below crest</td>
<td>posterior and middle greater trochanter</td>
</tr>
<tr>
<td>Gluteus Maximus</td>
<td>crest of the Ilium and posterior surface of</td>
<td>gluteal line and the iliotibial band of</td>
</tr>
<tr>
<td></td>
<td>the sacrum</td>
<td>fasciae latae</td>
</tr>
<tr>
<td>Piriformis</td>
<td>sacrum</td>
<td>greater trochanter</td>
</tr>
<tr>
<td>Obturator Internus</td>
<td>posterior portions of the ischium</td>
<td>greater trochanter</td>
</tr>
<tr>
<td>Obturator Externus</td>
<td>posterior portions of the ischium</td>
<td>greater trochanter</td>
</tr>
<tr>
<td>Gemellus Superior</td>
<td>posterior portions of the ischium</td>
<td>greater trochanter</td>
</tr>
<tr>
<td>Gemellus Inferior</td>
<td>posterior portions of the ischium</td>
<td>greater trochanter</td>
</tr>
<tr>
<td>Tensor Fasciae Latae</td>
<td>anterior iliac crest and surface of the</td>
<td>iliotibial band of fascia on thigh</td>
</tr>
<tr>
<td></td>
<td>Ilium</td>
<td></td>
</tr>
<tr>
<td>Rectus Femoris</td>
<td>anterior inferior iliac spine of Ilium</td>
<td>patella tendon to tibial tuberosity</td>
</tr>
<tr>
<td>Vastus Lateralis</td>
<td>upper intertrochanteric line and the</td>
<td>lateral border of the patella</td>
</tr>
<tr>
<td></td>
<td>greater trochanter</td>
<td></td>
</tr>
<tr>
<td>Vastus Medialis</td>
<td>intertrochanteric line and the medial lip</td>
<td>medial border of the patella</td>
</tr>
<tr>
<td></td>
<td>of the linea aspera</td>
<td></td>
</tr>
<tr>
<td>Vastus Intermedius</td>
<td>front and lateral surfaces of the body of</td>
<td>base of the patella</td>
</tr>
<tr>
<td></td>
<td>the femur</td>
<td></td>
</tr>
<tr>
<td>Articularis Genu</td>
<td>anterior surface of the lower part of the</td>
<td>synovial membrane of the knee-joint</td>
</tr>
<tr>
<td></td>
<td>body of the femur</td>
<td></td>
</tr>
<tr>
<td>Biceps Femoris</td>
<td>tuberosity of ischium, linea aspera and</td>
<td>lateral condyle of tibia, head of fibula</td>
</tr>
<tr>
<td></td>
<td>lateral condyloid ridge</td>
<td></td>
</tr>
<tr>
<td>Semitendinosis</td>
<td>Tuberosity of ischium</td>
<td>upper anterior medial condyle of tibia</td>
</tr>
<tr>
<td>Semimembranosis</td>
<td>Tuberosity of ischium</td>
<td>posterior surface of medial condyle of tibia</td>
</tr>
<tr>
<td>Pectineus</td>
<td>front of pubis about the crest</td>
<td>lesser trochanter to the linea aspera</td>
</tr>
<tr>
<td>Adductor Brevis</td>
<td>pubis</td>
<td>lesser trochanter and upper linea aspera</td>
</tr>
<tr>
<td>Adductor Longus</td>
<td>front of pubis</td>
<td>middle third of the linea aspera</td>
</tr>
<tr>
<td>Adductor Magnus</td>
<td>ramus and ischium</td>
<td>linea aspera and inner condyloid ridge</td>
</tr>
<tr>
<td>Gracilis</td>
<td>ramus of pubis</td>
<td>anterior medial surface of tibia</td>
</tr>
<tr>
<td>Sartorius</td>
<td>anterior superior iliac spine</td>
<td>medial surface of the body of the tibia</td>
</tr>
</tbody>
</table>
2.1.3 Movements of the Hip Joint

The hip joint is a ball and socket joint and it has 6 different movements. These are extension, flexion, abduction, adduction, internal rotation and external rotation. These can be seen in the Figure 2-3.

Figure 2-3 The six movements of the hip joint a) flexion b) extension c) adduction d) abduction e) internal rotation f) external rotation. (Harrison)
2.2 Reasons for Hip Surgery

When conservative treatments for relief of pain and restoration of range of motion in the hip joint fail, total hip replacement may be considered. Total hip replacement is performed on patients of all ages but because the prosthetic interface can deteriorate over time it is generally advised to delay the procedure until it is absolutely necessary. There are a number of pathologies which can lead to total hip replacement and these are discussed below.

2.2.1 Osteoarthritis

The loss of cartilaginous surfaces in the joint known as osteoarthritis is one of the most common pathologies leading to total hip replacement. Osteoarthritis is caused by everyday activities and thus commonly referred to as ‘wear and tear’ arthritis. It causes pain, swelling, stiffness and loss of motion to the extent that everyday activities such as walking or stair climbing become increasingly difficult. When cartilage is lost subchondral bone can become exposed and damaged. In response to this the bone reacts by forming fibro-cartilage which is harder than the hyaline cartilage it is replacing. This fibro-cartilage can cause further damage to the joint surfaces and cause the patient considerable pain and also accelerate the progression of the problem.

2.2.2 Post Traumatic Arthritis

This is symptomatically almost identical to osteoarthritis. The main difference is the instigating factor. In the case of post-traumatic arthritis it is instigated by trauma or injury. The trauma is typically a fracture or dislocation of the hip. The trauma alters the mechanical behaviour of the hip and over time the joint breaks down. The break down in the joint is usually not seen until years after the injury.

2.2.3 Rheumatoid Arthritis

Rheumatoid arthritis is an auto-immune disease. An auto-immune disease is an illness which causes the body to attack its own tissues, in this case cartilage. It causes all the same symptoms as is seen with other arthritic diseases such as pain, joint swelling and stiffness. In some more chronic cases the destruction of the joints can be so severe that they become visibly deformed. Rheumatoid arthritis is a systemic disease meaning it attacks the whole body and so symptoms will also be visible in other joints in the body.

2.2.4 Avascular Necrosis

This is also known as aseptic necrosis or osteonecrosis. It occurs when the blood supply to the femoral head is interrupted and the bone dies. It affects a person without cause however fractures and dislocations around the hip joint have been shown to increase the risk of occurrence. Also, patients who have been on corticosteroids or have a history of alcohol abuse are at greater risk. One of the main difficulties is that x-rays do not show any signs of the
disease at the early stages but it can be very painful. At the later stages the symptoms can be as severe as osteoarthritis and are clearly visible on X-rays.

2.3 Total Hip Replacement Surgery Failures

2.3.1 Introduction

Total Hip Replacement surgery is generally a successful procedure with at least 90% of the patients living normal, pain free lives for at least 10 years after operation (Mow and Huiskes, 2005, Havelin et al., 1995, Malchau et al., 1993, Concensus Development Panel, 1982). However there are numerous reasons for complications which may require further surgery as just mentioned. The cost of surgery is also expensive and therefore revision surgery or any complications straight after surgery can cost money and time. O’Shea et al. (2002) found that in 1999 the cost of surgery was 6,472 Irish punts. However this number has risen substantially in recent years during the Celtic Tiger. O’Shea et al. (2002) also found that only 2,785 hip replacements were performed within the Republic of Ireland. This number has also risen substantially. In the US Kim (2008) reported that during 2004 approximately 225,900 primary hip replacement surgeries and 37,115 revision replacements, were performed in the US. This is a 37% increase since the year 2000. The biggest increase was seen in the 45-64 age category. The study also predicts that 600,000 hip replacements will be performed in the year 2015. The American Academy of Orthopaedic Surgeons (AAOS) also found similar figures of 234,000 primary hip replacements and 46,000 revision hip surgeries. Watkins-Castillo (2006) predicts that the demand for hip replacements will grow exponentially in the next 10 years.

There are many different types of hip replacements on the market today and there are several different approaches to hip surgery however the main comparison is between cemented and non- cemented hip replacements. These two groups were analysed for both primary and revision hip surgery.

2.3.2 Cemented Hip Replacements

Cemented hip replacements were the first replacements to be used. Results for these were successful with the Charnley hip replacement becoming the gold standard in the 80’s and 90’s. Aamodt et al. (2004) conducted a systematic review of literature involving primary total replacements in Norway from 1996 to 2000. They found that Charnley cemented prostheses were used in 45% of the cemented surgeries in this study and had a survival rate better than 90% at ten years. Thereafter the survival rate was seen to decline by 10% in each of the next two decades. Allami et al. (2006) looked at cemented Charnley hip replacements with a ten year follow up. Of the 1001 hips implanted and followed up 62 (6.2%) had revision surgery. The reasons for revisions varied: 33 were for aseptic loosening, 14 for infection, 9 for dislocation and 6 for other reasons. These results were lower than other studies of similar replacements.
over the same time period. Beckenbaugh and Ilstrup (1978) looked at 301 cemented Charnley hip replacements over a four to seven year follow up. 23 of the 301 hips were revised during this period. 255 hips underwent roentgenography at an average of 5 years 9 months. It was found from this that 61 (21%) showed signs of loosening. Follow ups on these subjects were then carried out at 10 years by Stauffer and Henderson (1982) and at 15 years by Kavanagh et al. (1989). In this 15 year study the prosthesis survival rate was seen to be 89%. A study by Schulte et al. (1993) carried out a follow up of a minimum of 20 years of these same cemented Charnley hip prostheses. 32 (10%) of the 322 hips known at 20 years had been revised. Twenty one of these were due to aseptic loosening, 8 were due to loosening due to infection and 3 due to dislocation. Of the 98 hips that were in people still alive at the time of the follow up, 15 had been revised. 3 from loosening due to infection, 11 due to aseptic loosening and 1 due to dislocation. The rate of revision because of aseptic loosening of the femoral component from the whole group was 2% (8 hips), and for the 98 hips in living people at the time of follow up it was 3% (3 hips).

Espehaug et al. (2002) looked at cemented Charnley hip replacements over a ten year follow up. 17323 hips were studied and at 10 years 9.2% of the hip replacements had been replaced. It was found that 71% of the revisions were due to aseptic loosening of the femoral component. Berry et al. (2004) investigated the results of 6623 cemented Charnley prostheses at regular intervals. 1% dislocated at 1 month, 1.9% after 1 year and then 1% every 5 years thereafter, to 7% after 25 years. This study also found that females were 2.1 times more likely to dislocate than males. Mullins et al. (2007) carried out a 30 year follow up on 228 cemented Charley hip replacements. The survival rates were 93% at 10 years, 89.5% at 15 years, 84.1% at 20 years, 77.4% at 25 years and 73% at 30 years. 28 hips (12%) had been revised at the follow up and the main complication was aseptic loosening.

A study by Madey et al. (1997, 1997) looked at 356 cemented Charnley hip implants over an average 15 year period. In the 15 years 33 implants (9%) had to be revised. The most common reason was aseptic loosening with 19 hips (5%) and 6 (2%) of these were due to loosening of the femoral component. Loosening due to infection 7 (2%) and dislocation 7 (2%) also occurred. At the mean follow up of 15 years 142 hips were still in patients that were alive. 22 (15%) had been revised, 15 (11%) for aseptic loosening, with 5 (3%) of these due to loosening of the femoral component. 4 (3%) were due to dislocation and 3 (2%) for loosening due to infection. Radiography was done on 116 hips at 15 years and it was seen that 6 (5%) showed definite or probable loosening of the femoral part and 26 (22%) showed definite or probable loosening of the acetabular cup. Klapach et al. (2001) carried on from Madey et al. (1997) to look at a minimum 20 year period. Of the 356 cemented Charnley hip replacements 39 (11%) had been revised, 22 (6.2%) for aseptic loosening, 6 (1.8%) of these due to failure of the
femoral component, 7 (2%) for loosening due to infection and 10 (2.8%) for dislocation. Of the 91 hips in patients still alive 21 (23%) had been revised, 15 (16%) due to aseptic loosening and 5 (5%) of these due to femoral component loosening. 5 (5%) were revised for dislocation and 1 (1%) was revised for loosening due to infection. Buckwalter et al. (2006) carried out a further study on this same group at a 25 year follow up. Table 2-2 shows the revision rates for aseptic loosening of both the acetabular and femoral components. The survival rate for revision for any reason as an end point was found to be 80% (±7%), survival rate of revision due to aseptic loosening as an end point was found to be 86% (±7%), survival rate of revision due to aseptic loosening of the acetabular component as an end point was found to be 91% (±5%), survival rate of revision due to aseptic loosening of the femoral component as an end point was found to be 93% (±6%), survival rate of aseptic loosening of the acetabular component on a radiograph as an end point was found to be 61% (±3%), and the survival rate of aseptic loosening of the femoral component on a radiograph as an end point was found to be 80% (±10%).

Table 2-2 Revision rates due to aseptic loosening from Buckwalter et al. (2006) study

<table>
<thead>
<tr>
<th>Component</th>
<th>Surgery</th>
<th>All (356)</th>
<th>Alive (52)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Femoral</td>
<td>Revision</td>
<td>10 (2.9%)</td>
<td>5 (11%)</td>
</tr>
<tr>
<td></td>
<td>Radiograph</td>
<td>13 (3.8%)</td>
<td>3 (6%)</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td></td>
<td><strong>23 (6.8%)</strong></td>
<td><strong>8 (17%)</strong></td>
</tr>
<tr>
<td>Acetabular</td>
<td>Revision</td>
<td>18 (5.3%)</td>
<td>12 (26%)</td>
</tr>
<tr>
<td></td>
<td>Radiograph</td>
<td>27 (8%)</td>
<td>5 (11%)</td>
</tr>
<tr>
<td><strong>Total</strong></td>
<td></td>
<td><strong>45 (13.3%)</strong></td>
<td><strong>17 (36%)</strong></td>
</tr>
</tbody>
</table>

Other cemented hip replacements investigated were the Exeter prosthesis by Williams et al. (2002a, 2002b) who looked at the survival rate of 325 cemented Exeter universal hip prostheses at a 12 year follow up. The study showed good survivorship rates with only 8.26% in need of a revision surgery after 12 years. It also showed good survival rates for aseptic loosening of the femoral component with no revisions due to this type of loosening. Bourne et al. (1998) studied cemented Harris design 2 hip prostheses. 195 replacements were included in the study and had a mean follow up of 12 years. 5 (3%) of the patients had revision surgery all for aseptic loosening. The femoral component was found to be loose in all 5 cases with the acetabular cup loose in 2 cases. Other complications were dislocation 5 (3%) heterotopic bone formation 14 and others 3. 131 patients underwent radiography and it was seen that 9% showed signs of aseptic loosening in the acetabular cup with 2% aseptic loosening in the femoral component. Skutek et al. (2007) carried on from Bourne et al. (1998) and looked at the same 195 cemented Harris design 2 hip prostheses over a 20 to 25 year period. The 25 year survival rate for this study with revision as an end point was 83%. For aseptic loosening of the femoral component as the end point the survival rate was 86% and for aseptic loosening of the acetabular cup 93%. Datir et al. (2006) carried out a study of 269 cemented Harvard femoral stems with a mean follow up of ten years.
36 (13.3%) were revised due to aseptic loosening. The femoral component was revised in all of these hips. 11 (4%) were revised for loosening due to infection. 10 year survival rate for the Harvard stem was found to be 82.1%, the survival rate of the femoral component using aseptic loosening as an end point was 77.5% and for the acetabular component 91.1%.

The surface finish can have an effect on the survival rates of cemented prostheses as Field et al. (2006) showed in a 5 year follow up study on a tri tapered polished, cemented femoral stem with 54 hip replacements carried out. This study showed excellent results with 1 revision for dislocation and one due to infection. No revisions were carried out due to aseptic loosening nor were there any signs of the femoral component loose in the radiographs carried out at 5 years. The mean migration of the stem at 5 years was 1.9mm. Firestone et al. (2007) also looked at the difference between the surface finish of cemented prostheses to see if they reduce failure. Polished, matte finish and grit blasted, cemented hip replacements were the surface finishes tested and 149 cemented hips were followed for 10 years. None of the polished stems were revised due to aseptic loosening nor did any demonstrate radiographic loosening. 6 (2%) of the matte finish stems were revised due to aseptic loosening and a further 5 were radiographically loose. 11 (9.2%) of the grit blasted stems were revised due to aseptic loosening and another 4 were seen to be radiographically loose. Therefore out of the 149 hips, 11.4% were revised due to aseptic loosening and another 6% were loose radiographically.

To examine the failure rates from all of the cemented primary hip replacements studies in literature Table 2-3 was produced. This table was produced by finding the average yearly failure rate from each study and then from this finding the mean of these average values. This data was then used to predict the failure rates at 5 year intervals based on the literature. The same criteria has been used for tables 2-4, 2-5 and 2-6.

Table 2-3 Average failure results for primary cemented hip replacements at 5 year intervals

<table>
<thead>
<tr>
<th>Year</th>
<th>1</th>
<th>5</th>
<th>10</th>
<th>15</th>
<th>20</th>
<th>25</th>
<th>30</th>
</tr>
</thead>
<tbody>
<tr>
<td>Failure (%)</td>
<td>0.876</td>
<td>4.38</td>
<td>8.76</td>
<td>13.14</td>
<td>17.52</td>
<td>21.9</td>
<td>26.28</td>
</tr>
<tr>
<td>Aseptic Loosening (%)</td>
<td>0.539</td>
<td>2.695</td>
<td>5.39</td>
<td>8.085</td>
<td>10.78</td>
<td>13.475</td>
<td>16.17</td>
</tr>
<tr>
<td>Dislocation (%)</td>
<td>0.2167</td>
<td>1.0835</td>
<td>2.167</td>
<td>3.2505</td>
<td>4.334</td>
<td>5.4175</td>
<td>6.501</td>
</tr>
<tr>
<td>Infection (%)</td>
<td>0.265</td>
<td>1.325</td>
<td>2.65</td>
<td>3.975</td>
<td>5.3</td>
<td>6.625</td>
<td>7.95</td>
</tr>
<tr>
<td>Radiographically Loose (%)</td>
<td>0.167</td>
<td>0.835</td>
<td>1.67</td>
<td>2.505</td>
<td>3.34</td>
<td>4.175</td>
<td>5.01</td>
</tr>
</tbody>
</table>

2.3.3 Cementless Hip Replacements

Most of the results from studies of cementless prostheses show better long term results then the cemented prostheses. However, as the use of cementless replacements is more recent then the cemented prostheses the studies for cementless hip replacements have shorter follow ups and the performance of 20 years plus is unknown. Pieringer et al. (2006) carried out a follow up of 124 cementless ALLOCLASSIC hip arthroplasty systems at an average of 12 and a half years.
The survival rate at the time of follow up was found to be 95.6%. With aseptic loosening as the end point, survival rate was 98.4% and all of these were seen to be loosening of the acetabular cup and not the stem part. Archibeck et al. (2006) reported mean 9 year results of 100 second generation cementless prostheses in patients 50 years or younger (mean age 39). The study reported survival rates at 10 years as 87.5%, with aseptic loosening as the end point survival rates were 97.1% at 10 years. No femoral components were revised for loosening and none were seen to be radiographically loose. Froimson et al. (2007) reported a mean 11.5 year follow up study of a tapered, titanium hydroxyapatite coated cementless femoral hip prosthesis. 15 (15.6%) of the 96 hips in patients that were still alive at follow up had to be revised due to problems with the acetabular cup. 2 femoral implants had to be removed during the 11.5 years however none of these were due to aseptic loosening.

Chang et al. (2006) used the Omnifit hydroxyapatite coated prosthesis in 90 hips. A follow up of at least 7 years was carried out and the survival rate was found to be 85.7%. 8 acetabular cups had to be treated for aseptic loosening, while all femoral components were seen to be stable. Surdam et al. (2007) carried out a mean 9 years follow up study of second generation cementless hip prostheses. The ten year survival rate was found to be 92%. With aseptic loosening as the end point, survival rates were 98% for the femoral component and 99% for the acetabular component. Cameron et al. (2006) investigated cementless SROM stems implanted in 795 hips with a mean follow up of 11 years. The rate of aseptic loosening in the femoral component was found to be low at 0.25%. Gruen et al. (1979) investigated 471 uncemented hips and found the revision rate for these stems to be around 6%. They found uncemented femoral components to have a survival rate of between 96 and 98% at ten years. Lettich et al. (2007) looked at 700 uncemented hips at a follow up average of 4.35 years. 4 femoral components were revised due to aseptic loosening giving a survivorship rate of 99.4% with aseptic loosening as the end point.

The longest follow up for cementless studies found was Eskelinen et al. (2006) who investigated uncemented prostheses implanted in Finland between 1980 and 2003 and 100 surgeries were found to be suitable. The stem components were found to have a survival rate of over 90% at 10 years, with the Biomet Bi-Metric stem found to have a 95% survival rate at 15 years. Khalily and Whiteside (1998) carried out a follow up at between 8-12 years after surgery on 100 cementless prostheses. 24 (24%) had been revised due to aseptic loosening. They also found that radiographs at 2 years were a good indicator of hips that may loosen.


2.3.4 Revision Surgery

Revision surgery is carried out for primary or other secondary surgeries that ended in failure. Dobzyniak et al. (2006) investigated 745 subjects who underwent revision surgery to see what the cause for surgery was. 291 (39%) of the revisions were within the first 5 years after the primary THR surgery. 96 (33%) had revision surgery due to instability, 88 (30%) due to aseptic loosening, 41 (14%) due to infection, 14 (5%) for osteolysis and 52 (18%) for other reasons.

2.3.4.1 Revised Cemented Hip Replacements

The Charnley cemented hip replacement is also used for revision surgery however as revision subjects have already had one or more failed hip replacements the rate of revision and failure is a lot higher than with primary replacements. Echeverri et al. (1988) investigated 127 cemented Charnley low friction arthroplasties carried out on patients that required revision surgery. At a follow up of on average 10.4 years, 26 (20.5%) required revision surgery. 15 (11.8%) were due to deep infection, 4 (3.1%) for mechanical loosening and the others were due to other complications. Mulroy and Harris (1996) carried out a follow up of an average 15.1 years after 43 cemented revision surgeries were carried out. This continued the work from Rubash and Harris (1987) that carried out follow up studies at 6 years and Estock and Harris (1994) at 11.7 years. The 6 year follow up had shown that 4 of the 43 had to undergo revision surgery due to aseptic loosening. After 11.7 years another 3 prostheses had been revised for aseptic loosening. Mulroy and Harris (1996) found that after 15 years 7 (16%) prostheses were revised and 3 (7%) were radiographically loose, illustrating that 23% of the prostheses had become loose. The overall rate of revision for any complication was 43%.

Table 2-5 Average failure results of the revised cemented hip replacements at 5 year intervals

<table>
<thead>
<tr>
<th>Year</th>
<th>1</th>
<th>5</th>
<th>10</th>
<th>15</th>
<th>20</th>
<th>25</th>
<th>30</th>
</tr>
</thead>
<tbody>
<tr>
<td>Failure (%)</td>
<td>2.415</td>
<td>12.075</td>
<td>24.15</td>
<td>36.225</td>
<td>48.3</td>
<td>60.375</td>
<td>72.45</td>
</tr>
<tr>
<td>Aseptic Loosening (%)</td>
<td>0.679</td>
<td>3.395</td>
<td>6.79</td>
<td>10.185</td>
<td>13.58</td>
<td>16.975</td>
<td>20.37</td>
</tr>
<tr>
<td>Dislocation (%)</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Infection (%)</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Radiographically Loose (%)</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>
2.3.4.2 Revised Cementless Hip Replacements

Cementless replacements do not use cement to fill the gaps at the interface and revision cementless replacements already have a cavity in the bone from the previous replacement, therefore the prosthesis would have to have a larger stem than was used in the primary surgery to make sure the prosthesis was press fit into the cavity and that a close stem-bone interface was present. Kang et al. (2008) investigated the 39 cementless modular femoral components used in revision surgery with a 2 to 5 year follow up. 4 hips required revision one of these was due to the femoral component. Krupp et al. (2006) carried out a minimum 6.5 year follow up on 48 femoral component revisions using impaction grafting on subjects with significant bone loss. 46% of patients had complications with 21% of them requiring revision surgery. The main complications were fractures with 14%, dislocation with 13%, and infection with 8% and aseptic loosening was at 4%. Koster et al. (2008) evaluated 73 modular uncemented revision stems at an average of 10 years. The survival rate with aseptic loosening as the end point was 96% at 10 years. The overall survival rate at 10 years was 94%. Weiss et al. (2011) evaluated a cementless modular tapered stem in revision surgery for a minimum 5 year follow up. A 5 year survival rate was 98% with stem removal and 90% with any reoperation as the end point. Radiography was carried out on 68 subjects and the median stem migration was found to be 2.7mm at the 5 years.

Table 2-6 Average failure results for revised cementless hip replacements at 5 year intervals

<table>
<thead>
<tr>
<th>Year</th>
<th>1</th>
<th>5</th>
<th>10</th>
<th>15</th>
<th>20</th>
<th>25</th>
<th>30</th>
</tr>
</thead>
<tbody>
<tr>
<td>Failure (%)</td>
<td>1.97</td>
<td>9.85</td>
<td>19.7</td>
<td>29.55</td>
<td>39.4</td>
<td>49.25</td>
<td>59.1</td>
</tr>
<tr>
<td>Aseptic Loosening (%)</td>
<td>0.479</td>
<td>2.395</td>
<td>4.79</td>
<td>7.185</td>
<td>9.58</td>
<td>11.975</td>
<td>14.37</td>
</tr>
<tr>
<td>Dislocation (%)</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Infection (%)</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Radiographically Loose (%)</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

Figure 2-4 and Figure 2-5 show the primary and revision rates for the cemented and cementless hip replacements. These percentages are based on the results from table 2-3 to table 2-6. Although cemented prostheses have been the most common prostheses used for many years, the UK joint registry (National Joint Registry, 2011) found that within the last two years cementless prostheses are been used as much as the cemented prostheses. Therefore, the graphs use 50% of the cemented results and 50% of the cementless results in both cases. It can be seen that failure of the replacements is a big issue for subjects that have the prosthesis for many years and aseptic loosening plays a major role.
2.4 Gait

2.4.1 Introduction

Walking or any other motion may seem like a simple movement from one point to another; however this is not the case. There are many muscles and joints that play a part in any motion.
and these were discussed in the anatomy of the hip joint section previously. One of the aims of THR surgery is to return the gait of THR subjects to that of normal subjects. However, it has been reported that this is not the case and even young subjects gait does not return to that of the normal subjects of the same age group (Bennett et al., 2008b, Bennett et al., 2009). For this reason it is important to assess how gait is performed, what parts of the body are involved with gait and most importantly what are the determinants of gait.

2.4.2 Determinants of Gait

In 1953 Saunders et al. (1953) investigated what happens during the gait cycle. They came up with six determinants of gait which have been used in many studies to date. The main aim of the determinants is to reduce the movement of the centre of gravity which in turn reduces the energy cost of moving.

The first determinant of gait is pelvic rotation. Pelvic rotation is where the pelvis rotates alternately to the left and to the right, relative to the line of progression. Saunders et al. (1953) found the average rotation to be around 4°, which is in line with more recent studies like Crosbie and Vachalathiti (1997) who found healthy subjects to have a rotation of between 4.3° and 5.6°, Kerrigan et al. (2001b) who found rotation of 4.3° in healthy subjects and Stokes et al. (1989) who found a rotation of between 3.9° and 8.3°.

The second determinant of gait is pelvic tilt (also known as pelvic list). This is where the pelvis tilts downward on the leg that is in swing phase. An average of 5° angular displacement was found by Saunders (1953). This was similar to studies by Crosbie and Vachalathiti (1997) who found pelvic tilt of between 4.5° and 6.5°, and Gard and Childress (1997) who found an average angular displacement of between 5.5° and 6.5° for healthy subjects.

The third determinant of gait was seen to be knee flexion in the stance phase. The walking cycle starts at heel strike, at this time the leg is straight with no knee flexion. As the subject progresses through the stance phase the knee starts to flex and reaches an average around 15° of flexion (Gard and Childress, 1999).

The fourth and fifth determinants of gait are closely linked. They are the foot mechanics and the knee mechanics. The foot mechanics play an important part in the gait cycle. At heel strike the ankle is elevated but then falls flat on the ground as contact is made. As the body moves forward the heel starts to rise and the front of the foot pushes off. The knee mechanics play a very similar role and works with the foot to reduce energy. Generally when either the ankle or the front of the foot are off the ground during stance phase the knee joint is at full extension.
The sixth determinant of gait is the lateral displacement of the pelvis. There would be a large amount of lateral displacement of the pelvis without the existence of the tibiofemoral angle. This is where the knee joint bends inwards slightly and this reduced the lateral displacement.

All 6 determinants have been shown to play their part in reducing the energy cost of movement mainly by reducing the displacement of the centre of gravity. However the determinants do play other roles as well. Gard and Childress (1999) found that knee flexion only reduces the vertical displacement minimally as its main role was shown to be a shock absorber during the motion. Della Croce et al. (2001) investigated which determinant played the main role in reducing energy costs. They found that vertical displacement reduction was important to reduced energy costs and that the heel rise part of the foot mechanics was the main determinant for reducing the vertical displacement, as this was responsible for up to two thirds of the total reduction. This was in line with Kerrigan et al. (2000) who found the heel rise to play the most important role. Pelvic rotation was seen to account for around 10% of the reduction in vertical displacement.

Many different comparisons have been made between healthy subjects and other groups. Kerrigan et al. (2001a) found a reduction in hip extension between healthy subjects and elderly subjects. Kerrigan et al. (2003) found a modest improvement in a group of stretchers compared to a group of non stretchers. In 2010 Kiyama et al. (2010) found little difference in gait determinants between subjects that had undergone lateral surgery and post-lateral surgery after a minimum of two years post op.

2.4.3 Frequency of Everyday Motions

As this study investigates the everyday activities that are carried out by people, it is important to understand what motions are carried out during a day and how often these motions are carried out. Morlock et al. (2001) carried out a study investigating the movements that were carried out in 31 THR patients for 6 hours or more. The study found that the most common actions were sitting and standing. This accounted for nearly 70% of the time. Table 2-7 below shows a list of actions carried out in everyday activities and the percentage of time each action occupies during the day. Morlock et al. (2001) found that an average of 6,048 steps and 164 steps up stairs were carried out each day. This is a stairs step every 38 steps. Morlock et al. (2001) also predicted from their data an average yearly steps of 1.1 million steps per year. However Morlock et al. (2001) did state that this was higher than previous studies and that the subjects lived an active life.
Table 2-7 Percentage time everyday activities are carried out (Morlock et al., 2001)

<table>
<thead>
<tr>
<th>Action Carried Out</th>
<th>% Time</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sitting</td>
<td>44.3</td>
</tr>
<tr>
<td>Standing</td>
<td>24.5</td>
</tr>
<tr>
<td>Walking</td>
<td>10.2</td>
</tr>
<tr>
<td>Lying</td>
<td>5.8</td>
</tr>
<tr>
<td>Stair Climbing</td>
<td>0.4</td>
</tr>
</tbody>
</table>

2.4.4 Steps per Year

Several other studies have looked at the amount of steps per year. Wallbridge and Dowson (1982) studied 11 THR subjects. An equation for the amount of steps per day from their subject’s data was:

\[ S = 18000 - 175.6A \]  
(2.1)

The number of steps per year was also predicted in this study using the following equation:

\[ N = [3.29 - 0.03A]10^6 \]  
(2.2)

Where \( S \) is steps per day, \( A \) is the age of the subject and \( N \) is the number of loading cycles per year on each leg. Sequeira et al. (1995) tested 493 healthy subjects aged between 25 and 74. This study found an average of 10,400 steps per day for men and 8,900 steps per day for women. Goldsmith et al. (2001) compared his study of 63 THR subjects and 27 healthy subjects to the Wallbridge and Dowson (1982), and the Sequeira et al. (1995) studies and found new equations for both healthy and THR subjects. These are as follows.

For the healthy subjects:

\[ S = 14817 - 134.38A \]  
(2.3)

\[ N = [2.704 - 0.025A]10^6 \]  
(2.4)

For the THR subjects:

\[ S = 12863 - 89.4A \]  
(2.5)

\[ N = [2.35 - 0.016A]10^6 \]  
(2.6)

Schmalzried et al. (1998) and Zahiri et al. (1998) both tested the same population of subjects. Schmalzried et al. (1998) used 111 subjects, 97 of whom had THR with the remaining having Total Knee Replacement (TKR) and Zahiri et al. (1998) used 100 of these subjects in their study. Both studies showed results that were lower than the previous studies mentioned with an
average of 4,988, and 5,078 steps per day respectively. This represents just over 0.9 million steps per year. However a study by Storti et al. (2009) also found lower steps per day then the studies by Goldsmith et al. (2001), Wallbridge and Dowson (1982), and Sequeira et al. (1995). Stori et al. (2009) studied 2,604 subjects from ages 18 to 91. They found an average of 4,874 steps per day.

Wagenmakers et al. (2008) compared 273 THR subjects to 273 healthy subjects to investigate if there was any significant difference in physical activity between the two groups. The study found that the THR subjects were at least as active as the healthy subjects. However this paper did also state that only 51.2% of the THR subjects met the guidelines, and that the THR subjects need to be stimulated to become more physically active.

Most of the studies mentioned so far have investigated subjects over one day. Silva et al. (2005) investigated if just looking at the results of one days testing would under or over estimate the results if used to calculate several days. Silva et al. (2005) took 131 THR or TKR subjects and tested subjects for 4 days and 7 days. The average daily step count was 5,737 steps per day for the 4 day trial, and 5,464 steps per day for the 7 day trial. Silva et al. (2005) concluded that the 7 day trial gave a more representable daily average. However it was seen that with a large group the 4 day results found a daily average within 5% of the 7 day average. Another study that used a 7 day testing period was Ewald et al. (2009). They tested 684 subjects all over the age of 55 years old. The average for the 7 day trial was 7,698 steps per day. Ewald et al. (2009) also broke this down into age groups and found that for the 55-59 year olds the daily step rate average over the 7 days was 8,605. However in the over 80’s this dramatically drops to only 3,778 steps.

Other studies on healthy subjects were carried out by Wyatt et al. (2005), McCormack et al. (2006) and Dwyer et al. (2007). McCormack et al. (2006) also carried out a 7 day investigation of healthy subjects. They tested 428 subjects above the age of 18. The averages found in this study were quite a bit higher then seen in most studies to date with average daily steps of 10,079 for men and 9,169 for women. Dwyer et al. (2007) tested subjects over a two day period, both at the weekend. 1,126 subjects above the age of 25 were tested by Dwyer et al. (2007). These results were similar to McCormack et al. (2006) as they were higher than most other studies. They also found that the average daily steps for women were higher than that of men with 11,200 steps for women and 10,900 for men. Wyatt et al. (2005) tested 730 healthy subjects over a 4 day period. Their age group of 50-59 years old gave results of an average of 6,557 steps per day which is lower than the findings in Ewald et al. (2009). For subjects over 60 years, the average daily steps fell to 5,022 steps. A study that tested for a longer period was Harris et al. (2008) who tested 238 subjects over a 7 day period, all of whom were over 65 years of age. They found an average of 6,443 steps per day. Berlin et al. (2006) published data that compared the use of pedometers and accelerometers. They reviewed previous studies and found that young
healthy adults had an average of 7,000 to 13,000 steps per day, while older adults had an average of between 6,000 and 8,500 steps per day. Kinkel et al. (2009) analysed 105 THR subjects for an average of 9 days. An average of 6,144 steps per day was found, with the amount of steps decreasing with age. This ranged from 6,722 steps per day to 4,669 steps per day.

Huch et al. (2005) carried out an interesting study comparing 420 THR subjects and 389 TKR subjects. Of the 420 THR subjects 97% had performed sport during their lifetime, and 36% of them played sport up to the time of the surgery. After 5 years post operation 52% of the 420 THR subjects were playing sport. For the 389 TKR subjects 94% had performed sport during their lifetime, and 42% at the time of surgery. However after 5 years post operation only 34% of the 389 subjects were still participating in sport. This demonstrates that both THR and TKR subjects show a reduction in activity due to requiring surgery, however, THR subjects participation in sport increased 5 years post operation while TKR subjects further decreased.

In a paper investigating what loading configuration should be used for experimental modelling, Bergmann et al. (2010) looked into the fatigue of normal patient cycles and active patient cycles. 10 million cycles were used for both groups. This represented 7.3 years for the normal patients or 1.3693 million cycles per year, and 3.9 years for the active patients or 2.5534 million cycles per year.

### Table 2-8 The average number of steps per day from the literature for Healthy and THR subjects

<table>
<thead>
<tr>
<th>Study</th>
<th>Healthy</th>
<th>THR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Wallbridge and Dowson (1982)</td>
<td>-</td>
<td>18,000-175.6A</td>
</tr>
<tr>
<td>Sequeria et al. (1995) Men</td>
<td>10,400</td>
<td>-</td>
</tr>
<tr>
<td>Sequeria et al. (1995) Women</td>
<td>8,900</td>
<td>-</td>
</tr>
<tr>
<td>Goldsmith et al. (2001)</td>
<td>14,817-134.38A</td>
<td>12,863-89.4A</td>
</tr>
<tr>
<td>Schmalzried et al. (1998)</td>
<td>-</td>
<td>4,988</td>
</tr>
<tr>
<td>Stori et al. (2009)</td>
<td>4,874</td>
<td>-</td>
</tr>
<tr>
<td>Zahiri et al. (1998)</td>
<td>-</td>
<td>5,078</td>
</tr>
<tr>
<td>Silva et al. (2005)</td>
<td>-</td>
<td>5464</td>
</tr>
<tr>
<td>Ewald et al. (2009)</td>
<td>7,698</td>
<td>-</td>
</tr>
<tr>
<td>McCormack et al. (2006) Men</td>
<td>10,079</td>
<td>-</td>
</tr>
<tr>
<td>McCormack et al. (2006) Women</td>
<td>9,169</td>
<td>-</td>
</tr>
<tr>
<td>Dwyer et al. (2007) Men</td>
<td>10,900</td>
<td>-</td>
</tr>
<tr>
<td>Dwyer et al. (2007) Women</td>
<td>11,200</td>
<td>-</td>
</tr>
<tr>
<td>Kinkel et al. (2009)</td>
<td>-</td>
<td>6,144</td>
</tr>
</tbody>
</table>

### 2.5 Instrumented Prostheses

Instrumented prostheses have been in use since Rydell (1966b, 1966a) who used strain gauges on an implanted hip prosthesis to measure the forces at the replaced hip joint. These studies were followed up by Haggstrom (1974), English (1977), English and Kilvington (1979) and Davy et al. (1988) who investigated the forces using similar techniques. However most of these
studies report results shortly after implantation and therefore may not show a true indication into the forces when the hip joint has recovered. This is generally around 6 months post operatively.

Bergmann et al. (1993) published results of two patients over a 18 month and 30 month period. The study showed results for walking and running for these patients. The study found forces ranging from 2.8 times BW (Body Weight) to 4.8 times BW for walking at different speeds. The highest force found was 8.7 times BW and this was during stumbling. The devices used in this study were described in Bergmann et al. (1990), and had previously been tested in sheep implants (Bergmann et al., 1984, Bergmann et al., 1988). Since then Bergmann and his group in Berlin, Germany have led the way in this field. The newer devices used by this group were described in Graichen et al. (1999), Bergmann et al. (2008) and Damm et al. (2010).

Bergmann et al. (1995a) investigated the hip joint forces during stair climbing. They found the forces at the hip joint in two patients. While going upstairs at normal speed the hip joint force was seen to be 10% higher than walking at normal speed, while going downstairs increased the force by 20%. This study also found that the torsional moments in some of the movements were close to or exceeded torsional strengths of implants; indicating this may lead to problems with fixation and loosening. A second paper in 1995 by Bergmann et al. (1995b) investigated the influence of shoes on the hip joint forces. They found torsional moments during several everyday activities were in a critical range for loosening of cementless hip implants which Phillips et al. (1991) reported to be in the range of 19 to 52 µm with an average moment of 33 Nm.

Bergmann et al. (2001) increased their subject numbers to 4 patients. 2.38 times BW was the average hip joint force for walking, 2.51 times BW the average hip joint force for up steps and 2.6 times BW for down steps. This paper also reported large torsional moments. This study also contained a CD ‘HIP98’ that included all the numerical data collected during this study. Stumbling was reported by several publications from the Bergmann group to create large forces higher than all the other moments. Bergmann et al. (2004) investigated this and found that this had a peak force of 8.7 times BW. This study also found that the direction of peak contact forces were constant for all the activities tested.
Table 2-9 Summary of the peak normalised forces from instrumented prostheses studies

<table>
<thead>
<tr>
<th>Study</th>
<th>Motion</th>
<th>Peak Normalised Force</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bergmann et al. (1993)</td>
<td>Walking</td>
<td>2.4 to 4.8</td>
</tr>
<tr>
<td></td>
<td>Stumbling</td>
<td>8.7</td>
</tr>
<tr>
<td>Bergmann et al. (1995a)</td>
<td>Walking</td>
<td>3.07 to 4.09</td>
</tr>
<tr>
<td></td>
<td>Jogging</td>
<td>4.91 to 4.96</td>
</tr>
<tr>
<td></td>
<td>Up Steps</td>
<td>3.45 to 5.52</td>
</tr>
<tr>
<td></td>
<td>Down Steps</td>
<td>3.9 to 5.09</td>
</tr>
<tr>
<td>Bergmann et al. (1995b)</td>
<td>Walking</td>
<td>2.89</td>
</tr>
<tr>
<td></td>
<td>Jogging</td>
<td>4.72</td>
</tr>
<tr>
<td>Bergmann et al. (2001)</td>
<td>Walking</td>
<td>2.38 to 2.42</td>
</tr>
<tr>
<td></td>
<td>Jogging</td>
<td>2.5</td>
</tr>
<tr>
<td></td>
<td>Up Steps</td>
<td>2.51</td>
</tr>
<tr>
<td></td>
<td>Down Steps</td>
<td>2.6</td>
</tr>
<tr>
<td></td>
<td>Standing</td>
<td>1.9</td>
</tr>
<tr>
<td></td>
<td>Sitting</td>
<td>1.56</td>
</tr>
</tbody>
</table>

Bergmann et al. (2010) investigated if the ISO standards (ISO 7206-4, 2010, ISO 7206-8, 1995) simulated the real loads by comparing these to loads from realistic data obtained in previous studies. The average loads from all the 4 patients can be seen in Table 2-10. They found that the forces of the ISO standards were above the average peak loads, but were lower than the high peak loads found.

Table 2-10 Average and high peak loads from Bergmann et al. (2010). X is the lateral medial direction, Y is the anterior posterior direction and Z is the proximal distal direction.

<table>
<thead>
<tr>
<th>Activity</th>
<th>Average Peak loads</th>
<th>High Peak Loads</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>F (N)</td>
<td>Fx (N)</td>
</tr>
<tr>
<td>Walking</td>
<td>1800</td>
<td>403</td>
</tr>
<tr>
<td>Going Up Steps</td>
<td>1900</td>
<td>446</td>
</tr>
<tr>
<td>Going Down Steps</td>
<td>2000</td>
<td>370</td>
</tr>
<tr>
<td>Standing</td>
<td>1500</td>
<td>420</td>
</tr>
<tr>
<td>Sitting</td>
<td>1200</td>
<td>323</td>
</tr>
<tr>
<td>Standing on 1 Leg</td>
<td>1800</td>
<td>203</td>
</tr>
<tr>
<td>Knee Bend</td>
<td>1100</td>
<td>290</td>
</tr>
<tr>
<td>Stumbling</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

2.6 Musculoskeletal Modelling

As seen in the previous section contact forces at the hip joint can be measured using implanted measurement prostheses. However, such prostheses tend to be very expensive and the ethical approval needed to carry out implantation is rigorous. Musculoskeletal modelling is an alternative to this, which is both cheaper and of lower risk to the subject. Musculoskeletal modelling uses either computational or mathematical models to predict the forces throughout the body. The testing procedure and parameters used in this study are discussed in Chapter 3.
Mathematical models have been carried out for healthy hip joints since the 70’s. Crowninshield et al. (1978) found hip forces of between 3.3 and 5 times body weight (BW), with the average over 10 cycles to be 4.3 times BW. Crowninshield and Brand (1981) found values between 3 and 3.5 times BW. Rohrle et al. (1984) also looked at healthy subjects and found forces ranging from 2.9 times BW to 6.7 times BW. Brand et al. (1986) tested one male and one female subject and found values of 5.2 times BW and 5 times BW respectively which is at the higher end of the values found by other studies. This study also analysed the muscles involved with gait and the Physiological Cross Sectional Areas (PCSA) of the muscles. Other studies such as Friederich and Brand (1990) and Duda et al. (1996) also looked into the PCSA and the muscle attachment points before Heller et al. (2005) among others, used this information to remove muscles that played little or no part in the movement and that had little effect on the model. The results of this study have been used in both musculoskeletal models and numerical models.

Brand et al. (1994) compared measured hip contact forces from an implanted instrumented prosthesis to results calculated from a mathematical model for one subject. However the in vivo measurements were carried out 59 days after surgery while the gait analysis for the mathematical model was at 90 days. The results showed measured peak forces of between 2.5 and 3.5 times BW. The calculated values were seen to be around 0.5 times BW higher than the measured values. Heller et al. (2001) carried out gait analysis at the same time Bergmann et al. (2001) measured the hip contact forces. A musculoskeletal model was used to calculate the hip contact forces from these subjects. 4 subjects were analysed and the results compared. Walking differences of between 0.3% and 33% were seen between the peak forces, with an average difference of 12% with the musculoskeletal model predicting forces higher than measured. For stair climbing the differences were between 3% and 37% with an average difference of 14%. Stansfield et al. (2003) compared 2 subjects with implanted instrumented prostheses to musculoskeletal models. Peak hip contact forces of between 2 times BW and 4 times BW were found for walking with the differences in peak forces in the calculated and measured results between 6.2% and 32.9%. For sitting and standing the peak forces were found to be around 2 times BW. The difference between the two groups for standing was between 8.1% and 23.2% with sitting between 10.2% and 38.8%. These studies demonstrated that the calculated models tend to overestimate the hip contact force but do provide a reasonable representation of the values recorded from measurement devices.

Several studies have focused on Neck Length (NL) and Neck Shaft Angle (NSA). Lenaerts et al. (2008) in a study with 20 subjects found that an increase in NL affected muscle activation of hip abductors and increased the hip contact force in all directions and reduced vertical loading. The NSA was found to have a minor effect in this study. Jonkers et al. (2008) created 2 subject specific geometries with patient specific loading patterns from gait analysis. The patient specific
loads were then applied to the opposite model and found that this changed the pattern of the stress distributions and therefore showed that patient specific loads influence the results and therefore yield more accurate results. Lenaerts et al. (2009) carried out a similar study to their 2008 work and used the patient specific geometry and loading patterns but also included a model to include the hip joint centre location. The use of the hip joint centre location was seen to produce different results hinting that this may give greater accuracy.

Table 2.11 Summary of the peak normalised forces from musculoskeletal modelling studies

<table>
<thead>
<tr>
<th>Study</th>
<th>Motion</th>
<th>Peak Normalised Force</th>
</tr>
</thead>
<tbody>
<tr>
<td>Crownshield et al. (1978)</td>
<td>Walking</td>
<td>3.3 to 5</td>
</tr>
<tr>
<td>Crownshield and Brand (1981)</td>
<td>Walking</td>
<td>3 to 3.5</td>
</tr>
<tr>
<td>Rohrle et al. (1984)</td>
<td>Walking</td>
<td>2.9 to 6.7</td>
</tr>
<tr>
<td>Brand et al. (1984)</td>
<td>Walking</td>
<td>5 to 5.2</td>
</tr>
<tr>
<td>Brand et al. (1994)</td>
<td>Walking</td>
<td>2.5 to 3.5</td>
</tr>
<tr>
<td>Heller et al. (2001)</td>
<td>Walking</td>
<td>2.71</td>
</tr>
<tr>
<td></td>
<td>Stair Climbing</td>
<td>2.86</td>
</tr>
<tr>
<td>Stansfield et al. (2003)</td>
<td>Walking</td>
<td>2 to 4</td>
</tr>
<tr>
<td></td>
<td>Standing</td>
<td>2</td>
</tr>
<tr>
<td></td>
<td>Sitting</td>
<td>2</td>
</tr>
</tbody>
</table>

2.7 Numerical Modelling

2.7.1 Introduction

Numerical modelling has commonly been used for assessing hip replacements. Since 1972 numerical modelling has been used to evaluate stresses in human bones and since then it has been applied to stress analyses of bone and bone prosthesis structures as well as other implanted devices (Huiskes and Chao, 1983). Several different aspects of hip replacements, such as comparisons of different implants, have been investigated using numerical modelling and these sections are presented below.

2.7.2 Cemented Prosthesis vs. Cementless Prosthesis

Both cemented and cementless femoral prostheses are commonly in use today. Cemented implants were the first implants to be used for THR surgery. However, recently cementless prostheses are increasingly used and are reported as having higher survival rates especially in younger patients.

Numerical modelling of cemented hip replacements mainly investigate either new prosthesis designs such as Kayabasi and Erzincanli (2006), or the long term stability of replacements such as Lennon et al. (2007) who investigated the risk of aseptic loosening due to cement creep and cracks in cement using a continuum damage formulation. As there is no cement in cementless hip replacements, studies focus on loosening and instability at either the primary stability stage
such as Pettersen et al. (2009) who used micromotion to investigate the stability, or the secondary or long term stability that was investigated by Jung et al. (2004).

2.7.3 Modelling of Loosening in Cemented Replacements

Due to the lack of contact between the implant and the bone, loosening in cemented hip replacements occurs around the cement. Two modes of failure due to loosening can occur and these are crack formation in the cement and movement between either the cement-bone interface or more generally the interface between the cement and the implant.

Damage modelling is the most common approach to investigate fatigue modelling of the cemented replacements and this is carried out by using Continuum Damage Mechanics (CDM) with Finite Element Modelling (FEM). The general concepts of CDM and how to use it are described in Chaboche (1988a, 1988b) and Lemaitre et al. (1984). CDM allows heterogeneous microprocesses involved during straining materials to be described as structures at a macroscale. CDM was used by Verdonschot and Huiskes (1997) to investigate the effect a debonded implant-cement interface had on the fatigue process. They found that a debonded interface would increase the stress 2/3 fold and that it accelerated the failure process by a factor of 4 indicating that a bonded implant-cement interface is important for long term success.

Cement used in surgery can contain pores due to trapped air and this porosity is important to take account of when modelling the cement. Lennon and Prendergast (1994) presented an FEM model using CDM and random pores to predict the failure in elastic materials. Jeffers et al. (2005) carried out 4 point bend fatigue tests both experimentally and numerically including porosity using the Monte Carlo method. A good agreement was found between the experimental and numerical models and it was seen that porosity may reduce fatigue life. As Verdonschot and Huiskes (1997) showed that the interface is important, Mann et al. (2007) showed that porosity at the implant-cement interface may lead to early loosening and to reduce this risk surface porosity should be below 30%.

The inclusion of creep modelling into the CDM models is important as creep can occur in cement. Jeffers et al. (2007) investigated comparisons between experimental cement mantle models and their numerical models including creep by using an empirical creep law. Good agreement was seen between the two methods and it was concluded that the numerical model was capable of predicting cement mantle fracture. Stolk et al. (2004) and Lennon et al. (2007) both included the Maxwell creep model into their models. Stolk et al. (2004) found that this model was able to predict the locations of cement damage around the implants, and the amounts of implant migration attributable to creep. Lennon et al. (2007) used the FEM model with CDM and the Maxwell creep model to predict the aseptic loosening in 17 patients. Six of the patients had undergone early revisions (<10 years) and the numerical models found 5 of the 6 early
revision subjects had the highest migration within the fatigue models indicating that migration could be used to identify early failures using this method.

2.7.4 Primary Stability in Cementless Hip Replacements

Primary stability in cementless hip replacements is important for the prosthesis to have long term stability. If the primary fixation is not achieved the bone cannot remodel and form a bond with the prosthesis to perform secondary fixation. Micromotion for primary fixation should be between 50 and 150µm (Jasty et al., 1997, Pilliar et al., 1986). Several numerical modelling studies have been carried out to find the micromotion at the stem-bone interface after surgery. All studies demonstrated that micromotion was within the tolerances for osseointegration to occur. Validation studies such as Reggiani et al. (2007) have compared experimental and numerical results to ensure that numerical models give an accurate prediction of what happens in vitro and therefore in vivo.

The femoral part used in numerical modelling has been shown to affect the stress distributions within the model and consequently the micromotion results for primary fixation studies. Jonkers et al. (2008) carried out numerical modelling to investigate if subject specific models affect the stress distributions. They found that both the subject specific geometry and the subject specific loads influence the stress distributions. Subject specific models have been used in most studies examining the primary fixation such as Pettersen et al. (2009). However, some studies have used CT scans of their experimental studies to generate the geometry for their numerical models to carry out validation studies such as Viceconti et al. (2001).

The material properties used for numerical models of the hip joint vary between different studies. Some studies use the Hounsfield values from the CT scans to find apparent densities, Young’s Modulus and ultimate compressive strength to investigate cementless tibia implants such as Perillo-Marcone et al. (2004). Other studies, such as Andreaus and Colloca (2009), used simplified material properties as complex orthotropic material properties have only a small influence on finite element studies. Studies that did not use patient specific geometries cannot find values from scans and therefore have to use simplified material properties for numerical models. Wong et al. (2005) investigated the effects that different material properties had on the primary stability of cementless prostheses and found that a reduction in modulus of cortical and cancellous bone causes an increase in the micromotion and interface bone strain. Two studies that have looked into material properties of bone are Wirtz et al. (2000) and Burstein et al. (1976). Wirtz et al. (2000) carried out a literature review of the material properties of bone whereas Burstein et al. (1976) tested bone and listed material properties for bone with regards to age.
The contact parameters used to examine primary fixation of a cementless hip replacement are important as these can affect the micromotions at the stem-bone interface. Andreaus and Colloca (2009) used a fully bonded constraint at the interface however they looked at the shear stress in the vertical direction as a sign of micromotion initiation rather than micromotion. A fully bonded contact was also used by Behrens et al. (2008) to investigate stress shielding. Various different coefficients of friction (COF) have been used in studies. Viceconti et al. (2001) used a COF of 0.2 in most models to determine the effects soft tissue has on the primary fixation of cementless hip prostheses and found that stability of the implant is extremely sensitive to soft tissue at the stem-bone interface. In a numerical model of a push out test of porous coated stems Helgason et al. (2008) used a COF of 0.5. Several other studies such as Abdul-Kadir et al. (2008) and Wong et al. (2005) have used a COF of 0.4. Pettersen et al. (2009) used 0.4 as the COF as they modelled the interface using the augmented Lagrange contact algorithm. Mann et al. (1991) tested the mechanical characteristics of the stem-cement interface and found that using a COF gave a better representation than other methods.

2.7.5 Bone Remodelling

Bone remodelling or secondary stability is where the bone forms a bond with the cementless prosthesis and osseointegrates onto the stem. This is how cementless stems maintain long term stability. In vitro experimental testing of this process is not possible as this is a biological process. However, this process has been investigated using numerical models employing the FEM. Algorithms are generally used to model the bone remodelling process similar to the way that the cemented prostheses models used damage models discussed earlier.

Bone remodelling algorithms are complex code that change the elements or contact parameters in a model when certain defined thresholds are reached. These can be based on many different parameters. McNamara and Prendergast (2007) evaluated 4 algorithms that used strain, microdamage or a mixture of both as the stimulus. The study found that a model based on either strain or microdamage with a critical level for remodelling gave the best results. Testi et al. (2004) also compared algorithms to evaluate which model was better. One algorithm was based around density and the other around distance. Both algorithms were seen to be accurate from a computational point of view. Schmitz et al. (2004) investigated the first 9 months of a thrust plate prosthesis with is a short stemmed prosthesis. 6 bone remodelling algorithms were used and compared to a clinical study to see which gave similar results. Good agreement was found for a range of strain energy based signals and also deviatoric remodelling signals. However the compressive dilatational strain algorithm did not support the results. This work was based on earlier work by Ploeg et al. (2001) where the basic algorithm is discussed. Moreo et al. (2007) used an algorithm based on Continuum Damage Mechanics (CDM) with the internal variables
negative to allow the interface to osseointegrate. It was concluded that the results qualitatively agreed with clinical observations.

It has been seen in several studies that boundary conditions and muscle loadings can have an effect on the results from a model. Bone remodelling algorithms were found to be no different by Bitsakos et al. (2005). This study investigated the effect of muscle loading on bone remodelling using 3 different complexities of muscle configurations. It was seen that bone loss was affected and that simplified models over predict the amount of bone loss. This illustrates that the loading configuration plays an important part in bone remodelling.

As the threshold for bone remodelling has an upper and lower limit it is important to take into account both forces that go over the threshold and that are lower than the threshold. However, most studies only look at the upper value. Hazelwood et al. (2001) created a model taking into account both the dynamic responses to disuse and over load. A study by Gracia et al. (2010) did not use a bone remodelling algorithm but rather took DEXA scans preoperatively, at 6 months, 1, 3 and 5 years and then changed the material properties of the models according to the scans. If clinical evaluations of subjects are possible on a continuous basis this method would prove an asset and a good tool for both comparing and validating long term remodelling models.

Most bone remodelling studies are only concerned with secondary fixation which is the period from primary fixation to long term fixation and when the osseointegrated bond is formed. However, Jung et al. (2004) investigated the long term loosening of uncemented hip replacements due to compressive stress application while testing the effect of load angle, material properties of prostheses and stem length. A bone resorption threshold value of 50MPa was used which for cortical bone is equivalent to approximately 3600µ strain. Migration did not occur between 63° and 74° load angle due to the compressive stress not reaching 50MPa. However below 63° and above 74° bone resorption did occur. This shows that the forces in the transverse and axial direction, although not as big as the vertical load, may affect implant loosening. Stem length was also seen to have an effect on migration with short stems showing less migration.

2.7.6 Boundary Conditions for Numerical Models

Heller et al. (2005) carried out musculoskeletal models to find the effects of simplifying the muscle forces. 3 models were used with different muscles attached and the results were compared to in vivo hip joint contact forces measured in Bergmann et al. (1993). The study found a model simplified to a resolved abductor muscle, a resolved adductor muscle, the Ilio tibial tract and the tensor fascia latae gave the closest comparison to the in vivo results of the simplified muscle models. Phillips et al. (2007) investigated the effects of muscles on numerical models of the pelvis. They found that the inclusion of muscular and ligamentous boundary
conditions lowered the occurrence of stress concentrations within the cortex and therefore affects the results of the model. Another study that looked at the effect of muscle boundary conditions was Speirs et al. (2007). The study used a standard femur and varied the muscles in 5 models. Differences were found in the strains in the models and it was concluded that muscles play an important role in numerical modelling of the femur.

2.8 Experimental Modelling
The interface between either the implant and the cement, the cement and the bone or the implant and the bone to date is impossible to measure in vivo. However, these interfaces play an important role for implant fixation whether it is long term fixation in cement implants, primary fixation in cementless implants or fixation in other hip prostheses. The only way to measure the micromotions at the interface is either by numerical modelling as just discussed or by experimentally testing implants in vitro.

Strain gauges have been used to predict the stress and strain distributions in cemented models. Oh and Harris (1978) implanted 6 different femoral components (Charnley regular, Charnley extra-heavy Cobra, Mueller regular Protasul-10, Trapezoidal-28 large, mini CAD, and CAD standard straight stem) into fresh human cadavers and compared the results to that of an intact femur. The strain gauges were placed on the outside of the femur and were not at the interface. The results however showed a decrease in the strain in the femurs with prostheses implanted in them. Colgan et al. (1996a) also used strain gauges for investigating the stresses in the cement around implanted prostheses. Six gauges were strategically implanted within the cement part and the effects of the load especially the transverse load were evaluated. This was followed up by Colgan et al. (1996b) where a collar was applied to investigate any difference.

Strain gauges are quite complex and have to be precisely implanted to gain accurate results and although strain gauges are still being used, more and more studies are now using Linear Voltage Displacement Transducers (LVDT) to measure the migration at the interface as an alternative to using stress and strains within the cavity. LVDTs are used to measure linear mechanical displacement or position in control systems. The position of the core in the centre tube relative to the excitation coil and the two pick off coils changes the mutual inductance between the excitation coil and the pick-off coils. As one voltage drops the other rises and from this linear displacement can be measured.

For cemented prostheses the measurement of micromotion is important at the interface as this is where the problems for failure can arise. LVDT’s are used in models to measure the micromotion at the interface of these models in vitro. Maher et al. (2001) developed a device to measure the migration at one point. Instead of placing an LVDT into the interface a bar was attached to the prosthesis and 6 LVDT’s in the device were used to measure the micromotion of
the stem. The device can measure all six degrees of freedom of movement however it only measures at one point. This same device was used in Maher and Prendergast (2002) where they examined the effect of cyclic load of 2 million cycles on 2 different cemented prostheses. They found that micromotion and inducible displacement can be used to examine long term failure and that differences in migration can be seen after 1 million cycles. A study by Britton and Prendergast (2005) also used this setup to examine the results of 4 prostheses. They only carried out 1 million cycles due to the findings of Maher and Prendergast (2002). The other difference between the two studies was that Britton and Prendergast (2005) added 2 muscle forces to the rig setup. However they concluded that migration may not be the best way to compare prostheses designs.

A different approach is to use individual LVDT’s at different points within the model. This method was carried out by Bialoblocka-Juszezyk et al. (2009). They tested different types of cement for long term stability. The load used in this study was not from Bergmann et al. (1993) which is commonly used in experimental modelling of the hip joint but the loading cycle from stair climbing, the load from in and out of a car and the load from going in and out of the bath. The reason for this was that they believed their study had the highest forces which were more responsible for failure if it was to occur. One million cycles were applied and this was the equivalent to 24 years. Three other studies that used the same experimental set up and force were Cristofolini et al. (2007a, 2007b) and Cristofolini et al. (2010). The Cristofolini et al. (2010) study investigated the effects of undersizing and found that stems that are smaller than optimal may be prone to loosening and therefore the prosthesis size chosen is important. Sangiorgio et al. (2008) designed a new simulator with higher peak forces then the previous studies. 6 DVRT’s (Differential Variable Reluctance Transducers) were used, three at the stem-cement interface and three at the cement-bone interface. The results showed that higher peak joint forces and muscle forces had a significant effect on the femoral stem stability after 20,000 cycles.

The surface finish can be an important factor in the long term stability of cemented prostheses and Lennon et al. (2003) tested the effect surface finish had on cement fatigue damage over 2 million cycles, while comparing the results found to numerical modelling results. However the results showed there were no significant differences between any of the surfaces.

Cementless hip replacement experimental studies, although quite similar in setup to cemented experiments, normally study one of three areas. These are primary stability, secondary stability and long term stability. Primary stability is when the implant is press fit into place and is tightly fitting. Micromotion of between 50 and 150µm is the motion that is needed for an osseointegrated bond to occur.
Monti et al. (1999) investigated the primary stability of cementless hip replacements over 1,000 cycles using a stair climbing load and 4 LVDT’s at 4 different locations. Gotze et al. (2002a, 2002b) also investigated primary fixation by testing custom made cementless prostheses against normal cementless prostheses. They used 6 LVDT’s at 6 different locations throughout the model that used human femoral cadavers. Only 3 cycles were carried out and the micromotion was not seen to exceed 150µm for any of the prostheses tested. 150µm is seen in literature as the maximum motion that will not prohibit the bone remodelling and an osseointegrated bonding forming between the implant and the bone. Fottner et al. (2009) also used 6 LVDT’s to test primary fixation in the Thrust Plate Prostheses, Mayo prostheses and Metha prostheses implanted into Sawbone composite femurs. However, this study used the 6 LVDT’s in a similar device to Maher et al. (2001) where they measured all movements at one point. However they took readings at 5 different locations. All micromotions were found to be under 150µm indicating bone remodelling would occur.

Kassi et al. (2005) tested cementless prostheses on an experimental rig with 4 muscle attachments. This was a more complex setup compared to most other setups that have one or in some cases no muscle attachments. However the study tested the effect with and without all the muscle attachments and the results showed that all muscles should be replicated in vitro otherwise micromotion is underestimated and primary stability is overestimated. This study also compared stair climbing cyclic loads to that of the walking cycle and found that the stair climbing loading may compromise the osseointegration process through higher micromotion. Pettersen et al. (2009) investigated primary stability of cementless hip replacements using a specially designed fixture that contained 6 LVDT’s that could measure all the movement at one point. The fixture was then movable vertically so it could measure at different locations on the femur at different heights.

Other studies that investigated the primary fixation of cementless prostheses were Abdul-Kadir et al. (2008) and Westphal et al. (2006). Abdul-Kadir et al. (2008) only used 1 LVDT to measure the micromotion at the stem-bone interface and the experimental study was mainly a validation of the numerical work carried out in the study. Westphal et al. (2006) carried out a different approach to most studies as they used 3D video analysis and reflective markers to measure the migration. No measurements were taken at the stem-bone interface instead reflective markers were placed strategically to form coordinate systems for both the prosthesis and the bone. Measurements were taken for new short stemmed prostheses and compared to the standard prosthesis. Initial migration was found to be higher in the long stemmed prosthesis, however the cyclic motion was seen to be less.
2.9 Discussion
Total Hip Replacement (THR) surgery is a successful procedure and has good survival rates for both cemented and uncemented prostheses. The age profile of people is changing as people are now living to a greater age and are a lot more active as elderly people. The body mass of people today has also risen and people are now carrying more weight which puts more force on the hip joint. Due to these reasons the number of hip replacements is rising and with the number of operations rising the number of failures rises too. The AAOS reported 234,000 primary hip replacements were carried out in 2004 in the US and that 46,000 revisions were carried out in the same year. In Europe the figures would be similar. As the numbers are getting bigger even a small percentage of failures gives a large group of patients. From the review of the literature aseptic loosening is by far the most common reason for failure. The fatigue testing and investigation for aseptic loosening in cemented hip replacements has been extensively carried out both experimentally and numerically. Cementless hip replacements are more complex and there are several issues that may affect the fixation of the prosthesis and lead to loosening. The primary stability of cementless prostheses has been investigated and found to be important to the long term survival of the prosthesis. Secondary stability is complex as it is mainly a biological process within the body and can differ from subject to subject. However complex numerical models using different forms of algorithms have been successfully shown to predict this process within the body and the effect that parameters may have on the process.

The long term stability of the cementless prostheses is not well documented. *In vitro* studies of cementless prostheses that have shown primary stability of the prosthesis have not investigated the long term stability of the prosthesis. Of the numerical modelling studies only Jung *et al.* (2004) was found to evaluate long term stability and the loosening of cementless prostheses in the long term, with most other studies concentrating on primary or secondary stability.

In this review of the literature experimental setups have all been carried out using a ramp load at a constant angle. However the angle at which the force acts is not constant throughout a gait cycle, nor is the angle the same for all everyday motions. Therefore, an experimental setup is needed that will take into account the effect of all 3 load directions that will be independent of each other.

This study will investigate the effect of mechanical loading on aseptic loosening of cementless hip prostheses after primary and secondary stability has occurred. This will be carried out by experimental testing and numerical modelling.
2.10 Goals of the study
From the Literature review the main objectives of the study were identified and these are:

- Identify the major reason for failure after cementless total hip replacement surgery.
- Find representative loading configurations for 5 main motions, walking, sitting, standing, up steps and down steps.
- Find the maximum loading configurations for 5 main motions, walking, sitting, standing, up steps and down steps.
- Design, fabricate and commission an experimental simulator using realistic forces and with the ability to vary the loading directions.
- Carry out cyclic loading on representative Sawbone setups investigated different loading patterns.
- Develop a numerical model to investigate the stresses at the bone-stem interface using calculated hip force data.
- Investigate loading variations from realistic force data using numerical modeling to analyse the likelihood of aseptic loosening or instability occurring.
- Identify loading patterns that indicate loosening may occur.
CHAPTER 3

Musculoskeletal Modelling
Chapter 3  Musculoskeletal Modelling

3.1  Introduction
Gait analysis is a common procedure used to investigate the gait pattern and movements of subjects in many different areas of research. In chapter 2, good comparisons were seen between measured hip forces using musculoskeletal modelling and results from implanted instrumented prostheses. The motion analysis in this project was carried out with the help of the Physical Education and Sports Science department (PESS), who generously agreed to the use of their laboratory for this project.

3.2  Subject Recruitment
To carry out motion analysis in the biomechanics laboratory ethics approval was needed. This ethics approval was granted from the Faculty of Science and Engineering Research Ethics Committee from July 2008 for one year and extended in July 2009 for a further year. The Application No. was ULREC 08/52 and the approval letter can be seen in Appendix A.

Subjects were recruited through two means, firstly an email was sent around the University of Limerick recruiting volunteers, and secondly subjects were recruited through St Nessans Mid Western Orthopaedic Hospital in Croom. 23 patients (12 THR, 6 BIRM, and 5 Healthy) were recruited in total and subject details can be seen in Table 3-1 below. Of the 12 Total Hip Replacement (THR) subjects, 5 of these had their left hip joint replaced, 4 had their right hip joint replaced and 3 had both their left and right hip joints replaced. Of the 6 Birmingham Hip Replacement Surgery (BIRM), subjects, 3 subjects had surgery on their left hip joint 2 had surgery on their right hip joint and 1 subject had surgery on both their left and right hip joints. The remaining 5 were healthy subjects and were used as a control to gather information on subjects that didn’t have hip surgery. 15 of the subjects were male and 8 female. The surgery procedure used for each of the THR and BIRM subjects was recorded. In 14 subjects the posterior approach was used and in the other 4 an anterolateral approach was used. The average age of the subjects was 57 years old and this ranged from 23-74 years old. The only inclusion criteria used for the patient recruitment was that the patients with hip replacements were at least 6 months post operation and that all subjects were pain free in both hips at the time of testing.
Table 3-1 The subject information. M = Male, F = Female, THR = Total Hip Replacement, BIRM = Birmingham, H = Healthy, S = Standard, L = Large, AL = Anterolateral, P = Posterior, Lt = Left, Rt = right, B = Both, N/A = Not Applicable

<table>
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<th>Implant Head Size</th>
<th>Surgery Type</th>
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<td>L</td>
<td>P</td>
<td>Rt</td>
</tr>
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<td>B</td>
</tr>
<tr>
<td>14</td>
<td>M</td>
<td>558</td>
<td>H</td>
<td>N/A</td>
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</tr>
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<tr>
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<td>Rt</td>
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<td>Lt</td>
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<td>H</td>
<td>N/A</td>
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</tr>
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<td>590</td>
<td>BIRM</td>
<td>L</td>
<td>P</td>
<td>Lt</td>
</tr>
</tbody>
</table>

3.3 Motion Analysis Equipment

3.3.1 Cameras

6 Hawk cameras were used, these are digital infrared cameras, and a picture of one of these can be seen in Figure 3-1. The cameras were set up as in Figure 3-1 to create a capture volume. The capture volume is where the reflective markers are picked up by the infrared cameras. If the subject goes outside of the capture volume one or more of the 15 markers might not be picked up and the data may be missing markers. Two or more cameras must be able to see each marker.
3.3.2 Markers

15 reflective markers were used during each motion analysis. Each subject had to remove any shiny or reflective items such as watches, rings and chains. Any items that could not be removed were covered with black tape. The Helen Hayes marker set was used for all 23 subjects. The positions of each of the markers can be seen in Figure 3-2 below. The markers were situated at the left metatarsal, left heel, left ankle, left calf, left knee, left thigh, left pelvis, right metatarsal, right heel, right ankle, right calf, right knee, right thigh, right hip and at the sacrum. It is important to place the markers in the correct positions as studies have reported that errors can arise from the mis-placement of markers (Baker et al., 1999).

Figure 3-1 a) Schematic illustration of the positions of the 6 cameras b) one of the 6 cameras used
3.4 Testing Procedure

The testing procedure used for all 23 subjects was as follows:

- Subjects were asked to read and complete several forms. These included a patient information sheet, a patient consent form, a pre-test questionnaire, and a subject information sheet. A copy of each of these documents can be seen in appendix A.
- The subjects were shown the testing area and informed what motions would be carried out.
- Subjects were asked to change into cycling shorts in the changing area provided.
- Markers were applied using the Helen Hayes marker set described above.
- Subjects were asked to walk through the capture volume three times.
- A chair was placed in the capture volume and the subjects were asked to sit down from a standing position three times.
- Subjects were asked to stand up from a sitting position three times.
- The chair was removed and two aerobic steps were placed at different heights into the capture volume.
- Subjects were asked to walk up the steps three times.
- Subjects were asked to walk down the steps three times.
- The testing was completed.
3.5 Software
The software used for the motion analysis was Evart 5.04 (Motion Analysis Corporation, Santa Rosa, CA, USA). Evart carries out three main tasks. These are 1) calibrating the capture volume, 2) tracking the 3D marker locations and 3) the post processing of the motion data. Each marker is assigned a marker name in each frame, starting with the first frame. The data is then rectified, this means that the marker will keep the same marker name throughout the motion unless the marker disappears (ie the marker is blocked or not picked up by two or more cameras in that frame). When the marker reappears it is reassigned the marker name and then rectified again. This process continues until the final frame. If the gaps, where the markers are missing, are only a frame or two long ‘linear joint’ is used to join the gaps to complete the data. If the gaps are longer the ‘RB join’ is used. ‘RB join’ fills in the gap by using three or more other markers in the analysis to predict the movement of the missing marker.

3.6 Motion Analysis Results
Evart 5.04 was used to process the data obtained during testing. The software used the position data of each of the 15 markers at each frame during each motion. After the data was analysed it
was exported to the LifeMod software (LifeModeler Inc., San Clemente, CA, USA). Three types of data were exported for each repetition of each motion for each subject. These are marker position data, marker velocity data and marker acceleration data.

3.7 Data Formatting
When the 15 markers were complete, all frames of the data were exported as a .ts file. The LifeMod input file had to be in .slf format. Matlab (MathWorks, Natick, MA, USA) was used to change the format of the data. In the .ts file exported from Evart the motion data is in rows, from marker 1 to marker 15. For each marker, position data, velocity data and acceleration data is recorded in the file. The only data that is required for LifeMod is the marker position data. The .ts file is changed to a .csv file in excel and then put through the Matlab program. The data is then exported in column format as a .dat file. This is changed to a .txt file where the yaw pitch and roll is added. This is then copied and pasted into an empty .slf file. This data is then imported into the LifeMod software.

3.8 LifeMod Musculoskeletal Software
LifeMod is a virtual human modelling and simulation software solution. LifeMod has been used in industry in many fields such as orthopaedics and medical research, sports and equipment research and in the aerospace and automotive area. To create a model in LifeMod there are several parts that have to be carried out. These are 1) create the human model, 2) create and position the environment for the model, 3) run an inverse dynamic simulation, 4) run a forward dynamic simulation, 5) validate the model, 6) refine the model and 7) record results. The software uses the anthropometric data such as height and weight from each subject to create a skeleton that represents the subject been analysed. Using this anthropometric data and the environmental contacts that are put in place the model can calculate the forces at all the joints and muscles that are present within the model.

3.1 Governing Equations for Musculoskeletal Modelling
Multibody systems in LifeMod are modelled as a connected system of rigid and flexible bodies. Bodies may be connected to one another by kinematic constraints or forces. LifeMod is a subpart of the Adams software, which is an implicit nonlinear solver which models dynamic systems represented by a series of differential and algebraic equations (DAE). The governing equations have the following form:

\[ M(q)\ddot{q} - \mathbf{L}(q, \dot{q}, t) + C_q^T(q)\lambda = \mathbf{Q}(q, \dot{q}, t) \]  \hspace{1cm} (3.1)

\[ C(q, \dot{q}, \ddot{q}, t) = 0 \]  \hspace{1cm} (3.2)

Where:
$M$ is the system generalized mass matrix, $L$ contains Coriolis and related terms, $C$ is the set of constraint equations, $Q$ is the set of generalized inputs (forces), $q$ is the set of generalized states, $\lambda$ is the vector of Lagrange multipliers, and $t$ is time. A subscript denotes partial derivative with respect to a state.

### 3.1.1 Governing System Equations

For the multibody system in Equations 3.1 and 3.2 we consider the inclusion of one or more nonlinear single-physics parts whose governing equations are implicitly cast in general form as (Collingridge, 2010, Rennich and Zielonka, 2009):

$$R_i(q_i, \dot{q}_i, \ddot{q}_i, \lambda_i, Q_i, t) = 0$$  \hspace{1cm} (3.3)

Where $q$, $\lambda$ and $Q$ are defined as above and the subscript $i$ denotes a particular analysis component. Note that equations 3.1 and 3.2 can easily be cast in the form of equation 3.3 and simply considered to be another component. These sets are extended to couple the nonlinear equation set to the system as

$$R(q, \dot{q}, \ddot{q}, \lambda, Q, t) + C^T_q(q)\lambda_m = 0$$ \hspace{1cm} (3.4)

Where $R$ is now the aggregate set of $R_i$ and $\lambda_m$ is now the complete vector of Lagrange multipliers used to enforce system constraints internal to multi-physics components and between components of the system. In general, equation 3.4 is a set of Differential Algebraic Equations (DAE) that can be solved efficiently and accurately using implicit, variable step integrators with sparse matrix solvers. This is discussed further in Rennich et al. (2010)
3.1.2 Creating the Human Model

In this process a model of the lower body is created. Anthropometric data is used such as age, height, sex and weight to create a skeletal model to represent the subject that carried out the testing. Each subject’s anthropometric data can be seen in Table 3-2. There are three databases that LifeMod uses to create a human model, the GeBod Anthropometric database, the PeopleSize Anthropometric database and the US Army – Natick Anthropometric database. The GeBod database was chosen for this study as this provided options to use the anthropometric data collected from the subjects.
Table 3-2 Anthropometric data from the 23 subjects analysed during the motion analysis testing

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<th>Weight (kg)</th>
<th>Height (mm)</th>
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<td>861</td>
<td>72.2</td>
<td>1570</td>
</tr>
<tr>
<td>16</td>
<td>F</td>
<td>791</td>
<td>82.3</td>
<td>1652</td>
</tr>
<tr>
<td>17</td>
<td>M</td>
<td>619</td>
<td>76.2</td>
<td>1626</td>
</tr>
<tr>
<td>18</td>
<td>M</td>
<td>893</td>
<td>90.6</td>
<td>1795</td>
</tr>
<tr>
<td>19</td>
<td>M</td>
<td>718</td>
<td>80.3</td>
<td>1794</td>
</tr>
<tr>
<td>20</td>
<td>M</td>
<td>802</td>
<td>62.8</td>
<td>1599</td>
</tr>
<tr>
<td>21</td>
<td>M</td>
<td>790</td>
<td>105.3</td>
<td>1855</td>
</tr>
<tr>
<td>22</td>
<td>M</td>
<td>706</td>
<td>72.1</td>
<td>1700</td>
</tr>
<tr>
<td>23</td>
<td>M</td>
<td>590</td>
<td>100.2</td>
<td>1829</td>
</tr>
</tbody>
</table>

The joints at the hip, knee and ankle were added to the model. Stiffness and dampening values from Amankwah et al. (2004) and Al Nazer (2008a, 2008b) were used to define the parameters for the joints at the hip, knee and ankle. These can be seen in Table 3-3. During the inverse dynamics analysis these joints record the joint angulations while the model is being manipulated with motion agents. During the inverse dynamics simulation the passive stiffness/damping parameters and the motion agent stiffness’s will have an effect on the model. For example, if the joint stiffness’s were too excessive, the motion agent stiffness may not be sufficient to move the body segments. Conversely, if the motion agents were very stiff and the joint stiffness/damping parameters were small, too much oscillation during the simulation may occur. As a rule of thumb a damping value of 10% of the stiffness is sufficient in most cases. Soft tissues or muscles are added to the model. These are taken from the LifeMod database and are based on the anthropometric data recorded from each subject. The muscle data in LifeMod is generated from a series of studies from Schumacher and Wolff (1966c, 1966a, 1966b), Eycleshymer and Shoemaker (1970) and McMahon (1984).
3.1.3 Creating the Environment

In this section of the analysis all other parts of the model are added. For the present study, the hip replacement is the main component added here. The ground and any other items such as the stairs or the chair are also added as required. Two aerobic steps were used to create the up steps movement. The heights of these steps were 150mm and 250mm. A standard un-cushioned chair was used for the standing and sitting movements. The height of the chair was approximately 450mm. When environmental parts are added the contact between the human model and the environment part needs to be defined. This determines how the human model will react with the other parts of the environment.

3.1.4 Running the Inverse Dynamic Simulation

The motion data recorded on each subject is loaded into LifeMod. The inverse dynamic simulation uses this motion data for the subject for each motion that was performed. During this simulation the human model is controlled by the motion data. Both the soft tissues and the joints do not have an effect on the model but LifeMod records how they react and how they should move within the environment.

3.1.5 Running the Forward Dynamic Simulation

For the forward dynamic simulation the motion data is not used. Instead the LifeMod software runs the model using the results of the inverse dynamic simulation for the joints and soft tissues. The soft tissues and joints are responsible for moving the human model. The forces and other values required are then measured during this simulation.

3.1.6 Validation

Validation of the models is essential to check the accuracy of the simulation. This was done by using a force plate during all the motions carried out. Walking, up steps and down steps all had one foot stand on the force plate. The force from this foot ground reaction force in LifeMod was compared to the forces recorded from the force plate. For the sitting down and the standing up motions both the left and the right feet were on the force plate. Therefore both the forces for the left and the right ground reaction forces were added together and compared to the forces.
recorded on the force plate. Figure 3-5 shows an example of both the LifeMod results and forces recorded on the force plate for each of the 5 motions.

The peak forces for each of the LifeMod models and ground reaction force were compared and the % difference between them used to check accuracy. The % difference for each motion was averaged and these can be seen in Table 3-4 below.

The average differences between the peak forces recorded on the force plate and the LifeMod models for the walking, sitting and standing motions are less than 5%. Several papers have investigated the differences between the motion data and the calculated forces and they found results higher than 5% for similar comparisons (Heller et al., 2001, Heller et al., 2005, Stansfield et al., 2003). The up steps motion showed an average difference of just over 6%, whereas the down steps motion showed average differences of over 14%. The reason for this difference may be that the peak force for the motion only occurs on the force plate for a very short period of time. As the LifeMod data is filtered any sudden changes in the force will be
smoothed out. The graphs in Figure 3-5 show a good agreement between the curves for the force plate and the simulated models and they follow similar trends. Therefore, the results from the LifeMod software can be used with confidence that these results are an accurate model of the forces present during the motion analysis testing.

Table 3-4 Average % difference in peak forces for each motion compared to literature

<table>
<thead>
<tr>
<th>Motion</th>
<th>This Study</th>
<th>Heller et al. (2001)</th>
<th>Heller et al. (2005)</th>
<th>Stansfield et al. (2003)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walk</td>
<td>4.15%</td>
<td>12%</td>
<td>1%</td>
<td>3.2-12.2%</td>
</tr>
<tr>
<td>Sitting</td>
<td>4.26%</td>
<td>-</td>
<td>-</td>
<td>10.2%</td>
</tr>
<tr>
<td>Standing</td>
<td>1.91%</td>
<td>-</td>
<td>-</td>
<td>8.1%</td>
</tr>
<tr>
<td>Up Steps</td>
<td>6.07%</td>
<td>14%</td>
<td>4%</td>
<td>-</td>
</tr>
<tr>
<td>Down Steps</td>
<td>14.62%</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

3.1.7 Refining the Model

The models were refined if either the motion carried out by the subject did not match the motion of the model or the ground reaction forces in the model did not match the results of the readings from the force plate. The motions of the subject and model differed if the model jumped or a joint twisted in the wrong direction. These errors generally came from a missing marker or a marker being assigned the wrong name in the motion data. This was resolved by going back to the motion data and fixing the missing markers and then restarting the model with the new motion data. If the ground reaction forces did not match the foot ground contact forces in the model the parameters for the contact had to be changed or the height and position of the environment (the ground, chair or stairs) had to be changed. The inverse and forward dynamic simulations were rerun until the results were validated.

3.2 Musculoskeletal Modelling Results

23 subjects were tested and the subject details were presented earlier in Table 3-1. Each subject was instructed to carry out 5 everyday activities. 3 trials for each activity were recorded. The testing procedure was presented previously in this chapter. In total approximately 345 models were simulated (23 subjects x 5 movements x 3 trials). Results for each model were examined.

3.2.1 Defining Directions

It is important to define the force directions and co-ordinate system used. These will be described here and will be used throughout the rest of the study. A Cartesian co-ordinate system was used for this study. The positive X force is the medial to lateral direction, the positive Y force in the anterior to posterior direction and the Z force is in the proximal to distal direction. These force directions are illustrated in Figure 3-6. The results of the musculoskeletal modelling, the numerical modelling and the experimental modelling all investigate the effects of
the force in the X direction (Fx), the effects of the force in the Y direction (Fy), the effects of the force in the Z direction (Fz) and the influence of the force magnitude (Fm).

3.2.2 Walking Motion Results

As this study is primarily concerned with the hip joint this was the main focus of the results obtained. To obtain one cycle of the movement under investigation, one cycle had to be identified. For walking the cycle started at heel strike as can be seen in Figure 3-7. The cycle continued until heel strike occurred again on the same leg. The average length of the walk cycle was 1.24 seconds per cycle, with the shortest being 0.95 seconds per cycle and the longest nearly twice as long at 1.88 seconds per cycle.
Figure 3-7 Nine stages of the walking motion during the musculoskeletal modelling

Each of the models simulated were of different length and therefore had to be normalised to be draw comparisons. This was carried out in the freeware software Octave (Eaton, 2002). Data was interpolated with regards to the longest walk cycle and then normalised to represent a % of the cycle.
For every subject, 3 trials were carried out for each of the everyday activities. This gave a better understanding of the movement and limited the effect any outlier data would have on the results. The three trials were averaged for each subject. Figure 3-8 shows the force magnitude (Fm) on the left hip joint for the three trials and the average of the three trials for one subject.

![Hip Contact Force Graph](image)

**Figure 3-8** Force magnitudes from 3 trials of one subject during the walking motion and the calculated average

To compare the average of each subject to the other subjects the results again had to be interpolated as each average was a different length. The technique of Bergmann *et al.* (2010) was used to find the average peak loads. Once this was completed the results for all 23 subjects could be compared and this can be seen in Figure 3-9. This technique divides the force by Body Weight (BW) to normalise the data. The normalised data is indicated using an n subscript (ie $F_{xn}$, $F_{yn}$, $F_{zn}$ and $F_{mn}$). This graph only illustrates the overall pattern of the forces. In order to make comparisons between groups the average forces had to be found for each group. The 5 comparative areas of interest for these subjects were:

- Male vs. Female
- Standard prosthesis head size vs. Large prosthesis head size
- Posterior surgery vs. Anterolateral surgery
- THR vs. BIRM vs. Healthy subjects
- Healthy hip of THR subjects vs. Replaced hip of THR Subjects
Figure 3-9 Normalised force magnitude in the left hip joint for the 23 subjects during the walking motion

3.2.2.1 Male vs. Female

Of the 23 subjects analysed, 15 were male and 8 female. Figure 3-10 shows the average for the male subjects compared to the female subjects for the different forces on left and right hip joint.
Figure 3-10 The average forces for male and female subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the walking motion.

From the figure above it can be seen that there is similarity between the male and female results. The male results show higher values in all forces apart from $F_{x_m}$ for both the left and the right legs. In order to determine whether there were any significant differences between the two groups statistical analysis was carried out between both groups using the peak forces of each of the force directions. The T-Test and the Mann-Whitney test were carried out using the statistical analysis software SPSS (SPSS Inc., Chicago, IL, USA).
For the left hip joint male subjects had higher forces than female subjects in $F_{m,n}$, $F_{y,n}$ and $F_{z,n}$. For $F_{x,n}$ the female peak forces are higher. However, the differences between the male and female results were not significant $p>0.05$ in all of the force directions.

For the right hip joint male subjects again had higher forces than female subjects in the $F_{m,n}$, $F_{y,n}$ and $F_{z,n}$ and for $F_{x,n}$ the female peak forces were higher. As with the left hip the differences between males and females were not significant $p>0.05$ in any of the force directions. The effect size of all these forces was low to medium ranging from 0.17 to 0.28.

### 3.2.2.2 Standard Head Size Prosthesis vs. Large Head Size Prosthesis

The size of the prosthesis head in the THR and BIRM subjects, varied from subject to subject. 8 subjects had large head size prostheses and 10 subjects had standard size prostheses. The other 5 were healthy subjects and were excluded from this analysis. An analysis was carried out to see if any differences occurred within any of the force directions between the two groups. The average for the large prosthesis head size and standard prosthesis head size subjects can be seen in Figure 3-11 for the left and the right hip joints.
Figure 3-11 The average forces for the standard and the large head sized prostheses at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the walking motion.

It can be seen that for the left hip joint the forces show a good agreement in $F_{mx}$, $F_{my}$ and $F_{mz}$. A difference exists in $F_{mx}$, with the force in the standard head size subjects being higher than the large heads. For the right hip joint similar results can be seen with the only major difference in $F_{mx}$. Statistical analysis of the two groups was carried out using a standard T-Test and the Mann-Whitney test.
For the left hip joint large head size subjects had only slightly higher peak forces than the standard head size subjects in $F_{m\,n}$, $F_{y\,n}$, and $F_{z\,n}$. For $F_{x\,n}$ the standard peak forces were seen to be higher.

In the right hip joint, standard head size subjects had slightly higher peak forces than the large head size subjects in $F_{m\,n}$, $F_{y\,n}$ and $F_{z\,n}$. For $F_{x\,n}$ the standard peak forces were seen to be slightly higher. The differences between large and standard head size prosthesis subjects were significant $p<0.05$ in $F_{x\,n}$ for the left and the right hip joints but differing on each size. The effect size of these forces was 0.54 for the left hip joint and 0.61 for the right hip joint. None of the other directions were significant $p>0.05$.

### 3.2.2.3 Anterolateral Surgical Approach vs. Posterior Surgical Approach

Different surgery types affect different muscle groups. Consequently, surgery type has been investigated in previous studies (Foucher et al., 2011, Lugade et al., 2010) and may play a role in the muscles that are used to carry out certain movements. The subjects tested in this project are all 6 months or more post operation and should have fully recovered from surgery. However it is important to determine if there is any difference between the two groups based on surgical approach. Of the 23 subjects analysed, 14 subjects had undergone posterior surgery and only 4 anterolateral surgery. Figure 3-12 presents the average forces for the left and the right hip joints for both sets of subjects.
Figure 3-12 The average forces for anterolateral and posterior surgical approach subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the walking motion.

In Figure 3-12 it can be seen that for the left hip joint the posterior surgery approach subjects have a higher peak force in \( F_m \), \( F_y \) and \( F_z \), but there is a similar trend between the two groups. For \( F_x \) in the anterolateral surgery approach subjects show a higher force.

For the right hip joint similar results can be seen with the only major difference in the X direction. Statistical analysis of the two groups was carried out using a standard T-Test and the Mann-Whitney test.
For the left hip joint all force averages show a similar trend with posterior surgery type subjects having a higher mean in three of the four directions. However none of the forces in the left hip were found to be significant p>0.05. The forces in the Y direction did however show an effect size of 0.38 which shows a medium effect size. All the other forces had low effect sizes. The forces in the right hip joint also showed similarities between the two groups and none of the forces in the right hip joint were found to be significant p>0.05. The effect size however for \( F_{x_0} \) showed a medium effect of 0.37.

### 3.2.2.4 Total Hip Replacement (THR) vs. Birmingham Replacement (BIRM) vs. Healthy

Comparing results from the THR subjects, the BIRM subjects and the healthy subjects may indicate that THR or BIRM surgery may increase or indeed reduce the forces at the hip. Figure 3-13 shows the average forces for each group at the left and right hips joints.
The first point to note is that the healthy hip forces are the highest in all directions at both hip joints. In both the left and the right hip joints all forces seem quantitatively similar for $F_{mn}$, $F_{yn}$ and $F_{zn}$. In $F_{xn}$ a difference in peak forces occurs. To carry out statistical analysis an ANOVA and Kruskal-Wallis test were performed in the SPSS software.

At the left hip joint $F_{xn}$ was the only direction to have significant difference $p<0.05$. Gabriel’s test revealed this significant difference $p<0.05$ was between the healthy subject’s forces and the
Birmingham replacement subject’s forces, but also between the Birmingham group and the THR group. The differences in forces for the other directions were not significant p>0.05.

In the right hip joint there was a significant difference between the groups for Fx_n and Fz_n p<0.05. After carrying out Gabriel’s test the difference between the healthy and the Birmingham groups in Fx_n was significant p<0.05. In Fz_n the difference between the healthy group and the THR group was significant p<0.05. The differences in forces for the other directions were not significant p>0.05.

3.2.2.5 Healthy Hip of the THR Subjects vs. Replaced Hip of the THR Subjects

This project was concerned with investigating any difference between the sets of subjects that have just been discussed. The difference between the untreated hip and the replaced hip of the same subject is an important aspect to investigate to understand how surgery may affect a patient. Although the untreated hip cannot be guaranteed to be healthy the hip had to be pain free for testing. For this reason the untreated hip will be referred to the healthy hip. Of the 23 subjects, only 14 met the criteria. This was because 5 subjects were healthy subjects and 4 subjects had both the left and the right hips replaced. Of the 14 subjects 6 subjects had the right hip replaced and 8 subjects had the left replaced. Figure 3-14 shows the comparison between the average healthy force and the replaced force.

Figure 3-14 The average forces for the healthy (untreated) and replaced hip joints in terms of magnitude, X direction, Y direction and Z direction during the walking motion

Figure 3-14 shows a similarity between the forces on the replaced and healthy or the untreated hip joints in all force directions. To examine the statistical significance between the two hip
joints a paired T –Test and a Wilcoxon test were carried out. The results show a significant difference $p<0.05$ between the forces in the Y direction. However with a $p$ value of 0.042 this was only just significant. None of the other forces were seen to be significant.

3.2.3 Sitting Motion Results

It was stated earlier in chapter 2 that Morlock et al. (2001) believed sitting is one of the commonest everyday motions. Therefore it is important to assess the forces associated with this motion. To define a cycle for sitting the results had to be examined. All subjects carried out all of the sitting movement within 2 seconds. For sitting the most important part of the movement is when the forces on the chair start to rise and the ground reaction forces start to change. It was important to have all cycles for each subject having this change of force at the same time and therefore it was decided to take the start of the cycle one second before the forces on the chair was equal to the ground reaction force, and to finish the cycle one second later. The values were then normalised to show the % of the cycle that had been carried out. A model of sitting can be seen in Figure 3-15. The results of the sitting motion were analysed and effects of different groups were investigated. The results will be discussed using the same groups as the walking section.
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Figure 3-15 Nine stages from the sitting motion during the musculoskeletal modelling

3.2.3.1 Male vs. Female

Of the 23 subjects analysed, 15 were male and 8 female. Figure 3-16 shows the average for the male subjects compared to the female subjects for the different force directions on the left and right hip joint.
In Figure 3-16 the forces on the left hip joint are shown for the magnitude and the force directions. The male subjects showed a higher force for $F_{xm}$, $F_{xn}$ and $F_{yn}$. The right hip also shows higher forces for the male subjects in all directions. A T-Test and a Mann-Whitney test were carried out to analyse the significant differences between the two groups. For the left hip joint no differences in peak stresses were seen to be significant $p>0.05$. However the difference in forces in $F_{xn}$ had a low to medium effect size.
For the right hip joint $F_{x_n}$ was significantly different $p<0.05$. However all forces were seen to have a low to medium effect size with the X direction having a medium to large effect size.

3.2.3.2 **Standard Head Sized Prosthesis vs. Large Head Sized Prosthesis**

As mentioned previously, the size of the prosthesis head in the THR and BIRM subjects varied from subject to subject. 8 subjects had large head size prostheses and 10 subjects had standard size prostheses. The other 5 were healthy subjects. An analysis was carried out to see if any major differences occurred within any of the force directions. The average for the large and standard subjects can be seen in Figure 3-17 for the left and the right hip joints.
The forces in the left hip joint show very close averages and shows very little difference between the standard and large prostheses head groups for the left hip joint. The right hip joint does show a difference in the forces with the large head group having higher forces. The T-Test and Mann-Whitney test were used to test for significance between the two groups. For all the forces in the left and right hip joints there was no significance in the differences between the
large and standard prosthesis head size. The effect sizes for all the differences in forces were all low.

3.2.3.3 Anterolateral Surgical Approach vs. Posterior Surgical Approach
As already mentioned the surgery type chosen can affect the muscles around the hip joint and may cause a change in forces. 12 posterior surgery subjects and 4 anterolateral surgery subjects were tested. Figure 3-18 illustrates the average forces for both groups in the left and the right hip joints.
Figure 3-18 The average forces for anterolateral and posterior surgical approach subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the sitting motion.

Similar to the standard large analysis the forces at the left hip joint were seen to have very similar patterns and little differences. At the right hip joint there are differences between the forces but mainly in the magnitude of the forces and not the trend. A T-Test and Mann-Whitney test were again carried out for significant differences. The results showed that none of the differences in forces at the left or the right hip joint were significant p>0.05.
3.2.3.4 Total Hip Replacement (THR) vs. Birmingham Replacement (BIRM) vs. Healthy

The results of the sitting motion were broken into the three groups of subjects that were recruited, the THR subjects, the BIRM subjects and the healthy subjects. The average forces for each group can be seen in Figure 3-19 for the left hip joint and the right hip joint.

Figure 3-19 The average forces for the THR, BIRM and healthy subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the sitting motion

Again similar to the standard v large analysis the forces at the left hip joint show very similar patterns and little differences. At the right hip joint there are differences between the forces but
mainly in size of the force and not the trend. At this joint the Birmingham group’s forces are the higher in all force directions. Both an ANOVA and Kruskal-Wallis test were carried out to analyse if the differences in forces between groups was different. The results showed no significant differences between any of the three groups for either leg.

3.2.3.5 Healthy Hip of the THR Subjects vs. Replaced Hip of the THR Subjects

14 of the subjects had only one replacement in either the left or the right hip joints. Figure 3-20 illustrates the average forces in the healthy hip compared to the replaced hip.

![Figure 3-20 The average forces for healthy (untreated) and replaced hip joints in terms of magnitude, X direction, Y direction and Z direction during the sitting motion.](image)

As for walking, the average forces for sitting show similar trends and force magnitudes in the Y and Z directions. The X direction shows a difference in force between the healthy or the untreated hip joint and the replaced hip joint. A paired T-Test and Wilcoxon test were carried out to test significance. No differences in any of the forces in any of the directions were found to be significant. Although $F_{x}$ in Figure 3-20 appears to have a big difference between the healthy values and the replaced values these values are small and there is a large range in values for both the healthy and replaced groups.

3.2.4 Standing Motion Results

As with the sitting movement Morlock et al. (2001) stated standing was one of the most reoccurring everyday activities carried out in day to day living. Therefore it is important to assess the forces associate with this motion. The same method and theory behind defining the sitting movement was use to define the standing movement as they are in essence the opposite
of each other. However the forces associated with each movement differ. As for sitting the most important part of the standing movement is when the forces on the chair and the ground reaction forces start to change. The time for all the standing models was 2 seconds therefore the results did not need to be interpolated. The values were then normalised to show the % of the cycle that has been carried out. A model of standing can be seen in Figure 3-21. The results of the standing motion were analysed and effects of different groups were investigated. The results will be discussed using the same groups as the walking and sitting sections.
Figure 3-21 Nine stages during the standing motion from the musculoskeletal modelling
3.2.4.1 Male vs. Female

All 23 subjects carried out the standing movement. As mentioned before 15 male and 8 females participated in the study. The average forces for both males and females can be seen in Figure 3-22 for the left and the right hip joint.

Figure 3-22: The average forces for male and female subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the standing motion.

Figure 3-22 above illustrates the female group had higher average forces in the \( F_y \) and \( F_z \) for the left hip joint however \( F_x \) of the left hip joint and all the forces at the right hip joint had...
higher forces in the male group. There is a similarity between both trends. A T-test and Mann-Whitney test were carried out to test significance. There was no significance \( p > 0.05 \) found in differences between the male and female groups, however the effect size of the difference between \( F_{x,n} \) for the left and right hip joints were 0.39 and 0.41 respectively. These represent a medium to large effect size and would indicate that both were close to being significant.

### 3.2.4.2 Standard Head Sized Prosthesis vs. Large Head Sized Prosthesis

For the standing movement, the size of the prosthesis head was broken into two groups, a large head group and a group of standard head sized prostheses. 10 prostheses had standard sized heads and 8 had large sized heads. The two groups were compared to each other and the results were averaged together for each group. The comparisons can be seen in Figure 3-23 for the magnitude, X direction, Y direction and Z direction.
Figure 3-23 The average forces for the standard and large head sized prostheses subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the standing motion.

The forces at the left hip joint are equal for both prosthesis head size in the X and Z directions, while the Y direction shows the standard head sized group to have a higher peak force. For the right hip joint, $F_y$ and $F_z$ for both groups are equal, with $F_x$ showing the large head size to have a higher peak force. A T-Test and Mann-Whitney test were carried out and no differences in force were found to be significant $p>0.05$. The effect size for $F_y$ in the left hip and $F_x$ in the right hip were 0.3 and 0.31 respectively. This shows a medium effect size.
3.2.4.3 Anterolateral Surgical Approach vs. Posterior Surgical Approach

The posterior approach and the anterolateral approach to surgery were assessed for the standing movement. As previously mentioned 14 subjects underwent surgery using the posterior type and only 4 underwent the anterolateral approach. The averages were calculated and comparisons were made between the two groups. Figure 3-24 shows the comparisons between the groups for the left hip joint and the right hip joint.

**Figure 3-24** The average forces for anterolateral and posterior surgical approach subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the standing motion.
For both the left and the right hip joints the forces for the posterior and the anterolateral groups are similar. In the statistical analysis the difference in forces between the two groups was found not be significant in any direction. The effect size was also low in all force directions.

3.2.4.4 Total Hip Replacement (THR) vs. Birmingham Replacement (BIRM) vs. Healthy
The three groups, THR subjects, BIRM subjects and healthy subjects all carried out the standing movement. Any significant difference in forces between these groups may indicate that THR or BIRM surgery may increase or indeed reduce the forces at the hip. Figure 3-25 show the average forces for each group for the left and right hip joints.
Figure 3-25 The average forces for the THR, BIRM and healthy subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the standing motion.

Figure 3-25 illustrates that for all 3 directions in the left hip joint the forces for the BIRM subjects are slightly lower than the other two groups. However this is then reversed in the right hip joint where the BIRM subjects are higher in both $F_x$ and $F_y$. In the ANOVA and Kruskal-Wallis test no significant differences were found in any of the forces $p>0.05$. 

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3.2.4.5 Healthy Hip of the THR subjects vs. Replaced Hip of the THR Subjects

The healthy hip in the replacement subjects was compared to the replaced hip in the same subjects. 14 subjects were analysed in this way. The averages for each hip can be seen in Figure 3-26.

![Figure 3-26 The average forces for healthy (untreated) and replaced hip joints in terms of magnitude, X direction, Y direction and Z direction during the standing motion](image)

Both $F_{y_h}$ and $F_{z_h}$ show very similar trends and magnitudes for the replaced and healthy hips. $F_{x_h}$ is seen to be higher in the replaced hip. The results of a paired T-Test and Wilcoxon's test showed no significance in the differences between groups in any of the force directions. The differences in X direction forces may be due to the forces being very low (less than 10% of the magnitude) compared to the Y direction so the chances of noise or other errors are higher. Due to these forces being low the variety between subjects in the same group is a lot higher.

3.2.5 Up Steps Motion Results

Going up steps may not be the most common motion carried out in everyday life Morlock et al. (2001) found that on average only 0.4% of the day is spent going up or down stairs and that on average 1 in every 37 steps taken is up or down steps. As the up steps movement was not a cyclic motion it was not easy to determine one cycle. In several previous studies such as Bergmann et al. (1995a) the up steps cycle started at the heel strike on the first step. However it is felt that this did not take into account the lift and extension of the leg before this heel strike and therefore did not account for the full cycle. In this study it was decided that the cycle was started at the heel strike on the ground before starting to ascend the steps. This can be seen in a
LifeMod model in Figure 3-27. The cycle was finished half a second after the ground reaction forces for the left foot and the right foot are equal. This indicated that the subject had stopped climbing and both feet were on the top step. The half second was added as it was seen that most, if not all, subjects still had a small motion after this time to balance themselves before being stable. The finishing position for one cycle of the up steps movement can be seen in Figure 3-27. The average time for the up steps movement was 3.31 seconds per cycle. The longest time for one cycle was 5.3 seconds. The shortest time for one cycle was 2.15 seconds.

Figure 3-27 Six stages during the up steps motion from the musculoskeletal modelling
Similar to the walking movement, each of the up steps movement models simulated were of different length and therefore had to be interpolated so that they could be compared to each other. This was carried out in the freeware software Octave. Data was interpolated with regards to the longest up steps cycle and then normalised from time to the % cycle completed.

3.2.5.1 Male vs. Female

Out of the 23 subjects only 20 subjects completed the up steps movement. This was due to unavailability of the aerobic steps at the time of testing. Out of the 20 subjects that were analysed 12 were male subjects and 8 female. A comparison of the average forces at both the left and right hip joints can be seen in Figure 3-28.
Figure 3-28 The average forces for male and female subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the up steps motion

From the above graphs it can be seen that there is very close similarity between the male and female groups both in the left and in the right hip joints. The two groups are comparable in not only the magnitudes of the forces but also in the trends that they follow. A T-Test and a Mann-Whitney test were carried out to investigate if the differences in forces between the two groups were significant. The results showed that there were no significant differences between the two groups. Looking at the similarity in the graphs in Figure 3-28 this was to be expected.
3.2.5.2 Standard Head Sized Prosthesis vs. Large Head Sized Prosthesis

For the 20 subjects that carried out the up steps movement 2 were healthy subjects that did not have any prosthesis. Of the 18 remaining, 10 had a standard sized prosthesis head and 8 had large sized prosthesis head. The forces from these two groups were averaged together and then compared to each other. The comparison of forces can be seen in Figure 3-29.

Figure 3-29 The average forces for the standard and large head sized protheses subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the up steps motion.
On examination of the graphs no major differences can be seen between the standard head and large head groups. Both groups follow the same trend with no real differences. From the results of a T-Test and a Mann-Whitney test the differences in peak forces at the left hip joint were all found to not be significant $p>0.05$. For the right hip joint the statistical analysis results showed significance $p<0.05$ in the difference in peak forces for $F_x$. The effect size of this difference in force was 0.49 which shows a large effect size.

### 3.2.5.3 Anterolateral Surgical Approach vs. Posterior Surgical Approach

18 subjects which had either the posterior or anterolateral approach to surgery carried out the up steps movement. 14 of these were in the posterior approach groups and only 4 in the anterolateral approach group. Figure 3-30 shows a comparison between the average forces in these two groups in the left and the right hip joints.
Figure 3-30 The average forces for the anterolateral and posterior surgical approach subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the up steps motion.

For the left hip joint there was a difference in $F_{x_l}$ and $F_{y_l}$. The trends were similar but the magnitudes of the peaks were higher in both directions. From the statistical results no significant differences were found for the left hip joint. However the effect size of the differences in forces between groups in the X direction had a medium effect size. At the right hip joint the only major difference is seen in $F_{x_r}$. The other forces are quite similar. As with the left hip joint, none of the differences in forces for the right hip joint were significant $p>0.05$. 
The effect size of the differences in force between the posterior and anterolateral groups in the X direction were medium to high. This would indicate that these are close to being significant.

3.2.5.4 Total Hip Replacement (THR) vs. Birmingham Replacement (BIRM) vs. Healthy

3 healthy subjects did not carry out the up steps movement due to unavailability of the steps at the time of testing. This gave 20 subjects, 12 THR subjects, 6 BIRM subjects and 2 healthy subjects. The average forces of all three groups were calculated and these are shown in Figure 3-31 for the left and right hip joints. Due to only having 2 subjects for the healthy group statistical analysis was not carried out on this group.
Figure 3-31 The average forces for the THR, BIRM and healthy subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the up steps motion.

From the graphs in Figure 3-31 it can be seen that the trends for each group at the left hip joint are similar in all directions. However the magnitude of $F_{X_n}$ is different with the BIRM subjects having lower peak forces than the THR and healthy subjects. In the statistics the results showed a significant $p<0.05$ difference between the peak forces in the BIRM subjects and the THR subjects for the X direction. The healthy subjects were not included in the statistical analysis so no other comparisons could be drawn. For the right hip joint, the BIRM and THR subjects are
similar in all force directions and this is reinforced by the fact that none of the differences in peak forces for these groups are significant $p>0.05$. There is a difference between the healthy subjects and the other two groups in $F_{x_n}$. The healthy group only includes two subjects and due to the small forces involved may have included several errors. No statistical analysis was available to compare the groups and therefore no significant results can be taken from this.

### 3.2.6 Down Steps Motion Results

As with the up steps movement it was difficult to determine one cycle of this movement. The cycle was started half a second before the ground reaction force of either the left or right leg started to decrease. This represents the leg starting the descent and can be seen in Figure 3-32. The first heel strike after both feet have left the steps was the end of the cycle. This can be seen in Figure 3-32. The average time for one cycle of the down steps movement was 2.68 seconds. The longest time for one cycle taken was 3.72 seconds. The fastest cycle was 2.05 seconds.
Figure 3-32 Six stages of the Down steps motion from the musculoskeletal modelling

Similar to the walking and up steps movement, each of the down steps movement models simulated were of different length and therefore had to be normalised so that they could be compared to each other. This was carried out in the freeware software Octave. Data was interpolated with regards to the longest down steps cycle and then normalised from a time scale to a % of a cycle scale.
3.2.6.1 Male vs. Female

Out of the 23 subjects only 18 subjects completed the down steps movement. This was due to unavailability of the aerobic steps at the time of testing. Out of the 18 subjects that were analysed 11 were male subjects and 7 female. The comparison of the average forces at both the left and right hip joints can be seen in Figure 3-33.

Figure 3-33 The average forces for male and female subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the down steps motion
Figure 3-33 above illustrates a similarity between the male and female groups both in the left and right hip joints. The two groups have similarity in not only the magnitudes of the forces but also in the trends that they follow. $F_{x_n}$ shows a difference in direction of their peak force. A T-Test and a Mann-Whitney test were carried out to investigate if the differences in forces between the two groups were significant. The results showed that there were no significant differences between the two groups in any of the directions. However on the left hip joint the effect size of the differences in the X direction was low to medium and the Y direction in the right hip joint had a medium effect size.

3.2.6.2 Standard Head Sized Prosthesis vs. Large Head Sized Prosthesis

For the 18 subjects that carried out the down steps movement 2 were healthy subjects that did not have any prosthesis. Of the 16 left, 10 had a standard sized prosthesis head and 6 had large sized prosthesis head. The forces from these two groups were averaged together and then compared to each other. The comparison of forces can be seen in Figure 3-34.
Figure 3.34 The average forces for the standard and large head sized prostheses subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the down steps motion.

On examination of the graphs no major differences can be seen between the standard head and large head groups in most of the directions. However there is a difference between the two groups in \( F_{x_n} \) on the left hip joint. Both groups follow the same trend and have similar magnitudes. From the results of a T-Test and a Mann-Whitney test the differences in peak forces in the X direction for the left hip joint were found to be significant \( p<0.05 \). The forces in
the right hip joint were seen to not be significant $p>0.05$. However a medium effect size was seen in $F_{nx}$.

3.2.6.3 Anterolateral Surgical Approach vs. Posterior Surgical Approach

16 subjects that had either the posterior or anterolateral approach to surgery carried out the down steps movement. 12 of these were in the posterior approach groups and 4 in the anterolateral approach group. Figure 3-35 shows a comparison between the average forces in these two groups in the left and the right hip joints.
Figure 3-35 The average forces for the anterolateral and posterior surgical approach subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the down steps motion.

For the left hip joint there was a difference in $F_{x_n}$. The trends were similar but the magnitudes of the peaks were higher. From the statistical results no significant differences were found for the left hip joint. At the right hip joint the only major difference is $F_{y_n}$. The other forces are quite similar. As with the left hip joint, none of the differences in forces for the right hip joint were significant $p>0.05$. However the effect size of the differences in force between the posterior and anterolateral groups in both $F_{x_n}$ and $F_{y_n}$ were medium.
3.2.6.4 **Total Hip Replacement (THR) vs. Birmingham Replacement (BIRM) vs. Healthy**

3 healthy subjects and 2 Birmingham subjects did not carry out the down steps movement due to unavailability of the steps at the time of testing. This gave 18 subjects, 12 THR subjects, 4 BIRM subjects and 2 healthy subjects. The average forces of all three groups were calculated and these are shown in Figure 3-36 for the left and right hip joints. Due to having only 2 subjects for the healthy group statistical analysis was not carried out on this group.

![Figure 3-36](image)

**Figure 3-36** The average forces for the THR, BIRM and healthy subjects at the left and right hip joints in terms of magnitude, X direction, Y direction and Z direction during the down steps motion.
From the graphs in Figure 3-36 it can be seen that the trends in the left hip joint are similar in all directions. However the magnitudes of the forces are different between the groups. In the left hip joint THR subjects have a lower peak force than the BIRM and Healthy subjects. In the statistical analysis the results showed that there was not a significant p>0.05 difference between the peak forces in the Birmingham subjects and the THR subjects. The healthy subjects were not included in the statistical analysis so no other comparisons could be drawn. For the right hip joint, the Birmingham and THR subjects are similar in all force directions and this is reinforced by the fact that none of the differences in peak forces for these groups were significant p>0.05. There is a difference between the healthy subjects and the other two groups in Fx_n. The healthy group only includes two subjects and due to the small forces involved may have not give a proper representation. No statistical analysis was available to compare the groups and therefore no significant results can be taken for this.

3.3 Healthy and Replaced Hips of the Same Subject

All the above sections have compared the results of the left hip and the right hip between the different groups. In the groups of subjects that have one replaced hip it is possible to also compare the forces in the healthy or untreated hip to the forces in the replaced hip. There were 14 subjects that had one hip replaced and this group is discussed below.

3.3.1 Male vs. Female

Of the 14 subjects that had one hip replaced, 9 were male and 5 were female. Statistical analysis was carried out between the peak magnitude forces for each of these groups and also between the peak forces in the three force directions (Fx, Fy and Fz). This was done using a standard T-Test and the Mann-Whitney test. The peak forces of the healthy male hip subjects were compared to the healthy hips of the female subjects and the forces for the replaced hip also compared between the male and female groups. However, for the 5 motions carried out in the motion analysis none were found to be significantly different p>0.05 for either the healthy hip joint or the replaced hip.

3.3.2 Standard Head Sized Prosthesis vs. Large Head Sized Prosthesis

The 14 subjects with one hip replaced were made up of 7 standard sized heads and 7 large sized heads subjects. In the statistical analysis one force direction was found to be significantly different for the walking motion p<0.05. This was in Fx_n for the replaced hips. The subjects with a standard prosthesis head size were seen to be significantly higher in Fx_n for their replaced hip joints compared to their large prosthesis head sized counterparts. For the up steps motion two force directions were found to be significantly different p<0.05. These were Fx_n and Fy_n. Fx_n, similar to Fx_n in the walking motion, was significantly higher in the standard prosthesis
head size group, but the large prosthesis head size group was found to have a higher peak $F_{y_n}$. None of the other motions had significant differences in any force directions.

### 3.3.3 Anterolateral Surgical Approach vs. Posterior Surgical Approach

Any differences in results between groups of different surgery type is of great interest to surgeons in the area of hip replacement surgery as surgeons, like engineers, are trying to improve any parameter that may affect survival rates. Only 3 of the subjects that were available in this category had undergone anterolateral surgery, the other 11 had all undergone the posterior approach to hip surgery. For the replaced hip joint no force direction was found to be significantly different $p<0.05$ in any of the 5 motions carried out. For the healthy hip $F_{x_n}$ during the down steps motion was found to be of significant difference $p<0.05$ with the anterolateral approach having the higher force in this direction. All of the other forces in all the directions were not found to be significantly different. However due to the low numbers involved for the anterolateral group finding a significant value may not show an accurate indication and more numbers would be needed to reach any major conclusions between these two groups.

### 3.3.4 Total Hip Replacement (THR) vs. Birmingham Replacement (BIRM)

The final comparison for this group of subjects was between the THR subjects and the BIRM subjects. Of the 14 subjects 5 had undergone surgery and had the BIRM hip replacement implanted and the other 9 had undergone THR surgery. For the replaced hip $F_{x_n}$ was once again found to be significantly different during the walking motion. The force in the same direction during the up steps motion was found to also be significantly different. In both scenarios the THR subjects were seen to have a higher peak force. In the healthy hip the down steps motion was found to have significantly different forces in both $F_{m_n}$ and $F_{z_n}$. However unlike the replaced hips the BIRM subjects were found to have higher forces in both $F_{m_n}$ and $F_{z_n}$. This motion was the only one to have any significant differences between the forces.

### 3.4 Loading Configurations

One of the main aims of this project is to examine the mechanical loading using both numerical and experimental models. To carry out either of these procedures a realistic loading configuration needs to be applied that would represent the forces that act at the hip joint during everyday motions such as the motions analysed in this chapter. The force data from the motion analysis study can therefore be used to provide the loading configurations for the numerical and experimental modelling. The preparation of this loading configuration is discussed below.

#### 3.4.1 Averaging

To calculate a loading configuration for an implanted prosthesis only the forces from the THR and BIRM replacement subjects were used, ie as the inclusion of the healthy hips would give an unrealistic loading configuration. Five subjects were healthy, leaving 18 subjects available to
create the average curves. An average curve was created for each of the 5 motions carried out. These curves were created by taking the force curves from the replaced hip for each of the 18 subjects and getting the average curves following the process for each of the groups discussed previously. Where any subject had both hips replaced the average between the two hips was used. The resulting curves for the walking motion can be seen in Figure 3-37.

![Graphs of average curves for different motions](image)

**Figure 3-37** The average curves during the walking motion for the Magnitude, X, Y and Z directions taken from all THR and BIRM subjects

### 3.4.2 Static and Dynamic Modelling

For static modelling, which was used in the numerical modelling, the peak values from the average curves were used to represent the forces at the hip joint. To do this the peak values in the force magnitude recorded on the average curves graphs were identified and multiplied by Body Weight (BW). The X, Y and Z direction forces at this point were then used. The BW was calculated by taking the average weight of the 18 THR and BIRM subjects. These forces were then used for the loading configuration in the static numerical models.
Figure 3-38 The static loading configurations calculated from all the THR and BRIM subjects in the X, Y and Z directions for each of the 5 motions

For the experimental rig a dynamic load was required. However the electro-pneumatic regulators that controlled the pneumatic pistons on the rig could only change the applied force 3 times a second (this is discussed further in the experimental chapter). The average time of one gait cycle for all the subjects was 1.24 seconds and therefore the forces at 4 time points along the cycle were taken. As the peak force was the most important force this was taken as the first value. The rest of the values were then taken at 25% intervals of the rest of the cycle i.e. if the peak force was at 15% along the gait cycle the values used for the dynamic forces were taken at 15%, 40%, 65% and 90%.
Figure 3-39 The dynamic loading configuration calculated from the walking motions for all THR and BIRM subjects.

Figure 3-40 The dynamic loading configuration calculated from the up steps motion for all THR and BIRM subjects.
3.4.3 Comparison to Literature

Motion analysis has been carried out in previous studies and many of these have been discussed in the chapter 2. Gait analysis can vary between subjects and in a study by Brand et al. (1994) it was even shown to differ between different steps of the same subject. It is important however to make comparisons between the data recorded in this study and other previously published data.

As discussed in the literature review, the Bergmann group have carried out a substantial amount of research in gait analysis and recording hip joint forces. Figure 3-41 shows the results of 4 papers from this research group and compares them to the results of the average walking curve from this study. There is a good agreement between the results of the present study and the results seen in these published papers. One of the main reasons the results vary so much is due to the walking speed of the subjects. Slower gait patterns generally result in lower hip joint forces. During the present study the subjects were asked to walk at a pace comfortable to them, as everyday activities would be carried out at the comfortable pace. Figure 3-42 shows the % difference between the present study and the 4 published papers from the Bergmann group. It can be seen the results vary and that results from the same motion analysis lab can differ. This also indicated that motion analysis can have a wide variation between subjects but also within studies. It is important to note however that the results found in this study fall within the mid range of the four other previous studies.

Figure 3-41 The results of the present study compared to results from the Bergmann group. (Bergmann et al., 1993, Bergmann et al., 2001, Bergmann et al., 1995a, Bergmann et al., 1995b).
3.5 Discussion

This chapter illustrated the different forces for each of the groups in all the loading directions. The main aim of this motion analysis was to determine average forces at the hip joint for THR and BIRM subjects during different everyday activities. This data can be used as the hip contact forces in both the experimental and numerical modelling sections of this thesis. Data from 18 subjects was used to create the hip joint forces for the 5 different movements. Another aim of the motion analysis was to investigate if any differences occurred between any of the groups just discussed. As seen in the results most forces had good correlation between the groups.

The forces in the X direction, \( F_x \), were seen to have the most occurrences of statistical differences. All groups compared were found to have significant differences in \( F_x \) for one of the motions carried out. \( F_x \) is the force that is perpendicular to the sagittal plane acting in towards the hip joint and can be seen in Figure 3-6. This force generally is small in magnitude in the motions carried out, is significantly less than the forces in the other two directions and therefore would not create major differences to the magnitude of the force. However, significant differences could still create contrasting results between groups. To investigate if these findings cause differences in results the loads are applied to numerical models.

The two forces that were found to be significantly different in \( F_y \) and \( F_z \) between the left and the right hip were both found in the walking motion. A significant difference was seen between healthy and THR subjects in the right hip joint for \( F_y \). This demonstrates that there is a significant difference between the healthy and THR subjects and that \( F_y \) is significantly lower.
in the THR subjects at the right hip. The difference was however not seen to be significant at the left hip joint. However, a significant difference was seen in $F_{z_n}$ for walking between the healthy and replaced hips in the replacement subjects. The Z direction is the force perpendicular to the ground. The healthy hip was seen to have a higher force during walking than the replaced hip. One reason for this may be due to the subjects still not trusting the replaced hip joint and therefore not applying as much of a force to the joint while walking. Any differences in $F_{z_n}$ may also have implications to prosthesis loosening as it is stated by Fowler et al. (1988) and Colgan et al. (1996a) that implant fixation may be affected by loads in this direction. This issue will be investigated further in the experimental and numerical modelling chapters.

When the groups examined in this study are compared the male vs. female comparison only showed significant differences at the right hip joint in $F_{x_n}$ for the sitting motion. The comparison that demonstrated the greatest differences was the small head vs. large head comparison. This had significant differences in the walking, up steps and down steps motions. Both the left and the right hip joint groups were seen to be significantly different for the walking motion with a definite difference between the two groups with the standard group having higher forces in this direction. However, the large prosthesis head sized group had higher forces in the other two directions and therefore had only a slightly higher magnitude. Interestingly, there was no significant difference between the posterior approach and anterolateral approach groups. This indicates that the different surgery approaches should not affect the hip joint forces after full recovery. This study is limited to 18 treated subjects and only 4 of these subjects underwent surgery using the anterolateral approach so therefore to make any major conclusions to about these groups would be premature without further testing.

The healthy subjects were compared to the THR subjects and the BIRM subjects. Several significant differences were found between the groups, all during the walking motion. As mentioned earlier the right hip forces in $F_{y_n}$ were significantly different between the healthy subjects and the THR subjects. The healthy subject’s $F_{x_n}$ for the left and right hips joints were significantly different to the BIRM subjects. The healthy subjects had far higher forces than the BIRM subjects albeit $F_{x_n}$ is small but there is a 4 fold difference in peak average force. This difference is then further backed up with $F_{x_n}$ in the left hip joint also being significantly different between the THR subjects and the BIRM subjects, although the difference between these forces is only just over 2 fold. The other force significantly different between the THR group and the BIRM group is $F_{x_n}$ on the left hip joint while carrying out the up steps motion.
There were 20 significant differences within the analysis and these can be seen in Table 3-5. The majority of these are in $F_x$. Therefore, the effect this force has on the implant will be investigated using numerical modelling in future chapters. However, the statistical differences in each of the significant forces in $F_x$ were not investigated as the forces in the X direction were generally smaller than the other two directions. There were 5 significantly different forces that were not in the X direction and these were:

- Fy in the walking motion (Healthy v Replaced hips)
- Fz in the right hip for the walking motion (Healthy v THR Subjects)
- Fy in the replacement hips for the up steps motion (Standard v Large)


- Fm in the healthy hip for the down steps motion (BIRM v THR)
- Fz in the healthy hip for the down steps motion (BIRM v THR)

These 5 significant differences are in 4 different motions (Fm and Fz in healthy hip for the down steps are both in the same motion). Therefore, these 4 groups will be investigated further in chapter 7 to examine if any differences in stresses along the surface will occur.
CHAPTER 4

Numerical Modelling: Parameters and Boundary Conditions
Chapter 4  Numerical Modelling: Parameters and Boundary Conditions

4.1 Introduction
The parameters used within a numerical model are very important as these parameters can influence the model and results generated. This chapter will discuss all the sections of defining the numerical model, the parameters that were used and why these parameters were used.

4.1 Governing Equations for Finite Element Modelling
The development of numerical methods to solve solid mechanics problems took two distinctive paths, involving engineering and mathematical approaches. The two approaches involved the development of a number of techniques including structural analogue substitution, direct continuum elements, variational methods, weighted residuals, finite differences, variational finite difference and piecewise continuous trial functions. A combination of these methods led to the present day FEA. Mathematically, the FEM is used for finding an approximate solution to partial differential equations as well as integral equations such as the heat transport equation. The solution approach is based either on eliminating the differential equation completely (steady state problems), or rendering the partial differential equations into an equivalent ordinary differential equation, which is then solved using standard techniques such as finite differences.

In solving partial differential equations the primary challenge is to create an equation which approximates the equation to be studied, but which is numerically stable, meaning that errors in the input data and intermediate calculations do not accumulate and cause the resulting outputs to be meaningless. There are many ways of doing this, all with advantages and disadvantages. FEA is a good choice for solving partial differential equations over complex domains or when the desired precision varies over the entire domain (Callanan, 2009).

The governing equation used for finite element modelling can be simplified to:

\[ fKU = f \]  \hspace{1cm} (4.1)

Where U is the vector of unknown nodal displacements, f is the nodal force vector and K is the finite-element stiffness matrix. The solution of equation 3.5 is obtained by any appropriate numerical method, which involves the inversion of the stiffness matrix.

\[ U = K^{-1}f \]  \hspace{1cm} (4.2)

To include dynamic effects the equation becomes:

\[ M\ddot{U} + KU = f \]  \hspace{1cm} (4.3)
Where $M$ is the finite element mass matrix and $\ddot{U}$ are the nodal accelerations. However only static models are being considered in the study and therefore equations 3.5 and 3.6 are the governing equations used.

### 4.2 Creation of Femur and Prosthesis Geometries

To compare results to both *in vivo* and experimental studies, similar and therefore realistic geometries are needed. As the main comparisons will be made with the experimental setup the realistic geometry of Sawbone femurs was used. The Sawbone femurs are fourth generation composite femur bones (#3403 Sawbones Europe AB, Sweden) and are used in place of cadaver bones. They are designed to replicate the physical behaviour of human cadaver bones. They were used in the experimental part of this project as they give less variability than cadaver specimens and therefore the experiments would be repeatable and results comparable. These are discussed further in the experimental chapter. The Sawbone geometry file was taken from the Biomechanics European Laboratories repository (BEL Repository). This .stp file was of a full 3rd generation solid composite femur. ProEngineer 4.0 (PTC, Needham, MA, USA) was used to cut along the neck as would be the case in total hip replacement surgery. The prosthesis was positioned into the Sawbone femur using datum planes to fully constrain the part. Once the prosthesis was in position the cut out tool was used to remove the prosthesis geometry from the femur geometry creating the cavity. This file was then exported into the Abaqus software for the numerical analysis.

### 4.3 Grid Independence

There were three parts in the numerical models these were the prosthesis, the cortical bone and the cancellous bone. Two parts were used for the cortical and cancellous bone as they had different material properties. Grid independence was carried out on each of the three geometries to demonstrate that the results from the models were not dependent on or affected by the density of the mesh. Five models varying in mesh density were made for each part. The seeding density was varied to change the amount of elements and the density of the meshes in each model. A criterion of 5% difference between the stresses in differently meshed geometries was used to confirm the grid independence (Wong *et al.*, 2005). This criterion was used for all three parts in the model.

#### 4.3.1 The Prosthesis

The seeding density was varied between 5 and 2.5 for the implant geometry. A seeding density of 5 gave the lowest mesh density (10,131 elements) while 2.5 gave the highest mesh density (23,648 elements).
Figure 4-1 The prosthesis geometry with the red line indicating the area used for the Grid independence study

Figure 5-2 shows the Von Mises stresses for each of the mesh densities. A ±5% difference criterion was used to confirm the grid independence (Wong et al., 2005). From Figure 4-3 it can be seen that a seeding value of 3 gave a max % difference under the independence criteria. Therefore a seeding density of 3 was used for the prosthesis resulting in 17,835 elements.

Figure 4-2 Von Mises stress along the path for each mesh in the prosthesis part
4.3.2 The Cortical Bone

The seeding density used for the cortical bone was also varied between 5 (23,690 elements) and 3 (68,924 elements). Figure 4-4 shows the cortical bone geometry and the chosen line for the grid independence.

Figure 4-4 The cortical bone geometry and the chosen line for the grid independence

Figure 4-5 shows the Von Mises stress for each of the mesh densities. A ±5% difference criterion was used to confirm the grid independence as with the previous part (Wong et al., 2005). The results showed that a seeded mesh density of 4 met the % difference criteria. For this reason it was decided to use this seeding density for the model. This seed density will have 36,032 elements. The % differences along the chosen path can be seen in Figure 4-6.
Figure 4-5 Von Mises stress along the line in the cortical bone for each mesh

Figure 4-6 The % difference in Von Mises stress between the different meshes for the cortical bone

4.3.3 The Cancellous Bone

It was important to have a good mesh on the cancellous bone part as this is the area in contact with the prosthesis and is also the area in which any possible loosening or instability will occur. The seed densities were varied between 5 (14,304 elements) and 2 (78,330 elements).
Figure 4-7 The cancellous bone geometry with the red line indicating the area used for the grid independence

Figure 4-8 illustrates the Von Mises stresses for each of the seeded meshes. It was found that a seeding density of 2.5 met the grid independence criteria. This was therefore chosen as the seeding density for this part. This part then consisted of 49,611 elements. Figure 4-9 shows the % difference between the different meshes along the chosen path.
From the grid independence study the three geometries were meshed with a seed density that was grid independent. The prosthesis was meshed with a seed density of 3 giving 17,835 elements, the cortical bone was meshed with a seed density of 4 giving 36,032 elements and the cancellous bone was meshed using a seed density of 2.5 giving 49,611 elements. The final
model consisted of these three parts using these meshes and was therefore grid independent. The final model consists of 103,478 elements.

![Assembly of the prosthesis, cortical bone and the cancellous bone](image)

**Figure 4-10** The assembly of the prosthesis, cortical bone and the cancellous bone

### 4.4 Material Properties

Three materials were assigned to the three parts. All three materials were assumed to be homogenous isotropic materials for the numerical analysis in this study. The prosthesis was titanium and was assigned a Young’s Modulus of 110 GPa and Poisson’s Ratio of 0.32 (Jonkers *et al.*, 2008, Kayabasi and Erzincanli, 2006). The cortical bone was modelled with a Young’s Modulus of 17 GPa and a Poisson’s Ratio of 0.3 (Lennon *et al.*, 2007, Mann *et al.*, 2007). The final material was the cancellous bone. This was modelled with a Young’s Modulus of 1.5 GPa and a Poisson’s Ratio of 0.3. Table 4-1 shows the material properties used in this study.
4.5 Loads Applied

4.5.1 Hip Joint Force

It was seen in chapter 2 that studies have been carried out to investigate hip joint forces \textit{In vivo} using instrumented prostheses, and calculated, using musculoskeletal modelling. In an experimental setup dynamic loads should be applied to the model to replicate daily activities such as walking and stair climbing. However for numerical modelling many studies use peak static loads and this approach has been adopted for this study. A static load in the X, Y, and Z directions calculated from the musculoskeletal modelling results presented in chapter 3 earlier was used. Table 4-2 again presents the average peak forces for the 5 motions in the X, Y and Z directions.

<table>
<thead>
<tr>
<th>Motion</th>
<th>Fx (N)</th>
<th>Fy (N)</th>
<th>Fz (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walking</td>
<td>-400</td>
<td>-820</td>
<td>2060</td>
</tr>
<tr>
<td>Up Steps</td>
<td>-243.864</td>
<td>-752.629</td>
<td>1932.574</td>
</tr>
<tr>
<td>Down Steps</td>
<td>-92.369</td>
<td>-240.281</td>
<td>1303.208</td>
</tr>
<tr>
<td>Sitting</td>
<td>-595.046</td>
<td>-1211.91</td>
<td>720.53</td>
</tr>
<tr>
<td>Standing</td>
<td>-470.862</td>
<td>-1323.42</td>
<td>791.4838</td>
</tr>
</tbody>
</table>

4.5.2 Maximum Hip Joint Forces

The maximum forces found in the musculoskeletal modelling chapter for all 5 everyday motions carried out were assessed to see the differences and investigate what maximum stresses were involved in everyday movements and the effect of these loads in numerical models. These forces can be seen in Table 4-3.

<table>
<thead>
<tr>
<th>Motion</th>
<th>Fx (N)</th>
<th>Fy (N)</th>
<th>Fz (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walking</td>
<td>-223.08</td>
<td>-1653.2</td>
<td>4151.66</td>
</tr>
<tr>
<td>Up Steps</td>
<td>-419.06</td>
<td>-3487.3</td>
<td>4339.46</td>
</tr>
<tr>
<td>Down Steps</td>
<td>-1387.94</td>
<td>-592.09</td>
<td>5174.57</td>
</tr>
<tr>
<td>Sitting</td>
<td>-1498.27</td>
<td>-2728.84</td>
<td>1454.46</td>
</tr>
<tr>
<td>Standing</td>
<td>-1086.99</td>
<td>-2879.91</td>
<td>1564.38</td>
</tr>
</tbody>
</table>
4.5.3 Loads for Significantly Different Comparisons

In Chapter 3, twenty comparisons were found to be significantly different. Fifteen of these significant forces were found to be in the X direction and these forces were found to be substantially lower than the other two force directions. Therefore, five of these significant forces were chosen to be investigated further. These forces were applied to numerical models to investigate if the differences in forces varied the stress values and possibility of aseptic loosening. The forces for each of these models are seen in Table 4-4.

<table>
<thead>
<tr>
<th>Motion</th>
<th>Leg</th>
<th>Comparison</th>
<th>Fx (N)</th>
<th>Fy (N)</th>
<th>Fz (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Walk Fy</td>
<td>-</td>
<td>Healthy</td>
<td>-316.4</td>
<td>-824</td>
<td>2277</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Replaced</td>
<td>-406.4</td>
<td>-977.1</td>
<td>2487</td>
</tr>
<tr>
<td>Walk Fz</td>
<td>Right</td>
<td>Healthy</td>
<td>-766.3</td>
<td>-1067</td>
<td>3210</td>
</tr>
<tr>
<td></td>
<td></td>
<td>THR</td>
<td>-442.3</td>
<td>-873.9</td>
<td>2270</td>
</tr>
<tr>
<td>Up Steps Fy</td>
<td>Replaced</td>
<td>Standard</td>
<td>-514.2</td>
<td>-1201</td>
<td>2269</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Large</td>
<td>-310.9</td>
<td>-1703</td>
<td>2266</td>
</tr>
<tr>
<td>Down Steps Fz and Fm</td>
<td>Healthy</td>
<td>BIRM</td>
<td>-273.6</td>
<td>-511.3</td>
<td>2013</td>
</tr>
<tr>
<td></td>
<td></td>
<td>THR</td>
<td>-274.1</td>
<td>-376.6</td>
<td>1595</td>
</tr>
</tbody>
</table>

4.5.4 Muscles Forces

As already discussed in chapter 2 boundary conditions are an important part of the model. Studies such as Spiers et al. (2007) and Heller et al. (2005) have found that muscle attachments are important to model and that their exclusion can lead to models giving results which are different from those in vivo. This study used 2 muscle attachments, the Tensor Fascia Latae and a resolved abductor muscle force. These muscles were found to have the largest effect on the results for walking and therefore were used. The forces were taken from the average walking forces for the THR and BIRM subjects from the musculoskeletal modelling. For the static models the peak forces were taken and these can be seen in Table 4-5. The dynamic forces used the forces over the one full cycle.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Direction</th>
<th>Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Resolved Force</td>
<td>Fx</td>
<td>-612.84</td>
</tr>
<tr>
<td></td>
<td>Fy</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>Fz</td>
<td>-514.23</td>
</tr>
<tr>
<td>Tensor Fascia Latae</td>
<td>Fx</td>
<td>-123.3</td>
</tr>
<tr>
<td></td>
<td>Fy</td>
<td>0</td>
</tr>
<tr>
<td></td>
<td>Fz</td>
<td>699.21</td>
</tr>
</tbody>
</table>
These muscle forces in Table 4-5 were used in both the models with the musculoskeletal load configurations and the Bergmann studies load configurations. The muscles were not changed between models as the muscles were not currently one of the parameters that were under investigation.

4.6 Boundary Conditions
This study applied an encastre constraint boundary condition to the lower surface of the femur geometry and this can be seen in Figure 4-11, preventing movement in any direction. The ISO standards state that in an experimental setup the femur should be held in place no less than 100mm from the tip of the prosthesis (ISO 7206-4, 2010, ISO 7206-8, 1995). The constraint was therefore applied 100mm from the distal tip of the prosthesis stem.

4.7 Contact Parameters
Contact parameters had to be applied along the interface between any of the three parts. The cortical and cancellous bones are both part of the femur and therefore are the same part with two different material properties. A tie constraint was used to attach these two parts together. The other contact was between the prosthesis and either the cancellous or cortical part of the femur. Only one cycle was being analysed and therefore the interface was assumed to be fully bonded representing 100% bone ingrowth as the study is investigating long term loosening (Jonkers et al., 2008). To do this a tie constraint with a surface to surface discretization method was used between these surfaces.
4.8 Modelling Results
The static models took the peak force values from the average force cycle found from the musculoskeletal modelling of the THR and BIRM subjects. Several previous studies adapted bone resorption into the FEA code. Studies like McNamara and Prendergast (2007), McNamara et al. (2006) and Scannell and Prendergast (2009a, 2005) have used strain damage model and identified that bone resorption would occur under a strain of 1000µ, and over a value of 3500µ. The lower resorption would mainly affect secondary stability of a hip replacement due to stress shielding and the stiffness of the prosthesis. The higher value would however affect the long term stability of the prosthesis. Sugiura et al. (2000) found a similar value of 3600µ as the threshold value and this value was used in Jung et al. (2004). It was therefore decided that a value of 3600µ would be used as the threshold value to indicate if damage in the model would occur. Due to the materials being linearly elastic critical stress values can be calculated corresponding to the 3600µ strain value. The threshold stress was 61.2MPa for the cortical bone and 5.4MPa for the cancellous bone.

The results in chapter 3 indicated that certain force directions may play an important role in everyday motions. Therefore, throughout the numerical modelling chapters models without the X direction (YZ model) and models without the Y direction (XZ model) are included along with the model with the force in all three directions (XYZ model). These three models are investigated throughout all the analysis investigated.

4.8.1 Minimum Principal Stresses
To calculate the stresses in the models the minimum principal stress was used. The reason for this was due to damage being caused by compressive stresses. The peak minimum principal stresses in each part can be seen for both the cancellous and cortical bone in graphical form in Figure 4-12 below.
Figure 4-12 The peak stresses within the cancellous and cortical bone for the three loading configurations with the lines indicating the threshold values for both the cancellous and cortical bone.

From Figure 4-12 it can be seen that all three models reach values over the threshold for both the cancellous and cortical bone with the exception of the cortical bone in the YZ model (it is worth noting that this value is less than 6% away from reaching the threshold stress). The results iterate that the stresses are greatest in both the cancellous and cortical bone for the XYZ model compared to the XZ and YZ models. From the results it can also be seen that the threshold values are exceeded within the bone. However, most of the peak stresses do not occur at the surface boundary between the bone and the prosthesis stem and therefore this interface requires further investigation.

4.8.2 Peak Forces along the Bone-Implant Interface

The previous results show high minimum principal stresses in both the cancellous and cortical bone. However, as stated above, these maximum stresses may not occur near the bone-stem interface. Therefore, the peak minimum principal stresses along the bone-stem interface are evaluated to determine if values exceed the threshold. These stresses from the surface of the bone prosthesis interface are shown in Figure 4-13.
### Numerical Modelling: Parameters and Boundary Conditions

<table>
<thead>
<tr>
<th>Stress (MPa)</th>
<th>Walk XYZ</th>
<th>Walk YZ</th>
<th>Walk XZ</th>
</tr>
</thead>
<tbody>
<tr>
<td>-5.00189</td>
<td>-5.82598</td>
<td>-6.99203</td>
<td></td>
</tr>
<tr>
<td>-25.1608</td>
<td>-30.9166</td>
<td></td>
<td>-17.1526</td>
</tr>
<tr>
<td>-65.0000</td>
<td>-85.0000</td>
<td></td>
<td></td>
</tr>
<tr>
<td>-85.0000</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

**Figure 4-13** The peak stresses within the cancellous and cortical bone at the surface interface for the three loading configurations with the lines indicating the threshold values for the cancellous and cortical bone.

Contrary to the results of the stresses throughout the bone presented earlier, the results at the bone-stem interface for the cancellous bone show higher stresses in the XZ and YZ models rather than the XYZ model. For the cortical bone the XZ model produces the lowest stresses of the three models with the YZ model producing the highest stresses. However, all these stresses are well below the threshold.

#### 4.8.3 Methodology Used to Generate Results

To examine the full surface at the interface between the bone and prosthesis the surface has to be unwrapped. The surface is selected in Abaqus and exported as a .vrml file. This is then opened in the freeware software Blender (Blender Foundation, Amsterdam, The Netherlands). In this software another surface is created that surrounds the imported surface. This surface is then unwrapped to form a flat surface. The surface was “cut” along the back surface of the prosthesis as is indicated by the red line in Figure 4-14. The colours are then projected onto this unwrapped surface using the BAKE tool in Blender. This can then be saved as a picture and the flat surface plot is formed.
To better understand the effects of different forces it is important to know the locations where peak forces exceeding the threshold occur in the models. The models were unwrapped and can be seen in Figure 4-15 as a flat surface using the results from the average walking models as an example. The white lines shown represent the Gruen zones and these are identified in Figure 4-15 (Gruen et al., 1979). The hip replacement model is broken into 12 sections, zones 1, 2 and 3 are on the medial side of the model and 5, 6 and 7 are on the lateral side. Zone 4 is from the tip of the prosthesis downwards and as only the surface is unwrapped zone 4 is not included. To identify if the stresses occurred at the front or back of the model each zone was further split into an anterior and posterior zone. In zones 1, 2, 6, and 7 the prosthesis is in contact with the cancellous bone whereas in zones 3 and 5 it is in contact with the cortical bone. As zone 4 is below the prosthesis it does not have a surface in contact with the prosthesis and is therefore not discussed within the results.
Figure 4-15 demonstrates that the peak stresses in all three models occurred at the neck of the prosthesis. The XYZ model and the YZ model show the peak stresses in the posterior region of Gruen Zone (GZ) 7P but the XZ model shows high peaks in GZ 7A. For this model the GZ 7A
is the only zone where the stresses are higher than the other two models as the stresses are clearly lower in all the other zones. From comparing the XYZ model to the YZ model it can be seen that there is a region of high stresses in GZ 1A. These are not present in the XZ model suggesting that this load direction would play an important part in these stresses. Although these surface plots show good detail of the stresses and where they occur it is important to understand what effects these stresses have on the model. This will be discussed in the next section.

4.8.4 Threshold Grouping

The results above show the stresses on the surface and in the bone around the prosthesis through both peak stresses and surface plots. It is important however to indicate what effects these stresses and strains may have on the interface. Instead of plotting a surface plot of the stresses involved, using different categories to represent different effects on the interface would better represent the results. To do this the different stresses have to be categorised. It has been stated in several papers (McNamara and Prendergast, 2007, McNamara et al., 2006, Scannell and Prendergast, 2009a, Scannell and Prendergast, 2005) that the following categories show what effect the stress/strain has on the model. However, it is important to note that these stresses only indicate that these effects would occur over a long period of time if the model was subjected to these stresses. Table 4-6 describes these effects.

Table 4-6 The four thresholding groups used to examine the models

<table>
<thead>
<tr>
<th>Strain (µE)</th>
<th>Cancellous Stress (MPa)</th>
<th>Cortical Stress (MPa)</th>
<th>Effect on Model</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>0</td>
<td>0</td>
<td>Bone Resorption</td>
</tr>
<tr>
<td>1000</td>
<td>-1.5</td>
<td>-17</td>
<td>No effect (lazy Zone)</td>
</tr>
<tr>
<td>2000</td>
<td>-3</td>
<td>-34</td>
<td>Bone Remodelling</td>
</tr>
<tr>
<td>3600</td>
<td>-5.4</td>
<td>-61.2</td>
<td>Bone Interface Threshold</td>
</tr>
</tbody>
</table>

Figure 4-16 shows the stresses found in the three models broken into the groups in the table above. Blue represents a region of bone resorption due to low stress. A stress of between 0 and -1.5MPa for cancellous bone and between 0 and -17MPa for cortical bone fall into this category. The green represents a group where the stresses have no effect on the model. A stress of between -1.5 and -3MPa for cancellous bone and -17 and -34MPa for cortical bone fall into this category. The red represents an area where the bone remodelling would be initiated. A stress of between -3 and -5.4MPa for cancellous bone and -34 and -61.2MPa for cortical bone fall into this category. The final group is where the stress is over the threshold at which damage may be initiated over a long period of time. A stress over -5.4MPa in cancellous bone and -61.2MPa in cortical bone would cause this.
Figure 4-16 The threshold surface plots for the three models with different loading configurations. [Blue is Bone Resorption (less than 1000), green is Lazy zone (1000-2000), red is Bone Remodelling (2000-3600) and black is over Damage initiation threshold (over 3600)]
The surface plots in Figure 4-16 give a good representation of what stresses occur throughout the models and the effects these stresses have on the models. The majority of the surface is blue in all three models. The red and black regions only occur in the proximal region close to the neck region. To investigate the amount of the unwrapped surface that is in each of the groups Table 4-7 was created with a percentage for each of the groups, while Figure 4-17 show these results as a bar chart.

Table 4-7 The percentage area for each of the threshold groups in the average walking models for each loading configuration

<table>
<thead>
<tr>
<th>Stress</th>
<th>Strain(µE)</th>
<th>XYZ (%)</th>
<th>YZ (%)</th>
<th>XZ (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 to -1.5MPa</td>
<td>&lt;1000</td>
<td>88.94</td>
<td>88.11</td>
<td>92.45</td>
</tr>
<tr>
<td>-1.5 to -3MPa</td>
<td>1000 - 2000</td>
<td>9.18</td>
<td>9.29</td>
<td>3.69</td>
</tr>
<tr>
<td>-3 to -5.4MPa</td>
<td>2000 - 3600</td>
<td>0.86</td>
<td>1.81</td>
<td>1.31</td>
</tr>
<tr>
<td>Higher -5.4MPa</td>
<td>&gt;3600</td>
<td>0</td>
<td>0.002</td>
<td>0.07</td>
</tr>
</tbody>
</table>

Figure 4-17 The percentage area for each of the threshold groups in the average walking models with the three loading configurations

From the graph and the table above the blue region is at least 88% of the area in all three models. Although the XZ model has the highest percentage of area over the threshold, it also has the highest amount of the blue region. This would suggest that the stresses in this model are lower in this model compared to the other two models and that the area over the threshold is just a local peak stress on the anterior side of the neck region. To identify more local differences between the models the Gruen zones for each model need to be examined.
4.8.5 Threshold per Gruen Zones
To identify differences in the stresses and the effects these stresses have on the model in certain parts of the model the Gruen zones were individually checked. As can be seen in the surface plots of the thresholding carried out in Figure 4-16 many of the Gruen zones are very similar and do not vary much between the different load models. Zones that showed differences between the models were 1A, 5P, 6P, 7A and 7P. The area for each Gruen zone is graphed in Figure 4-18.
**Figure 4-18** Graphical bar charts for each Gruen zone for the average walking load configurations (Yellow indicates Gruen zones of interest)
Figure 4-18 shows the results of the XYZ model and the YZ model to have stresses that would encourage bone remodelling in Gruen zone 1A. This cannot be seen in the third model (XZ model) indicating the Y force applied would affect the stresses in this Gruen zone.

The stresses in Gruen zones 5P and 6P show the XZ model to be lower than the other two models and also indicate that the zone would be in the lazy zone. This is expected as the Y load would push the implant onto this part of the bone.

Figure 4-18 also shows that in Gruen zone 7A the stresses are sufficiently high for some bone remodelling to be initiated in all three models. The stresses in the XZ model show the highest stresses in this zone and this may be due to the presence of only a small amount of cancellous bone between the implant and the cortical bone. In the other two models the Y load is pushing the implant away from this area and therefore reduced the stresses.

The results from Gruen zone 7P also highlight the XZ model is the only model not to have stresses sufficiently high to initiate bone remodelling. The YZ model has a very small area over the threshold for damage initiation. However, this is not present in the other two models indicating that the X load significantly affects the stresses in this region. Therefore both the X and Y loads play a part in controlling the stresses in this zone.

From these results it can be seen that using the thresholding values helps identify the areas that stresses may exceed the thresholding value. The surface plots and the graphing of each of the Gruen zones clearly identifies these areas and therefore this method of presenting the data will be used in the future modelling results.
CHAPTER 5

Numerical Modelling Validation
Chapter 5  Numerical Modelling Validation

5.1 Introduction
An experimental setup that was used for previous testing was available to base numerical models on. The setup was a 3 times scale model of a cemented Exeter hip replacement, the details of which are explained in the previous work section. It was decided to use this model and the published results available in a comparison to the results from the numerical modelling of the same setup. Numerical modelling plays an important role within engineering. Modelling can save on both time taken to find the stresses and forces involved, and expense that goes with building experimental rigs and equipment needed. It is important however to make sure the results retrieved from models are accurate and are not simplified too much. The following work was carried out to examine a numerical model against previous work carried out using an experimental rig using strain gauges. This work also looked at the role which the transverse load component plays within the hip joint.

5.2 Previous Work
McTague et al. (1992) manufactured a 3 times scale model based on the geometry of an implanted left femur. This was done by scanning an implanted left femur using computerised tomography (CT) to replicate the geometries of both the cement mantle and the cortical shell in the proximal femur. A filled epoxy with the elastic modulus of 13.4kN/mm$^2$ was used as the cortical shell, while an Araldite epoxy (CT200) was used to model the cement mantle. The model used 6 embedded strain transducers. The six sites chosen, three in the proximal region and three in the distal section, were embedded where the adverse effects of strain gradients were predicted to be negligible. The strain transducers were orientated relative to the prosthesis so as to measure hoop, radial and longitudinal strains at each site. The positions of the 6 transducers can be seen in Figure 5-1.
Colgan et al. (1996b, 1996a, 1995) used the McTague (1992) model and carried out a simulated load. The simulated joint and muscle forces were taken from several papers (Bergmann et al., 1993, Jacob et al., 1982, McLeish and Charnley, 1970, Paul, 1976). The hip load was applied at 6.5° anteriorly in the sagittal plane for the three-dimensional loading configuration, and 21° medially for both the three-dimensional and two-dimensional loading configurations. The abductors and ilio-tibial tract muscles were simulated for both load cases as a single resultant muscle force acting from the greater trochanter. The force was 0.6 times body weight and acted 75° medially in the coronal plane. Both the hip contact load and the muscle force can be seen in Figure 5-2.
5.3 Materials and Methods

5.3.1 CT scan and Mimics

A numerical investigation of the experimental rig used in previous studies (Colgan, 1995, Colgan et al., 1996a, McTague, 1992, O'Driscoll et al., 2009, Colgan et al., 1996b) was carried out. CT scan was used on the experimental model in the Midwestern Regional Hospital in Limerick to get precise realistic geometries. The experimental rig can be seen in Figure 5-3.
Figure 5-3 The rig used in previous experimental studies

CT scans were obtained using a Somotom Plus 4 (Siemens AG, D-91052 Erlangen, Germany). The mean pixel size of the CT scans was 0.675mm with all scans taken using a 3 mm slice increment. A total of 654 slices were taken for the model. A 3D reconstruction was carried out using the reconstruction software Mimics v11.11 (Materialise, Belgium). This software allows the transformation of 2D CT scans into realistic 3D models of exact geometry. The software uses a marching squares algorithm to threshold and segments the regions of interest using Hounsfield values. Using this software the three parts (femur, cement and implant) were identified by using Hounsfield values. The gauges were assumed to be part of the cement part and Figure 5-5 shows the position of 4 of the gauges within a scan.
A threshold Hounsfield value of between 519 and 2346 was used to find the femur geometry. A threshold Hounsfield value between -350 and 466 was used to find the cement geometry. The implant could not be CT scanned as it was made of stainless steel. The implant geometry was therefore found by filling in the gap left by the implant in the cement. Once the geometries were defined in each of the slices a 3D surface geometry was constructed for each of the three parts. A mesh was then added to this 3D surface geometry. This is done by using the remesh tool. The shape quality threshold parameter was used to determine the density of the mesh and the amount of elements in each part. The geometries were then exported for use in the numerical modelling software Abaqus. The cement part can be seen selected in Figure 5-4 from the scans and the geometries exported for Abaqus can be seen in Figure 5-5.
5.3.2 Model Construction

The 3D surface geometries from mimics were only surfaces and needed to be changed into volume geometries and constructed into one part. This was carried out in Abaqus using the tri to tet meshing tool. All three of the 3D volume geometries were copied into one file to create one single model.

5.3.3 Grid Independence

Grid independence was carried out on each of the three geometries. The reason for grid independence is to show that the results from the models were not dependent on or affected by the density of the mesh. Four models varying in mesh density were made for each part. The shape quality threshold was varied to change the amount of elements and the density of the meshes in each model.

Figure 5-5 a) A CT scan showing 4 of the gauge positions b) The 3D geometry of all three parts
5.3.3.1 The Implant

The shape quality threshold was varied between 0.3 and 0.45 for the implant geometry. This varied the number of elements from 49,559 to 122,370. The results were taken from each of the four models along 3 paths. A 30mm path in the X direction, a 50mm path along the Y direction and a 50mm path along the Z direction. The path in the X direction was smaller than the other two paths due to the small distance in the X direction. Figure 5-6 shows the position of the paths in the model, the Von Mises stress along the X direction and the % Difference between meshes in this direction.

![Figure 5-6 a) The positions of the three paths, b) Von Mises stress along the X direction and c) the % difference between different meshes along this path](image)

Table 5-1 The average % differences between meshes

<table>
<thead>
<tr>
<th></th>
<th>49,559 v 58,211</th>
<th>58,211 v 93,400</th>
<th>93,400 v 122,370</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>X</strong></td>
<td>0.813768</td>
<td>1.323947</td>
<td>2.003887</td>
</tr>
<tr>
<td><strong>Y</strong></td>
<td>1.39039</td>
<td>2.720738</td>
<td>1.323158</td>
</tr>
<tr>
<td><strong>Z</strong></td>
<td>1.614635</td>
<td>2.042817</td>
<td>0.936473</td>
</tr>
</tbody>
</table>

Table 5-1 above shows the average % differences between the different mesh densities. A ±2% difference criteria was used to confirm the grid independence. This criteria has been used in
other modelling studies (Helgason et al., 2008). From Table 5-1 it can be seen that all mesh densities agreed with the criteria, therefore a shape quality threshold of 0.3 and 49,559 elements was used for this part.

5.3.3.2 The Femur

The shape quality threshold was varied between 0.3 and 0.45 for the femur geometry. This ranges from 254,105 elements to 493,518 elements. The results were taken from each of the four models along 3 paths. A 10mm path in the X direction, a 50mm path along the Y direction and a 50mm path along the Z direction. The path in the X direction was smaller than the other two paths due to the small distance in the X direction. Figure 5-7 shows the position of the paths in the model, the Von Mises stress along the X direction and the % Difference between meshes in this direction.

![Three paths used](image1.png)

Figure 5-7 a) The three paths used, b) Von Mises stresses in the X direction and c) the % difference between meshes along one path

Table 5-2 The average % differences between meshes

<table>
<thead>
<tr>
<th></th>
<th>254,105 v 271,997</th>
<th>271,997 v 324,183</th>
<th>324,183 v 493,518</th>
</tr>
</thead>
<tbody>
<tr>
<td>(X)</td>
<td>1.005415</td>
<td>0.89606</td>
<td>2.274815</td>
</tr>
<tr>
<td>(Y)</td>
<td>1.093064</td>
<td>1.217233</td>
<td>1.2283</td>
</tr>
<tr>
<td>(Z)</td>
<td>2.130473</td>
<td>1.688426</td>
<td>2.133541</td>
</tr>
</tbody>
</table>
Table 5-2 above shows the % differences between the different mesh densities. A ±2% difference criteria was used to confirm the grid independence as with the prosthesis part. Table 5-2 showed that the meshes with a shape quality threshold below 0.4 were not grid independent in all directions. However the % difference between the 0.35 and 0.4 shape quality threshold models was independent in all three directions. For this reason it was decided to use a shape quality threshold of 0.35 and a mesh density of 271,997 elements.

5.3.3.3 The Cement

As the cement part was the primary section under investigation and the part that contained the gauges, it was decided to use high shape quality threshold values. These were between 0.3 and 0.465. This varied the elements between 191,240 and 529,966. The results were taken from each of the six models along 3 paths. A 50mm path in the X direction, a 50mm path along the Y direction and a 50mm path along the Z direction. Figure 5-8 shows the position of the paths in the model, the Von Mises stress along the X direction and the % Difference between meshes in this direction.

![Figure 5-8](image)

Figure 5-8 a) The three paths used for the cement part, b) Von Mises stresses along the X direction for each mesh and c) the % Difference between the meshes along one of the paths

143
Table 5-3 The average % differences between meshes

<table>
<thead>
<tr>
<th></th>
<th>191,240 v</th>
<th>222,081 v</th>
<th>274,443 v</th>
<th>406,071 v</th>
<th>482,802 v</th>
<th>529,966 v</th>
</tr>
</thead>
<tbody>
<tr>
<td>X</td>
<td>2.403405</td>
<td>2.601159</td>
<td>3.220104</td>
<td>2.363383</td>
<td>1.501444</td>
<td>765</td>
</tr>
<tr>
<td>Y</td>
<td>2.307389</td>
<td>2.469346</td>
<td>2.842503</td>
<td>2.439718</td>
<td>1.388988</td>
<td>892</td>
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<tr>
<td>Z</td>
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<td>2.538829</td>
<td>3.086941</td>
<td>1.351216</td>
<td>1.996707</td>
<td>36</td>
</tr>
</tbody>
</table>

Table 5-3 above shows the % differences between the 6 different mesh densities. A ±2% difference criteria was used to confirm the grid independence as with the other parts. Table 5-3 showed that the meshes with a shape quality threshold of 0.45 or below were not grid independent in all the three directions. However the % difference between the 0.46 and 0.465 shape quality threshold models was independent in all three directions. For this reason it was decided to use a shape quality threshold of 0.46 and a mesh density of 482,802 elements.

From the grid independence study the three parts chosen were an implant with a shape quality threshold of 0.3 with 49,559 elements, an outer part with a shape quality threshold of 0.35 with 271,997 elements and an inner part with a shape quality threshold of 0.46 with 482,802 elements. Each part was found to have less than 2% difference between the mesh chosen and the other denser meshes. The final model consisting of these three parts is therefore grid independent. The final model consists of 804,358 elements.

### 5.4 Hip Joint Loading Conditions

The loading conditions in the numerical model for the hip joint were applied similar to the experimental setup (Colgan et al., 1996a). The force is broken into X, Y and Z loads using the angles given in Colgan et al. (1996a). The point at which the force acted on the prosthesis head was at the same dimensions as the experimental setup. The hip joint forces can be seen in
Table 5-4.

5.5 Muscle Loading
The experimental model in Colgan et al. (1996a) included a muscle force. This force was applied using a pulley system and a metal plate attached to the femur. This muscle force was replicated in the numerical models. The surface the metal plate was estimated and a surface set was selected of nodes in this area. The muscle force was then applied to this region using the X, Y and Z directions and the angles described in the experimental setup. The forces applied can be seen in
Table 5-4.
Table 5-4 The hip joint and muscle forces created to replicate the forces from Colgan et al. (1996a)

<table>
<thead>
<tr>
<th>Direction</th>
<th>Hip Force (N)</th>
<th>Muscle Force (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Fx</td>
<td>179.6</td>
<td>0</td>
</tr>
<tr>
<td>Fy</td>
<td>609.22</td>
<td>985</td>
</tr>
<tr>
<td>Fz</td>
<td>1576.88</td>
<td>264</td>
</tr>
</tbody>
</table>

5.6 Boundary Conditions and Limitations
A boundary condition was applied to the distal end of the femur model in the numerical models. An encastre condition was applied to the bottom surface of the femur model meaning the nodes on this surface could not move in any direction. The experimental experiments used a base to hold the femur in place at the distal end. This setup enabled some movement within the base of the femur and was complex to model and replicate. This boundary condition was simplified for the numerical models and therefore is a limitation to the results. One other limitation of the numerical models was as the prosthesis was not able to be CT scanned, its geometry had to be created. The geometry of the prosthesis within the femur was calculated from the cavity left within the model. However, the neck of the prosthesis had to be estimated and as the neck thickness may be different it may increase or decrease the stiffness of the model and therefore affect the results.

5.7 Results
The numerical model was set up with 6 individual coordinate systems one for each gauge from the experimental rig. The position of each gauge was identified by examining the CT Scans in the mimic’s software. The transverse load is thought to play an important part in the loosening of hip prostheses. Figure 5-9 and Figure 5-10 show the significant stresses in both the 2D and 3D load configuration experimental models for each gauge in the experimental work of Colgan et al. (1996a) and the corresponding numerical values from work done in this study.

The experimental models in Colgan papers used both a 2D loading configuration and a 3D loading configuration. The 2D loading configuration consisted of the load only been applied in the vertical and lateral directions with no transverse force applied. The 3D loading configuration included all three loading directions. The numerical modelling used the same loading configurations for both the 2D and 3D loads.
## 3D Significant Values

![3D Diagram](image)

<table>
<thead>
<tr>
<th>E</th>
<th>$\sigma_E$</th>
<th>$\sigma_L$</th>
<th>$\tau_{EL}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>E</td>
<td>-0.104</td>
<td>0.251</td>
<td>0.100</td>
</tr>
<tr>
<td>N</td>
<td>0.0095864</td>
<td>0.8665622</td>
<td>0.215214</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>E</th>
<th>$\sigma_E$</th>
<th>$\sigma_L$</th>
<th>$\tau_{EL}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>N</td>
<td>-0.329856</td>
<td>0.920576</td>
<td>-0.09325</td>
</tr>
</tbody>
</table>

$E = \text{Experimental}$  
$N = \text{Numerical}$  
Significant in Bold Italic

Figure 5-9 3D significant values from Colgan *et al.* (1996) and the Numerical values that correspond
The effects of the transverse load were seen in both models at site 1. However, the numerical model and Colgan et al. (1996a) experimental model differed in directions with the experimental model predicting tensile stress and the numerical model a compressive hoop stress. The numerical model in this study showed the prosthesis being forced towards the posterior part of the cavity. The experimental results of Colgan et al. (1996a) show a tensile hoop stress as the prosthesis is forced into the proximal-posterior aspect of the cavity. Colgan et al. (1996a) found the $\tau_{\theta R}$ stress to be negative similar to Allison et al. (1992) who looked at torsional loading of fully bonded cylinders. The numerical model however found this to be a very small positive shear stress. The $\tau_{\theta L}$ stress in the experimental study was found to be significant with a positive stress induced due to torsion of the model. The numerical model also showed this shear stress as positive.

To fully understand the effects of the transverse force the differences between the 3D and 2D models have to be examined. For hoop stress the experimental results showed a big drop from a tensile stress to a value close to zero, whereas the numerical results showed little difference between 2D and 3D hoop stresses. The $\tau_{\theta L}$ stress also dropped from a significant positive stress to a very small negative value compared to a very small increase in the numerical models. The
numerical model showed little difference between the 3D and 2D tensor or shear stresses at this site with only very small differences between values. All 2D stresses in the experimental model were very low and none were significant whereas the numerical model showed similar stresses to the 3D results and indicated the prosthesis being forced into the posterior aspect of the cavity.

Figure 5-11 The path from A to B in both the experimental and numerical models

To investigate the effects of the transverse load on site 1 the radial stresses were examined along the proximal-medial cement mantle surface. This can be seen in Figure 5-11 for both the experimental and numerical models. In the experimental model four rosettes were mounted radially from the stem-cement interface (A) to the cement-bone interface (B). In the numerical model a path was created along the cement mantle in the same position as the four rosettes used in the experimental work. From Figure 5-12 it can be seen the radial stress in the 2D and 3D numerical models showed an increase in stress as the path gets closer to point B. The 3D experimental model shows a similar trend to the numerical models as it increases towards B. The 2D experimental model shows a reduction in stress along the path. The magnitude of the stresses seen in the numerical model were substantially higher than the experimental results however this was seen to be the case at most of the gauge sites.
Site 2 is located in the proximal-posterior region of the model. The numerical results showed a sizable compressive radial stress, as the prosthesis is pushed towards this site due to the transverse load. The model with only the 2D load configuration showed a reduction in the compressive radial stress. Colgan et al. (1996a) experimental results showed a significant tensile longitudinal stress at site 2 however this value was not large in the numerical models. The $\tau_{RL}$ stress was a significant positive stress in both the numerical and experimental models with both models showing a drop for this stress in the 2D loading models. This shearing stress is due to the frictional shear at the interface. The increase in the 3D loading configuration stresses occurs due to the transverse load component. Colgan et al. (1996a) and Fowler et al. (1988) stated that this shear induced by torsion may affect implant fixation. A negative $\tau_{RL}$ stress occurred in the numerical and experimental models.

The results of Colgan et al. (1996a) did not show any results to be significant for the 3D load configuration at site 3 and this was due to the prosthesis being pushed away from this area. The 2D load configuration did however have significant tensile radial stresses. The numerical model also showed similar large tensile forces in the radial direction. Longitudinal forces were also significant in the experimental work however these were not replicated in the numerical model.

Transducer sites 4, 5 and 6 were towards the distal end of the models, located close to the tip of the prosthesis. Site 4 is located on the anterior side of the prosthesis. The radial stress for this site in the 3D experimental models was compressive and as the radial stress at site 2 is also compressive this shows the twisting movement of the prosthesis due to the transverse force.

Figure 5-12 Radial stresses from A to B for the 2D and 3D loading configurations for the numerical and Colgan et al. experimental models.
However the radial stress at this site in all the other models was seen to be tensile indicating the prosthesis is being pushed away from this site. The low radial values for all models would suggest that a good stem/cavity engagement and the rectangular proximal cross sectional prosthesis geometry may play a part in preventing the transmission of the transverse load to the cement mantle. The transverse load component was seen to have a large effect on the longitudinal stress at site 4. The 3D load configuration models both showed tensile longitudinal stresses. When only the 2D load configuration was applied the longitudinal stresses changed to compressive stresses in both the experimental and numerical models. Brown et al. (1988) carried out Finite element models that found similar results.

For sites 5 and 6 longitudinal stresses were both high and significant in all models. At site 5 the longitudinal stresses were high compressive stresses in all models. Colgan et al. (1996a) stated that this was induced due to the eccentricity of the axial load coupled with the joint load component in the medial lateral direction. The radial stress in the numerical and experimental 3D load configuration models were both tensile with similar values found for both. A significant positive $\tau_{\theta L}$ stress was found by Colgan et al. (1996a) at this site and the stress found in the numerical model was similar but lower. However it was the other two shear stresses that were significant for the experimental 2D load configuration. The $\tau_{LR}$ stresses in the 2D models gave similar negative stresses for both experimental and numerical, however the $\tau_{RL}$ stresses were found to be negative in the experimental models but positive in the numerical 2D load configuration model.

The results from site 6 showed tensile stresses in the longitudinal direction, and this was seen in both models with both load configurations. The hoop stresses in the numerical models were found to be large tensile stresses. These may be explained by studies by Dragoni et al. (1992) and Brown et al. (1988). Both suggested that it was due to the wedging or pistoning of the taper on the cement mantle. The experimental work showed the 2D loading configuration to have a significant tensile hoop stress however the 3D loading configuration was tensile but not significant. Unlike site 5 the radial stresses in both the 2D and 3D models were both small and insignificant, suggesting there is not much movement in the distal part of the prosthesis due to the load. The $\tau_{RL}$ stress showed that introducing the transverse load in the 3D model the frictional shear was reduced. This was not seen in Colgan et al. (1996a) experimental model with only the 3D model having a significant stress.

## 5.8 Discussion
The 2D loading model in the numerical analysis shows, in the proximal area, the prosthesis is pushed away from the lateral side and towards the site 2 aspect of the cavity, and distally towards the posterior aspect as well. Colgan et al. (1996a) experimental model agreed by
finding the prosthesis moving away from the lateral aspect of the cavity in the proximal part, and towards the medial aspect distally. The main differences between the 2D and 3D loading configurations were seen at sites 2 and 4. This would be expected with the additional transverse force in the same direction as the radial directions of these two gauges. In the numerical model it was seen that large stresses that were parallel to the transverse force were reduced in the 2D loading configuration model, indicating that reducing the transverse force would reduce the stresses in this direction. In the proximal part of the cavity (sites 1, 2 and 3) the 3D load configuration models show the transverse force pushing the prosthesis towards the posterior aspect of the cavity but also creating a twisting moment. In the 2D load model the twisting moment is still apparent but has been reduced due to the lack of the transverse force.

From the study it can be seen that the transverse load component does play an important role within the load component. The load was seen to affect both the tensor stresses and the shear stress with the shear stresses showing the biggest differences between 2D and 3D load configuration models. Shear stresses due to the transverse force have been shown by Wroblewski et al. (1980) to induce loosening within the cemented cavity. At site 2 the $\tau_{RL}$ stress was seen to decrease showing a decrease in the friction between the prosthesis and the cement. The frictional shear was also seen to be affected by the inclusion of the transverse force at site 4 with a reduction in the negative stress.

### 5.9 Conclusion

This study has shown that the transverse force through considerable shearing stresses may play a factor in aseptic loosening and therefore may induce instability within the hip joint. The numerical models examined only investigate the transverse load during walking however the forces at the hip joint can be higher during the other everyday activities during the day such as stair climbing. The forces associated with this and other movements should be investigated in future studies. Applying a Dynamic load configuration may also give a better insight into the changes in stress at the interface as large changes in stress would indicate movements that may lead to instability and loosening. Although the magnitude of the stresses found were not always in agreement both studies show similar trends for the inclusion/exclusion of a transverse force and the effect this has on the stress along the cement-stem interface.
CHAPTER 6

Experimental Rig: Design and Results
Chapter 6  Experimental Rig: Design and Results

6.1  Introduction
To examine the effects of forces on the stability of hip replacements an experimental rig had to be designed, fabricated and commissioned with the ability to apply forces similar to the forces found in the body. The integral part of stability within a cementless hip replacement is at the bone-prosthesis interface and this was previously discussed in the literature review chapter. The micromotions in this area require investigating and stability of the prosthesis would depend on an increase in this micromotion.

6.2  Loading Configuration

6.2.1  Musculoskeletal Modelling
As described in chapter 2, studies have been carried out to investigate forces In vivo using instrumented prostheses and calculated using musculoskeletal modelling. In an experimental setup dynamic loads should be applied to the model to replicate daily activities such as walking and stair climbing. Previously published experimental setups simply used a cyclic ramped load from a low force to the peak force and back. However, realistic loads behave differently and stay close to the peak load for an extended period of time. Also, the load can have more than one peak during activities such as walking. This experimental setup used a realistic force in the X, Y, and Z directions calculated from the musculoskeletal modelling results presented earlier in chapter 3. The average was taken from all THR and BIRM subjects for each activity tested during the motion analysis. The force for the walking motion calculated can be seen in Figure 6-1.
6.2.2 Pneumatic Pistons.

The forces calculated in the musculoskeletal models were applied to the experimental setup by using pneumatic pistons. 5 pistons were used in total, 3 to replicate the realistic force in the X, Y and Z directions, one to apply the resolved abductor muscle force, and the other to apply the tensor fasciae latae muscle force. The force in the vertical direction i.e., the Z direction in this experimental setup was the greatest and therefore needed a 100mm bore cylinder (C95SDB100-160, SMC Pneumatics (Ireland) Ltd, Dublin, Ireland). The bigger bore size was a bigger area within the cylinder and therefore can create a higher force. The other 4 forces in the setup were all created using a 63mm bore cylinder (C95SDB63-160, SMC Pneumatics Ireland Ltd, Dublin, Ireland). The maximum force applied by a piston can be calculated using equation 6.1.

Electro-pneumatic pressure regulators were used to control the air supply to the pneumatic pistons (ITV1030-31F2N-Q, SMC Pneumatics (Ireland) Ltd, Dublin, Ireland). The air was supplied to these regulators and they controlled the air and therefore the force the piston created. The electro-pneumatic regulators were controlled by the software Labview (National Instruments, Newbury, UK) which will be discussed further on in this chapter. The maximum pressure available in the air supply was 0.7MPa. However, the pneumatic regulators used were only to provide a maximum pressure of 0.5MPa.

6.2.2.1 Piston Calibration

The force created by the pistons can be calculated by the equation below:
\[ F_{Piston} = P_{Cylinder} \left( \pi \left( \frac{d_c}{2} \right)^2 \right) \]  

(6.1)

Where:

\( F_{piston} \) = Piston force (N), \( P_{Cylinder} \) = Air pressure (Pa), and \( d_c \) = bore diameter of the piston (m)

This equation can be used to calculate the air pressures required to apply the forces needed for each of the 5 pistons used. However, it has been stated previously in literature that the pistons are only 95% efficient (Kavanagh, 2001). Therefore, it was decided to calibrate the pistons to gain an accurate reading of the air pressures needed. This is discussed further in this chapter.

The pistons apply a force using air pressure. The force of the piston can be calculated using the formula above and the SMC documentation also indicated general forces from the air pressure applied. However, it has been reported that there can be a 5% loss in air pressure and therefore a reduced load applied (Kavanagh, 2001). The forces applied by both piston sizes were investigated using a Tinius Olsen testing machine (Surrey, UK). The voltage was varied from zero to 5V in 0.2V increments and then back down to zero again in 0.2V increments. Both tensile and compressive forces were applied to check any variation that may occur as the 3 hip joint forces will be applying compressive or pushing forces and the muscle forces applying tensile or pulling forces. The pistons were also checked at a low stroke length and at a high stroke length. The results of this testing can be seen in Figure 6-2 to Figure 6-5.

![Graph of 63mm Piston Compressive forces](image)

**Figure 6-2** The calibration chart for the 63mm bore cylinder for compression

159
Figure 6-3 The calibration chart for the 63mm bore cylinder for a tensile load

Figure 6-4 The calibration chart for the 100mm bore cylinder for compression
Figure 6-2 to Figure 6-5 demonstrate that a different force is applied from the same voltage depending on whether the force is going increasing or decreasing. Therefore, when calculating the voltage to apply for a certain force different curves were used depending whether the force is increasing or decreasing. There was also a difference between the pistons in compression and tension and the corresponding forces were calculated from Figure 6-2 and Figure 6-3 for the 63mm piston and Figure 6-4 and Figure 6-5 for the larger piston. This difference is expected as the volume of the front air chamber of the piston is less than the back chamber due to the piston rod.

### 6.2.3 Muscle Attachments

Muscle forces were applied to the femur by the pneumatic cylinders. Polyethylene rope (Dyneema, Toyobo Ltd., Japan) was used to attach the pistons to the femur where the muscles are applied. The rope was attached to the Sawbone femur at the attachment points using a high strength epoxy adhesive (Araldite Precision, Bostick Findley Ltd., UK). This form of muscle attachment has been used in similar studies such as Britton and Prendergast (2005). The forces for the two muscles applied were taken from the musculoskeletal modelling similar to the hip joint contact forces.
6.3 Micromotion Measurement

As presented earlier in chapter 2, micromotion at the interface has been used to predict both bone growth and implant failure. This was done with the use of LVDTs. These can be attached to some form of device which uses a number of LVDTs at one point along the interface to find the movement in all directions at that point. This technique was used in Britton and Prendergast (2005) and Pettersen et al. (2009). An alternative approach was taken by Abdul-Kadir et al. (2008) among others, who used single LVDTs at different locations along the interface. The present study employed the approach taken by Abdul-Kadir et al. (2008) and used 3 LVDTs in three different locations. After examining the positions of the LVDTs in other studies the three positions chosen were one distally near the tip of the stem, one slightly distal to the lesser trochanter and the final one slightly distal to the greater trochanter. The LVDTs chosen were GH-SM_2.5A LVDTs (Singer Instruments and Control Ltd). These were micro-sized transducers with a M4 threaded body so the transducer could be directly attached to the bone by threading the body of the transducer into the Sawbone. The LVDTs are gauge head LVDTs with a stroke length of 2.5mm. A USB-10-4 signal conditioner with non-linearity correction was also used. This has the capacity to run four channels and converts the readings using a 21 bit analog to digital (A/D) converter and then communicates with information to the PC via USB. An operating and data collection software package was also supplied. The LVDTs had a custom made 20mm thread for attachment to the Sawbone femurs.

An LVDT uses the position of the core to measure the micromotion. The position of the core in the centre tube relative to the excitation coil and the two pick off coils changes the mutual inductance between the excitation coil and the pick-off coils. As one voltage drops the other rises and from this linear displacement can be measured. A picture of the LVDT can be seen in Figure 6-6.

![Figure 6-6 One of the LVDT's used to measure micromotions](image-url)
6.4 Femur Base
When the experiment is running the femur has to be firmly secured to the rig so that the only movement is within the femur. ISO standard (ISO 7206-4, 2010) has criteria for testing THR prostheses and how they are secured to the base. The material used should meet the following criteria:

- Not break or crack under the load applied during the testing
- Not exhibit excessive deformation or creep
- Be reproducible in strength and other characteristics

The material chosen for this study was a polyester based body filler (Easy One, U-Pol Ltd, Wellingborough, Northamptonshire, UK). A polyester based material was used by Colgan et al. (1995, 1996b, 1996a) also for securing experimental femurs. It meets all the criteria from the ISO standard and is commercially available at a reasonable price. ISO 7206-4 (2010) also identifies the specific angles relative to the applied load. Figure 6-7 shows these angles as $\alpha$ and $\beta$. When the distance between point C and point T is less than or equal to 200mm then $\alpha$ should equal 10°±1 and $\beta$ should equal 9°±1. Should the distance between C and T be higher than 200mm, then $\alpha$ should be equal to 0° and $\beta$ equal to 4°.

Figure 6-7 The angles indicated by the ISO standards for securing the femur
A mould identical to the holding clamp was made to set the material being used and to hold the material in place during this period. The polyester based filler needs 20-30 minutes to set properly but was given overnight to set for each experiment. The femurs were held in place at the specified angles using clamps.

### 6.5 Composite Bone

Fourth generation composite femur bones (#3403 Sawbones Europe AB, Sweden) were used in place of cadaver bones. They are designed to replicate the physical behaviour of human cadaver bones. They were used in this study as they give less variability than cadaver specimens and therefore the experiments would be repeatable and results comparable. Cortical bone is replicated using a mixture of epoxy resin and glass fibers. The glass fiber epoxy resin has a compressive strength of 157MPa and a tensile strength of 106MPa. The Cancellous bone material is made up of solid rigid polyurethane foam. The polyurethane foam has a compressive strength of 5.4MPa and a Poisson’s Ratio of 0.3. Figure 6-8 shows composite bone cut in half.

![Figure 6-8 Sawbone femurs used for experimental testing cut in half](image)

### 6.6 Prosthesis

An orthopaedic company supplied 12 femoral stems for testing in this study. A non-disclosure agreement was agreed and the details of this can be seen in Appendix B. The stems are cementless with a proximal roughened surface onto which the hydroxyapatite coating would be applied. The prostheses used in this study did not have hydroxyapatite applied to them. However, as this experimental setup does not replicate any biological processes, and only concentrated on the mechanical aspect of the experiment, this should not affect the results. All prostheses have a double wedge, metaphyseal filling body with a neck angle of 132° and a neck
length of 30mm. They have a taper lock head, onto which a ceramic based head is usually attached. A head was not used in this experimental setup and a loading block was designed around the taper lock of the prosthesis stem.

6.7 Labview Programming

A PCI card from National Instruments (PCI 6703, National Instruments, Newbury, UK) was installed on the PC to be used to control the rig. This card allowed a connection to be made to the shielded connector block (SCB 68, National Instruments, Newbury, UK) from which the electro-pneumatic regulators were controlled. The regulators are controlled by a 0-10VDC signal which is generated from the PCI card through the connector block.

The values from the musculoskeletal chapter were converted to a Fourier transformation using the Matlab r2009a (Mathworks, Natiks, Massachusetts, USA). The Fourier series was then imputed into the Labview software (National Instruments, Newbury, UK) to create a signal and this can be seen in Figure 6-9. The ‘Create Analog signal’ Virtual Instrument (VI) in Labview (National Instruments, Newbury, UK) was used and the data for the Fourier series was applied to the formula section. In this VI it was also possible to adjust the frequency, offset and amplitude of the signal. However they were all left as the default. The ‘create analog signal’ VI can be seen in Figure 6-10.

![Figure 6-9 The Fourier series used to control the X, Y and Z forces for the three pistons](image)
The Labview software (National Instruments, Newbury, UK) was used to control all 5 pistons. 5 signals were created, 3 for the hip contact joint and 2 for the muscle attachments. A counter and a waveform graph were added to each force. The counter indicated how many cycles the pistons had applied and the waveform graph showed the voltages that were applied to each channel in the connector block as these controlled the forces applied. Figure 6-11 shows the block diagram of the Labview programme used (National Instruments, Newbury, UK).
6.8 Calibration of Experimental Setup and Electro-Pneumatic Regulators

The system used voltages created in Labview to control the pressure in the electro-pneumatic regulators which control the forces applied by the pistons. To investigate if the system produces the forces required and to assess the response time of the electro-pneumatic regulators a Tinius Olsen tensile testing machine (Surrey, UK) was used. A triangular waveform was used varying the voltage from minimum to maximum. The changes per second were varied to investigate if the regulators were able to change the air pressure from minimum to maximum within the required period of time. The slowest change in voltage was a change every 2 seconds with the fastest change being 6 times per second. The calibration tests demonstrated that the applied voltages in Labview did change the forces as needed. The optimum response time of the regulators was found to be 3 times per second, as any changes faster than this resulted in the force the piston applied not reaching the maximum force as the regulator could not change the air pressure quickly enough. For the walking force the one cycle was 1.24 seconds in length, therefore the regulators were able to change 4 times per cycle.
6.9 Rig Design

6.9.1 Holding Clamp
The femur was set using the polyester based filler in a mould of similar dimensions to the holding clamp within the rig. The holding clamp was located in the centre of the four main posts and was in a 2mm deep cavity that was machined into the base plate. This added stability to the clamp during testing. A counter sunk bolt was also used to attach the holding clamp to the base plate. Femoral clamp inserts were used to hold the potted cement into the holding clamp. Screws were used to tighten the inserts to the cement in case the cement shrank during moulding or had room to move within the cavity. The inserts applied this pressure over a wide area and therefore minimised the risk of damage to the cement. Figure 6-12 shows the femoral clamp with the femoral clamp inserts.

Figure 6-12 Femoral clamp including the femoral clamp inserts

6.9.2 Piston Movement
The holding clamp was secured to the base. Therefore the pistons were designed to be movable in both the vertical and horizontal directions so that the load was applied at the correct location on the model.
Muscle Attachment

The two muscle forces were applied using two pistons, one for the resolved abductor force and one for the tensor fasciae latae muscle force. The two pistons were held in a rigid position with the pistons in opposite directions as the resolved abductor force acts upwards from below the greater trochanter, and the tensor fasciae latae acts downwards.

Both muscle forces employed pulley systems to ensure the forces were applied at the correct angle. The pulley systems for both forces can be seen in Figure 6-14. For the resolved muscle force two pulleys are used, one attached to the top plate and one attached to top slide plate. The tensor fasciae latae muscle force pulley system also uses two pulleys which are both attached to the base plate.
6.9.4 Air filter and Tubing

An air filter (AW3000, SMC Pneumatics Ireland Ltd, Dublin, Ireland), was used to filter the compressed air supplied to ensure that the air was pure. A divider was used to split the air supply into 5 channels one for each electro-pneumatic regulator that feeds each piston. The air tubing used to connect all the equipment was 6mm SMC Ltd tubing. The full rig with the tubing can be seen in Figure 6-15.
6.9.5 Loading Block

The centroid of the loading block was calculated to limit moments that would be created by the pistons and the weight of the block. The head of the prosthesis had to slot into the loading block at the centroid otherwise the weight of the block would cause a moment about the prosthesis. The three pistons also had to act through this point as they would also create unwanted moments that did not represent those in the body. The centroid was calculated using the following equation 6.2.

\[
y = \frac{\sum A_i \bar{y}_i}{A_i} \tag{6.2}
\]

The Centroid was calculated to be at \( x = 46.14 \text{mm} \), \( y = 54.83 \text{mm} \) and \( z = 75 \text{mm} \) and Figure 6-16 shows a diagram of the loading block.
The loading block was manufactured from acrylic. The face was angled to be perpendicular with the neck of the prosthesis. A hole where the taper lock attaches to the block was located at the centroid to ensure that there are no moments created by the weight of the block. To reduce the occurrence of any moments created by any of the three pistons all forces acted at the centroid. Consequently the pistons did not create any moments about the head of the prosthesis as would be the case in vivo. The loading block can be seen in Figure 6-17.

### Figure 6-16 An illustrated diagram of the loading block (not to scale)

<table>
<thead>
<tr>
<th>Centroid (mm)</th>
<th>Area (mm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>A</td>
<td>(4.75, 75)</td>
</tr>
<tr>
<td>B</td>
<td>(61.10, 75)</td>
</tr>
<tr>
<td>C</td>
<td>(44.76, 75)</td>
</tr>
</tbody>
</table>
CHAPTER 6 - EXPERIMENTAL RIG: DESIGN AND RESULTS

6.10 Osseointegration Bond Material

6.10.1 Introduction
This study concentrates on long term stability of cementless hip replacements. Other studies have carried out testing for primary and secondary stability of cementless hip replacements through both experimental and numerical modelling. Long term stability is however not well understood. To investigate failure or loosening in an osseointegrated replacement, an osseointegrated replacement has to be replicated. Testing was carried out in conjunction with (Behan, 2009) to investigate a material suitable to replicate the osseointegrated bond.

6.10.2 Materials
To limit the number of adhesives that would be investigated a material selector software (Cambridge Engineering Selector) was used. Taking Poisson’s Ratio of human bone to be 0.3964 (Reilly and Burstein, 1974) and plotting this against a fracture toughness range of between 0.3-0.5MPa.m^{1/2} (Wang and Agrawal, 1997, Wang et al., 1994) three adhesives were chosen. These were an unfilled epoxy resin (Araldite Precision, Bostick Findley Ltd., UK), Devweld 531 Adhesive (ITW Devcon, Rushden, UK) and a high strength epoxy adhesive.
The Araldite Epoxy had a work time of about 2 hours, needed a support time of 6 hours and full cure time was 16 hours although 24 hours were given to allow for full curing in laboratory temperatures. The Devweld 531 adhesive had a work time of only 10-11 minutes, support time was 30-35 minutes and 24 hours was needed for full cure time. The high strength epoxy adhesive had a working time of 15-20 minutes and curing times were unspecified so 24 hours was allowed for full curing with a two hour support time.

6.10.3 Tests Carried out

6.10.3.1 Push-Out Test

The push-out test has been used for years to calculate the interfacial shear strength of the osseointegrated bond formed at the bone-implant interface. Plugs are implanted into animal specimens, often canine, and allowed to integrate for any time from 4 to 52 weeks or more. The shear strength is calculated by equation 6.3.

Shear Strength (MPa) = Force to failure / bone-implant contact area  \hspace{1cm} (6.3)

The push-out test has been extensively used to calculate the interfacial shear strength of the osseointegrated bond formed at the bone-implant interface. Plugs are implanted into animal specimens, often canine, and allowed to integrate for any time from 4 to 52 weeks or more. The shear strength is calculated by equation 6.3.

Figure 6-18 shows the assembly of the designed push-out. This is an experimental set up where two chocks are used to hold the Sawbone sample square. The titanium sample is inserted into the Sawbone using the appropriate adhesive. The titanium sample was 15mm taller than the surface of the Sawbone sample which allowed a compression test jig on the Tinus Olsen machine to be used as opposed to manufacturing a custom push-out jig. The Sawbone samples were machined from a biomechanical testing block which was purchased from Sawbones (Sawbones Europe AB, Sweden). This biomechanical testing block was composed of the same polyurethane foam that is used in the composite femurs.
The objective of the push out test was to quantify the shear strength of the bond formed between a femoral stem and the biomechanical testing block using various adhesives. The following method was used. Firstly, the surface of the femoral stem was replicated on the cylindrical test pieces. Secondly, the test pieces were implanted into the biomechanical testing block using various adhesives. Thirdly, the push-out test was performed using a Tinius Olsen tensile tester. And finally the force required to push out the test piece from the testing block was recorded. The adhesive with shear strength closest to those found in the literature should be used for securing the femoral stem in the composite Sawbones. Typically the osseointegrated bond is measured by looking at its microscopic structure or by mechanically quantifying its shear strength. From previous studies that carried out in vivo tests it can be inferred that the shear strength of the interfacial bond between bone and a HA coated implant ranges from approximately 5MPa to 17MPa (Hayashi et al., 1994, Muller et al., 2006, Vercaigne et al., 1998, Xue et al., 2005, Xue et al., 2004).

6.10.3.2 Tensile Test

The tensile test is one of the most established mechanical tests in engineering today. It is used to evaluate a huge variety of materials from steel to vascular tissue. Equation 3.10 shows how the tensile strength was calculated.

\[
\text{Tensile Strength (MPa) = Peak Force/ Area of bone-implant contact.} \quad (6.4)
\]

Figure 6-19 shows the assembly of the tensile test employed in this study. The Sawbone samples are simple rectangular shaped blocks. The titanium sample has machined flat sides which are bonded to the Sawbone samples. Four clamps were designed. The clamps were designed so that the thickness could be maximised to avoid any deformation during the testing. The Sawbone samples were machined from a biomechanical testing block which was purchased from Sawbones. This biomechanical testing block was composed of the same polyurethane foam that is used in the composite femurs.
The objective of the tensile test was to quantify the tensile strength of the bond formed between a femoral stem and the biomechanical testing block using various adhesives. Firstly, the surface of the femoral stem was replicated on the test pieces. Secondly, the flat sides of the test pieces were bonded to the biomechanical block using various adhesives. Thirdly, the tensile test was performed using the Tinius Olsen tensile tester with a claw type attachment and finally the force required to pull the foam from the test piece was recorded. The adhesive with a tensile strength closest to those found in the literature should be used for securing the femoral stem in the composite Sawbones. From previous studies it can be inferred that the tensile strength of the interfacial bond between a HA coated implants and bone is between 1 and 2MPa (Lin et al., 1992, Lin et al., 1998).

6.10.3.3 Fracture Toughness Test

The interfacial fracture toughness (Kc) can be calculated by the following equation:

\[
K_c = \frac{\lambda^3 P Y}{B \sqrt{W}}
\]  \hspace{1cm} (6.5)

Where \( W \) = sample width, \( B \) = sample thickness, \( P \) is the load, \( Y \) is a constant and \( \lambda \) is a scale factor which is determined by the elastic properties of the two materials in the compact sandwich.

\[
\lambda = \sqrt{\frac{1 - \alpha}{1 - \beta^2}}
\]  \hspace{1cm} (6.6)
α and β are commonly referred to as Dundurs’ parameters and they estimate the elastic mismatch across an interface of two dissimilar materials. Their equations are given in equations 6.7 and 6.8.

\[
\alpha = \frac{E_1\left(1 - \nu_2^2\right) - E_2\left(1 - \nu_1^2\right)}{E_1\left(1 - \nu_2^2\right) + E_2\left(1 - \nu_1^2\right)} \tag{6.7}
\]

\[
\beta = \frac{E_1\left(1 - \nu_2^2 - 2\nu_2^2\right) - E_2\left(1 - \nu_1 - 2\nu_1^2\right)}{2E_1\left(1 - \nu_2^2\right) + 2E_2\left(1 - \nu_1^2\right)} \tag{6.8}
\]

Where \(E_1, E_2, \nu_1\) and \(\nu_2\) represent the elastic moduli and Poisson’s Ratio of the bilayer materials.

\(\psi\) is a correction factor and is a function of the interlayer thickness and width.

\[
\psi = 1 - 14.07\left(\frac{h}{W}\right) + 136\left(\frac{h}{W}\right)^2 - 538.8\left(\frac{h}{W}\right)^3 \tag{6.9}
\]

Where \(h\) = adhesive layer thickness

\(Y\) is a constant and is a function of the crack length and sample width.

\[
Y = \left(2 + \frac{a}{W}\right) \frac{0.886 + 4.64\left(\frac{a}{W}\right) - 13.32\left(\frac{a}{W}\right)^2 + 14.72\left(\frac{a}{W}\right)^3 - 5.6\left(\frac{a}{W}\right)^4}{\left(1 - \left(\frac{a}{W}\right)^\frac{3}{2}\right)^\frac{3}{2}} \tag{6.10}
\]

Where \(a\) = crack length,

The compact sandwich method for evaluating fracture toughness of the bone-implant bond is relatively well-defined in the field of orthopaedic and dental research. The sandwich test involves a layer of one material ‘sandwiched’ between two layers of another material with a precrack in place at one of the interfaces. This test was originally developed for testing adhesive joints and interfacial bonding strength of composite laminates and has since been adopted by the biomechanical field. Figure 6-20 shows a fracture toughness model.
Figure 6-20 The fracture toughness model

Figure 6-21 shows the assembly of the fracture toughness test. The Sawbone sample and the titanium test piece both had 2mm holes drilled through. This allowed a wire rope to be attached to both sample in order to apply the forces to pull them apart. The holes were lined up so the force was applied from the same distance to the applied crack. The titanium sample had a flat surface machined onto it which was the interface for bonding to the Sawbone. The Sawbone samples were machined from a biomechanical testing block which was purchased from Sawbones. This biomechanical test block was composed of the same polyurethane foam that is used in the composite femurs.

Figure 6-21 The assembly diagram of the fracture toughness test

The objective of the fracture toughness test was to quantify the fracture toughness of the bond formed between a femoral stem and the biomechanical testing block using various adhesives.
Firstly, the surface of the femoral stem was replicated on the test pieces. Secondly, the flat side of the test pieces were bonded to the polyurethane foam using various adhesives. Thirdly, a measured crack was created at the bond interface using a surgical blade and the fracture toughness test was performed using the Tinius Olsen tensile tester with a wire rope attachment. Finally, the force required to pull the foam and the test piece apart was recorded. The adhesive with a fracture toughness at least higher than those found for the bone-bone cement interface should be used for securing the femoral stem in the composite Sawbone (Wang and Agrawal, 1997, Wang et al., 1994).

6.10.4 Test Specimen Preparation
The following steps were used to prepare each test specimen:

1. The Sawbone samples were shaped to the specifications for each test.
2. The samples were then rinsed under running water to remove any dust or debris left over from the machining and shaping.
3. The samples were dabbed dry and placed on a wire tray (Figure 6-22) and allowed to dry for 12 hours. In the case of the push-out specimens the walls of the holes were dabbed dry using cotton buds.
4. The titanium samples were cleaned down with alcohol wipes to remove any residue from the machining process and also placed on the wire tray and allowed to dry.

6.10.4.1 Push Out Specimen
The following steps were used to prepare each specimen for the push out test:

1. The Sawbone sample and titanium sample were both marked with a letter corresponding to the type of adhesive to be used and a number to track each specimen.
2. The adhesive was then applied as per the instructions. A small plastic rod was used to apply the adhesive to the inner walls of the hole. The excess adhesive was removed using cotton buds.
3. The time and date were noted to monitor curing time.
4. The specimen was then returned to the wire tray and allowed to cure.

6.10.4.2 Tensile Specimen
The following steps were used to prepare each specimen for the tensile test:
1. The Sawbone sample and titanium sample were both marked with a letter corresponding to the type of adhesive to be used and a number to track each specimen.

2. The adhesive was then applied as per the instructions and the excess was removed using cotton buds.

3. The time and date were noted to monitor curing time.

4. The two pieces were held together using a 100mm quick ratcheting bar clamp and allowed to cure for the support time.

5. After the support time had elapsed the ratcheting clamp was removed and the second Sawbone piece was attached in the same fashion as the first.

6. The tensile ‘sandwich’ was then held once more using a ratcheting bar clamp for the support time.

7. After that time had elapsed the ratcheting clamp was removed and the specimen was returned to the wire tray for further curing.

6.10.4.3 Fracture Toughness Specimen

The following steps were used to prepare each specimen for the fracture toughness test:

1. The Sawbone sample and titanium sample were both marked with a letter corresponding to the type of adhesive to be used and a number to track each specimen.

2. The adhesive was then applied as per the instructions and the excess was removed using cotton buds.

3. The two pieces were held together using a 100mm quick ratcheting bar clamp and allowed to cure for the support time.

4. The time and date were noted to monitor curing time.

5. After the support time had elapsed the ratchet clamp was removed and the specimen was returned to the wire tray for further curing.

6. After curing was complete a mark was made at a measured distance into the bond.

7. Using a surgical scalpel a cut was made to the mark.

8. This cut was the induced crack.
6.10.4.4 Push-Out

The following steps were preformed for the push out tests:

1. An M16 threaded bar with a diameter eight hole through the centre was secured into the base of the Tinus Olsen machine to the point that the top of the bar was flush with the base.

2. The test specimen was lined up so that the titanium rod was directly in line with the diameter eight hole in the base.

3. The crosshead was jogged into position just above the titanium piece (Figure 6-23).

4. The testing was carried out at a rate of 2mm/min.
6.10.4.5 Tensile

The following steps were performed for the tensile test:

1. One set of clamps was secured to base fitting.

2. The second set of clamps was attached to the crosshead fitting.

3. The crosshead was lowered to the point at which the two sets of clamps met.

4. At this point the test piece was slid into place.

5. The crosshead was jogged back to remove some slack (Figure 6-24).
6. The testing was carried out at a rate of 2mm/min.

6.10.4.6 Fracture Toughness Test

The following steps were performed for the fracture toughness test:

1. The wire rope was passed through the titanium test piece, then through the jig attached to the base of the testing machine.

2. The two ends of the wire rope were overlapped by approximately 10cm.

3. Two wire rope grips were used to secure the wire rope and were tightened with an adjustable spanner.

4. The same procedure was followed for the Sawbone portion and attached to the top portion of the Tinus Olsen machine.
5. The setup was jogged into a position such that some of the slack was removed from the wire rope (Figure 6-25).

6. It was estimated that a low force would be required to complete the test and so it was not feasible to remove all of the slack from the wire ropes. For this reason the testing was run at 10mm/min.

Figure 6-25 (A) Wire rope grips in place, (B) Testing set up, (C) Load Cell

6.10.5 Surface Finish

The stems that were provided did not have the HA coating but were roughened using titanium arc deposition. It was the aim of this study to have the test pieces for the bond testing arc
deposited so that the surfaces would be equated, however this was not possible. The arc deposited surface is shown below in Figure 6-26. The surface of one of the test pieces is also shown in Figure 6-26. It is obvious that these two surfaces are not similar and so the purpose of the testing is to prove the viability of the tests and also to a certain extent, to compare the adhesive types. The values themselves are not suitable for input into the finite element model but the results should give an indication as to which adhesive would be closest to an osseointegrated bond and therefore give the best representation.

![Figure 6-26](image)

Figure 6-26 (A) Arc deposited surface of the femoral stem, (B) Machined surface of a titanium test piece, (C) Lower magnification of the test piece

### 6.10.6 Material Testing Results

#### 6.10.6.1 Introduction

Three tests were investigated namely the push out test, the tensile test and the fracture toughness test. For each test three adhesives were tested and for each adhesive three samples were used. In total 27 tests were carried out initially with 3 more push out tests carried out at a later stage to confirm the results of the Araldite epoxy.

#### 6.10.6.2 Push Out Test

The height and diameter were measured with a digital caliper to calculate the surface area that was in contact between the two parts and the peak force was recorded from the Tinius Olsen tensile testing machine. From these values the shear strength of the bond was calculated. Samples A1 and C3 were discounted as there was an error with the load cell. Also sample A3
was found to be misaligned and was believed to have caught the edge of the cavity into which the samples were pressed and so this test resulted in high values.

Table 6-1 Push out test results

<table>
<thead>
<tr>
<th>Sample No.</th>
<th>Height (mm)</th>
<th>Diameter (mm)</th>
<th>Area (mm^2)</th>
<th>Peak Force (N)</th>
<th>Shear Strength (MN/m^2)</th>
<th>Standard Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Araldite1</td>
<td>16.6</td>
<td>6</td>
<td>312.7</td>
<td>-</td>
<td>-</td>
<td>10.8505805</td>
</tr>
<tr>
<td>Araldite2</td>
<td>16.91</td>
<td>6</td>
<td>318.6</td>
<td>1422</td>
<td>4.46</td>
<td></td>
</tr>
<tr>
<td>Araldite3</td>
<td>16.6</td>
<td>6</td>
<td>312.7</td>
<td>6195</td>
<td>19.81</td>
<td></td>
</tr>
<tr>
<td>Devweld1</td>
<td>17.7</td>
<td>6</td>
<td>333.5</td>
<td>633</td>
<td>1.9</td>
<td>0.29813786</td>
</tr>
<tr>
<td>Devweld2</td>
<td>17.7</td>
<td>6</td>
<td>333.5</td>
<td>613</td>
<td>1.84</td>
<td></td>
</tr>
<tr>
<td>Devweld3</td>
<td>16.4</td>
<td>6</td>
<td>309</td>
<td>418.5</td>
<td>1.35</td>
<td></td>
</tr>
<tr>
<td>Radionics1</td>
<td>16.2</td>
<td>6</td>
<td>305.2</td>
<td>315.2</td>
<td>1.03</td>
<td>2.05816593</td>
</tr>
<tr>
<td>Radionics2</td>
<td>17</td>
<td>6</td>
<td>320.3</td>
<td>1263</td>
<td>3.94</td>
<td></td>
</tr>
<tr>
<td>Radionics3</td>
<td>18.4</td>
<td>6</td>
<td>346.7</td>
<td>-</td>
<td>-</td>
<td></td>
</tr>
</tbody>
</table>

The results in Table 6-1 show that the Araldite epoxy provided the closest results of the three adhesives to the shear strength values for an osseo integrated bond in literature. The other two adhesives values were low but because only one of the samples was tested properly testing was repeated for the Araldite epoxy. The results of this are shown in Table 6-2 below and the values are lower than the original samples tested. These results for the Araldite epoxy are still the highest of the three adhesives and are closest to the values for the osseo integrated bond.

Table 6-2 The push out results of the Araldite epoxy

<table>
<thead>
<tr>
<th>Sample No.</th>
<th>Height (mm)</th>
<th>Diameter (mm)</th>
<th>Area (mm^2)</th>
<th>Peak Force (N)</th>
<th>Shear Strength (MN/m^2)</th>
<th>Standard deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Araldite4</td>
<td>9.65</td>
<td>6</td>
<td>173</td>
<td>544.5</td>
<td>3.14</td>
<td>0.35002</td>
</tr>
<tr>
<td>Araldite5</td>
<td>9.84</td>
<td>6</td>
<td>177</td>
<td>646.5</td>
<td>3.66</td>
<td></td>
</tr>
<tr>
<td>Araldite6</td>
<td>9.72</td>
<td>6</td>
<td>175</td>
<td>664.5</td>
<td>3.81</td>
<td></td>
</tr>
</tbody>
</table>

6.10.6.3 Tensile Test

As in the push out test, the length and width were measured with a digital calliper to calculate the surface area that was in contact between the two parts and the peak force was recorded from the Tinius Olsen tensile testing machine and can be seen in Table 6-3. Samples B2, B3 and C1 all failed prematurely and the bond became loose with very little force applied. Both the Devweld 531 and the Radionics high strength epoxy showed low peak forces and therefore low tensile strength. The Araldite epoxy again performed better than the other two with values within the values seen in literature for an osseo integrated bond.
Table 6-3 The results from the tensile test

<table>
<thead>
<tr>
<th>Sample No.</th>
<th>Length (mm)</th>
<th>Width (mm)</th>
<th>Area (mm(^2))</th>
<th>Peak Force (N)</th>
<th>Tensile Strength (MPa)</th>
<th>Standard Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Araldite1</td>
<td>30.72</td>
<td>4</td>
<td>122.9</td>
<td>202.5</td>
<td>1.647949219</td>
<td>0.743289</td>
</tr>
<tr>
<td>Araldite2</td>
<td>32.22</td>
<td>4.34</td>
<td>139.8</td>
<td>173.3</td>
<td>1.23931954</td>
<td></td>
</tr>
<tr>
<td>Araldite3</td>
<td>31.9</td>
<td>4.55</td>
<td>145.1</td>
<td>389.2</td>
<td>2.681456475</td>
<td></td>
</tr>
<tr>
<td>Devweld1</td>
<td>31.9</td>
<td>4.5</td>
<td>143.6</td>
<td>85.8</td>
<td>0.597701149</td>
<td>-</td>
</tr>
<tr>
<td>Devweld2</td>
<td>30.7</td>
<td>4.5</td>
<td>138.2</td>
<td>-</td>
<td>-</td>
<td></td>
</tr>
<tr>
<td>Devweld3</td>
<td>29.9</td>
<td>4.5</td>
<td>134.6</td>
<td>-</td>
<td>-</td>
<td></td>
</tr>
<tr>
<td>Radionics1</td>
<td>30.4</td>
<td>4.5</td>
<td>136.8</td>
<td>-</td>
<td>0.094684</td>
<td></td>
</tr>
<tr>
<td>Radionics2</td>
<td>30</td>
<td>4.5</td>
<td>135</td>
<td>82.7</td>
<td>0.612592593</td>
<td></td>
</tr>
<tr>
<td>Radionics3</td>
<td>30.5</td>
<td>4.5</td>
<td>137.3</td>
<td>65.7</td>
<td>0.478688525</td>
<td></td>
</tr>
</tbody>
</table>

6.10.6.4 Fracture Toughness Test

To calculate the fracture toughness of the bond the equations discussed earlier were used. The elastic modulus was 110GPa for the titanium pieces and 15GPa for the Sawbone samples. Poisson’s Ratio was 0.1 for titanium and 0.3 for the Sawbone. The adhesive layer thickness was given as 0.05mm for each sample. Sample width, and thickness as well as the crack length were needed and were calculated using a digital calliper. The peak load was recorded from the Tinius Olsen testing machine and can be seen in Table 6-4. Similar to the tensile test the Devweld 531 samples did not create a good bond and all three samples failed as the initial load was applied. The high strength epoxy also failed for one sample, but did have one value within the fracture toughness values seen in literature and one value close to these values. The Araldite epoxy showed a good bond being created between the two surfaces and the values were within or slightly higher than the values seen in literature. This test however has not been well documented in literature for bone and therefore these values are only an indication of the fracture toughness of the bond.

Table 6-4 The results of the fracture toughness test

<table>
<thead>
<tr>
<th>Sample No.</th>
<th>Sample Width (mm)</th>
<th>Sample Thickness (mm)</th>
<th>Crack Length (mm)</th>
<th>Peak Load (N)</th>
<th>Kc (Mpa/m(^{1/2}))</th>
<th>Standard Deviation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Araldite1</td>
<td>24.46</td>
<td>6.7</td>
<td>2.34</td>
<td>37</td>
<td>1.553</td>
<td>0.4148994</td>
</tr>
<tr>
<td>Araldite2</td>
<td>24.06</td>
<td>6.4</td>
<td>1.74</td>
<td>32</td>
<td>2.125</td>
<td></td>
</tr>
<tr>
<td>Araldite3</td>
<td>25.7</td>
<td>6.25</td>
<td>3.13</td>
<td>36</td>
<td>2.359</td>
<td></td>
</tr>
<tr>
<td>Devweld1</td>
<td>25.1</td>
<td>6.12</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Devweld2</td>
<td>25.27</td>
<td>5.9</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Devweld3</td>
<td>25.46</td>
<td>6.36</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Radionics1</td>
<td>24.45</td>
<td>6.45</td>
<td>3.19</td>
<td>35</td>
<td>2.183</td>
<td>1.1256733</td>
</tr>
<tr>
<td>Radionics2</td>
<td>24.86</td>
<td>6.51</td>
<td>4.48</td>
<td>9.6</td>
<td>0.591</td>
<td></td>
</tr>
<tr>
<td>Radionics3</td>
<td>24.31</td>
<td>6</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>
6.10.7 Material Testing Discussion

A decision had to be made which adhesive would be used for the experimental setup. From the tests carried out it was seen that the Araldite epoxy gave the closest results to an osseointegrated bond in all three tests. For the fracture toughness test it was seen that the Araldite samples all broke not only at the adhesive interface but also into the Sawbone as seen in Figure 6-27. This would suggest the bond would be of similar strength as the Sawbone samples. This also occurred for the pushout tests for the Araldite. The surface finish differs in this study compared to actually finish of a prosthesis stem, but if the failure occurs within the Sawbone and not at the actually bond interface then the surface finish may not play as an important part in the failure and therefore the comparisons to literature would be beneficial and more representative. The Devweld 531 showed in the fracture toughness test and the tensile test that the bond formed was not a secure bond and although a rough surface, which would be the case with the prostheses, the bond would just not be able to replicate the properties needed. The Radionics high strength epoxy showed results similar to the literature for the fracture toughness test, however the results in the push out and tensile test were lower than the literature values for the osseointegrated bond and this could give loosening and instability in the experimental setup that would occur at lower loads then in vivo. Therefore the Araldite epoxy was the best adhesive tested and was chosen to be used to create the osseointegrated bond in the experimental setup.
6.11 Experimental Results

6.11.1 Introduction

Numerical modelling was used in the previous chapters to assess the stresses occurring in an implanted prosthesis. This was done by examining the stresses along the surface of the interface. The experimental setup was used to investigate the effects of the mechanical load on the stability of the implant by applying the load over an extended period of time. Linear Voltage Displacement Transducers (LVDT) were used to measure the micromotion between the Sawbone and the surface of the prosthesis at three locations. After examining the positions of the LVDTs in other studies the three positions chosen were one distally near the tip of the stem, one slightly distal to the lesser trochanter and the final one slightly distal to the greater trochanter.

The setup of the experimental rig has been discussed earlier in this chapter. Three load configurations were applied on the rig on three different Sawbone femur models. The loads used were taken from the average values from all the Total Hip Replacement (THR) and Birmingham hip Replacement subjects from the results of the musculoskeletal work carried out earlier in this
thesis. The load used consisted of the walking and stair climbing loads. The stair climbing load was applied once every 39 walking cycles (Britton and Prendergast, 2005, Morlock et al., 2001). Three load configurations were applied, the first had the load in the X, Y and Z directions, the second configuration did not include the load in the X direction and the third did not include the load in the Y direction. The results from these models are discussed below.

6.11.2 Micromotion results
Micromotion readings were taken at three locations along the surface of the bone stem interface. The position of these LVDT’s was decided from the investigation into the literature from previous studies. The position of the LVDT’s were the same as displacement transducers 2a, 2b and 4b in the study by Kassi et al. (2005), and similar positions to those used in other studies (Abdul-Kadir et al., 2008, Britton and Prendergast, 2005, Fottner et al., 2009, Maher et al., 2001). The positions for the LVDT’s can be seen in figure 6-28. LVDT 1 was also in the proximal stem region however this was positioned on the anterior surface. This transducer measured the movement in the anterior posterior direction. Figure 6-28. LVDT 2 was located at the medial surface in the proximal stem region. This transducer measured the lateral medial movement. The final transducer, LVDT 3, was positioned on the lateral surface of the distal stem region. This transducer also measured the medial lateral movement but at the distal region of the prosthesis.

![Figure 6-28 The positions of the 3 LVDT's in the experimental model](image-url)
6.11.2.1 Elastic and Plastic Deformations and Displacements
There are two types of deformation that can occur and these are expressed by displacement. It is therefore important to understand what the displacements show. The two types of deformation are elastic and plastic. Elastic deformation occurs when the material moves back to its original shape. The displacement that expresses this can be found by comparing the minimum and maximum displacements during one cycle. The plastic deformation occurs when the stress is sufficient to cause permanent deformation to the material. The displacement that can identify this is the difference between the average displacement in the first cycle and the average displacement in the last cycle investigated.

6.11.2.2 Elastic Deformation
As explained the elastic deformation can be expressed by comparing the minimum and maximum displacements during one cycle. This will show how much the implant is moving throughout the cycle. The minimum and maximum displacement values were recorded for the first 24 cycles. This was again carried out at every 0.5 million cycles. From these minimum and maximum displacements the difference or movement per cycle was calculated. Figure 6-29 below shows the differences for each of the LVDT’s for the three different loading configurations.
The elastic micromotion shows the maximum amount of movement that occurs during each cycle. In all three LVDT’s the YZ model has a very large micromotion at the start of the testing in comparison to the other two models. This would indicate that without the presence of the X direction force the model is able to move more due to the effect of the Y direction load. However the elastic movement in all three models starts to flatten out after 1 million cycles for LVDT 1 and LVDT 2, and after 1.5 million cycles for LVDT 3. This would indicate that after a million cycles or so the elastic displacement stays the same for the rest of the cycles. This is

Figure 6-29 The elastic micromotion for all three models at LVDT 1, LVDT 2 and LVDT 3
seen to be the case in the XYZ model after 5 million cycles. However, this does not imply that there is no loosening occurring. To investigate if there was movement within the model that was not elastic, the minimum displacements were investigated.

6.11.2.3 Plastic Deformation

**Minimum Displacement**

Table 6-5, Table 6-6, Table 6-7 and Figure 6-30 graphically present the minimum displacements for each of the LVDT locations.

Table 6-5 The minimum displacements in the XYZ model at all three LVDT locations

<table>
<thead>
<tr>
<th>Cycles (Million)</th>
<th>LVDT 1 (µm)</th>
<th>LVDT 2 (µm)</th>
<th>LVDT 3 (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>-39.8</td>
<td>-162.5</td>
<td>121.8</td>
</tr>
<tr>
<td>0.5</td>
<td>-69.9</td>
<td>-168.9</td>
<td>212.7</td>
</tr>
<tr>
<td>1</td>
<td>-76.5</td>
<td>-168.6</td>
<td>209.5</td>
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<td>1.5</td>
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<td>-170.1</td>
<td>335.5</td>
</tr>
<tr>
<td>2</td>
<td>-93.7</td>
<td>-173.8</td>
<td>355</td>
</tr>
<tr>
<td>2.5</td>
<td>-98.3</td>
<td>-173</td>
<td>371.9</td>
</tr>
<tr>
<td>3</td>
<td>-134.8</td>
<td>-178.3</td>
<td>428</td>
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<tr>
<td>3.5</td>
<td>-134.1</td>
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<td>4</td>
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</tr>
<tr>
<td>4.5</td>
<td>-171.4</td>
<td>-220.8</td>
<td>660</td>
</tr>
<tr>
<td>5</td>
<td>-164.3</td>
<td>-227</td>
<td>664.6</td>
</tr>
</tbody>
</table>

Table 6-6 The minimum displacements in the XZ model at all three LVDT locations

<table>
<thead>
<tr>
<th>Cycles (Million)</th>
<th>LVDT 1 (µm)</th>
<th>LVDT 2 (µm)</th>
<th>LVDT 3 (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>-136.9</td>
<td>24</td>
<td>927.7</td>
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<tr>
<td>0.5</td>
<td>-140.3</td>
<td>28.1</td>
<td>1078.6</td>
</tr>
<tr>
<td>1</td>
<td>-150.8</td>
<td>23.1</td>
<td>1145.6</td>
</tr>
<tr>
<td>1.5</td>
<td>-152.4</td>
<td>22.4</td>
<td>1200.5</td>
</tr>
<tr>
<td>2</td>
<td>-202.4</td>
<td>12</td>
<td>1414.3</td>
</tr>
</tbody>
</table>

Table 6-7 The minimum displacements in the YZ model at all three LVDT locations

<table>
<thead>
<tr>
<th>Cycles (Million)</th>
<th>LVDT 1 (µm)</th>
<th>LVDT 2 (µm)</th>
<th>LVDT 3 (µm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>-40.7</td>
<td>-130.6</td>
<td>986.5</td>
</tr>
<tr>
<td>0.5</td>
<td>-38.1</td>
<td>-226.7</td>
<td>1183.3</td>
</tr>
<tr>
<td>1</td>
<td>-76.5</td>
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<td>1399.3</td>
</tr>
<tr>
<td>1.5</td>
<td>-82.7</td>
<td>-279.2</td>
<td>1434.6</td>
</tr>
<tr>
<td>2</td>
<td>-149.7</td>
<td>-298.9</td>
<td>1739.4</td>
</tr>
</tbody>
</table>
From the tables and graphs above it can be seen that there is a change in the minimum displacements in all the models. For LVDT 1 and LVDT 2 all three models have their minimum displacements decreasing indicating that the LVDT’s are under compression with the prosthesis being pushed towards these locations. LVDT 3 shows the minimum displacements increasing and this would indicate that the prosthesis is moving away from this LVDT. For LVDT 1 and LVDT 3 all three models show similar change in displacement. However LVDT 2 shows the
YZ model decreasing in minimum displacement at a higher rate than the other two locations. This will be investigated further in the plastic deformation section.

**Average Displacement**

Plastic deformation occurs when the stress is sufficient to cause permanent deformation. This is measured by getting the difference between the average displacement in the first cycle and the average displacement in the last cycle. The difference between these average displacements is permanent or plastic deformation. Figure 6-31 show the differences between the average displacements in all three models at each of the LVDT’s.
LVDT 1 is located at the proximal posterior site of the femur. This location is directly parallel to the Y direction load and would be the main area this force affects. From Figure 6-31 it can be seen that model YZ shows a lot of movement at this location indicating that the Y direction load is causing the prosthesis to permanently move. As the force is decreasing it indicates that the prosthesis is being pushed against the posterior wall. The other two models do not show significant movement at this location. This may indicate that the X direction load plays a part in
reducing the affects of the Y direction as the XYZ model does not show similar displacement to the YZ model.

LVDT 2 is located in the proximal neck region on the medial side. This LVDT is parallel to the X direction load. It can be seen from Figure 6-31 that all three models show a decrease in the difference between the average displacements. Again as the micromotion is moving in the negative direction it would indicate that the prosthesis was compressed against the medial wall which is the neck region. Although all three models show signs of movement, the YZ model is the model with the greatest change after 2 million cycles. The XYZ model indicates that this trend may continue as this model continues to show movement. The Z direction load tends to be the largest load of the three directions and this would be one of the main contributors to the stresses in the neck region and as the Z load is included in all three models this may explain the movement in all three models.

LVDT 3 is located in the distal part of the prosthesis cavity, on the lateral side. This location measured any movement in the latero-medial direction. All three models showed a change in their displacements and all three can be seen to have the same pattern of change. As the micromotion is getting larger the motion for all three models can be seen to be moving away from this area indicating that the prosthesis is being pushed towards the medial region distally. No major differences can be seen between the models in the first 1.5 million cycles. However, both the XZ and YZ models show a bigger increase between 1.5 and 2 million cycles in comparison to the XYZ model. As the former models only went to 2 million cycles it is difficult to draw any conclusions from this.

The displacements of the LVDT 3 are at least three times that of the other locations in both elastic and plastic deformation. One of the main reasons for this is the differences between the numerical and experimental models in the distal area. The experimental models used Sawbones to represent the cadaver bone and these include a reamed out hole to represent the Haversian canal. In the distal region this then means there is not a contact between the prosthesis and bone as the distal part of the prosthesis is smaller than the canal diameter. This is not represented in numerical models as the prosthesis is used as a template to create the cavity and no canal is present.
6.12 Discussion

The numerical modelling chapter 4 investigated the occurrence of stresses at the stem-bone interface for one cycle and chapter 7 will investigate if these stresses were over the threshold value proposed by other studies for damage initiation (McNamara and Prendergast, 2007, McNamara et al., 2006, Scannell and Prendergast, 2005, Scannell and Prendergast, 2009a, Scannell and Prendergast, 2009b). To identify if the mechanical loading would cause the damage to initiate and aid loosening to occur, an experimental rig was designed, fabricated and commissioned. The experimental setup described earlier in this chapter applies the load in the three directions X, Y, and Z. This allowed for the angle of the load to be changed at any time during the cycle. The rig therefore gave realistic loads and displacements were calculated accordingly.

For the elastic deformation, a lot more movement occurs in the YZ models than the XYZ or XZ models indicating that the X direction plays a part in controlling the effect of both the Y and Z directional loads and that without this there is more movement at the surface interface. However, results for all the models start to flatten out and if the models followed the trends of the XYZ model then the elastic movement will become constant. Plastic deformation was thought to occur as changes in the minimum displacement, while the elastic deformation became constant would suggest this being the case. The differences in average displacement confirmed this was the case. Similar to the elastic deformation the greatest changes in the plastic deformation were seen in the YZ model.

The main goal of the experimental setup was to investigate the displacement and possible loosening that would occur after the hip joint was subjected to the forces in everyday activities. The results show that there are both elastic and plastic displacements that occur in the hip joint due to the mechanical loading. At the proximal medial location of LVDT 2 and the distal lateral region at LVDT 3, progressive movement was seen with all three load configurations. Although LVDT 1, located on the proximal posterior surface, only showed small movements in the XYZ and XZ models, both do show a slow increase in movement. The YZ model at LVDT 1 showed a large change between the average displacements, indicating that both the Y and Z forces are important factors in causing loosening but also that the X direction plays a role in controlling the effects of these two forces.

The displacements of the LVDT 3 are at least three times that of the other locations in both elastic and plastic deformation. The canal was mentioned as one reason for this difference as the numerical models did not show high stresses that would indicate the biggest movement in this area. Another reason for this was the surface fit of the prosthesis when implanted into the femur. The numerical models had a perfect fit as the prosthesis was used as the template for the
creation of the cavity. In the experimental setup the cavity was created using a rasp and then the prosthesis was press fit into the cavity. This method would not always give a perfect surface finish and would contain imperfections that would increase the chance of movement within the model.

The displacements recorded in this set of results indicate signs that the mechanical loading could lead to the loosening of the prosthesis. However, it is important to identify that none of the three models has become unstable at the end of its testing whether it was run for 2 million or 5 million cycles. What these results do show is that the mechanical loading can change the position of the implanted prosthesis over a long period of time.

Furthermore, these experimental results do show that the typical stresses seen in numerical models of this experimental setup, created by applying everyday movement loading patterns also cause experimentally measured significant micromotions between the implant and the bone surface. These results indicate that using numerical models to predict stress distributions in various implanted femur models and evaluating the predicted stresses in terms of magnitude and distribution, is an appropriate approach to determine if post operative hip patients have a typical gait that is an aseptic loosening producing motion as the predicted stresses can be related to interface micromotions and consequently aseptic loosening.

There are limitations that are associated with the experimental models. The main limitation is the in vitro nature of the model. As the model is not in vivo the biological behaviour at the hip joint cannot be examined or taking into account. This biological behaviour has been found to be a factor in loosening of hip replacements. However, this study is only concerned with the effects the mechanical loading has on the stability of the prosthesis and not the overall biological bone remodelling that occurs.
CHAPTER 7

Numerical Modelling Results
Chapter 7 Numerical Modelling Results

7.1 Introduction
The setup and parameters used for the numerical modelling have been described in chapter 4. The main aim of this chapter is to investigate the effects the mechanical force has on the stability of the prosthesis and to further investigate if certain forces play important roles in the loosening of hip prostheses. The models all consisted of the same geometry with only the load being varied. The geometry can be seen below in Figure 7-1. It can be seen that the prosthesis makes contact with both the cancellous and cortical bone.

![Figure 7-1 The three parts created in the numerical modelling of the hip joint](image)

7.2 Maximum Load Configurations
The average loading configurations give a good indication of the effects that everyday activities have on the interface between the stem and bone. However the gait cycle for a person can vary substantially over 9 or ten gait cycles. It was therefore decided to investigate the worst case scenario and examine the surface interface of models with the maximum loads for each everyday activity. Each subject’s gait data was investigated to see what the maximum
normalised force calculated was. The maximum values were then applied to the numerical models using the average body weight from the 23 subjects tested. This body weight was used as the Sawbone femur represents an average femur geometry and therefore using the body weight of the subject that created the force could under or over estimate the stresses along the surface.

7.2.1 Walking
Walking is one of the most common everyday motions to be carried out. It is important to investigate the stresses that are caused during the motion along the surface. Figure 7-2 shows the surface plots of the walking load with the XYZ load, The YZ load and the XZ load applied. Each Gruen zone is then examined in Figure 7-3 to compare the differences in stress between the different models.
Figure 7-2 Surface plots of the thresholding groups for the max walking load configurations
Figure 7-3 Graphical bar charts for each Gruen zone for the max walking load configurations
The results show that the presence of the load in the Y direction increases the stresses found throughout the middle and distal parts of the model. The main differences in stress are seen in the posterior region along Gruen Zones 5P, 6P and 7P. The graphs in Figure 7-3 also show that the threshold is only exceeded within three Gruen Zones. These are 1A, 7A and 7P. These are all proximal zones along the neck of the prosthesis and although the stresses along 5P and 6P are increased in the XYZ and XZ models they do not show any areas where damage may occur. The area over the threshold in the model without the Y direction load is higher than the other two models suggesting that the Y load may also play a role in reducing the effect of the Z direction load which is generally the greatest load.

7.2.2 Up Steps
Previous studies have investigated the effects of walking up and down stairs and have found that stair climbing can cause some of the biggest hip joint contact forces. It is important to investigate the stresses that are caused during this motion along the surface. Figure 7-4 shows the surface plots of the up steps load with the XYZ load applied, The YZ load and the XZ load. Each Gruen zone is then examined in Figure 7-5 to compare the differences in stress between the different models.
Figure 7-4 Surface plots of the thresholding groups for the max Up Steps load configurations
Figure 7-5 Graphical bar charts for each Gruen zone for the max up steps load configurations
The results of the maximum up steps load configuration show very different results to the walking cycle. Although like the maximum walking cycle load configuration the proximal Gruen Zones show high stresses with areas over the threshold, distally the results are different. It can be seen that large areas in Gruen zones 5P, 6P and to some extent 2P show that the stress is over the threshold for both the models with the Y load. Seven zones have stresses that are over the threshold including the zones around the neck of the prosthesis and the zones down the posterior side of the surface. It can clearly be seen that the Y direction plays an important role in increasing these stresses as the model without the Y direction shows stresses greatly reduced in all zones with the exception of 7A.

7.2.3 Down Steps
As mentioned in the up steps section previous studies have investigated the effects of the up and down stairs motions and have found that stair climbing can cause some of the biggest hip joint contact forces. It is important to investigate the stresses that are caused during this motion along the surface. Figure 7-6 shows the surface plots of the down steps load with the XYZ load applied, The YZ load and the XZ load. Each Gruen zone is then examined in Figure 7-7 to compare the differences in stress between the different models.
Figure 7-6 Surface plots of the thresholding groups for the max down steps load configurations
Figure 7-7 Graphical bar charts for each Gruen zone for the max down steps load configurations
The maximum down steps loading configuration showed high stresses in the proximal regions along the neck of the prosthesis. The three Gruen zones to breach the threshold were 7A, 7P and a very small area in 1A. The stresses distally were a lot lower and unlike the up steps motion did not reach the damage initiation threshold. Interestingly the stresses in the load configuration without the X load showed higher stresses than the other two models indicating that the presence of the X load reduces the stresses in the proximal region. This is expected as the X load pushes the prosthesis away from the neck region and therefore would reduce the stress.

7.2.4 Sitting

Sitting and standing are two motions that cause the lowest hip joint contact forces of the 5 motions investigated. Although this load is lower in magnitude, the loads in the Y direction are as big as or higher than the Y loads associated with the other motions. Therefore, it is important to investigate the stresses that are caused during these motions along the surface. Figure 7-8 shows the surface plots of the sitting load with the XYZ load applied, The YZ load and the XZ load. Each Gruen zone is then examined in Figure 7-9 to compare the differences in stress between the different models.
Figure 7-8 Surface plots of the thresholding groups for the max sitting load configurations
CHAPTER 7 – NUMERICAL MODELLING RESULTS

Figure 7-9 Graphical bar charts for each Gruen zone for the max sitting load configurations
The maximum sitting load showed similar results to the up steps motion with high stresses along the distal posterior side. The main area of the stresses over the threshold was between Gruen Zones 6P and 2P. The stresses were over the threshold in all of the proximal Gruen Zones (1A, 1P, 7A, 7P). In the model without the Y load the stresses reduced significantly and no Gruen zone was over the threshold. This therefore further indicated that the Y load plays an important part in the damage initiation.

7.2.5 Standing
Figure 7-10 shows the surface plots of the standing load with the XYZ load applied, The YZ load and the XZ load. Each Gruen zone is then examined in Figure 7-11 to compare the differences in stress between the different models.
Figure 7-10 Surface plots of the thresholding groups for the max standing load configurations
Figure 7-11 Graphical bar charts for each Gruen zone for the max standing load configurations
The results of the maximum standing load were quite similar to the sitting results. As with the sitting and up steps motions the main areas where the stresses were over the threshold were seen in Gruen Zones 2P, 5P and 6P all in the posterior region. Threshold breaching stresses were also seen along the neck region in GZ 1A, 1P, 7A and 7P in the two models with the Y load applied. The model without the Y load applied again had no stresses over the threshold value and therefore again show the importance of the Y direction load.

7.3 Average load models
The average loads were applied for each of the 5 motions carried out. The walking motion was discussed in chapter 4 and the other four motions are discussed here. The up steps and down steps motions showed similar results to the three models for the walking motion seen in Figure 4-16. These models all showed the main stresses to be proximal with only low stresses in the middle and distal areas. Similar to the walking motion the models without a Y direction load were seen to have increased stresses at the proximal neck region. High stresses were also seen along the neck region in all the models. However, only the model without a Y direction load was seen to exceed the threshold value.

For the sitting and standing models the stresses were seen to be more spread out throughout the model. Similar to all the other motions the neck region on the surface plots was seen to have high stresses. However unlike the other motions the middle and distal regions were seen to have higher stresses. Stresses in these regions reached the threshold value in small areas in the neck distal regions. Even though the load magnitude for sitting and walking motions is the same or lower than other motions the stresses are higher in the distal regions. One main reason for this would be the Y direction load as this is higher in these motions than in the walking, up steps and down steps motions. This is further proven by looking at the model for these motions without the Y direction load. For these models the stresses are very low throughout the model. The surface plots for the three models for each of these motions discussed here can be seen in Appendix C.

7.4 Significant Values from Musculoskeletal Modelling Results
In the musculoskeletal modelling in chapter 3 statistical analysis was carried out between five main categories. These categories were gender (Male v Female), prosthesis head size (standard v Large), surgical approach (Anterolateral v Posterior), subjects after one surgery (Healthy hip v Replaced hip) and type of overall surgery (Healthy v THR v BIRM). When investigating the significant differences between these groups in the left and right hip joints 12 comparisons were found to be significant. For the 14 subjects that had one replaced hip, comparisons were made for each of the above mentioned groups for the healthy hip and the replaced hip. Eight of these comparisons were found to be significantly different. This gave 20 significantly different forces.
from the musculoskeletal modelling. Five of these significant forces were chosen to be investigated further, which two of these differences being in the same comparison and therefore four models were run. These forces were applied to numerical models to investigate if the differences in forces varied the stress values and possibility of aseptic loosening.

The model with the greatest difference between the two groups was found in the comparison between the standard prosthesis head size and the large prosthesis head size in the replaced hip for the up steps motion. This showed the large head size group having higher stresses along the interface surface with the posterior side of the surface showing the highest stress differences. The bar charts in Figure 7-17 indicate differences between the two groups in GZ 5P, 6P and 7P on the posterior side but also in 1A and 6A on the anterior side. The large prosthesis head size group shows stresses over the threshold values in several Gruen zones.

The other three models that were examined only showed small changes. The comparison between the healthy group and the THR group in the right hip for the walking motion demonstrated small changes along the posterior side in Gruen zones 2P, 3P and 5P, with the healthy group showing a small amount of the surface in GZ 7P to be over the threshold. The comparison between the healthy and replaced hips in the same subject for the walking motion only indicated that GZ 2P had a change in stresses and these were still low. The final comparison did not show any differences between the two groups and the forces were seen to be low. Figure 7-12 to Figure 7-20 show the surface plots and the stresses in each Gruen zone for the four comparisons analysed.
Figure 7-12 the surface plots for the healthy group and the THR group for the walking motion in the right hip
Figure 7-13 Graphical bar charts for each Gruen zone for the healthy and THR groups during the walking motion in the right hips
Figure 7-14 the surface plots for the healthy hip and THR hip in subjects with one hip replaced for the walking motion.
Figure 7-15 Graphical bar charts for each Gruen zone for healthy and replaced hips during the walking motion.
Figure 7-16 Surface plots for the up steps motion in the replaced hips comparing standard prosthesis head size and the large prosthesis head size.
Figure 7-17 Graphical bar charts for each Gruen zone in the standard and large prosthesis head size during the up steps motion.
Figure 7-18 Surface plots for the BIRM and THR subjects in the healthy leg during the down steps motion
Figure 7-19 Graphical bar charts for each gruen zone for the BIRM and THR subjects for the healthy leg during the down steps motion.
7.5 Identifying Threshold Force

The values of -5.4MPa for cancellous bone and -61.2MPa in cortical bone are important threshold values for the long term stability of a hip prosthesis. The peak loads from the average and max loads for each of the five motions have been applied in the models shown in previous sections and it was identified that stresses along the surface do reach the threshold in many of these models. The load in the Y direction has been shown in these models to be one of the main contributors to the stresses reaching the threshold value and plays an important part in the load’s effect on the long term stability. As the threshold is important and stresses need to be below this threshold and that the Y direction plays such an important part in this, it was decided to identify the maximum load in the Y direction that would give stresses under the threshold along the surface. As the walking motion is the most common the peak loads from the average walking motion were chosen. The loads in the Y direction were then altered between models. The first model started with no Y direction force and then one eighth of the original Y direction load was added on to each consecutive model until the final model which had a Y load 1 and a half times the original average walking Y direction load. For the average walking model the stresses closest to the threshold were identified in the proximal region and therefore the peak stresses were identified for the cancellous bone. The results can be seen in Figure 7-20.
Figure 7-20 The effect the Y direction load has on the peak stress recorded at the interface

Figure 7-20 shows the effect of the Y direction load on the stresses along the surface. A value over -5.4MPa is over the stress threshold for cancellous bone. When the Y load direction was zero a stress of -7MPa was recorded this was over the threshold value. The peak stress however reduced the higher the Y direction load became up to around half the original average Y direction load. At this point the peak surface stress was -3.54MPa well under the threshold stress value for cancellous bone. From this point the stresses started to rise with the increase of the load in the Y direction. This then keeps rising up until the last model that was modelled with 1 and a half times the original average walking Y direction load. These results indicate that without a force in the Y direction the threshold would be exceeded. One reason for this would be that this Y direction load would reduce the effect of the vertical Z direction load on the proximal surface along the neck of the prosthesis, and as the Y direction load is increased it reduces the effect of the Z direction load up until a point where the Y direction begins to influence the stresses. The peak force calculated from the average walking motion data showed no stresses over the threshold however a Y direction load just one eighth bigger was found to be over the threshold. Therefore the walking gait for most people is under the threshold but anything that would increase the Y direction load would bring the stresses over the threshold value.
7.6 Discussion

One of the main aims of the study was to investigate the effects the mechanical loading has on the bone-stem interface, and the overall effect this may have on the long term stability of the implanted prosthesis. This has been shown here with both the average and maximum peak loads for each of the 5 motions investigated. The neck of the prosthesis was seen to be an area where stresses are high relative to the other parts of the surface. This can be explained by the vertical force in the Z directions for most motions being the predominant load direction. The average forces with all three load directions applied for all the motions showed the stress levels to be under the threshold stress values used in previous studies (McNamara and Prendergast, 2007, McNamara et al., 2006, Scannell and Prendergast, 2005, Scannell and Prendergast, 2009a, Scannell and Prendergast, 2009b). This indicates that if subjects do not exceed the average forces for the everyday activities tested there would be a high possibility for good stability in the replaced hip. However, most of the motions showed stresses close to the threshold level for the forces from average motion loads. With gait differing from cycle to cycle even between the gait of the same person then there is a good possibility of some motion causing forces that would cause stresses that exceed the threshold value.

The X direction load was the lowest load direction for all the motions carried out in the testing. However during the statistical analysis this direction was found to be the direction with the most significantly different forces between the compared groups. All the comparisons that were found to have significantly different forces were modelled but the results were similar and did not show any significant changes in the stresses along the surface. One reason for this may be that the X direction loads were the smallest forces. Although, the X direction force did not affect the stresses to a great extent, they were seen to play a part in reducing the stresses along the surface in the proximal region. This would be due to the X direction reducing the effect of the Z direction load on the neck region of the surface and therefore reducing the high stresses in this region. Therefore although the stresses are small in the X direction it does play a part in reducing the stresses.

The Y direction load was found to have much more of an influence on the stresses. This load was found to affect both the proximal and distal regions. In the proximal region when the Y direction load was not applied the stresses were seen to rise above the threshold value. The influence of the Y direction was investigated for the peak stresses which occurred at the proximal region for the average walking motion. It was seen that the bigger the load in this direction the lower the peak stress became indicating that it also played a part in reducing the effect of the Z direction load from pressing on the neck region. However, this is only until just over half the average walking Y direction load was applied the effect of the Y direction then changed, and the increase in Y was then seen to start to increase the peak stress. At around 1.1
times the average walking Y direction load the threshold was breached and all loads after this were over the threshold.

The maximum load models for each of the motions showed the effects the loads can have on the stresses at the bone-stem interface. All 5 motions showed large areas of stress over the threshold. The up steps model showed the greatest effect of the load as threshold levels were seen in the proximal, middle and distal areas. The sitting and standing motions showed surprisingly large areas of stress over the threshold value. Although these were maximum loads and would not apply on a regular basis the stresses are very high and show that even smaller loads would still give some stresses that would be over the threshold. The stresses over the threshold were seen on all motions in the proximal neck region but the sitting, standing and up steps motions all showed distal stresses over the threshold. The distal stresses over the threshold in these three motions were all found to be on the posterior side of the surface. This is explained by the Y force acting in an anterior to posterior direction and therefore forcing the prosthesis into this area of the bone surface.

The load magnitude was the highest in the maximum up steps motion models and this model showed the highest stresses and had the highest area over the threshold. However the two motions with the lowest magnitude loads were the sitting and standing motions. However these motions both showed very similar results to the up steps motion results, which had the highest magnitude load. These two motions did however have high Y direction loads. The results from the maximum models show how important the Y direction load is by comparing the models with and without this load and the stresses are drastically different. Therefore an important finding from this study is that the magnitude of the overall resultant is not the most important load to identify stability issues but rather the size of the Y direction load.
CHAPTER 8

Discussion and Conclusions
Chapter 8 Discussion and Conclusions

8.1 Discussion
Total Hip Replacement (THR) surgery has proven to be a successful procedure and both cemented and uncemented prostheses have good survival rates. However, as people are living longer they are becoming more active in their old age. An average increase in body mass has also had an impact as people are carrying more weight which places increased forces on hip joints. Both of these factors have led to an increase in the number of hip replacements and with the number of operations rising the number of failures rises too. It is predicted that the demand for hip replacements is to grow exponentially in the next 10 years (Watkins-Castillo, 2006).

From the review of the existing literature, aseptic loosening is by far the most common cause of failure. The literature shows that there are several reasons for this including biological factors that cause the loosening and the increased forces at the hip. However, mechanical loading plays an important role in both processes. In cemented hip replacements aseptic loosening usually occurs due to cracks in the cement and movement at the cement-stem interface. However, aseptic loosening in cementless hip replacements occurs differently and at different stages. Primary stability can be an issue immediately after surgery and the prosthesis needs to be press fit securely so that secondary stability can occur. Secondary stability is a biological process where the bone forms an osseointergrated bond with the stem and secures the implant in place, known as osseointegration. This stability becomes an issue when strains are either too high or too low for the biological bonding process to occur. Loosening of cemented hip replacements has been well documented, and for cementless replacements primary stability and secondary stability have been investigated. However, very few studies have investigated the effects mechanical loading had on long term stability associated with cementless replacements after the osseointegrated bond has been formed.

There are limitations associated with loading patterns in the existing published data as most previous research has used a dynamic ramp load from zero to peak and back to zero as their cyclic load. However this does not take into account the realistic forces created at the hip joint from everyday motions. Many studies have documented how the walk cycle often has 2 peaks during the cycle. This study created a realistic dynamic hip joint force by carrying out motion analysis on 23 subjects. Motion analysis has been carried out previously in many studies (Bergmann et al., 2001, Bergmann et al., 1993, Bergmann et al., 1995a, Bergmann et al., 2004) and the motion analysis from this study compared favorably with this literature. Figure 3-42
shows the % difference between the present study and work from the Bergmann group. The present study fits well within the work from the Bergmann group and shows that musculo skeletal modeling is a good tool for identifying the forces at the hip joint. Dynamic cycles were created for all 5 everyday motions tested in the motion analysis. Figure 8-1 shows the loads for the walking data. Maximum loads are an important factor as these are very often the loads associated with mechanical loading which causes loosening. The maximum normalized to Body Weight (BW) hip joint forces for each of the five motions were also identified. Statistical analysis was carried out to identify any differences between five comparable groups. However, only 20 statistical differences were found. Five of these were applied to numerical models the difference in stresses between groups was very small. Most of the significant differences were seen between groups in the X direction load. The load is generally the smallest in the X direction and this was seen in the numerical models to mainly reduce the effects of the Y and Z load directions on the surface stresses.

![Figure 8-1](image.png)

**Figure 8-1** The calculated average forces for the walking motion in each of the loading directions

To separate the three load directions the experimental rig was designed and built to apply the forces at the hip joint through the use of three pneumatic pistons. Other experimental models used tensile testing machines that applied the load through one point by tilting the model at certain angles. However the angle was always constant throughout the cycle and could not be changed. It was seen from the motion analysis study that the angle of the loads would not be constant throughout the motion nor would the angles be the same for different motions. Bergmann et al. (2001, 2004) investigated the angles and found similarities with all groups in
the frontal plane. However in the transverse plane there was a larger difference with the up steps motion, creating the greatest angle which was confirmed in this study. For the sitting and standing movements the angles in the transverse plane were found to be very low in the previous studies but in this study both were found to have a large angle as the force in the Y direction was seen to be the greatest of the three load directions. The other motions investigated in this study all fell close to the values found in the literature.

The experimental rig was designed to accommodate five pneumatic pistons, three for the hip contact force and two to apply the resolved muscle forces. The three hip contact force pistons were attached to a loading block and all acting at the centroid of the block. The head of the hip prosthesis also acted at this point so no additional moments were added to the experimental setup. The important aspect of this setup was the ability to vary the angle and therefore vary the direction of the force being applied to the head of the prosthesis. The results from the numerical modelling demonstrated that the Y direction was the most important load direction in terms of stability and that this load is a lot more important than the magnitude of the force in terms of exceeding the threshold value. From observations of the experimental setups in the literature, most studies identify higher magnitude loads as higher risks to stability; but this may not be the case. Lower magnitudes with higher Y direction loads were seen in the numerical modelling chapter to have higher areas that exceeded the threshold value for loosening. A material had to be used to represent the osseointegrated bond that occurs between the bone and the stem in long term prosthesis. The Cambridge Engineering Selector material selector software was used to both find and limit the testing to three adhesives. After material properties tests had been carried out on the three adhesives, Araldite Precision, an unfilled epoxy resin, was chosen. This adhesive was the best representation of material properties found in the literature from in vivo testing.

The experimental results investigated both elastic and plastic deformation. For the elastic deformation there was a lot more movement in the YZ models than in the other two models. This indicated that the X direction plays a part in controlling the effect of both the Y and Z directional loads and that without this more movement at the surface interface would occur. However, all the models start to flatten out and if all the models follow the XYZ model then the elastic movement will become constant in all models. The plastic deformation can indicate the possibility of loosening and instability at the interface. At the proximal medial location of LVDT 2 and the distal lateral region at LVDT 3, progressive movement was seen with all three load configurations. Although LVDT 1, located on the proximal posterior surface, only showed
small movements in the XYZ and XZ models, they did show a slow increase in movement. The YZ model at LVDT 1 showed a large change between the average displacements, indicating that both the Y and Z forces are important factors in causing loosening but also that the X direction plays a role in controlling the effects of these two forces. The displacements recorded in this set of results indicate that the mechanical loading could lead to the loosening of the prosthesis. However it is important to identify that none of the three models has become unstable at the end of its testing, regardless of whether it was run for 2 million or 5 million cycles. What these results do show is that the mechanical loading can change the position of the implanted prosthesis over a long period of time. Unfortunately due to time restraints it was not possible to run the models for a longer period and as a result it was not possible to observe if the progression of the displacements lead to the loosening and instability of the prosthesis.

Experimental results from a series of work by Colgan et al. (1995, 1996b, 1996a) were used to form a validation between the numerical modelling techniques used in this study and the results found in Colgan et al. (1996a). There were similarities between the two groups which indicated some agreement, but a lot of the results did not agree and therefore validation could not be proven. It was decided that the experimental rig designed and created in this study was to be used as a further back up to the numerical modelling results. Although the numerical modelling results could not be directly compared to the experimental rig, as the numerical models were not run over millions of cycles, the experimental results demonstrated that when the threshold value was exceeded in the numerical model movement was recorded after several million cycles.

The long term stability of cementless prostheses has not been well documented. in vitro experimental studies of cementless prostheses have all shown primary stability of the prosthesis but not investigated the long term stability of the prosthesis nor evaluated the effects that mechanical loading has on the stability of the prosthesis. Of the numerical modelling studies only Jung et al. (2004) was found to evaluate long term stability and the loosening of cementless prostheses in the long term, with most other studies concentrating on primary or secondary stability. For this reason the numerical models concentrated on long term stability.

The threshold value for an initiation of loosening and stability in the cortical and cancellous bone was taken from previous literature (McNamara and Prendergast, 2007, McNamara et al., 2006, Scannell and Prendergast, 2009a, Scannell and Prendergast, 2005). These studies used
stress values as a criteria in algorithms used for numerical studies of mainly secondary stability. This is because they were primarily focused on bone remodelling and the creating of the bond between the stem and the bone. The results throughout chapter 7 show the Y direction playing an important part in causing the increase in stress. The average walking load results in chapter 4 show that with all three load directions applied no stresses were seen over the threshold value and this would indicate that no loosening would occur due to this force. However, when either the X or Y loads are not included the stresses exceeded the threshold at some point along the surface, although these were in very small locations. These results show that even with normal walking the stresses are close to the threshold value. The results from the maximum forces in chapter 7 show a lot higher areas over the maximum and would indicate that if these loads occurred over a continuous period then loosening could occur.

The numerical results investigated all five motions that were carried out during the motion analysis using both the average and maximum loads from the THR and BIRM subjects tested. The maximum and average results showed similar patterns for each of the motions, with the maximum having higher stresses and more areas along the surface interface that were over the threshold value. For the maximum load models the up steps motion had the greatest area over the threshold value and this is in agreement with other studies (Bergmann et al., 1995a). However, the sitting and standing models were found to have areas in the middle and distal regions that were over the threshold values. These were not as great as the up steps areas but they were higher than the walking and the down steps models which only had stresses over the threshold in the proximal neck region. When the Y direction force was removed for the sitting and standing models the stresses were greatly reduced throughout the models and no stresses along the surface were over the threshold value. When the Y direction was removed the other motions also had a reduction in the stresses in the distal and middle region however, the stresses at the neck region were still over the threshold values.

As already mentioned the load in the Y direction played an important role in increasing the stresses along the surface interface. The Y direction is in the anterior – posterior direction and previous studies have commented on how this may be a cause for loosening in all hip replacements (Colgan et al., 1996a, Fowler et al., 1988, Wroblewski, 1980). In chapter 7 the load in the Y direction was varied from zero to one and a half times the normal load for the average walking motion and the results showed the Y direction plays an important role when large or small. An increase in the Y load when the load is low is seen to reduce the affect of the Z direction force on the stresses at the surface interface, especially around the neck region, but
when the force is increased over the average Y load it is seen to cause stresses that go over the stress threshold for the stability in areas on the surface and this is one of the main reasons for loosening. A means of reducing or controlling the force direction would be beneficial to all THR subjects.

The magnitude of the load in the sitting and standing models was lower than both the walking and down steps motions. However, the stresses were a lot higher in the middle and distal regions of the surface plots as a result of the force in the Y direction being higher in the sitting and the standing motions. Therefore, load magnitude is not always the best indicator to future loosening as the load in the Y direction provides a better indication. These models also show that the sitting and standing motions can play a part in the loosening of hip replacements. Although it has to be taken into account that these were the maximum loads found and as such are the worst case scenarios. Having said this, all these maximum loads did occur during the motion analysis carried out as part of this study.

### 8.2 Conclusions

- Designed, fabricated and commissioned an experimental rig with realistic loads configurations with the ability to vary the loading pattern

- Calculated representative and maximum loading configurations for 5 motions using THR data.

- Found hip joint force values from musculoskeletal modelling gave a good representation of the values found using implant instrumented prosthesis.

- The load in the Y direction was identified as the main force associated with high stresses that may lead to instability and aseptic loosening.

- The load in the Y direction is a higher indicator of possible loosening then the magnitude of the load.
CHAPTER 9

Future Work
Chapter 9  Future Work

9.1  Introduction
This study has covered a small amount of the possible areas for research in the stability of hip prostheses. The advantages of developing a calibrated experimental rig become apparent when assessing the possible applications of the rig. Proposed below are some additional experimental models that could be investigated. There are also further areas that may be developed in the motion analysis and computational modeling areas. These are also presented below.

9.2  Motion Analysis
Motion analysis is an important tool to assess the gait of subjects and with the aid of musculoskeletal modeling software the forces at the hip joint can be analyzed. Surgeons and engineers are both interested in comparisons between various groups for statistical analysis. Further motion analysis focusing on surgical approach and prosthesis type would be a beneficial exercise and would give insight into these areas. After the recall of DePuy ASR acetabular systems in August 2010 the study of this prosthesis would be of great interest to both surgeons and companies alike. Comparisons between this prosthesis and others in terms of gait pattern and forces created at the hip joint could be investigated. Extending the numbers of subjects analysed would make the statistical analysis become more relevant.

9.3  Experimental Modelling
One of the major achievements of this study was to design, create and calibrate the experimental setup able to create a hip contact force that is capable of loading in any possible direction. There are many areas of hip replacements that this setup could be used to investigate and two main areas would be to use the rig to experimentally investigate the long term stability of cemented hip replacements and to investigate primary stability of cementless hip replacements. Cemented hip replacements have been investigated experimentally in literature to date (Britton and Prendergast, 2005). However the experimental rig in this study uses a more realistic loading configuration. Comparison could be made between different experimental setups to investigate the limitations associated with certain setups. Similar to the cemented hip replacements, cementless primary stability has been investigated previously. However primary stability is an integral part of the stability of a cementless hip replacement and it is important to fully
understand the movement of a press fit replacement in a cementless hip replacement straight after surgery.

One of the benefits of this experimental setup is that the hip contact force is split into the 3 load directions and each load is very adjustable. For this reason it is very easy to investigate the effects of a certain loading direction or the effects of a certain load direction reduced or not applied at all. This study concluded that the Y load direction was an important load direction and played a major role in the long term loosening of a prosthesis. Future work could investigate this further but also investigate if this force direction may affect the stability in both primary and secondary stability after THR surgery.

9.4 Computational Modelling
The THR prosthesis used during surgery can vary depending on the patient or even the surgeon. These can range from totally different stem designs to just a change in stem size. Computational models could be used to investigate the different prostheses and investigate the stresses and strains that these would have on the surface interface. These may lead to certain prostheses reaching the stress threshold values in important locations or reducing the stresses to a value that does not exceed the threshold values.
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APPENDICES
17th September 2009

Dr. Michael Walsh
Department of Mechanical and Aeronautical Engineering
University of Limerick

Re: Ref: DE0613 - Extension Request to U. ERC Application No. 06:02 "An Investigation into Aneurysm Healing After Total Hip Replacements"

Dear Michael,

The Faculty of Science and Engineering Research Ethics Committee, at its meeting of 15th September 2009, approved the above extension request. Please resubmit the original application with the amended dates.

Yours sincerely,

[Signature]

Prof. Tim McGloughlin
Chair
Science & Engineering Research Ethics Committee
HIP PROTECTOR PROJECT (INVESTIGATION OF ASEPTIC LOOSENING)

I _______________________________________ have read and understand the patient information sheet provided for this study and understand that participation in this study is entirely voluntary. Any questions I have in relation to this study have been answered to my satisfaction.

Volunteer _______________________________ Date _______________
(Please Print)
Signature _________________________________

Witness _________________________________ Date _______________
(Please Print)
Signature _________________________________

Investigator ______________________________ Date _______________
(Please Print)
Signature _________________________________
PRE-TEST QUESTIONNAIRE

As you are to be a subject in this laboratory/project, would you please complete the following questionnaire. Your cooperation in this is greatly appreciated.

Please tick appropriate box

YES
NO

Has the test procedure been fully explained to you?

Any information contained herein will be treated as confidential

1. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?

2. Do you feel pain in your chest when you do physical activity?

3. In the past month, have you had chest pain when you were not doing physical activity?

4. Do you lose your balance because of dizziness or do you ever lose consciousness?

5. Do you have a bone or joint problem that could be made worse by a change in your physical activity?

6. Is your doctor currently prescribing drugs for your blood pressure or heart condition?

7. Do you know of any other reasons why you should not undergo physical activity? This might include severe asthma, diabetes, a recent sports injury, or serious illness.
8. Have you any blood disorders or infectious diseases that may prevent you from providing blood for experimental procedures?

If you have answered NO honestly to all questions then you can be reasonably sure that you can take part in the physical activity requirement of the test procedure.

I ………………………………… declare that the above information is correct at the time of completing this questionnaire Date ……/……/…….

Please Note: If your health changes so that you can then answer YES to any of the above questions, tell the experimenter/laboratory supervisor. Consult with your doctor regarding the level of physical activity you can conduct.

If you have answered YES to one or more questions:
Talk with your doctor in person discussing with him/her those questions you answered yes. Ask your doctor if you are able to conduct the physical activity requirements.

Doctor’s signature ……………………………………… Date ……/……/……

Signature of Experimenter……………………………… Date ……/……/……
Subject Information Sheet

Section A: To be filled out by the Subject:

Any information contained herein will be treated as confidential.

Q1. Name: ____________________________________

Q2. Date of Birth: __ / __ / ____

Q3. Gender:

Male: ☐ Female: ☐

Q4. Weight: ____________ kg

Q5. Height: ____________ cm

Q6. Do you have a Hip Prosthesis Implanted?

Yes: ☐ No: ☐

If Yes please fill out the next few questions, if No please skip questions 7-10.

Q7. What is the name of the implanted prosthesis?

_________________________________________

Q8. What side of the body is the prosthesis implanted?

A-8
Left:  □  Right:  □

Q9. Have you had any revision surgery on the replaced hip?
Yes:  □  No:  □

Q10. How long have you had the implant?

Less then 6 months:  □  7 months – 1 year:  □  1 year – 2 years:  □

2 years – 3 years:  □  3 years – 5 Years:  □  Greater then 5 Years:  □

Signature and Date

Signature: ___________________________  Date: _____________________

Section B: To be filled out by the principal investigator:

(i) Test procedure: _____________

(ii) Date of Procedure: __ / __ / _____
To whom it may concern,

We are currently conducting a joint IRCSET and BeoCare funded research project to access aseptic loosening (loosening without infection) of a hip prosthesis after total hip replacement surgery. The project aims to access everyday motions, such as walking, and to find the stresses and strains associated with these motions at the prosthesis - femur interface.

This study will involve carrying out several activities such as walking, sitting on a chair and stepping onto a step. All motions to be performed are everyday activities and there is no risk involved. Motion analysis will be carried out using infrared motion analysis cameras and reflective markers. Motions of both people with and without hip replacements will be accessed and a comparison will be made between the motions recorded. These motions will then be used in computational models to further access the effects of these motions on the loosening of the prosthesis over longer periods of time.

Participation in this study is voluntary. You will be asked to sign a consent form to indicate that you understand what is involved and are happy to take part. The information obtained will give us a better insight into the motions of people with and without implanted hip prostheses. A greater
understanding of the motions involved may help us to improve existing products and designs, but also to increase the life span of a hip prosthesis.

Your involvement in this will be of great benefit to our ongoing research. I will be happy to discuss any further questions you may have.

Yours sincerely,

Dr Michael Walsh

Centre for Applied Biomedical Engineering Research

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University of Limerick

Prof. Tim McGloughlin

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University of Limerick
Project Brief;

An evaluation of commercial adhesives for representation of an osseointegrated bond in an in vitro model so as to quantify secondary stability micromotions at the bone-implant interface and to use such information to validate a finite element model.

Many preclinical studies of cementless hips have concentrated on primary stability as a measure of success. Our study is concentrating on secondary stability of an osseointegrated hip.

To measure secondary stability we will be monitoring micromotion at the bone-implant interface in an in vitro model. This model will apply complex dynamic forces to the femoral head and also include the two primary muscle reactant forces acting from the greater trochanter region of the femur.

One of the biggest challenges of this study is to replicate a successfully integrated bond between a composite femoral Sawbone and an implanted hip stem. Some form of adhesive will have to be used. To choose the most suitable material we will test the bond strengths using Ti6Al4V sample pieces with an equivalent surface finish to the hip stems which will be implanted. The bond strengths will be measured using shear strength, tensile strength and fracture toughness tests. The recorded values will be compared to those found in literature for osseointegrated bonds and the most suitable will be chosen.

Because this project is primarily concerned with a replicated bond and not an actual clinical in vivo bond the properties and characteristics of the hips and their coatings will not be influential in the results. Any publishable results will not refer to the specific hip used nor will any images of the unimplanted hip be used in any relevant publications.

Project Manager; Kieran Behan

Project Supervisor; Dr. Michael Walsh

Stryker Representative;
APPENDIX B2
APPENDIX C
Average Down Steps Surface Plots

Average Down Steps XYZ

Average Down Steps YZ

Average Down Steps XZ
Average Down Steps Gruen Zone Graphs
Average Sitting Surface Plots

Average Sitting XYZ

Average Sitting YZ

Average Sitting XZ
Average Sitting Gruen Zone Graphs
Average Standing Gruen Zone Graphs
CONFERENCE PUBLICATIONS


Donoghue M.F., D.S. O’Reilly, M.T. Walsh, The Effect of Computer Use on Carpal Tunnel Syndrome, World Congress of Medical Physics and Biomedical Engineering 2009.

CONFERENCE PROCEEDINGS


