University of Southern Queensland
Faculty of Engineering and Surveying

The Investigation into Modes of Failure of Total Hip Replacements

A dissertation submitted by

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In fulfilment of the requirements of

Bachelor of Engineering (Mechanical)

October 2005

ABSTRACT

The investigation into modes of failure of total hip replacements was the purpose of this project. This involved the development of a finite element model of a simplified un-cemented, simplified cemented and a geometrically real hip in order to study the behaviour of loosening of the prosthesis.

Loosening of the femoral and acetabular components in total hip replacements are major long-term complications of hip replacements. Such loosening usually produces substantial loss of supporting bone making revision much more difficult and less successful than primary surgery.

Implant loosening can be caused from resorption of the bone around the prosthesis from wear debris, insufficient osseointegration of the bone into the prosthesis and brittle failure of the bone cement.

The solid modelling packages ProEngineer 2001 and Wildfire2.0 were used to create models of all components. Solid parametric feature-based models of the un-cemented and cemented prostheses were created from actual orthopaedic implants measured using a coordinate measuring machine. Geometrically simplified models of both prostheses were then created for the initial stages of the finite element analysis.

The commercial FE package ANSYS was used to create all meshed volumes and contact areas and to apply the loading and constraints acting on the system. Loosening occurring for un-cemented and cemented femoral stems were analysed though contact pressure at the interface, the stress in the components and the deformation of the whole system.

Therefore, the purpose of this project was considered valuable when trying to increase the longevity of the THR from implant loosening.

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ACKNOWLEDGEMENTS

This research was carried out under the principal supervision of Chris Snook, a lecturer at the University of Southern Queensland.

I would like to thank Chris for his continued efforts throughout the duration of my project, Toowoomba Metal Technologies for the use of their reverse measurement machine and my brother David French for the use of various prostheses.

CERTIFICATION

I certify that the ideas, designs and experimental work, results and analysis and conclusions set out in this dissertation are entirely my own efforts, except where otherwise indicated and acknowledge.

I further certify that the work is original and has not previously submitted for assessment in any other course or institution, except where specifically stated.

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NOMENCLATURE

The following have been used throughout the text and references:-

Anterior	Toward the front, in front of
CCD	Cervical-Diaphyseal
Distal	Farther from any point of reference
Dorsal	Pertaining to the back
FEA	Finite Element Analysis
In vitro	In an artificial environment
In vivo	Within the living body
Lateral	Denoting a position farther from the median plane or midline of the body or of a structure. Pertaining to a side
Medial	The side of the body or body part that is nearer to the middle or centre of the body
PMMA	Polymethamethylcrylate
Posterior	Situated in back of or in the back part of
Prosthesis	An artificial substitute for a missing body part
Proximal	Nearest, closer to any point of reference, e.g. the joint centre
Sagittal	Relating to, situated in the median plane of the body or any plane parallel thereto
THA	Total Hip Arthroplasty
THR	Total Hip Replacement
Transverse	Acting or being across the body

CHAPTER 1

INTRODUCTION

1.1 Introduction

The aim of this project was to investigate the possible modes of failure of a Total Hip Replacement (THR) by modelling and analysing one of the more prominent failures. This was done using appropriate three-dimensional modelling and Finite Element Analysis (FEA) software packages.

The failure of a THR can cause the patient extreme pain, discomfort and temporary immobilisation. Treatment for this can be costly and traumatic making it an undesirable outcome of a hip replacement. Implants on average last between 10 and 15 years, so unless the implant outlives the patient, THR failure is unavoidable.

Therefore, the purpose of this project was considered valuable and important when trying to increase the life of a THR especially in the younger active patients that will outlive the life of the implant.

1.2 History/Background

A THR is the surgical modification of the hip joint where prosthetic components replace damaged parts of the natural hip. When a THR fails a revision hip replacement is needed. This is the surgical modification of a hip that has previously had a THR. Between 500,000 and 1 million hip replacements are carried out worldwide each year (http://www.engin.umich.edu/class/bme456/artjoint/artjoint.htm).

The first THR was designed and inserted in the 1930's by Philip Wiles in London. Prior to this date hip replacement surgery was of the hemi-arthroplasty type (replacing the femoral head), which had unsatisfactory results.

George Kenneth McKee was a trainee with Wiles and began development of THR designs. He developed various un-cemented prototype hip replacements in the 1940's and 1950's. The results in those early days were initial relief of pain followed by loosening and mechanical failure.

In 1953 polymethylmethacrylate (PMMA) bone cement was introduced for fixation of hip implants, which was then popularised by Professor Sir John Charnley. In 1960 the McKee cemented hip replacement became the first widely used and successful THR. This THR had a Thompson stem, which is a chrome-cobalt metal on metal articulation and both the acetabular and femoral components were fixed with cement.

In 1962 Charnley designed and tested a cemented femoral stem, a 22.25 mm femoral head and a high-density polyethylene cup inserted into the acetabulum. This proved successful in the elderly inactive patients treated and has been the basis for cemented hip replacements developed since.

Peter Ring designed the first THR that did not require the use of the bone cement. This is where the bone osseointegrates (grows) into the prosthesis. By the 1970's three types of total hip replacements were in common use, the McKee, Charnley and Ring types (http://www.midmedtec.co.uk/total_hip_history.htm).

1.3 Medical Conditions

People young and old may need to get a THR for various reasons. It is however more common for the older patients due to the deterioration of the bone and cartilage that can occur over time. The most common medical conditions that inhibit the function of the natural hip are:

- Arthritis – This is where the cartilage lining is thinner than normal or completely absent. This causes the joint to become stiff and limits activities due to pain and fatigue.
- Osteoarthritis – This is where the surface cartilage is damaged or the joint surfaces are not aligned properly, which causes the bone to rub against each other due to the cartilage wearing out.
- Damage to the joint caused by an accident.
- Still's disease – This is joint pain also called Systemic Juvenile Rheumatoid Arthritis.
- Ankylosing Spondylitis – This is a chronic inflammatory condition of the joints.
- Congenital dysplastic or dislocated hips.

- Paget's Disease – This is excessive resorption of the bone.
- Trauma or Avascular Necrosis – This is bone death that is generally associated with the femoral head.

There are many complications that can occur after undergoing hip replacement surgery. Some of these complications lead to failure of the THR and some are just conditions that cause slight discomfort/pain for the patient that can be treated without the need for a revision. These complications are:

- Leg length discrepancies
- Dislocation
- Loosening of the implant
- Infection
- Implant wear
- Heterotopic ossification– This is where bone forms in the soft tissues and muscles surrounding the hip joint. Mild cases do not cause stiffness and pain.
- Femoral bone fractures

There are medical conditions that can preclude a THR from taking place. These include skeletal immaturity, active infection, progressive neurological disease and muscle weakness.

1.4 Components making up the Prosthesis

The hip is made up of two main components, the acetabulum (the socket) and the femur head (the top of the thigh bone). The head of the femur articulates with the cuplike acetabulum of the pelvic bone. These two bones are lined with surface cartilage. Ligaments that form a complete sleeve around the joint, which creates a capsule, hold the head of the femur and the acetabulum together. This capsule has a thin lining of synovial cells that produce a thin layer of lubrication film. As cartilage does not show up on x-rays, a space is seen between the femoral head and the acetabulum. This 'joint space' is approximately ¼ inch wide and reasonably even in outline.

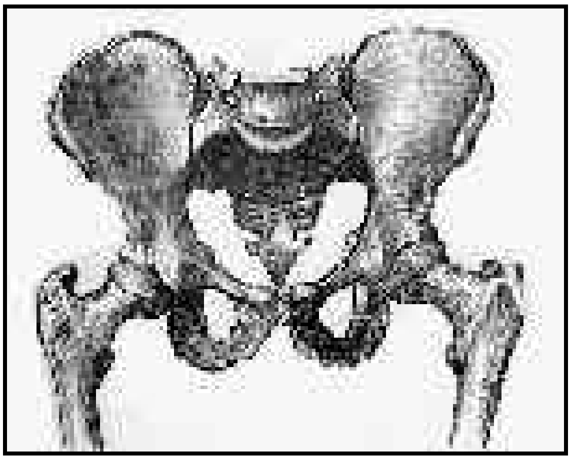

Figure 1.1 - A skeletal view of a human hip

Source: (http://www.engin.umich.edu/class/bme456/artjoint/artjoint.htm)

In a THR there are three main components that make up the prosthesis, the femoral stem, femoral head and the acetabular cup. In figure 1.2 you can see a femoral stem with a femoral head attached to it on the left, an acetabular cup in the middle and a whole THR on the right.

Figure 1.2 - The components that make up a total hip replacement

Source: http://www.lima.it/english/english.html

Originally femoral components were made of stainless steel that was replaced by a Cobalt-Chromium-Molybdenum (Co-Cr-Mo) alloy. Femoral components are now made of stainless steel, Co-Cr alloys and Titanium (Ti) allows. The cup component can be made of ceramic, high-density polyethylene or metal (Katti, KS. 2004).

A number of individual pieces can make up the femoral component of the prosthesis, the femoral stem, the femoral head and, depending on the type of hip, a separate modular neck. Figure 1.3 shows a femoral stem of the left, a ceramic and a metal femoral head in the middle and a sketch of a femoral stem with a modular neck on the right. The femoral head fits onto the top of the femoral neck. Both have tapered so that a press fit is secure enough for application.

Figure 1.3 - The individual parts that can make up the femoral component of a prosthetic hip
Source: http://www.lima.it/english/english.html

The acetabular component is a semi-spherical cup that is bonded to the bone. Sometimes the cup has a separate liner that articulates with the femoral head. These can be seen in figure 1.4.

Figure 1.4 - The individual parts that can make up the acetabular component of a prosthetic hip
Source: http://www.lima.it/english/english.html

1.5 Types of Total Hip Replacements

There are two types of hip replacements, hemi-arthroplasty and a total hip replacement. When a hemi-arthroplasty hip replacement is carried out, just the femoral component is replaced. With a THR both of the femoral component and the acetabular cup are replaced (http://www.hipsknees.infio/thr.asp).

There are two types of prosthesis currently used, cemented and un-cemented. A hybrid hip is where one component of the prothesis is cemented and the other is not. A cemented hip is where the components are bonded to the bone using a polymethylmethacrylate (PMMA) bone cement. Fixation using PMMA allows the patient to bear weight on the sooner than fixation from osseointegration (Katti, KS. 2004).

An un-cemented hip is where the components have a porus coating, which enables the bone to osseointegrate (grow) into the bone, thus no cement is required. The porus coating can either be Hydroxy Apatite (HA) or porus metal. Un-cemented hip replacements are more common with younger patients because a tight bond of scar tissue is formed, which anchors the metal to the bone and has less chance of loosening compared to the cemented prosthesis.

This type surgery is technically more sensitive, requiring a more exact fit of the metal component to the femur. Generally prosthetic implants used in the non-cemented procedures are larger than those used with cement but are still proportional to the size of the individual bone.

Prosthetic hips come in many different sizes and shapes to suit the patient. Some hips are made as right and left or are symmetrical and can be used in either side. The taper of the femoral neck can also vary. Centralisers and plugs are used in the distal and proximal areas of the femoral stem for accuracy of alignment and to increase cement pressurisation.

During surgery the hip joint is exposed and the head and neck of the femur are removed. The shaft of the femur is then reamed to accept the femoral stem and then the acetabulum is reamed to accept the cup. The new hip joint is then replaced to normal position (http://www.bosmc.com.au/patienteducation_hip.htm).

Figure 1.5 shows a cross section of a cemented THR that has used a polyethylene acetabular cup, a metal femoral stem and head. You can see the outer (cortical bone) and inner (cancellous bone) regions of the femur that are different in physical and mechanical properties.

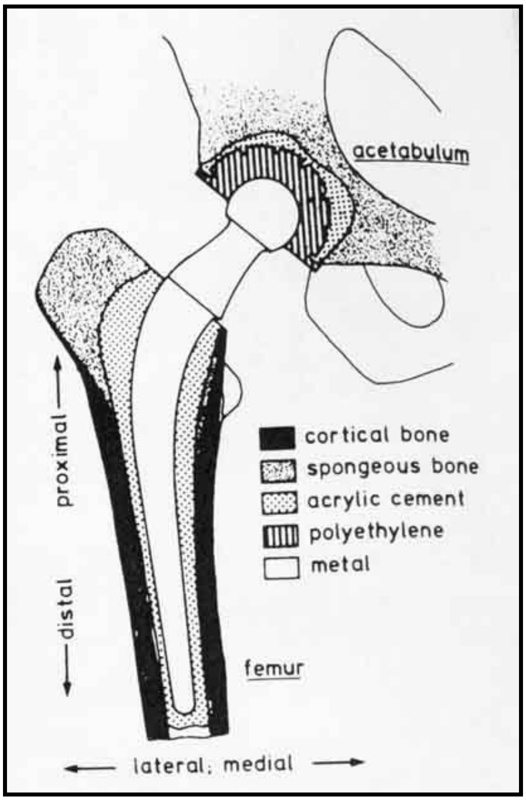

Figure 1.5 - Cross section of a cemented THR
Source: http://www.engin.umich.edu/class/bme456/artjoint/artjoint.htm

1.6 Project Description

The aim of the project was to investigate the modes of failure in total hip replacements. Background information, prosthetic component material properties, loading and movement of prostheses, and THR failure types were researched and documented as a literature review.

From these findings, a decision was made to model and analyse the loosening of the femoral stem using the finite element method. Simplified and geometrically real solid models were created for the cemented and un-cemented hip components for the purpose of failure analyses.

The objectives for completing finite element analyses were to analyse the stress behaviour of the cement (if present), the femoral stem and femur, the deformation of the whole system and the contact pressure between the implant and the bone. Variation in loading, constraints and friction coefficients at the bonding areas were varied and compared in order to ascertain design, modelling and analysis recommendations.

1.7 Project Outline

The project outline begins with the background, history and general information on total hip replacements in chapter 1. Literature reviews on component material properties, prosthesis loading, prosthesis movement and modes of failure can be found in chapters 2 and 3 respectively.

The solid modelling of the geometrically real hip components are documented in chapter 4 with the finite element analyses on simplified un-cemented and cemented hips following in chapters 5 and 6. Chapter 7 outlines the attempt made at analysing the real prosthesis using the finite element method.

Design, modelling and analyses recommendations can be found in chapter 8 with the project summary and conclusions following in chapter 9. The project specification and FEA reports can be found in the appendices.

CHAPTER 2

MATERIAL PROPERTIES, LOADING AND MOVEMENT OF PROSTHESES

2.1 Material Properties for Prostheses, Bone and Bone Cement.

The components that make up a THR can be made from many material types. When selecting the material for a THR component, careful consideration must be given to its application and purpose. For example, if the THR is going to be cemented or un-cemented and what other components will be in immediate contact. When completing this literature review, material property data for particular implant types were difficult to find.

2.1.1 The Titanium Alloy Hip

All components in a prosthetic hip can be made form Titanium (Ti) alloy metal. Titanium alloys have high strength to weight ratio and excellent corrosion resistance, however it does suffer from relatively low fracture toughness and poor wear properties. (http://www.materials.qmul.ac.uk/casestud/implants/).

Titanium alloys do not bond directly to bone, thus the application of bioactive coatings enhances the adhesion of the implant to the surrounding bone when in an un-cemented hip (Katti, KS. 2004). For the Ti alloy 'Ti-6A-4A', which is used in THR components today, the following material properties apply, which were published in the Journal of Materials Processing Technology.

- Young's Modulus = 100 GPa
- Max Tensile Strength = 105.49 MPa

(El-Sheikh, HF, MacDonald, BJ, Hashmi, MSJ. 2001)

2.1.2 The Cobalt-Chromium Alloy Hip

Cobalt Chromium (Co-Cr) alloy's have good wear properties and are somewhat resistant to scratching. The implants can directly bond to bone, however the surface of the implant needing to do this has to be roughened to increase friction and stability. Other surfaces that do not need to bond to the bone should be polished to prevent the stem from rubbing against the inside of the bone canal, which may lead to wear debris (http://www.materials.qmul.ac.uk/casestud/implants/).

A common Co-Cr alloy used in hip replacements is Cobalt-Chromium-Molybdenum (Co-Cr-Mo). This metal consists of 27-30% Cr, 5-7% Mo and the rest is Co. Co-Cr-Mo implants obviously do not contain nickel; this means it can be used in patients who have nickel sensitivity (http://www.materials.qmul.ac.uk/casestud/implants/).

Material property data for Co-Cr alloy was obtained from three difference sources. The Journal of Materials Processing Technology published the following:

- Young's Modulus = 196 GPa
- Maximum Tensile Strength = 155.9 MPa

(El-Sheikh, HF, MacDonald, BJ, Hashmi, MSJ. 2001)

The Journal of Biomechanics published the following:

- Young's Modulus = 210 GPa
- Poisson's ratio of 0.3

(McNamara, BP, Cristofolini, L, Toni, A, Taylor, D. 1996)

The John Stocks Society web page published the following:

- Young's Modulus = 225 GPa
- Ultimate Strength = 750 MPa
- Yield Point = 550 MPa
- Density = 8700 Kg/m3

(http://www.kodeks.com/levite/cocr1htm)

The more reliable source of data would be from either journal listed above. Therefore, those material properties will be used in further analysis if necessary.

2.1.3 The Stainless Steel hip

Stainless steel is an alloy of iron, chromium, nickel and molybdenum. With an extremely high resistance to corrosion it does not degrade in the body. When manufacturing stainless steel hip components, the production costs can potentially be kept to a minimum because it can be shaped easily. However the negative aspects to stainless steel implants are its relatively high stiffness and the nickel content, making it incompatible with people with nickel sensitivity (http://www.materials.qmul.ac.uk/casestud/implants/).

2.1.4 Ultra High Molecular Weight Polyethylene

Ultrahigh-Molecular-Weight Polyethylene (UHMWPE) is a high performance polymer. It has excellent abrasion resistance, a low coefficient of friction, high impact strength and good chemical resistance. The crystalline melting point of the material is approximately 130.6 degrees Celsius. Due to a high coefficient of thermal expansion, the maximum service temperature is approximately 93.3 degrees Celsius (http://www.machinedesign.com/BDE/materials/bdemat2/bdemat2_21.html).

High-density polyethylene with a molecular weight greater than 3,100,000 is described as UHMWPE. In 1981, the standard of the molecular weight equalling 3,100,000 or higher for UHMWPE was approved by ASTM, which is now defined in ASTM D4020. The material costs for UHMWPE rise rapidly above the dividing molecular weight value due to the difficultly in manufacturing process.

Processing difficulties of UHMWPE are due to the fact that conventional moulding and extrusion processes would break the long molecular chains that give the material its excellent and unique properties. Current methods include compression moulding, ram extrusion, and warm forging. Developmental work is being done on injection moulding of UHMWPE resins, but so far the molecular structure cannot be maintained.

The following states the material properties for UHMWPE:

- Tensile strength = 21.374 – 37.921 MPa
- Yield strength = 20.684 – 27.579 MPa
- Compressive strength = 18.616 – 24.821 MPa
- Tensile modulus = 414.685 – 1241.057 MPa

(Hill 1998)

2.1.5 Ceramics

Ceramics generally have good biocompatibility but poor fracture toughness and tend to be brittle. They have excellent wear resistance and have the ability to be polished to a high surface finish.

Traditionally a true ceramic prosthetic hip required the femoral head and the acetabular component to be made out of alumina ceramic (aluminium oxide ceramic). This can be seen in the figure 2.1. When a polyethylene on ceramic prosthetic hip is used, a ceramic (alumina or zirconia) femoral head articulates with a polyethylene acetabular component. Under no circumstances does a ceramic on metal coupling exist.

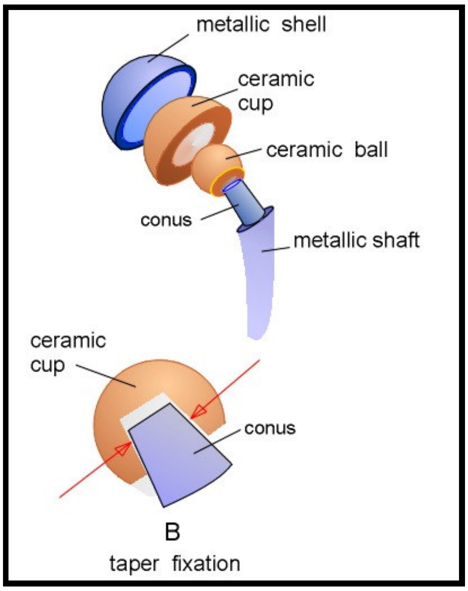

Figure 2.1 - Ceramic on ceramic hip prosthesis

Source: http://www.totaljoints.info/CERAM_HIP2.jpg

For the least amount of wear, ceramic on ceramic couplings gives the best results. The wear rates for ceramic on UHMWPE are considerably lower than that for metal on UHMWPE. Figure 2.2 shows the different wear rates for different articulating surface material combinations. As you can see, ceramic on ceramic articulation gives the lowest volume of wear debris.

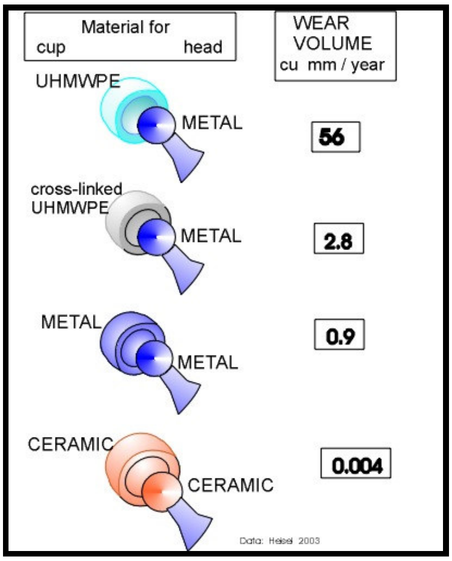

Figure 2.2 - Wear rates for different articulating surfaces
Source: http://www.totaljoints.info/BearingCombin.jpg

Ceramics are characterised by fine-grained microstructures and also have the ability to be polished to produce a fine, hard surface finish. With the appropriate porous form, mechanical fixation to the bone is promoted allowing the bone to grow into the implant.

Two sources were found that published the material properties for ceramics. The Colloid and Surfaces B: Biointerfaces journal stated the following:

- Zirconia ceramics
    - Ultimate Compressive Strength = 2000 MPa
    - Ultimate Tensile Strength = 820 MPa
    - Young's Modulus = 220 GPa

- Alumina ceramics
    - Ultimate Compressive Strength = 4000 MPa
    - Ultimate Tensile Strength = 300 MPa
    - Young's Modulus = 380 GPa

(Katti, KS. 2004)

The Journal of Materials Processing Technology stated the following for alumina ceramics:

- Young's Modulus = 400 GPa,
- Maximum Tensile Stress = 222.32 MPa

(El-Sheikh, HF, MacDonald, BJ, Hashmi, MSJ. 2001)

Comparing the data for alumina ceramics, you can see that the results are reasonably similar. Both journals are reliable sources of information.

2.1.6 Bioactive Materials

Hydroxy Apatite (HA) is a bioactive coating used for the osseointegration of the bone to the implant, giving it excellent fixation. It's primarily consists of calcium phosphate, a mineral that forms one of the prime constituents of bone. Although it is a relatively weak and brittle material it does have good bioactivity (http://www.materials.qm-ul.ac.uk/casestud/implants/). It is applied by spaying onto an existing porus Ti surface and if less than 400 microns the body is able to completely resorb the calcium phosphate and then bond is directly onto the implant.

The properties found for HA are as follows:

- Ultimate Compressive Strength = 600 MPa
- Ultimate Tensile Strength = 50 MPa
- Young's Modulus=117 GPa

(http://www.materials.qmul.ac.uk/casestud/implants/)

2.1.7 Polymethylmethacrylate Bone Cement

Polymethylmethacrylate (PMMA) bone cement is a self-curing, acrylic polymer that is made up of a catalyst (powder) and an accelerator (liquid). It generally takes between 5 and 15 minutes to cure. Because it acts more as a grouting agent, it supports the implant rather than attaching it to the bone. It does not bond to polished surfaces but does bond weakly to the slightly roughened surfaces, such as non-polished implants, polyethylene and pre-coated implants.

Heat stable antibiotics can be added in the catalyst powder. When adding 0.5-2 grams of antibiotic powder per 40 grams of packet cement, the static strength of PMMA does not change, however the fatigue strength can be decreased. This is not considered significant by most surgeons when performing the operating.

PMMA is applied to the bone as partially polymerised dough and is then pressurised. This is where complete polymerisation occurs. The average thickness of the cement layer is 2mm (Pipino 1999, p109). Strength properties are excellent in compression but weak under tension. Adding chopped carbon fibres can increase the cement toughness and material strength. (El-Sheikh, HF, MacDonald, BJ, Hashmi, MSJ. 2001)

The failure of PMMA bone cement is often initiated by propagation of cracks, usually at pre-existing flaws in the material. The presence of the carbon fibres inhibits cracking by increasing the amount of energy needed to cause crack propagation. (El-Sheikh, HF, MacDonald, BJ, Hashmi, MSJ. 2001)

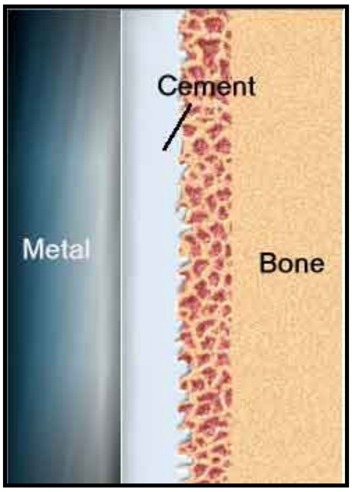

Figure 2.3 - The layers between the metal prosthesis and the bone in a cemented implant
Source: (http://www.engin.umich.edu/class/bme456/artjoint/artjoint.htm)

The advantages of PMMA bone cement are that it is tough, resistant to impact and is easily fabricated. A major disadvantage with PMMA bone cement is that a significant amount of heat is released to the surrounding bone during the curing process. This can causes cell death if the temperature becomes too high, which results in the bone having poor resistance to fracture. Other disadvantages include the shrinkage of the cement and the release of toxic monomers into the blood stream (http://www.materials.qmul.ac.uk/casestud/implants/). It also has a low maximum working temperature of 50 degrees Celsius (Hill 1998).

Only one source was found that published the material properties for PMMA bone cement used for the fixation of hip replacements. The results were as follows:

- Young's Modulus = 3.5 GPa
- Maximum Strength = 70 MPa
- Poisson's Ratio = 0.3

(http://www.totaljoints.info)

2.1.8 Bone

Bone properties can depend on function, age, disease and use. It is an anisotropic and heterogeneous material, which means that the properties depend on direction and location (McNamara, BP, Cristofolini, L, Toni, A, Taylor, D. 1996). Natural bone is a bio-nanocomposite system where a colagen fibre matrix is stiffened by hydroxyapatite crystals that account for 69% of the weight of bone. (Katti, KS. 2004)

Figure 2.4 – Coronal cut through the proximal femur showing the distribution of cancellous bone and cortical bone

Bone is composed of two distinct forms of hard connective tissue. In figure 2.4 you can see the cortical (or compact) bone on the outer region of the section and the cancellous (or trabecular) bone in the inner region of the section (Nigg, MacIntosh, Mester 2000).

Finding accurate mechanical and physical properties for bone was extremely difficult due to the lack of information available. Two sources however were found that published some of the properties. The Journal of Materials Processing Technology published data that assumed the bone material was isotropic (properties that do not vary with direction or location). This proved to be useful when performing a simplified FEA in chapter 5 and 6.

- Cortical bone
    - Young's modulus is 16.2 GPa
    - Poisson's ratio is 0.36

- Cancellous bone
    - Young's Modulus = 32 MPa

(El-Sheikh, HF, MacDonald, BJ, Hashmi, MSJ. 2001)

The second set of data was for cortical bone of the femur and gave properties for the longitudinal (cylinder axis of the bone) and transverse directions.

- Longitudinal
    - Tensile Strength = 133 MPa
    - Compressive Strength = 193 MPa

- Transverse
    - Tensile Strength = 51 MPa
    - Compressive Strength = 133 MPa
    - Shear Strength = 68 MPa

(Nigg, MacIntosh, Mester 2000, p291)

From the second set of data you can see that cortical bone of the femur is stronger in the longitudinal direction compared to the transverse direction. This is true for both compression and tension.

2.1.9 Friction Between Components

The more lubrication and the more separated the components, the less the coefficient of friction. There are five types of lubrication, which are hydrodynamic, elastohydrodynamic, mixed, boundary and weeping. The following data shows the friction coefficients between the articulating surfaces of the femoral head and the acetabular cup. You can see that both metal-to-metal and metal to UHMWPE friction coefficients are not as low as the real hip situation (cartilage to cartilage).

- Cartilage to cartilage coefficient of friction = 0.01-0.02
- Metal to Metal coefficient of friction = 0.4 (dry), 0.15 – 0.35 (wet)
- Metal to UHMWPE coefficient of friction = 0.05 – 0.15

(Kuiper JH, Huiskes, R. 1996)

2.2 Loading of the Prosthesis

The loading and movement on a THR depends on what the individual is doing at the time. The primary factors influencing both the magnitude and the direction of the compressive forces acting on the femoral head are the position of the centre of gravity, the abductor muscles and the magnitude of body weight (http://www.aboutjoints.com/physicianinfo/topics/anatomyhip/biomechanicship.htm). Up to 5-6 times the body weight of a person can be loaded on a THR simply during walking (El-Sheikh, HF, MacDonald, BJ, Hashmi, MSJ. 2001), and as high as 10 times the body weight when jumping (Katti, KS. 2004).

For the purpose of this project, the loading on the total hip replacement will be analysed as if the person is standing straight with the body weight evenly distributed though both legs.

When the weight of the body is being supported on both legs, the centre of gravity is central and is located above the pelvis. The force is then distributed equally on both hips (Figure 2.5). Under these loading conditions, the weight of the body minus the weight of both legs is distributed equally on the femoral heads. The resultant vectors acting on both femoral heads are vertical.

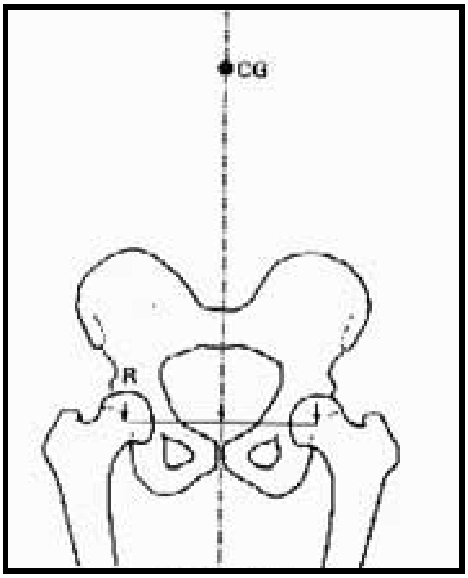

Figure 2.5 - Schematic showing the direction and magnitude of the load on the femoral head in symmetrical two-leg stance

Source: (http://www.aboutjoints.com/physicianinfo/topics/anatomyhip/biomechanicship.htm)

When viewing the pelvis in the sagittal plane (side on) with the centre of gravity directly over the centres of the femoral heads, no muscular forces are required to maintain the equilibrium position. However minimal muscular force will be necessary to maintain balance. If the upper body is leaned slightly back so that the centre of gravity comes to lie forward to the centres of the femoral heads, the compressive forces acting on each femoral head represent approximately one-third of body weight.

When standing on one leg, the centre of gravity moves away from the supporting leg since the non-supporting leg is now calculated as part of the body mass acting upon the hip bearing all of the weight. Due to the leg that is not grounded, a turning motion around the centre of the femoral head is created and must be offset by the combined abductor muscle group forces acting into the lateral femur. The resultant force is no longer vertical; this can be seen in figure 2.6.

In the erect position, the abductor muscle group includes the upper fibres of the gluteus maximus, the tensor fascia lata, the gluteus medius and minimus, and the piriformis and obturator internus.

Figure 2.6 – Resultant force on the hip in single leg stance

Source: (http://www.aboutjoints.com/physicianinfo/topics/anatomyhip/biomechanicship.htm)

The combined loading of body weight and the response from the abductor muscle group when standing on one leg, results in the loading of the femoral head to be approximately 4 times the body weight. This means that in normal walking the hip is subjected to wide swings of compressive loading from one-third to four times the body weight.

2.3 Movement of the Prosthesis

When considering the swinging movement of the leg when walking, the unloaded femoral head in hip joint slightly separates from the centre of the acetabulum and moves to the upper outer side (http://www.totaljoints.info/THPconstruction.htm). When the leg takes the body's weight again, the femoral head returns to a snug fit in the acetabulum. This piston like movement can be seen in figure 2.7.

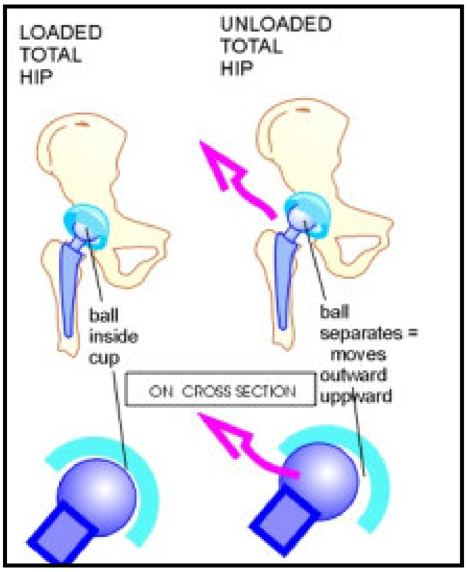

Figure 2.7 – Location of the femoral head when the leg is unloaded

Source: http://www.totaljoints.info/THPconstruction.htm

Stair climbing or raising from a chair is another type of movement that has an effect on the operation of the hip joint. On average patients with a THR perform 45 stair climbs and 70 chair-rises per day (http://www.totaljoints.info/THPconstruction.htm). This can pose a problem if the femoral neck comes into contact with the rim of the acetabular cup (figure 2.8).

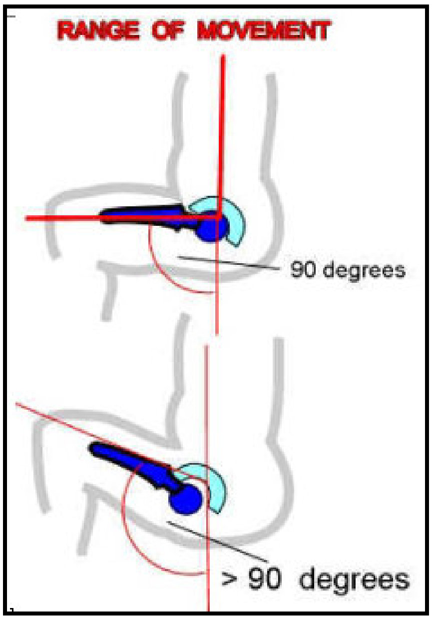

Figure 2.8 – The range of movement in a THR

Source: http://www.totaljoints.info/THPconstruction.htm

When the neck of the femoral stem is forced into this rim, the femoral head can be mechanically levered out of the cup. This is a major cause for dislocation in Hip replacements. Using a ceramic on ceramic coupling when this lever action occurs increases the risk of fracture to the ceramic liner.

From figure 2.9 it can be seen that the smaller the diameter of the femoral head the less stable the hip joint is. New designs are now offering femoral heads with large diameters (>30mm). These implants allow flexion above 90 degrees without increasing the risk of dislocation. The advantage of these implants obviously allow greater stability of the hip, however, the larger the femoral head, the greater the wear, especially if the articulating surfaces include the polyethylene acetabular cup and a metallic femoral head.

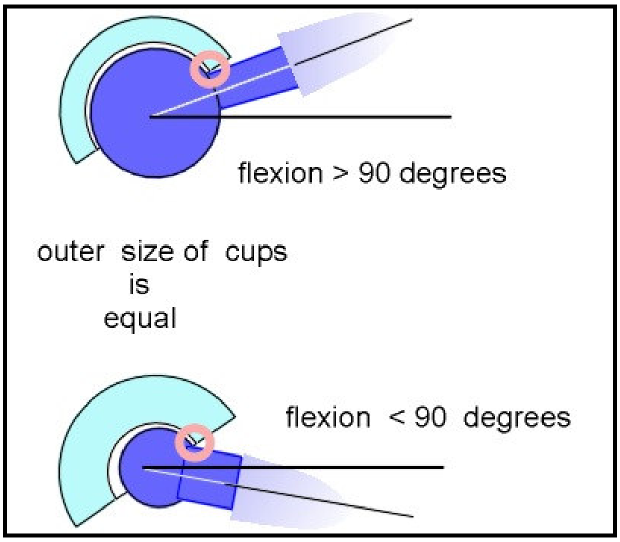

Figure 2.9 – The range of movement in a THR

Source: http://www.totaljoints.info/LARGE_ball.jpg

CHAPTER 3

REVIEW OF POSSIBLE FAILURES IN A TOTAL HIP REPLACEMENT

3.1 Introduction

For a THR to be classified as 'failed', it no longer able to serve the purpose that is was originally implemented for. The main causes of THR failure are listed below:

- Dislocation
- Loosening
- Fracture of the THR components
- Disease and infection
- Unexplained increased pain levels

This chapter will investigate the first four causes listed above. From this, one particular type of failure will be chosen to model and analyse.

3.2 Dislocation

A dislocation of a THR is where the femoral head is no longer securely positioned correctly in the acetabular cup. This can cause extreme pain and dysfunction to the patient experiencing the dislocation. Dislocation of a THR can occur due to poor prosthetic design, surgical procedure, and instrument design and from the patient not being compliant with the post-operative restrictions (http://www.bosmc.com.au/patienteducation_hip.htm).

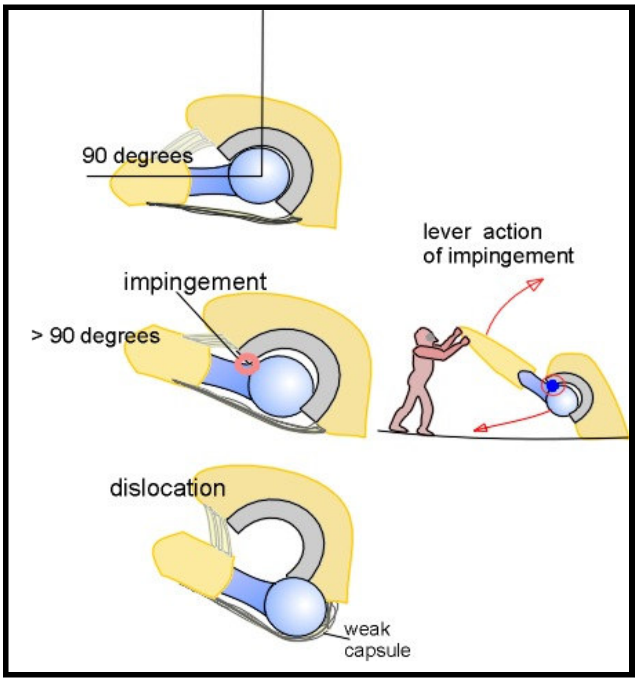

Figure 3.1 - THR dislocation

Source: http://www.totaljoints.info/Dislocation.jpg

When researching the statistics for dislocation of hip replacements, different sources gave different statistical data. The web page ‘About.com’ stated that hip replacement dislocations occur in 4% of first-time surgeries, and 15% of revision hip replacements (http://orthopedics.about.com/od/hipkneereplacement/a/dislocation.htm). The USA orthopaedic manufacturing, distributing and designing company Endotec published that primary hip replacements dislocate at a rate of 1 to 3%, while revision hip replacements dislocate at higher rates of up to 30% (http://www.endotec.com/r&b_for_hip_replacements.htm). The Brisbane Orthopaedic and Sports Medicine Centre published that dislocation has a low incidence of approximately 3% (http://www.bosmc.com.au/patienteducation_hip.htm).

From the information found, it would be reasonably accurate to say that for first time THR patients the risk of a dislocation occurring would be approximately 3%.

To reduce the risk of experiencing a hip dislocation, the following should be avoided:

- Crossing your legs
- Bending out legs up beyond 90 degrees
- Siting on sofas or in low chairs
- Sleeping on your side

These activities place the hip joint in a position where the femoral head may be forced out of the acetabulum. Other, less obvious factors that can contribute to THR dislocations include excessive alcohol intake, neuromuscular problems (e.g. Parkinson's disease) and developmental hip dysplasia. Sometimes patients have no identifiable cause for a dislocation of their hip replacement.

Treatment for a THR dislocation includes the reposition of the hip joint. This is called a reduction of the hip replacement and is performed under anaesthesia, during which the surgeon will manipulate the leg to reposition the femoral head in the acetabulum. Once the surgeon believes the hip is in the correct position, X-rays will be taken for conformation and to work out the possible causes (http://orthopedics.about.com/od/hipkneereplacement/a/dislocation.htm).

Another cause of dislocation is the mal-position of the acetabular or femoral components. If this is the case they may need to be removed and repositioned. Another choice of operative treatment is the use of constrained acetabular cups. This can be seen in figure 3.2.

Figure 3.2 - Constrained Acetabular Cup

Source: http://www.totaljoints.info/CONSTRAINT_cup.jpg

3.3 Loosening of the Prosthesis

Loosening of the prosthesis is where the fixation of the implant to the bone is no longer as strong as it needs to be. When this happens, a soft fibrous tissue develops at the interface that allows more relative motion between the prosthesis and the bone under loading. This allows the implant to shift and causes pain to the patient. When the pain becomes intolerable, the patient will have to undergo revision surgery. Approximately 3-5% of hip replacements result in this particular type of failure (http://www.engin.umich.edu/class/bme456/artjoint/artjoint.htm).

There are a number of factors that may encourage loosening of the prosthesis. These include:

- Avascular necrosis – This is where the blood supply to the bone stop, which causes death of the bone around the implant.
- Mechanical damage done during surgery.
- Wear debris - This can cause bone resorption around the implant.

- Mechanical loosening at the interface.

(http://www.engin.umich.edu/class/bme456/artjoint/artjoint.htm)

The main types of loosening that will be investigated in this chapter will be from wear debris and mechanical loosening at the interface. Loosening of the femoral and acetabular components of the hip are major long-term complications of hip replacements. Such loosening usually produces substantial loss of supporting bone making revision much more difficult and less successful than primary surgery.

3.3.1 Implant Loosening Due to Wear Debris

Wear debris can cause bone resorption, which is otherwise known as osteolysis. This is where the bone has been removed or lost. This can cause loosening between the bone-implant interfaces. Wear debris includes particles from bone cement, polyethylene, ceramics and metal. There are five types of wear that can occur. Adhesion, abrasion, transfer, fatigue and third body wear.

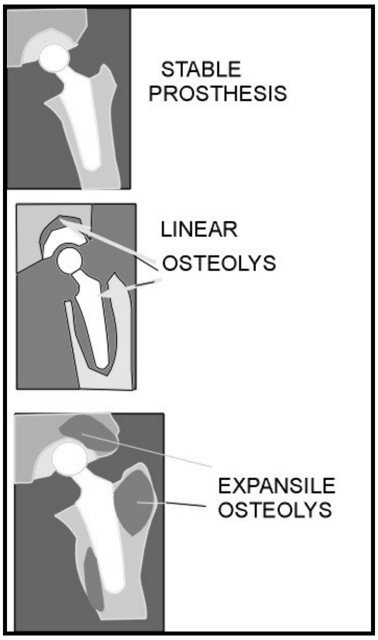

Figure 3.3 – Bone resorption (osteolysis) from wear debris

Source: http://www.totaljoints.info/LOOSENINGTOTALJOINTS.htm#3

Reducing the amount of wear debris in a THR *in vivo* comes down to the design and mechanical properties of the implant components (Katti, KS. 2004). Efforts to reduce wear in hip replacements have been a major area of research and development activity for at least the last decade. These two major areas of research have been in:

1. To develop an improved UHMWPE
2. To develop alternate combinations of articulating surfaces that have a reduced amount of wear

Currently no improved the UHMWPE has yet been demonstrated clinically. Attempts that have been made at polyethylene improvement have resulted in clinical failure, where the new material had greater wear clinically than conventional UHMWPE. There are also claims of improved cross-linked UHMWPE, however this has not yet been proven clinically.

Ceramic-on-ceramic articulating surfaces can essentially reduce the wear problem. Design difficulties include the avoidance of excessive bone removal for implantation and the fracture associated with the brittleness of ceramics (http://www.endotec.com/r&b_for_hip_replacements.htm).

Metal-on-metal also has its disadvantages, which is the femoral neck and acetabular cup impingement that results in increased metal wear debris. Long term metal wear debris may have more negative effects than polyethylene wear debris (http://www.endotec.com/r&b_for_hip_replacements.htm). There is currently debate on the significance of increased metal ions in blood serum, which has been recorded on the use of modern metal-on metal couplings.

Ceramic-on-polyethylene is also an alternative explored and has been shown to substantially reduce polyethylene wear. Usually in this situation a ceramic femoral head is assembled onto a metallic femoral stem that articulates against a polyethylene acetabular cup. The ceramic implementation problems are greatly reduced, compared to ceramic-on ceramic (http://www.endotec.com/r&b_for_hip_replacements.htm).

3.3.2 Mechanical Loosening of the Implant

In an un-cemented prosthesis, the interface between the bone and the implant can loosen due to insufficient osseointegration occurring. The bone itself fails to attach itself adequately to the surface of the implant (http://www.bosmc.com.au/patienteducation_hip.htm).

Osseointegration relies on a stable connection between prosthesis and bone. If relative motion is not eliminated, ingrowth will not occur and the bond between the bone and implant will not be achieved. Controlling this relative motion depends on the design of the prosthesis, the surgical instruments and the surgical procedure used to implant the prosthesis and the skill of the surgeon doing the implantation.

In a cemented THR, failure can occur in the bone cement and on the interface between the cement, implant and the bone. Cement acts simply as a grouting agent to fill the gaps between the implant and the bone, helping the bone to support the implant (http://www.endotec.com/r&b_for_hip_replacements.htm). Flaws in the cement generally initiate failure of the bone cement when the shear, tensile and compressive stress is too high (El-Sheikh, HF, MacDonald, BJ, Hashmi, MSJ. 2001).

3.4 Fracture of Components

Failure of components is a reasonably uncommon cause of failure these days. Improvements in design, manufacture and metallurgy have reduced this failure mode. In the past, femoral stem fracture was the most common form of failure of the prosthesis. Femoral neck fractures and acetabular fractures have also been observed (http://www.endotec.com/r&b_for_hip_replacements.htm). The reason why some of these THR components fail can be due to stress shielding.

Stress shielding (figure 3.4) is where the body weight is transferred incorrectly through the hip area, resulting in the femoral stem being overloaded and the skeletal regions surrounding the prosthesis underloaded (http://www.totaljoints.info/). The femoral stem is much stiffer than the skeleton and will take the greater part of the body weight load.

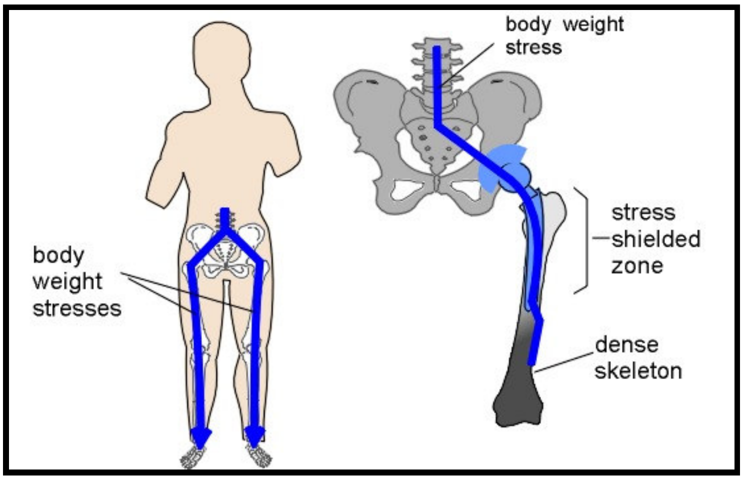

Figure 3.4 – Stress shielding

Source: http://www.totaljoints.info/

In the area where the load has been diminished the skeleton retains only so much bone tissue that is necessary to sustain the diminished load, making it weaker. Whenever there is a composite system (the bone plus the femoral stem can be classified as this), the stiffer material of the two will sustain the greater part of the load (http://www.totaljoints.info/).

3.5 Disease and Infection

Disease and infection to a total hip replacement is said to account for 1 to 2% of THR failures. The Brisbane Orthopaedic Sports and Medicine Centre published 1% (http://www.bosmc.com.au/patienteducation_hip.htm) and the USA manufacturer, distributor and designer of orthopaedic implants, Endotec, published 2% (http://www.endotec.com/r&b_for_hip_replacements.htm).

There are two basic types of infection:

1. Superficial infection (infection of the skin and tissue near it), and
2. Deep infection (infection of the bone supporting the prosthesis).

Superficial infection is normally caused by contamination of the wound from the outside. This can be prevented by particular care taken by the patient after surgery. Inadequate blood supply resulting from the surgical incision contributes to the problem. Usually this is not as serious as a deep infection as antibacterial drugs and protection of the wound can control it fairly easily. However, if it is not controlled, it may become a deep infection where control becomes much more difficult (http://www.bosmc.com.au/patienteducation_hip.htm).

Deep infection is very dangerous and is difficult to manage. It can progress to the loss of the leg or even death. The presence of the THR increases the danger and difficulty to the point where sometimes re-operation is required to remove the THR before the infection can be controlled. This procedure and problem can result in substantial loss of bone. If the infection can be controlled and bone loss is not excessive, an implant can often be re-implanted. The prediction for success of the revision is dependant on the damage done as a result of the infection (http://www.endotec.com/r&b_for_hip_replacements.htm).

3.6 Discussion of Findings

From the four types of causes of THR failure investigated in this chapter, the one that will be incorporated in the modelling and analyses will be the loosening of the femoral stem. This is because it is one of the more common forms of failure affecting the life of the implant. Two aspects of loosening will be modelled, loosening at the bone-implant interface of an un-cemented prosthesis and the loosening of the cemented prosthesis due to cement failure and interface loosening.

Loosening occurring for an un-cemented femoral stem at the bone-implant interface will be modelled to analyse the contact pressure, the stress in the bone and implant and the deformation of the whole system. The failure of the cement will be analysed though the stress intensity in the cement layer around the femoral stem, the bone and the implant, the deformation of the whole system and the contact pressure at each interface.

CHAPTER 4

THREE DIMENSIONAL SOLID MODELLING OF THE PROSTHETIC COMPONENTS

4.1 Choice of Program

When choosing the solid modelling computer package, the decision was based on availability and ability to model more complex shapes. ProEngineer and Solid Works were both available, however, due to the fact that ProEngineer has greater capabilities when creating and modifying more complex shapes, it was chosen as the primary solid modelling program.

Consideration also had to be given to how solid models can be imported into FEA programs. Most FEA programs accept IGES files, and this is achievable when using ProEngineer.

Initially the 2001 version of ProEngineer was used for the creation of the implant components. However, midway though the project the latest version of ProEngineer (Wildfire2.0) became available. There was a considerable difference in the usability of the programs. ProEngineer Wildfire2.0 was far easier to use, because of this, it was then used for the remainder of the project.

4.2 Prosthetic Components

All of the solid models, excluding the femur, were created using real prostheses. All co-ordinates for all of these components were then measured using a Co-ordinate Measurement Machine (CMM) at Toowoomba Metal Technologies (TMT) QLD. A similar CMM to the one used at TMT can be seen in figure 4.1.

Figure 4.1 – Coordinate Measurement Machine

Source: http://www.johnhart.com.au/products/cmms/one_cmm/index.html

The co-ordinates and dimensions for outer surfaces of the prostheses were obtained with the assistance of the machine technician. The implants were then modelled in Pro-Engineer using the measurements obtained.

General difficulties encountered when measuring the components were finding an appropriate origin for all of the co-ordinates and realising which dimensions on the component were critical and should have a low tolerance. This was overcome with the aid and expertise of the machine technician and understanding how the components were to be used.

4.2.1 Cemented Femoral Stem

The cemented femoral stem that was chosen to model was made out of Co-Cr-Mo and had a Cervical-Diaphyseal (CCD) angle of 135 degrees. This component proved to be a difficult shape to model due to the complexity of the shape (complex blended curves). Unlike the un-cemented femoral component, the femoral neck was not of the modular style and could not be detached from the stem. This gave the impression at the beginning that it would be the easier femoral component to model, however this was not the case.

The initial model created was done as one part. This can be seen in the figure 4.2. The problem with this model is that the smooth surfaces around the femoral neck could not be achieved to the correct dimensions. To overcome this, the cemented femoral stem was then created as two parts, the neck (figure 4.3) and then the stem (figure 4.4). It was then put together as one component as an assembly (figure 4.5).

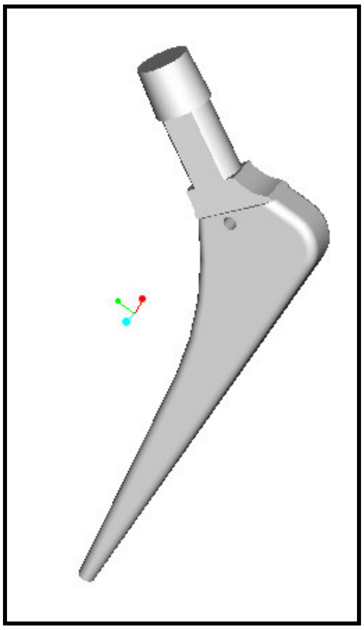

Figure 4.2 - The uncompleted cemented femoral stem created as one part

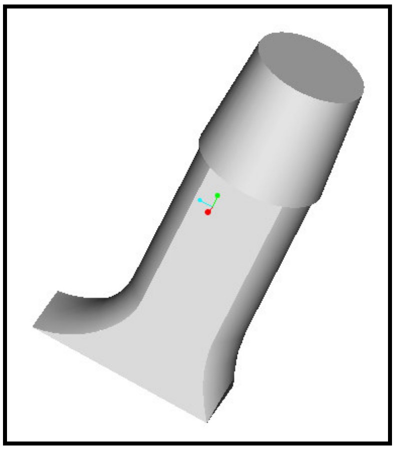

Figure 4.3 - The femoral neck for the cemented prosthesis

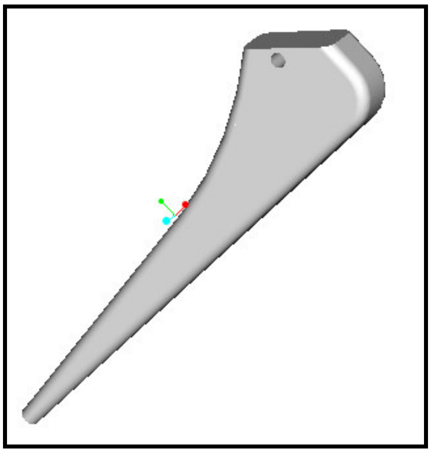

Figure 4.4 - The femoral stem for the cemented prosthesis

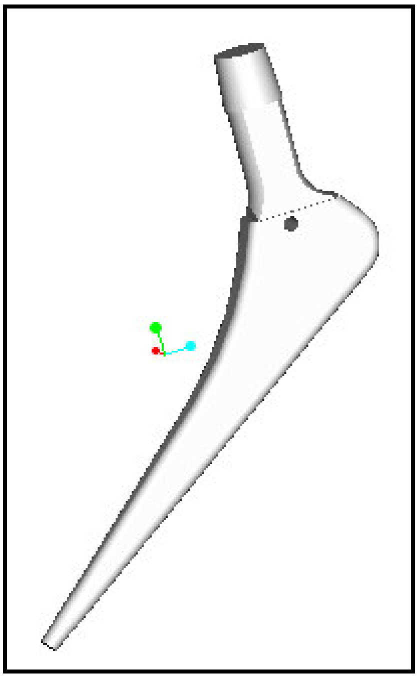

Figure 4.5 - The complete assembly of the cemented femoral stem

Due to the FEA being carried out on the femoral component, the model had to be as close to the real shape as possible. Every shape/curve on the part was carefully measured and then created as close to the real shape as possible. When comparing the created ProEngineer model (figure 4.5) to a photo of the real prosthesis (figure 4.6), you can see that there is little difference between the two.

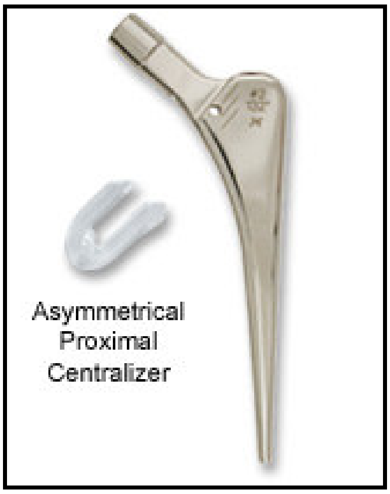

Figure 4.6 - A photo of the actual cemented femoral stem that was replicated as a solid model
Source: http://www.lima.it/english/english.html

4.2.2 Un-cemented Femoral Stem

The un-cemented femoral stem that was chosen was made out of Ti alloy and had a HA coating in the proximal third. The un-cemented femoral component still had a high degree of difficulty when it can to modelling. Every shape/curve on the part was also carefully measured due to it being the main component in the FEA.

This part ended up being an assembly because it was made up of two components, the femoral stem and the modular neck. For simplicity a straight modular neck with a CCD angle of 135 degrees was chosen. The ProEngineer models can be seen in figures 4.7 and 4.8.

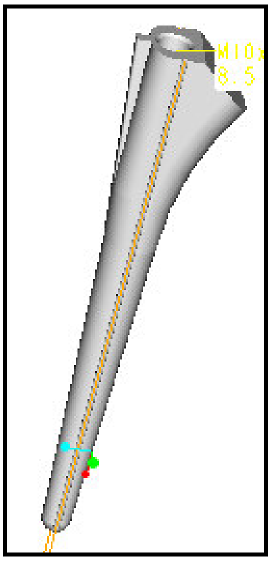

Figure 4.7 – The ProEngineer model created of the un-cemented femoral stem without the modular neck

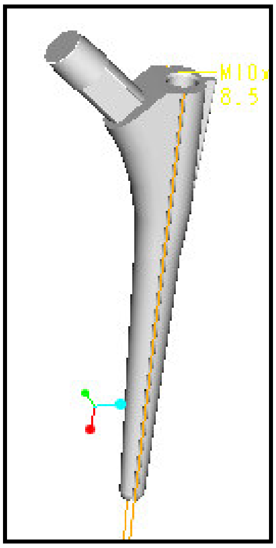

Figure 4.8 – The ProEngineer model created of the un-cemented femoral stem with the modular neck

4.2.3 Femoral Head

The femoral head was one of the easier parts to model. The 28mm diameter head was chosen and is compatible for both the un-cemented and cemented components that were modelled. The head joins with the femoral neck via a taper Morse connection of 12/14. The modelled femoral head can be seen in figure 4.9.

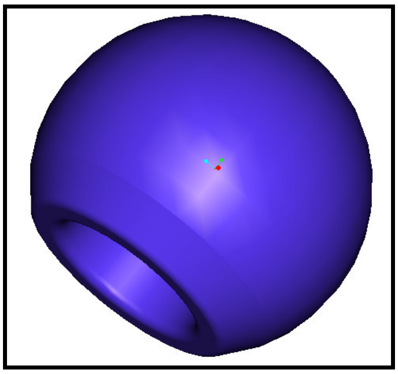

Figure 4.9 – The femoral head that was created in ProEngineer

4.2.4 Acetabular Cup

The acetabular cup chosen to model was Ti alloy with a HA coating on the area needing to osseointegrate with the bone (figure 4.10). Along with this component came a polyethylene liner (figure 4.11), which fitted into the cup. These components were not included in the FEA, thus the dimensions used for modelling were not critical.

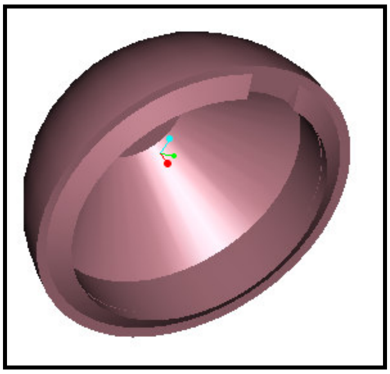

Figure 4.10 – The acetabular cup that was modelled in ProEngineer

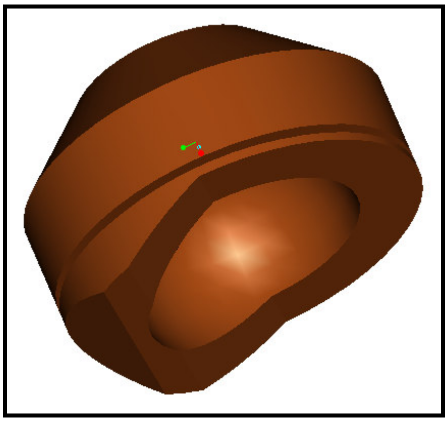

Figure 4.11 – The acetabular liner that was modelled in ProEngineer

4.2.5 The Femur

A model of the femur (figure 4.12) was obtained from the Internet as a surface in Solid Works. The surfaces then had to be changed to a solid volume. This step proved to be extremely difficult due to the lack of solid modelling experience in both Solid Works and ProEngineer. Eventually this problem was overcome in Solid Works (version 2005). A cavity of the bone surface was created in a solid volume in an assembly file. The solid volume was then split and new files were created with the cortical and the cancellous bone regions as solid parts. These were then saved as an ACIS file and then imported into ProEngineer.

The cancellous bone model (which is the bone type that is in direct contact with the femoral stem *in vivo*) was then cut down to the required dimensions as if it were cut in surgery (figures 4.13). The femoral head was cut at a 45-degree angle. This is because the CCD angle of the femoral stem was 135 degrees. The volume where the femoral stem would be situated was removed.

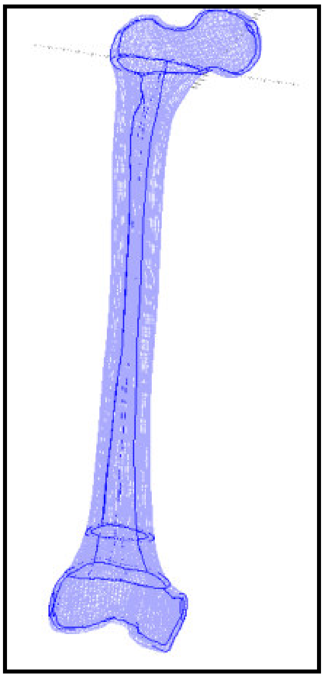

Figure 5.12 – The surfaces of the cancellous and cortical bone of the femur viewed in AusCad2000

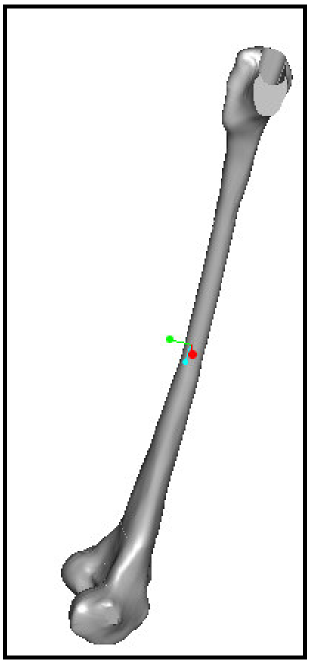

Figure 4.13 – The model of the cancellous bone cut down in preparation for the implementation of the femoral stem

4.3 Models of the Whole THR System

Assemblies in ProEngineer were created that modelled both the cemented (figure 4.14) and un-cemented (figure 4.15) hip replacements. No problems were encountered when creating the assemblies.

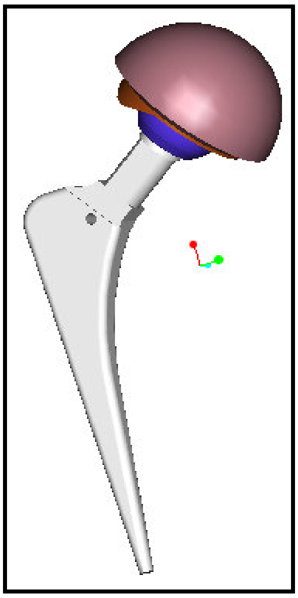

Figure 4.14 – The assembly created of the cemented hip

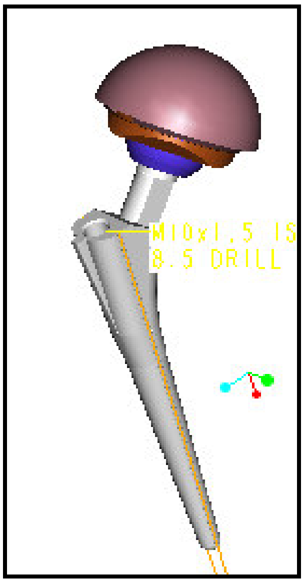

Figure 4.15 – The assembly created of the un-cemented hip

CHAPTER 5

FINITE ELEMENT ANALYSIS OF A SIMPLIFIED UN-CEMENTED PROSTHESIS

5.1 Introduction

A finite element analysis of a simplified un-cemented femoral stem bonded to the femur was the first step in modelling the behaviour of the prosthesis. The main reason for simplifying the geometry and the material properties of the model was initially to understand the FEA process and to become proficient in the analysis procedure. Finite element modelling can be difficult due to the bones having irregular geometry and complex material properties. Along with this the bonding between the bone and the prosthetic implant can also be difficult to model. (McNamara, BP, Cristofolini, L, Toni, A, Taylor, D. 1996)

The un-cemented hip is slightly simpler compared to the cemented hip for the fact that it only has two volumes needing one pair of contact surfaces. The purpose of modelling the un-cemented hip was to ease into the FEA process in preparation for the cemented hip analysis in Ansys. The FEA also aims to investigate the bonding between the implant and the bone, the deformation in all components and the stress in the femoral stem and femur.

Two finite element analyses were completed for the un-cemented hip. The first one with the bonding between the bone and the implant having a singular high value for the coefficient of friction and the second having two different (high and low) values for the coefficient of friction on different areas of the bond. This aims to simulate the instance where half of the prosthesis is bonded sufficiently and the other half is not.

5.2 Choice of Program

The choice of program primarily depended on the programs available at the university. Ansys (version 9) and Abacus were both available for students to use, however, Ansys was chosen because of prior experience and tuition gained from past subjects. The FEA abilities of both programs were considered when making this decision. Due to the fact that simplified models of the femoral systems were used in the FEA, both programs would be more than able to produce the trends needed from the analysis. Therefore the choice was based on user familiarity.

5.3 Modelling Techniques of the Simplified Un-cemented Femoral Stem

The simplified femoral stem was created first, keeping the overall dimensions realistic. The CCD angle of the femoral stem was kept as 135 degrees, which is the same as the un-cemented prosthesis that was modelled earlier in chapter 4. The femoral head was included with the stem so that fewer parts had to be inserted into the assembly. The model of the simplified femoral stem can be seen figure 5.1.

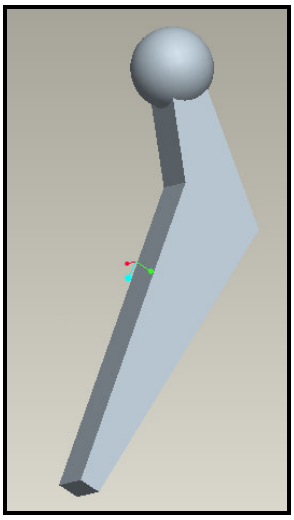

Figure 5.1 - The simplified un-cemented model of the femoral stem

The simplified model of the femur (figure 5.2) only represented the proximal portion of the bone. This would assume that when the load was applied to the femoral head, the distribution of stress would become almost obsolete in the distal regions of the femur. Because of this assumption, only half the bone was modelled. The femoral head of the bone was cut on a 45-degree angle and the area where the femoral stem would reside was removed as if it were in surgery. Due to the geometry being symmetrical, half of the femur was created first. This made it easier to remove the cavity of material. The full femur was then obtained by mirroring the part about the appropriate plane.

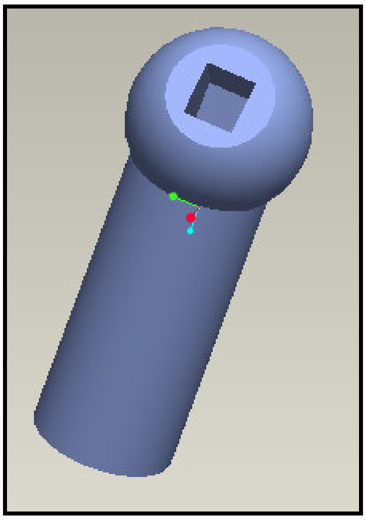

Figure 5.2 – The simplified model of the femur

An assembly representing the whole un-cemented hip was created (figure 5.3). The outer surfaces of femoral stem were mated to the inner cavity surfaces on the femur; this fully constrained the femoral system.

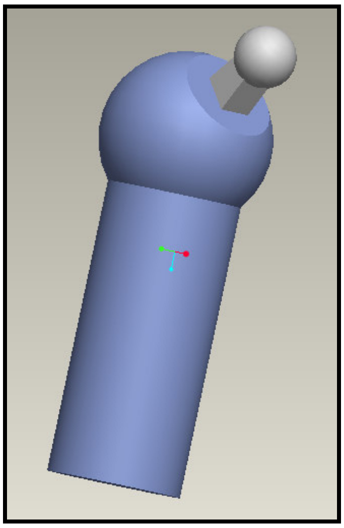

Figure 5.3 - The simplified assembly of the un-cemented femoral system

This ProEngineer assembly was then saved as a solid IGES file in preparation of the FEA analysis in Ansys.

5.4 Type of Analyses

The aim of this analysis is to investigate the stress distribution through the femur and the femoral stem, the contact pressure at the bone-implant interface and the deformation of the whole system.

The type of analysis in Ansys is a static structural analysis. This is to represent the hip when standing still with all of the body weight distributed evenly through both legs. The IGES file was imported as a two solid volumes using the default tolerances.

5.5 Choice of Elements

The choice of elements for meshing both the femur and the prosthesis were kept as simple as possible. Because of the rounded geometrical features present in the models, the solid 8-node brick element '8brick185' was selected for the bone and the implant. Having 8 nodes on the element allows the shape to be more flexible, which is what was trying to be achieved when selecting the type of element.

5.6 Material Types

The two different material types were the bone and the femoral stem (Co-Cr-Mo alloy). In reality, there would be two different types of bone that would be making up the femur, which would each have their separate anisotropic material properties. However, for the simplification of the system, it was assumed that only the cortical bone made up the volume of the prosthesis. For further simplification purposes, it was also assumed that the femoral bone was isotropic, with a Young's Modulus is 16.2GPa and a Poisson's Ratio of 0.36. These material properties were found earlier in the literature review (chapter 2).

The Co-Cr-Mo alloy femoral stem was assigned with a Young's Modulus of 210 GPa and a Poisson's Ratio of 0.3. This is true to what the actual prosthesis material properties would be (according to the literature review in chapter 2), assuming that there were no material or manufacturing processing defects to the implant.

5.7 Meshing

When creating the meshed volumes, each component had to be meshed separately so that the correct material properties could be assigned to each. The solid 8-node brick elements (SOLID185) were used with the default hexagon shaped elements meshed freely. This can be seen in figure 5.4.

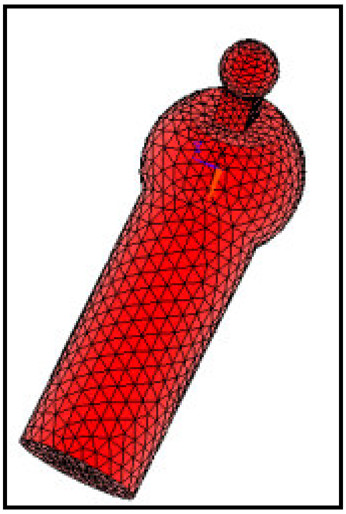

Figure 5.4 - The meshed volumes of the un-cemented hip

To obtain more precise results, the meshing of the volumes would be a very critical step in the analysis process. You would map the mesh instead of mapping it freely. You would also have smaller more densely spaced elements in the areas that would be likely to have higher stresses, or in the regions that are of more interest to the analysis. Due to the fact that this analysis is simplified, free meshing was adequate.

5.8 Contact Areas

When trying to replicate the bond between the bone and the implant, contact pairs need to be created in Ansys. The target areas were selected first. These were the inner surfaces of the cavity on the femur that were in contact with the implant. Selecting these areas proved to be somewhat difficult as some of the areas on the prosthesis had exactly the same geometry as the bone. Each surface area had a separate number in the system, so by trial and error, the correct areas were eventually chosen.

The next step was to select the contact areas. These were the areas on the prosthesis that were in contact with the femur. The same problem occurred when trying to select the correct areas on the prosthesis. Again trial and error was used to select the correct surfaces that were the same as the cavity in the femur.

Once the target and contact areas had been selected, material properties and a coefficient of friction had to be assigned to the bond at the contacting interface. For the two analyses, the material properties of the bone were selected to replicate the bone growing into the metal implant.

A high coefficient of friction of 0.98 was selected for all contact areas for the first analysis. This assumed a good bond had been achieved between the bone and the implant. Figure 5.5 shows the contact pair that was created in the contact wizard.

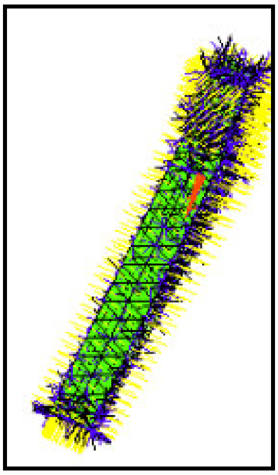

Figure 5.5 - The contact pair that was created for the bone between the bone and the implant

The second analysis had two separate contact pairs. The first pair with a coefficient of friction equalling 1 (strong bond) and the second pair with a coefficient of friction equalling 0.2 (week bond).

5.9 Appling Loads and Constraints

Applying the loads and constraints are essential for any analysis. Without each other, the system would not be fully constrained, and there would be no results. The constraints were originally on all of the outer areas of the femur, however, further thought and research proved that that would be far from what was happening in the real system. The femur would actually be allowed to move within the muscles and tissue surrounding it.

The constraints were then changed to just the bottom area where it had been chopped in half for the purpose of simplification. This assumes the femur would not move along the cylindrical axis of the leg, i.e. from the hip to the knee, and in the x and y directions at the constrained location. The constraints were set so there were no degrees of freedom for the area selected.

The loads acting on the femoral system simulate the person standing still with all of the body weight distributed through both hips. The resultant force in this situation acts vertically downwards onto the femoral head. To simulate this, equal forces were assigned to act on 26 different nodes on the top of the femoral head (figure 5.6). The total force was made equivalent to 105kgs (1030.05N), so this divided equally over the 26 nodes gave 40N on each node (figure 5.7).

The load on the head is approximately four times the load that would be acting if the person was of average weight (70kg) and had the load distributed evenly through both hips. The reason for this larger load was to simulate a worse case scenario.

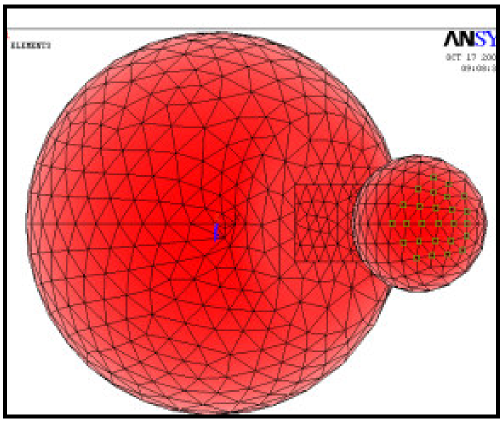

Figure 5.6 – The nodes selected on the femoral head for the application of loading

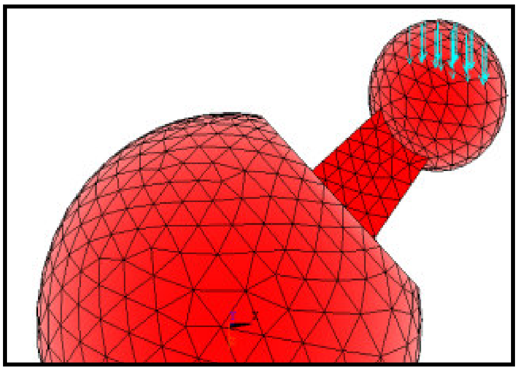

Figure 5.7 – The loading applied to the femoral head

5.10 Results

Being able to solve both systems showed that the components were constrained, the forces had been correctly applied and that the contact pair was working. When initially looking at the deformed shapes and the distribution of stress, everything appears to be in order.

5.10.1 Results for the Analysis with One Contact Pair

The results for the first analysis, where there was only one contact pair, showed realistic deformation for the direction of the loading and the constraints made on the model. The maximum deformation in the system was 0.00413mm. Figures 5.8 and 5.9 show the deformed and undeformed shape in the different viewing planes.

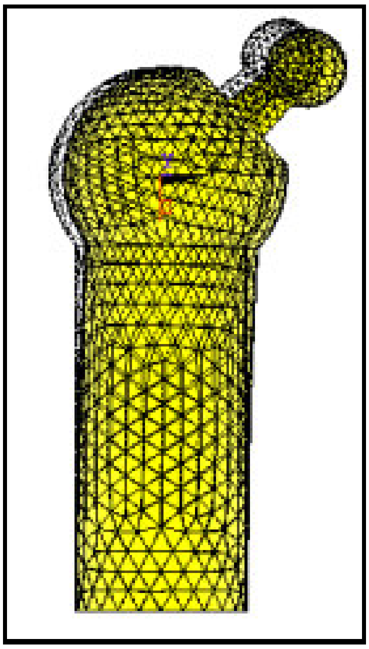

Figure 5.8 – The deformed and undeformed shape for the un-cemented FEA in the z-x plane

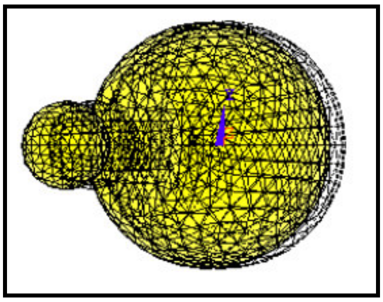

Figure 5.9 – The deformed and undeformed shape for the un-cemented FEA in the x-y plane

When looking at the distribution of stress in both of the components, again the areas where the high and low stress initially appears to be reasonable. There were practically no stress concentrations on the outer areas of the femur. This would be due to the considerable amount of material between the prosthesis and the outer surfaces of the bone (figure 5.10). There are however stress concentrations in the areas on the femur that are being compressed by the prosthesis. You can see a small amount of this in figure 5.11. The maximum amount of stress in the whole system was 47.407 KPa.

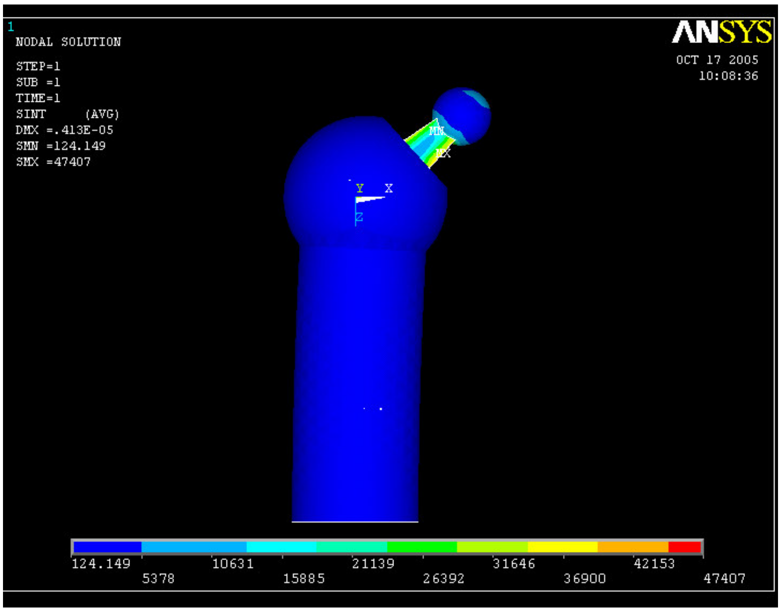

Figure 5.10 – The stress intensity in the un-cemented hip components in the z-x plane for 1 contact pair

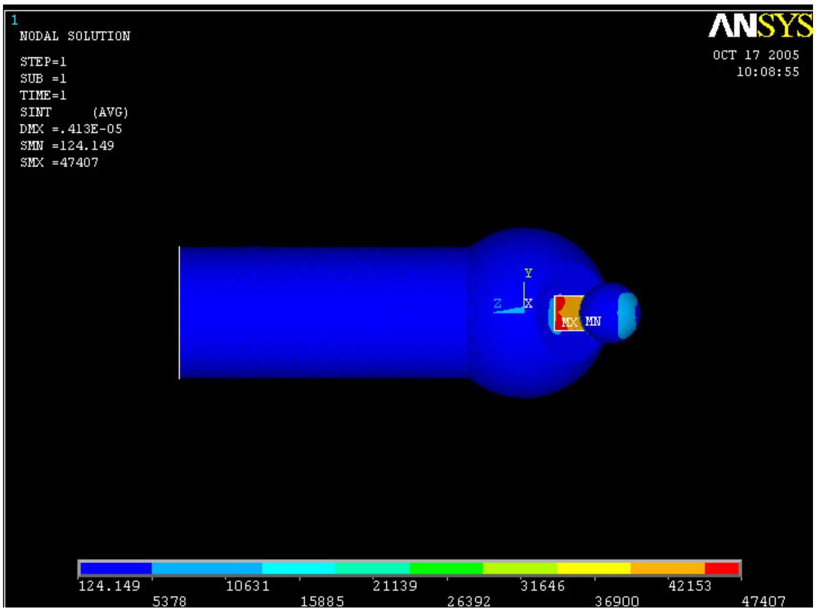

Figure 5.11 – The stress intensity in the un-cemented hip components in the x-y plane for 1 contact pair

The distribution of stress within the femoral stem was far more obvious (Figure 5.12). The areas of higher stress were realistic given that the prosthesis had sharp square corners. The neck of the femoral stem was also subjected to greater stress. This is due to the bottom of the femoral stem being pushed into the bone by the force acting on the femoral head. This was stressing the underside of the femoral neck.

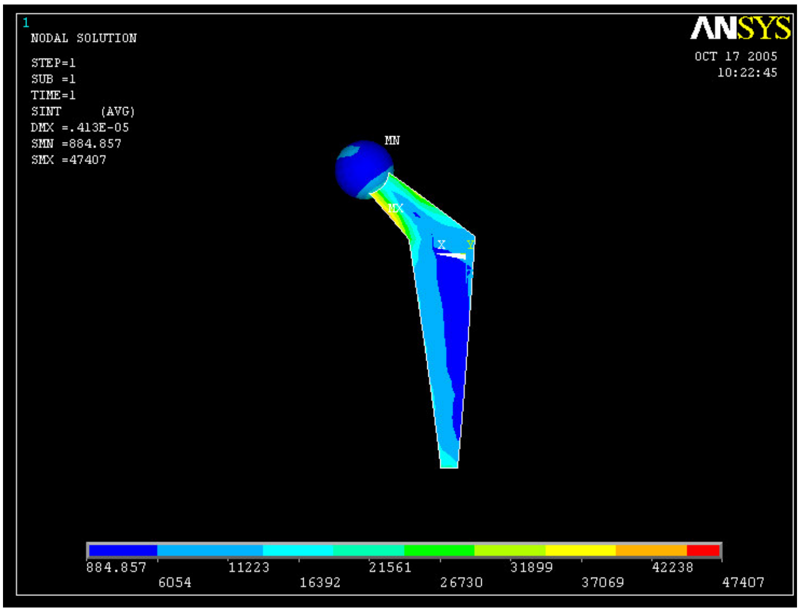

Figure 5.12 – The distribution of stress in the femoral stem in the un-cemented FEA in the z-x plane for 1 contact pair

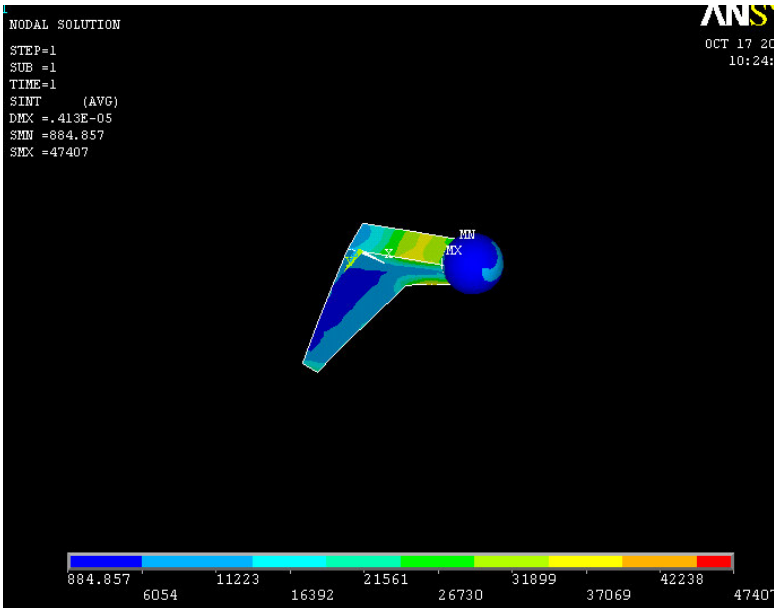

Figure 5.13 – An isometric view of the femoral stems showing the stress on the lateral side for 1 contact pair

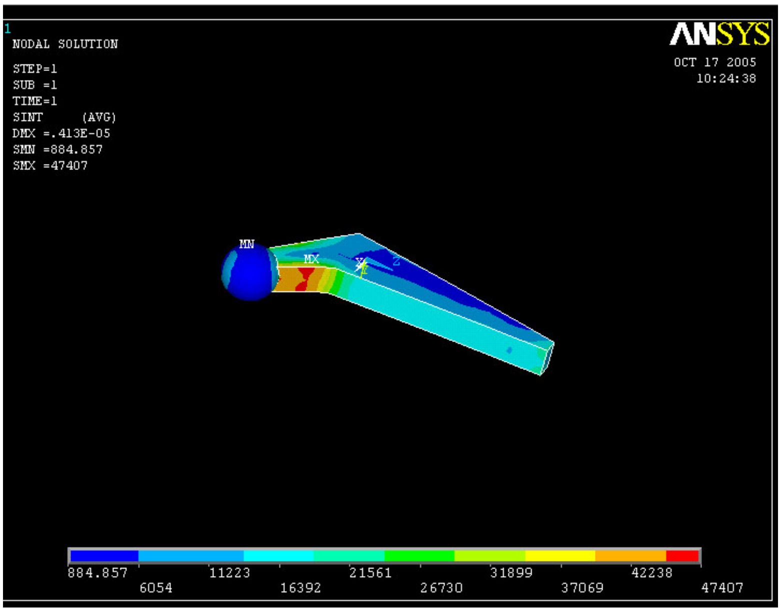

Figure 5.14 – An isometric view of the femoral stem showing the stress on the medial side for 1 contact pair

5.10.2 Results for the Analysis with Two Contact Pairs

The second analysis, where there were 2 contact pairs, showed similar results. The way in which the components deformed were similar, however the maximum value for deformation was 0.00105mm. The deformed shapes can be seen in figures 5.15 and 5.16.

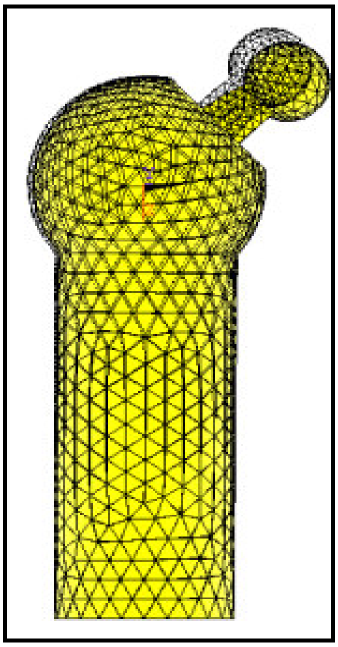

Figure 5.15 – The deformed and undeformed shape of the un-cemented hip in the z-x plane with 2 contact pairs

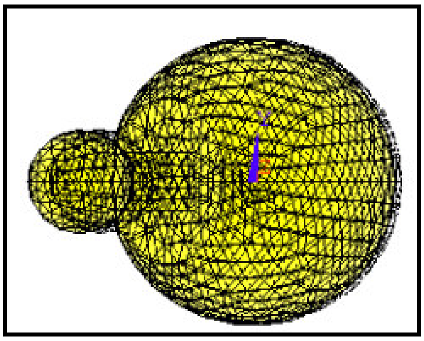

Figure 5.16 – The deformed and undeformed shape of the un-cemented hip in the x-y plane with 2 contact pairs

The stress throughout the system was generally greater with the 2 contact pairs. There were more evident changes in stress in the area on the femur directly under prosthesis. This can be seen in figure 5.17.

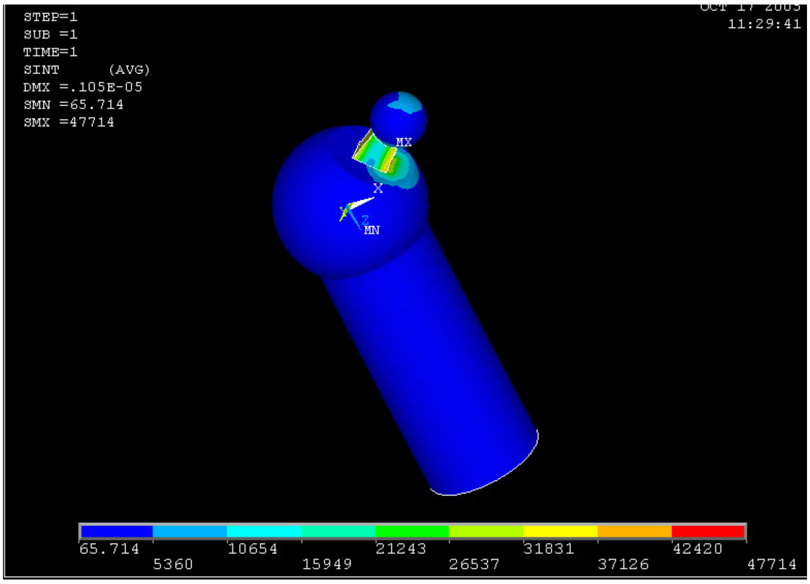

Figure 5.17 – The stress intensity of the un-cemented hip with 2 contact pairs

The maximum value of stress was 47.714 KPa. This higher value of stress was occurring on the underside of the femoral stem. You can see from figure 5.18 that there was greater stress where the higher coefficient of friction was assigned to the contact elements for that side.

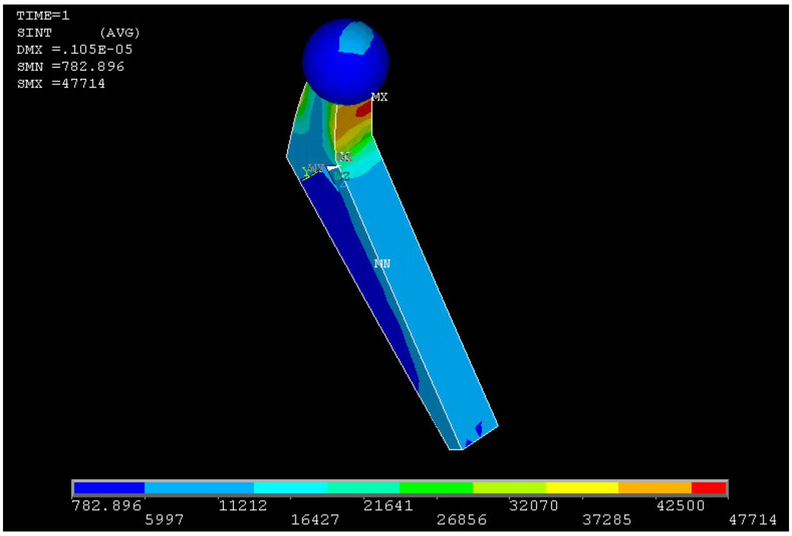

Figure 5.18 – View 1 of the stress intensity through the femoral stem for the un-cemented hip with 2 contact pairs

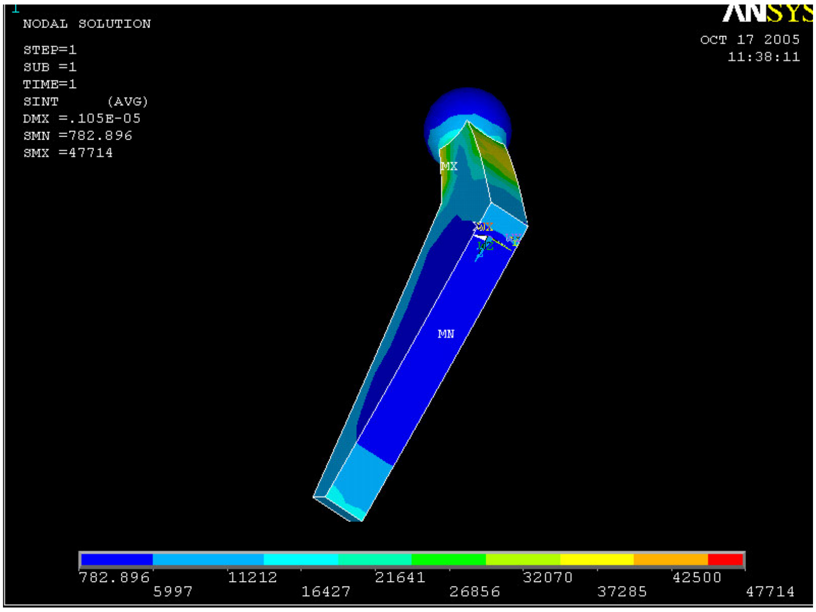

Figure 5.19 - View 2 of the stress intensity for the un-cemented femoral stem with 2 contact pairs

Again there are larger stresses in the areas with sharp corners. The underside of the prosthesis had larger stress values due to the way it was being pressed down from the force on the femoral head and constrained by the femur at the distal end of the implant. Both areas that were being pushed upon, the area on the femoral head where the force was acting and the area on the prosthesis being levered into the femur, were experiencing larger stress values.

5.10.3 Contact Pressures for both Analyses for the Un-cemented Hip

The contact pressures for both analyses were non-existent (figure 5.20). This suggests that the force applied was too great and had broken the bond between the implant and the prosthesis. For the value of force applied in both analyses (equivalent to 105kg on each hip) the bond between the implant and the bone could well have been broken. Thus, this is not such an unrealistic result.

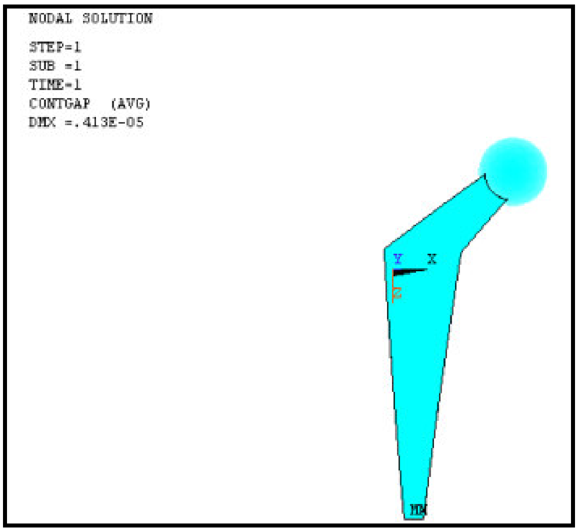

Figure 5.20 - The non-existent contact pressure on the un-cemented femoral stem

5.11 Discussion of Findings

Comparing the analyses and assessing the results obtained, certain trends and observations could be made.

- Areas with sharp corners had higher stress concentrations. This was expected because of basic stress analysis theories.

- Contact areas that have a stronger bond put a greater stress on the implant and would more than likely have a higher contact pressure.

Reasons for not having any contact pressure between the bone and the implant in the analyses performed would be that the loading was too high and/or the geometry of the components were too far from the real system, which affected the bonding between the implant and the bone.

- In the system with the lower coefficient of friction present, more stress was placed on the femur.

This could be explained by the movement of the prosthesis being greater with a lower coefficient of friction, thus applying more force onto the bone, which is in direct contact.

- The analysis with two contact pairs had a lower deformation for the whole system compared to the analysis with one contact pair.

This is a realistic trend when talking about the prosthesis. When more movement of the femoral stem is allowed (i.e. when the bond is weaker), there is more relative movement of the whole implant, rather than deformation.

When investigating the maximum stresses in the femoral stem for analyses 1 and 2 (47.407 KPa and 47.704 KPa respectively), it was evident the Co-Cr-Mo implant was far from its yielding point of 550 MPa (literature review, chapter 2). This would indicate that failure would not be occurring in the femoral stem.

CHAPTER 6

FINITE ELEMENT ANALYSIS OF A SIMPLIFIED CEMENTED PROSTHESIS

6.1 Introduction

Three finite element analyses of simplified cemented hips were also completed in Ansys. The reasons for simplifying the geometry and the material properties for the cemented hip are the same for that of the un-cemented hip, to understand the procedure, processes and trends of the FEA. The cemented hip had three volumes (femur, cement and femoral stem) needing two pairs of contact surfaces.

The objectives for completing finite element analyses for the cemented hip was to analyse the stress behaviour in the cement, the femoral stem and the femur, the contact pressure between the implant, cement and bone and the deformation of the whole system.

6.2 Modelling Techniques of the Simplified Cemented Femoral Stem

The simplified femoral stem used in the un-cemented analysis was also used as the femoral stem in the cemented analysis (figure 5.1).

The simplified femur (figure 6.1) again only represented the proximal portion of the bone. The outer dimensions for the bone were kept the same as the femur used in the un-cemented analysis. However the cavity area, where the prosthesis would be located when inserted, was enlarged so that a 2mm layer of cement could be added to the system. The femoral head of the bone was again cut on a 45-degree angle and the area where the femoral stem would reside was removed as if it were in surgery. The femur was created as a half and then mirrored about the appropriate plane.

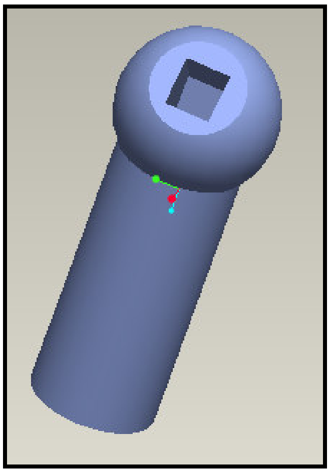

Figure 6.1 - The simplified femur that was created for the cemented analysis

The 2mm thick layer of cement was created in ProEngineer by using the outer dimensions of the femoral stem and then protruding 2mm outwards in all directions (figure 6.2). One half of the cement volume was created first and then mirrored about the appropriate plane.

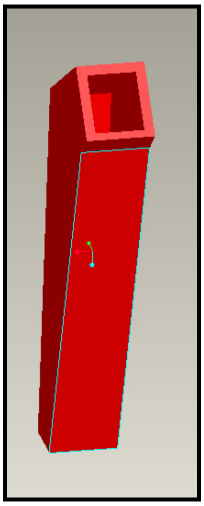

Figure 6.2 - The simplified cement volume used in the cemented hip analysis

An assembly representing the whole cemented system was then created. The outer surfaces of femoral stem were mated to the inner cavity surfaces on the cement volume. The inner surfaces of the cavity in the femur were then mated to the outer surfaces of the cement volume. This fully constrained the cemented hip modelled. An assembly and a section of the assembly can be seen in the figures 6.3 and 6.4 respectively.

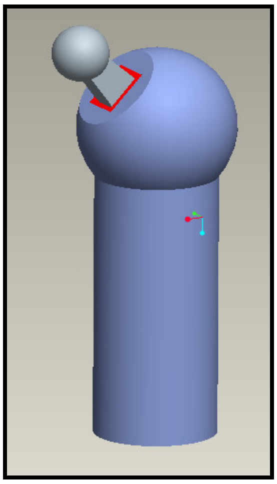

Figure 6.3 - The simplified assembly of the cemented femoral system in ProEngineer

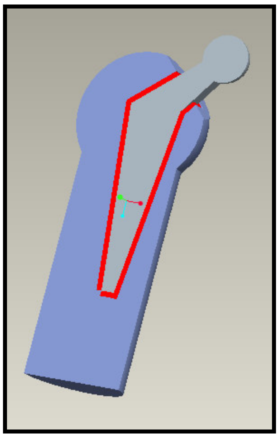

Figure 6.4 - A sectioned assembly of the simplified femoral system in ProEngineer

This assembly was then saved as a solid volume in IGES format so that it could be imported into Ansys for the FEA.

6.3 Types of Analysis

The aim of these analyses was to investigate the stress distribution through the femoral stem, the cemented volume and the femur, the deformation of the whole system and the contact pressures between the bone, cement and implant. Two of the analyses were kept the same with only the load on the femoral head varied. The third analysis varied the constraints acting on the system.

The types of analyses in Ansys were static, structural analysis. This is to represent the cemented hip when standing still with all of the body weight distributed evenly through both legs. The IGES file was imported with three solid volumes using the default tolerances. This can bee seen in figure 6.5.

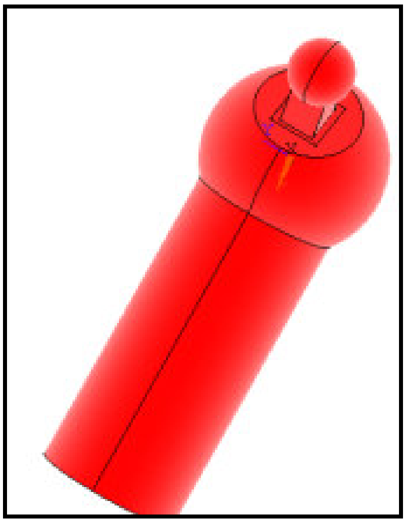

Figure 6.5 - The simplified cemented femoral system imported into Ansys

6.4 Choice of Elements

The choice of elements for meshing the femur, the cement volume and the prosthesis were kept as simple as possible. Because of rounded geometrical features present in some of the models, the solid 8-node brick element ‘8brick185’ was selected for all three volumes. Having 8 nodes on the element again allows the shape of the element to be more flexible.

6.5 Material Types

The three different material types were bone, PMMA bone cement and the Co-Cr-Mo alloy implant. The properties for the femur were again simplified to an isotropic material with a Young’s Modulus is 16.2 GPa and a Poisson’s Ratio of 0.36. The femoral stem remained the same with the properties of Co-Cr-Mo alloy (Young's Modulus of 210 GPa and a Poisson's Ratio of 0.3).

The properties for the PMMA bone cement used were taken as isotropic, which is a true representation of the material when assuming the cement has no impurities or exterior substances altering its composition. The material properties found in the literature review (chapter 2) showed the Young's modulus equalled 3.5 GPa and the Poisson's Ratio equalled 0.3.

6.6 Meshing

When creating each separate meshed volume, the appropriate material properties for each component were appropriately assigned. The solid 8-node brick elements (SOLID185) were used with the default hexagon shaped elements meshed freely (figures 6.6 and 6.7).

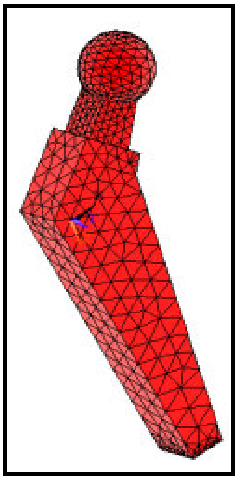

Figure 6.6 - The meshed volumes of the femoral stem and the cement

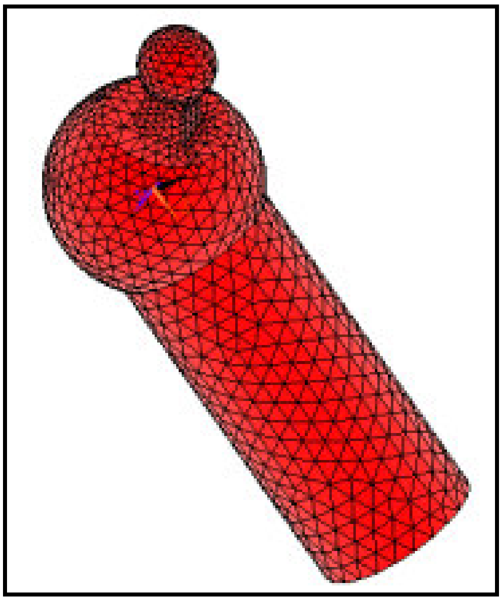

Figure 6.7 - The meshed volumes of all three components of the simplified cemented femoral system

6.7 Contact Areas

Replicating the bond between the bone, cement and the implant required 2 contact pairs. The target areas were selected first on the inner surfaces of the cavity on the femur that are in contact with the cement. The contact areas on the outer surface of the cement volume were then selected to complete the first contact pair. The material properties assigned to this contact pair were for that of bone and the coefficient of friction was 0.95, assuming a tight bond between the bone and the cement.

The second contact pair was then created by first selecting the target areas on the inner surfaces of the cement volume. The contact surfaces selected next were the outer surfaces of the femoral stem. The material properties assigned to this contact pair were for that of cement and the coefficient of friction was 0.95.

The same problem occurred for the cemented hip when trying to select a surface on one volume where another surface, exactly the same, existed in the same location. By keeping track of the surface numbers, this problem was overcome again by trail and error. Figures 6.8 and 6.9 show the contact pairs created in Ansys.

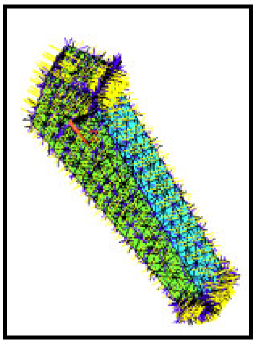

Figure 6.8 - The contact pair between the femur and the cement

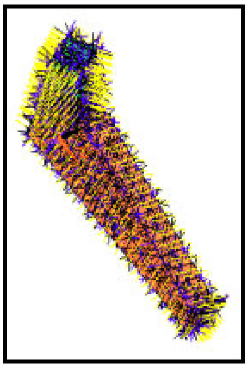

Figure 6.9 - The contact pair between the cement and the femoral stem

6.8 Applying Loads and Constraints

The structural constraints on the analyses were all assigned to the base area of the femur (where it appears to be chopped off). This area was assigned with zero displacement in all degrees of freedom. The third analysis had an additional constraint to the femoral head. After doing the first two analyses it was then thought that in reality the femoral head would be restricted from moving by the acetabular cup. Six nodes (figure 6.10) on the medial side of the femoral head were constrained in all degrees of freedom.

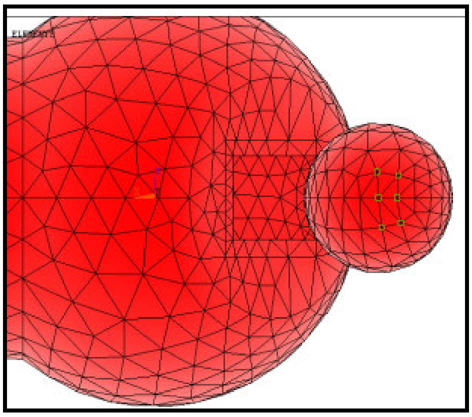

Figure 6.10 – The nodes on the femoral head that were constrained for the third analysis of the cemented hip

The static load applied vertically on the femoral head was again distributed evenly through 26 nodes on the top of the femoral component. For the first cemented analysis, the total force acting on the system was made equivalent to 26.5 Kg (260N). This divided equally over the 26 nodes gave 10N on each node (figure 5.6). This would be a reasonable load of average body weight minus the weight of the legs acting on one hip.

The second cemented analysis had a total force of 53kg (520N). This divided evenly over the 26 nodes gave 20N on each node. The third analysis had the 10N on each of the 26 nodes (same as the first cemented analysis) but also had the extra constraints on the femoral head.

6.9 Results

6.9.1 Results for the First Cemented Hip Analysis

The results from the first cemented analyses (smaller load without constraints on the femoral head) showed that the maximum deformation to the whole system was 2.676mm. This can be seen in the figures 6.11 and 6.12.

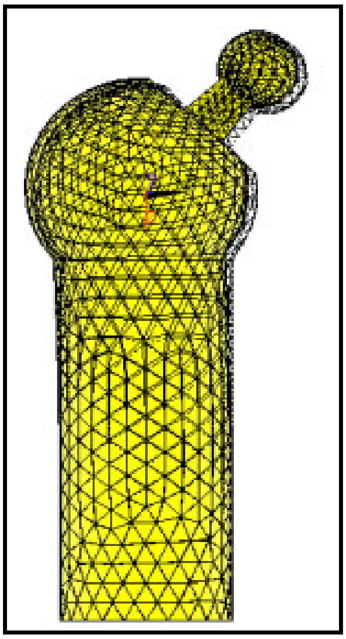

Figure 6.11 - The deformation in the z-x plane for the first analysis of the cemented hip

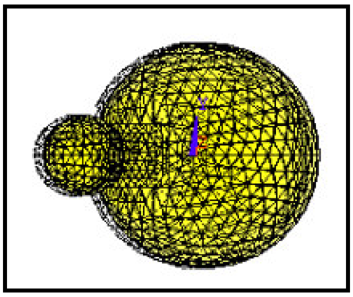

Figure 6.12 - The deformation in the x-y plane for the first analysis of the cemented hip

The stress intensity in the cement for the first analysis showed maximum stress values of 38 MPa. The areas showing such values were right at the base and were extremely small (Figure 6.13). Most of the higher stress values were situated in the distal third of the cement, however stresses of 5.75 MPa were found up on the top areas of the cement where the sharp corners were present (figure 6.14).

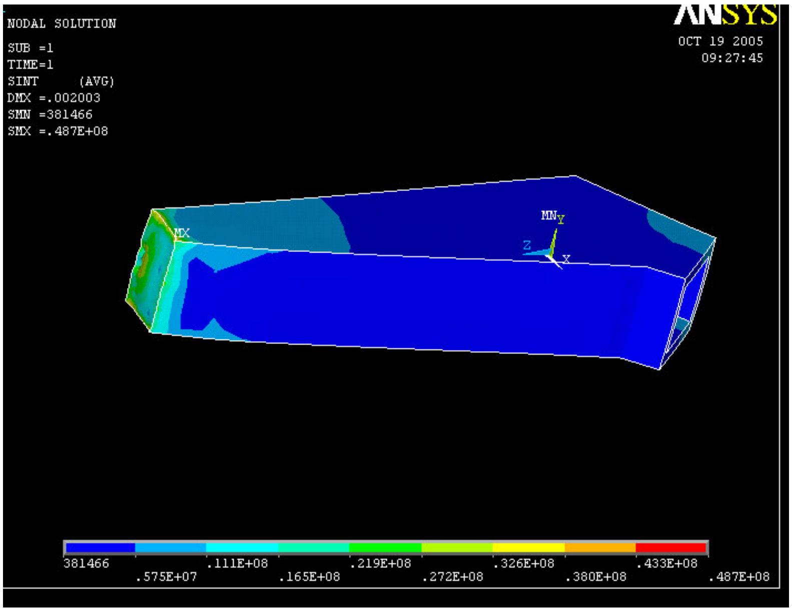

Figure 6.13 – View 1 of the stress distribution in the cement for the first cemented hip analysis

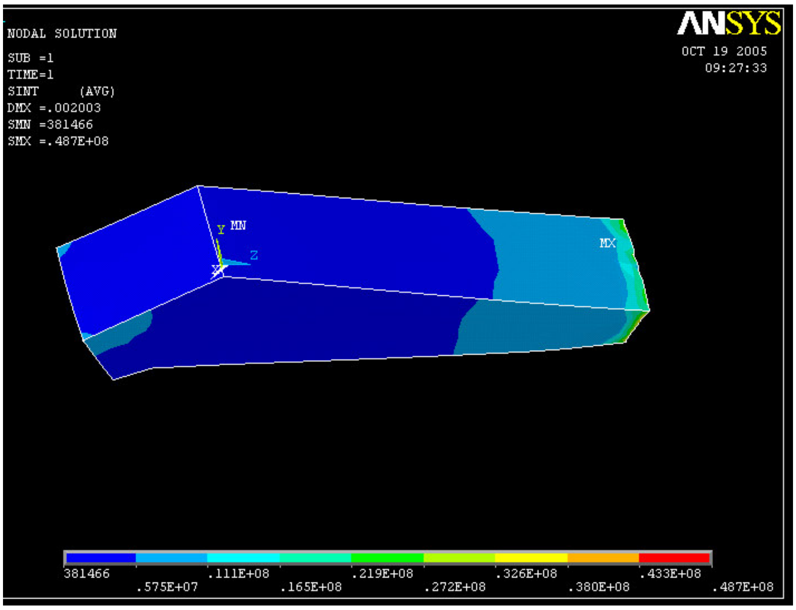

Figure 6.14 - View 2 of the stress distribution in the cement for the first cemented hip analysis

The stress in the prosthesis had the highest values for the system (48.7 MPa). These larger stress values were only at the very bottom of the femoral stem (figure 6.15). The distal area of the femoral stem was where the larger stress values were found.

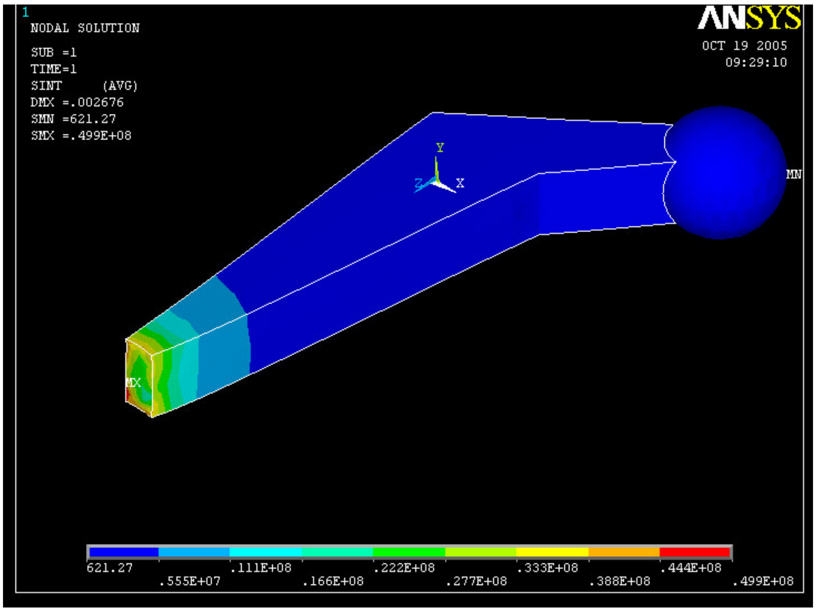

Figure 6.15 - The stress distribution of the femoral stem for the first analysis of the cemented hip

6.9.2 Results for the Second Cemented Hip Analysis

The results obtained for the second analysis with the larger load showed a maximum deformation on the whole system of 3.68mm. Figures 6.16 and 6.17 show the way deformation was occurring.

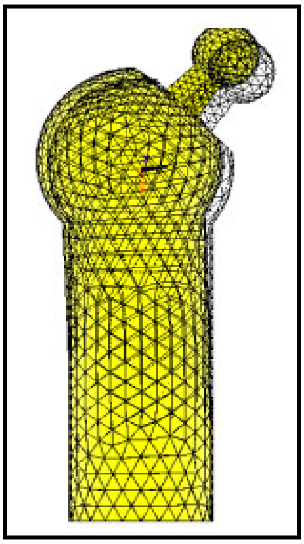

Figure 6.16 - The deformation of the cemented hip in the z-x plane

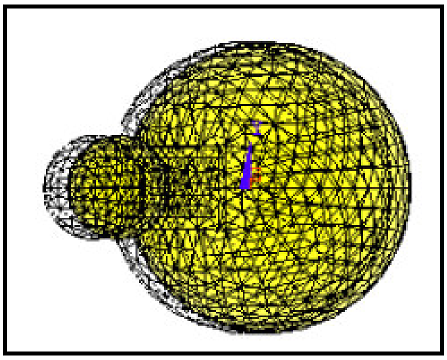

Figure 6.17 - The deformation of the cemented hip in the x-y plane

When considering the way in which the components were loaded and constrained, the maximum value and the way in which the components were deformed seam realistic.

The maximum stress intensity over all three components was 3.65 MPa. Figure 6.18 shows the stress distribution in the whole system.

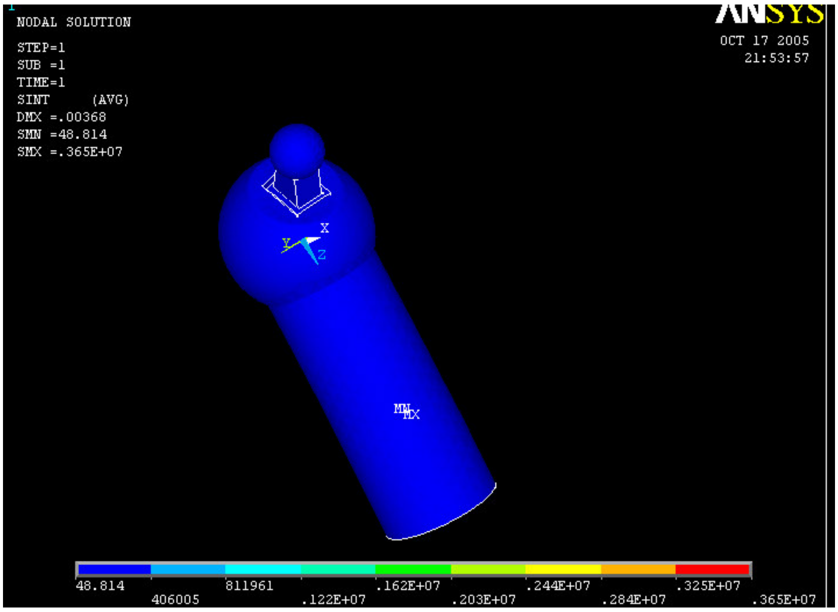

Figure 6.18 - The distribution of stress in the whole cemented hip for the second analysis

When looking at the stress in the cement volume (figures 6.19 and 6.20), you can see that the areas with larger stress values were the same as the first analysis. The only difference is that the value for stress is considerably lower compared to that of the first analysis. The maximum stress in the cement for second cemented analysis was 0.9437 MPa.

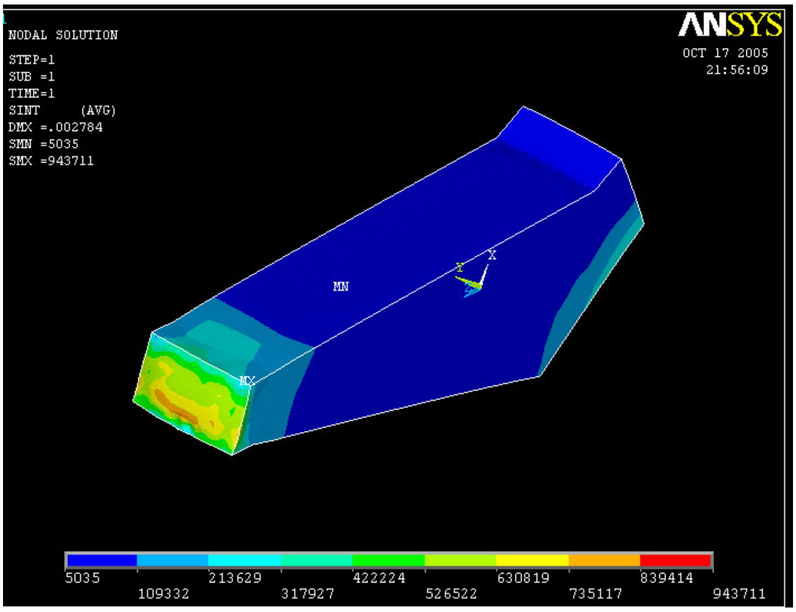

Figure 6.19 - View 1 of the distribution of stress in the cement for the second analysis of the cemented hip

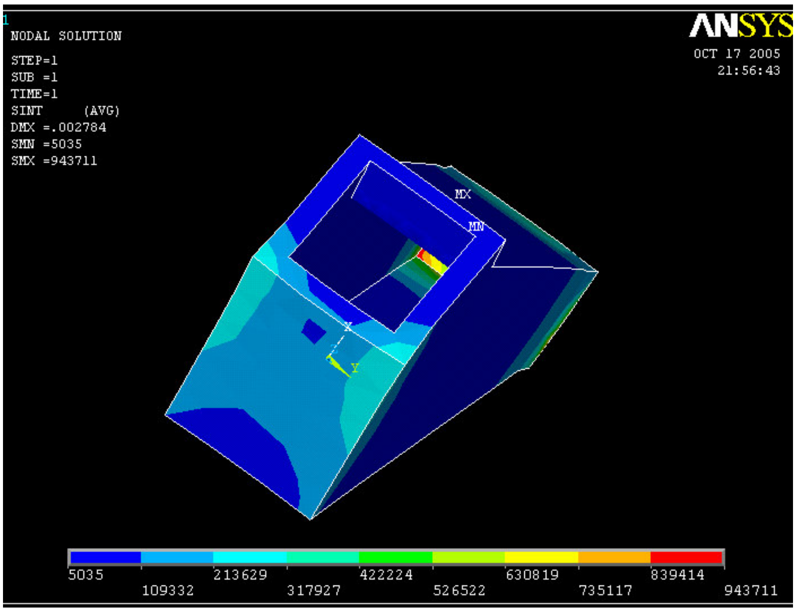

Figure 6.20 - View 2 of the distribution of stress in the cement for the second analysis for the cemented hip

The stress intensity throughout the femoral stem for the second analysis also had the higher values in the distal area of the prosthesis (figure 6.21). The maximum value of stress in this component was 3.16 MPa.

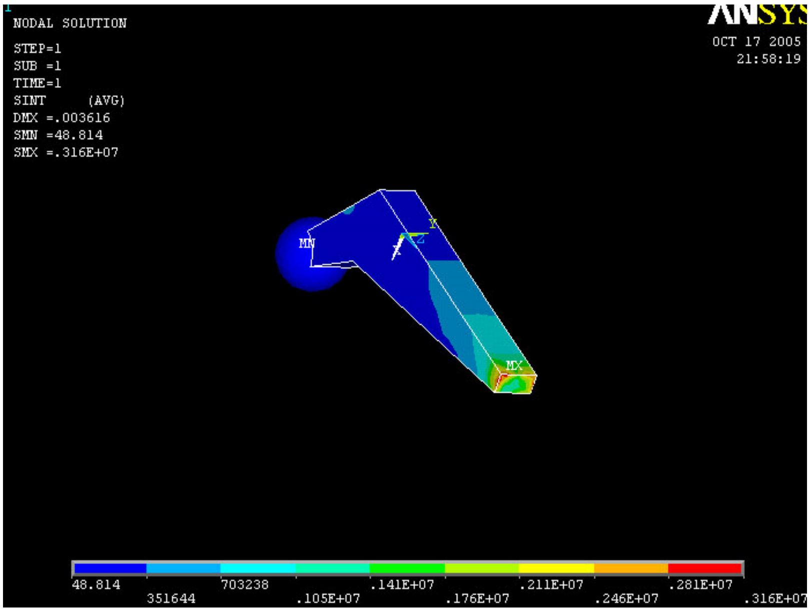

Figure 6.21 - The distribution of stress throughout the femoral stem for the second analysis of the cemented hip

6.9.3 Results for the Third Cemented Hip Analysis

The third analysis with constraints on the femoral head showed quite different results from the other two cemented hip analyses. The maximum deformation in the whole system was only 0.0000238mm (figures 6.22 and 6.23).

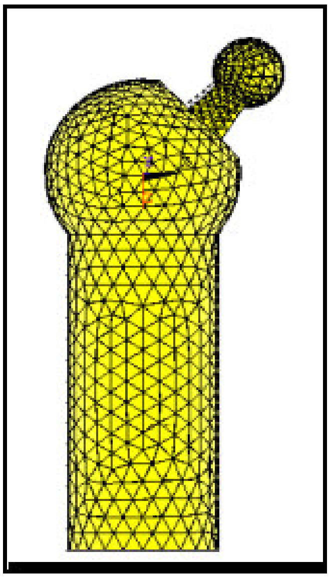

Figure 6.22 - The deformation for the third analysis done on the cemented hip in the z-x plane

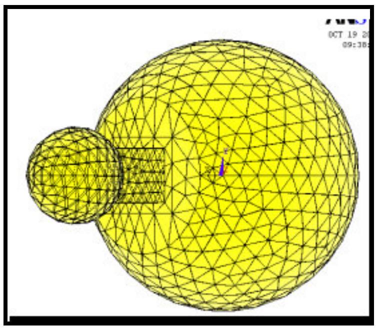

Figure 6.23 - The deformation for the third analysis done on the cemented hip in the x-y plane

The stress through the whole system (figure 6.24) had a maximum value of 24.169 Pa. This is significantly small compared to the first analysis of the cemented hip. When looking at figure 6.24 however, you can see that the bond between the femoral stem, cement has been broken.

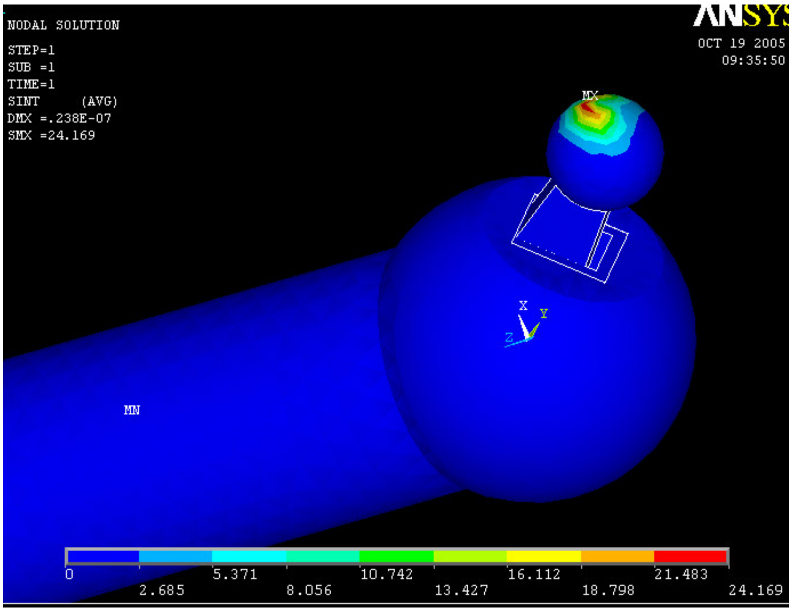

Figure 6.24 - The stress for the third analysis for the cemented hip

The only significant form of stress on the femoral stem was where the load was being applied. This can be seen in figure 6.25. The rest of the prosthesis is practically under no stress at all.

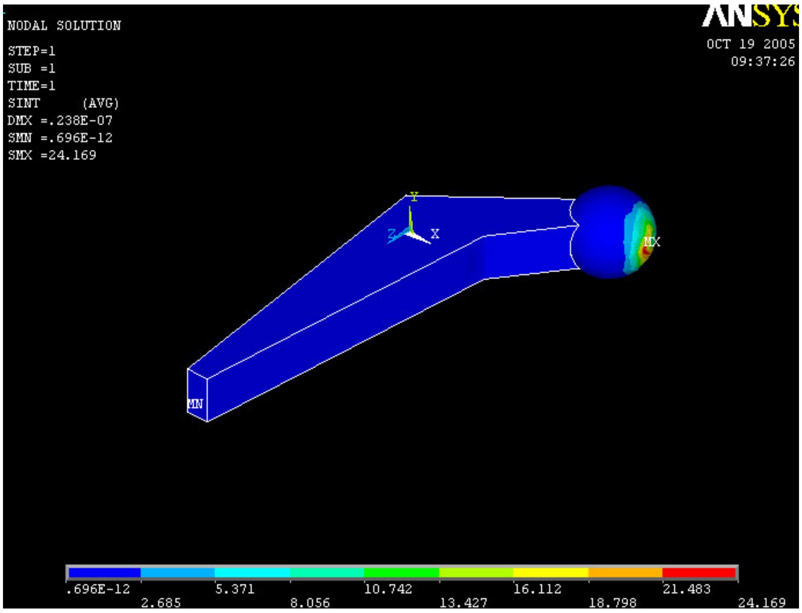

Figure 6.25 - The stress in the femoral stem for the third analysis of the cemented hip

6.9.4 Results Showing the Contact Pressures for the Three Analyses

The contact pressure for the cement in all analyses was again non-existent (figure 6.26). The bond between all of the components may have had to been broken to explain this.

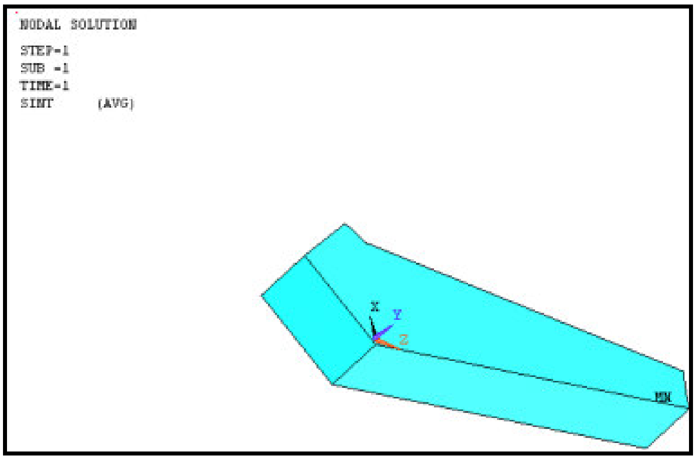

Figure 6.26 - The contact pressure on the cement for the all analyses for the cemented hip

6.10 Discussion of Findings

By comparing the first two analyses for the cemented hip some realistic and unrealistic trend were shown.

- By doubling the load on the femoral head, whilst keeping the constraints constant, the maximum deformation increased from 2.62mm to 3.68mm. This was a realistic result for the increased load.

- The maximum stress in the cement dropped from 38 MPa to 0.9437 MPa. This would be an unrealistic trend. When increasing the load, the stress in the cement would be expected increase also.

- In the femoral stem the maximum value of stress decreased from 48.7 MPa to 3.16 MPa when the load increased. Again this would be an unrealistic trend for the same reasons as the cement.

The load being too large on the second analysis can explain the decrease in stress in the components when the load is increased. The larger load may have separated the contact elements at the component interfaces, which would then change the way in which each separate component in the system was constrained.

The results obtained from the third analysis with the extra constraints on the femoral head were dramatically smaller.

- The maximum deformation in the system was 0.0000238mm. This is considerably smaller than the first analyses (2.62mm) with the same loading values.

- The maximum stress in the femoral stem did not exceed 24.169 Pa. This is also significantly smaller than the first analysis where the maximum stress in the implant was 48.7 MPa.

The deformation and the stress values were so small because of the area on the femoral head being constrained in all degrees of freedom. This shows that a slight change in the constraints can have a dramatic effect on the results obtained from the FEA.

Comparing the stress in the cement to the maximum strength of PMMA, which was found in the literature review in chapter 3 to be 70 MPa, shows that the cement has a safety factor of less than 2 before it would fail. This would be unacceptable, as the load on the hip was representing a person of average weight with all of the weight distributed evenly on both hips, remembering that at any given time the load on the hip joint can be up to four times the body weight. The stress values in the Co-Cr-Mo prosthesis were far below the yield stress value that was found in the literature review in chapter 2 of 550 MPa.

CHAPTER 7

FINITE ELEMENT ANALYSIS ON REAL PROSTHESES IN *VIVO*

7.1 Introduction

When trying to analyse the real components of an un-cemented hip, the solid models that were created in chapter 4 were used. The un-cemented femoral stem with the modular neck (Figure 7.1), the femoral head (Figure 7.2) and the cut down model of the femur (Figure 7.3) were brought together in an assembly in ProEngineer.

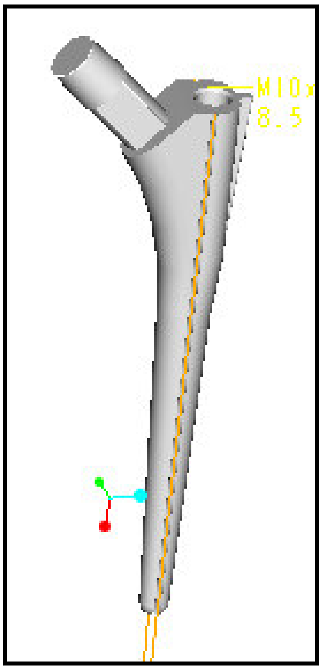

Figure 7.1 – The un-cemented femoral stem modelled in chapter 4

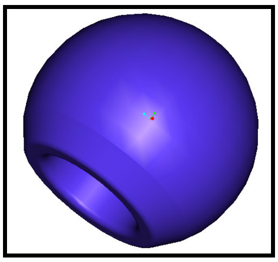

Figure 7.2 – The femoral head modelled in chapter 4

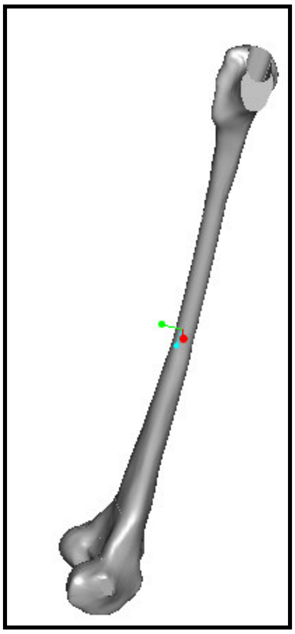

Figure 7.3 – The cut down model of the femur that was created in chapter 4

Once the assembly of the 3 components was complete, it was then saved as an IGES file so it could be imported into Ansys for the FEA.

7.2 Importing the Real Hip Model into Ansys

Unfortunately the IGES file of the real hip model could not be imported into Ansys. Several attempts were made, however the geometry of the femur and the femoral stem were far too complex for Ansys to import. Small surface areas and irregular shapes on the models were the main reason for this occurring.

The next task was to import another type of hip using a real prosthesis that had simpler geometrical features and a simpler representation of the femur. The model of the femoral stem was found on the Internet. Figure 7.3 shows the geometry of the components.

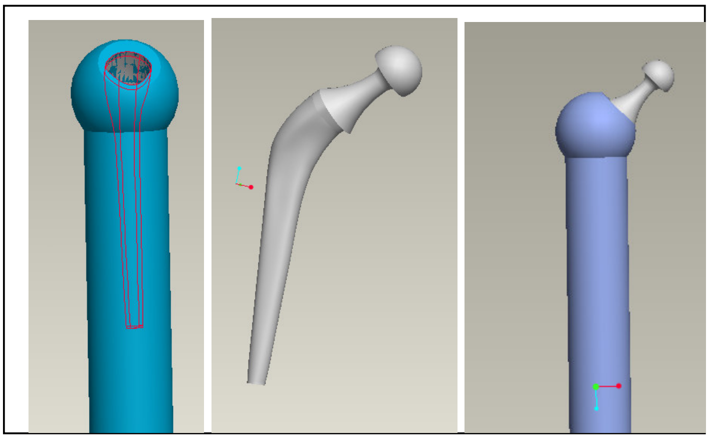

Figure 7.3 – Models of the half real, half simplified hip in ProEngineer

Once saved in IGES format, importing the assembly into Ansys was attempted. This again was unsuccessful due to the geometry of the femoral stem still being too complex.

7.3 Discussion of Findings

By attempting to analyse a real hip *in vivo*, it became evident that the model to be imported needed to be created in a way that keeps the dimensions true, but has no overlapping, small or extremely complex surfaces.

The un-cemented femoral stem that was created in chapter 4 had far too many small surfaces, which were created from cutting the volume down to get the desired geometry. The femur and the femoral stem that was found on the Internet had too many complex surfaces that were blended into each other. A way in which to overcome this problem would be to create the model of the femur from cross sections at particular intervals and then use a blended sweep in ProEngineer to obtain the outer surface.

When the models can be imported into Ansys, the processes outlined in chapter 5 and 6 would then be implemented. The loading, constraints, bone material properties and contact properties may be different depending on what the analysis was trying to achieve. The two volumes of bone (cortical and cancellous) would have to be modelled separately with their own anisotropic material properties. If the anisotropic properties would unable to be replicated, then different x, y and z properties could be assigned.

CHAPTER 8

RECOMMENDATIONS AND FURTHER WORK

8.1 Modelling and Design Recommendations

When trying to create solid models of any of the hip components, it is important to know what the models are going to be used for. If they are just for geometrical representation, then it does not matter how the final shape is obtained. However, if the models are being used for analysis or if the surfaces are being referenced in another solid model, then it is important to create the model with the least amount of complexity and small surfaces.

If modelling the femur for analyses, it would be recommended that the whole femur be modelled. This would allow more accurate positioning of constraints and forces acting on the system. It will also allow for the bone to deform as far down axially as it needs to below the base of the femoral stem.

When designing prosthetic implants it is important to remove any sharp corners and large changes in volume/area. This will reduce localised stress concentrations in the implant and in the mating cement or bone.

The bond made between the implant, cement and the bone needs to be uniformly strong between the contact areas. Contact areas that have a smaller coefficient of friction will place greater stress on the prosthetic components involved and will more than likely fail under smaller loads.

8.2 Analysis Recommendations

When undergoing a FEA on either an un-cemented or cemented THR it is important to have the constraints as true as possible to the real system. Any small adjustment to the constraints can dramatically change the results from the analysis, even if all the other input data is kept constant.

The material properties of bone should be represented as anisotropic if possible. The way in which the bone is securing the implant or cement will have a large effect on the deformation and the stress in all components.

The contact element properties should be assigned with caution. Changes to the coefficient of friction and the material properties of the contact surfaces can affect the type of bond being analysed. When implanting a THR a tight bond around the whole prosthesis to either the cement or the bone is critical to the life and the strength of the implant. Small areas of interface loosening are examples where the whole system can be affected which may lead to the failure of the implant.

8.3 Further Work

To continue on with this project there are many more ways a THR can be analysed to confirm and expand on the results already obtained. Initially the simplified finite element analyses would need to be improved so that constraints, contact properties and bone properties are as close to the real system as possible.

When improving the constraints on the models, the whole femur should be present for the analysis instead of the proximal half used in chapter 5 and 6. The femur can then be constrained at the base of the bone where it would attach to the knee. The constraints on the femoral head should be in the same location as used in chapter 6, however, the nodes should only have zero displacement in the x and y directions and not in the z direction (axially). Muscular constraints around the remaining surface areas of the femur should also be investigated and incorporated into the analysis if found necessary.

The contact properties may also need investigation. The actual coefficient of friction between the bonding interface and the material properties of the bond would need to be confirmed and changed if necessary.

Modelling and analysing the femur should include both the cortical and cancellous bone regions. The material properties would need to be assigned as anisotropic so the properties are not the same in all directions of the bone. This would be an extremely hard task to complete successfully as all bones are different and grow in response to the loading that acts upon them.

Once the simplified analyses achieve results that are reasonably true to the hip *in vivo*, the geometrically real models of the hip should then be analysed. Before the analyses takes place the models need to be re-created so that the hip assembly can be imported in Ansys in IGES format. Once imported successfully, the same process for the simplified hip analyses can then be used on the real system.

The only problem that may be perceivable in doing this would be selecting the target and contact surfaces for the bonding. If there are many surfaces making up the outer area of the femoral stem, then there will be many areas to select for the contact surface wizard in Ansys. This however isn't impossible; it will just take more time and precision.

CHAPTER 9

SUMMARY AND CONCLUSION

9.1 Summary and Conclusions

From the investigation into modes of failure in total hip replacements, the conclusions made are as follows:

- From the literature review in chapter 3, the most common form of mechanical failure in a total hip replacement is the failure of the bond between the implant and the bone. This could be either the failure of the cement, the failure of the bond between the cement, implant and bone or the failure of the osseointegration of the bone to the implant.

- When analysing the failure between the implant and the bone, the results needed for comparison are deformation, stress intensity in all components (implant, bone and cement) and the contact pressure at all bonding interfaces.

- When creating solid models of THR components and surrounding bone for the purpose of any kind of analysis or geometrical reference to other solid models, it is important they are created without extremely small, complex and/or overlapping surface areas.

- From the finite element analyses in chapters 5 and 6 it is shown that rounding the geometry of the prosthetic implants will reduce localised stress concentrations in the components.

- Interface bonding with a lower coefficient of friction (representing a weaker bond) will place higher stress on the femoral stem and the femur.

- When more movement of the femoral stem is allowed (i.e. when the bond is weaker), there is more relative movement of the whole implant, rather than deformation.

- Increasing the load on the femoral head, whilst keeping all other variables constant, increases the amount of deformation in the whole system.

- Any small changes to the constraints on the hip analysis can dramatically change the deformation and stress results. Careful consideration needs to be made when applying constraints on a THR finite element analysis.

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APPENDIX A – PROJECT SPECIFICATION

University of Southern Queensland

Faculty of Engineering and Surveying

ENG4111/4112 Research Project

Project Specification

FOR: **Vivienne French**

TOPIC: **Investigation of Modes of Failure in Total Hip Replacements**

SUPERVISORS: Chris Snook

BACKGOUND: A Total Hip Replacement (THR) is the surgical modification of the hip joint where prosthetic components replace damaged parts of the natural hip. The first THR was designed and inserted in the 1930's. Since then there has been many developments in the THR design.

Essentially there are two types of hip prosthesis, cemented and un-cemented. The cemented components are bonded to the bone using a bone-cement and the un-cemented components rely on the bone to osseointegrate to a bioactive or porus coating on the prosthesis.

PROJECT AIM: To investigate the modes of failure in hip replacements by literature reviews and research. Solid modelling and finite element analysis will be carried out on the cemented and un-cemented hip for a particular failure type.

PROJECT OUTLINE:

1. Conduct literature review of hip implant terminology and normal methods of operation.

2. Conduct literature review of common modes of failure of hip implants.
3. Compare and critique the two implant technologies (cemented and osseointegration).
4. Create parametric feature-based solid models of the un-cemented and cemented implants.
5. Conduct a literature review on the physical and mechanical properties of various hip implants components, implant movement and the loading that acts on them.
6. Conduct a FEA on the hip component. This will be focusing on the bonding between the femoral stem and the bone for both the un-cemented and cemented hip.
7. Develop results showing stress, deflection and separation of and the pressure on contact elements and the components.
8. Complete FEA reports on all analysis performed. State all assumptions and conditions.
9. Comment on and analyse the results obtained.
10. Make design, modelling and analysis recommendations regarding shape, loading conditions, constraints and bonding techniques between the bone and the femoral stem.
11. Prepare and conduct an oral presentation of the project.
12. Include and report all project aspects in a dissertation.

AGREED:

______________(Student) _______________, ________________(Supervisors)

___/___/___ ___/___/___ ___/___/__

APPENDIX B – FINITE ELEMENT ANALYSIS REPORTS

FEA Model Report

Job Name: Un-cemented simplified hip: 1 contact pair

Job Number: 1

Analyst: Vivienne French

Problem Description

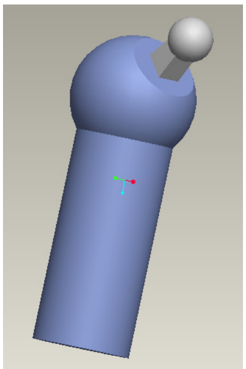
Related Drawings:

This analysis concentrates on a FEA of a simplified model of the un-cemented femoral system. The load on the system will act on the femoral head and will be counter-acted by a constraint in all degrees of freedom on the bottom are of the femur (where it appears to be chopped off). There will be contact pairs created to replicate the bond between the bone and the implant. The properties of the contact pair will have the material properties for the bone and the coefficient of friction will be 0.95, which indicates that the bond is strong. Both volumes will have there separate material properties (structural, linear, elastic, isotropic):

Implant (Co-Cr-Mo):	Bone:
E=210GPa	E=16.2GPa
V=0.3	v=0.36

Meshing the volumes will be done using the solid brick elements through a free mesh. The results obtained should show the stress intensity throughout the system and the contact pressure between the bone and the element.

The loading in the femoral head will be 40N on 26 separate nodes.



Approach to Analysis

Analysis Type: Structural

Package Used: Ansys v9

Elements Used:

SOLID185

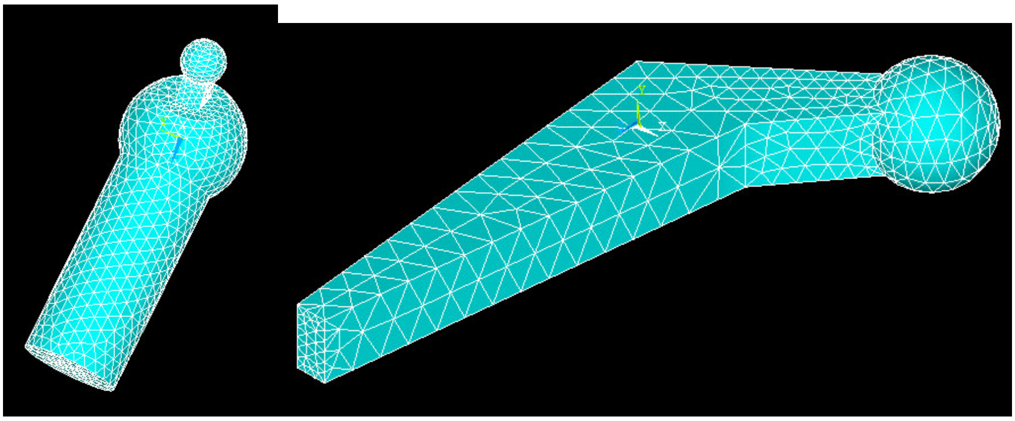
TARGE170 CONTA174

Modelling Assumptions:

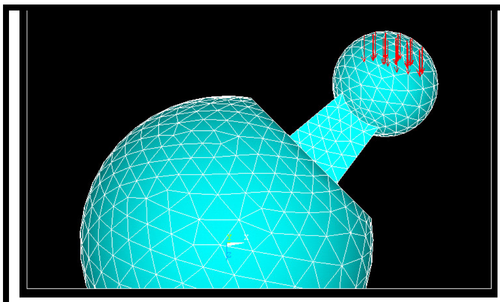
- The system is constrained at the base of the bone, where it has been cut off in the solid model
- The bone is isotropic (it is actually anisotropic)
- The contact properties are for that of the bone and the COF is 0.95 (strong bond)
- The load is spread over 26 nodes on the femoral head

Basic Model (including mesh generation):

The mesh for both components are as follows:



The loading on the system was as follows:



Default Freedom Conditions: ______UX, UY, UZ_____

(e.g. UX,UY for 2D)

Model Summary
Number of Nodes: 3760
Number of Elements: 18525
Number of DOF: 11280
Maximum Half-Bandwidth (or Wavefront):
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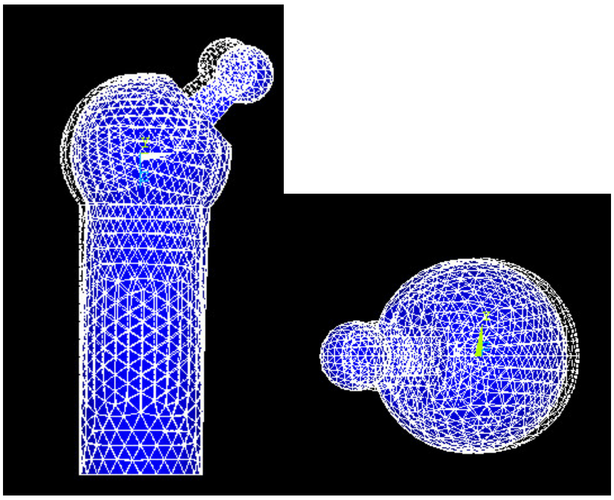
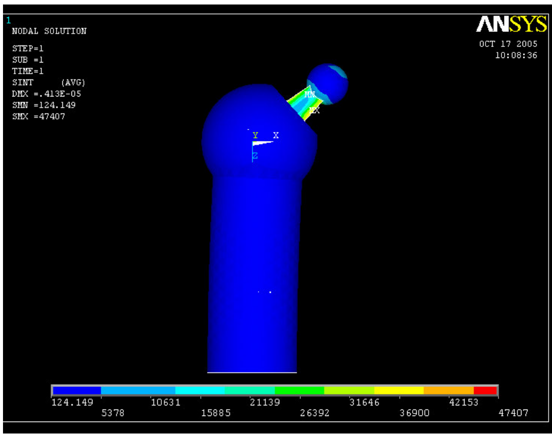
Disk Required: 11.079 Mb
Solution Time (CPU seconds): 4 minutes and 20 seconds
Errors/Warnings

Results





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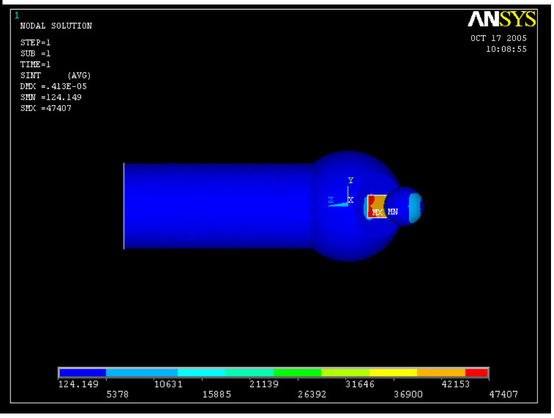

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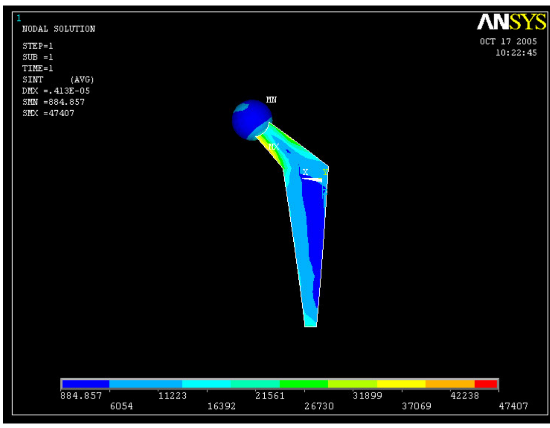

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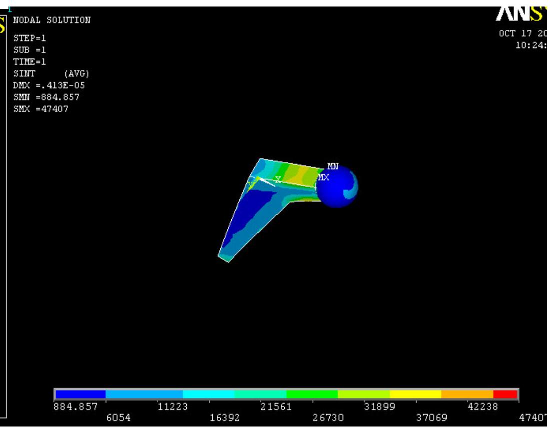

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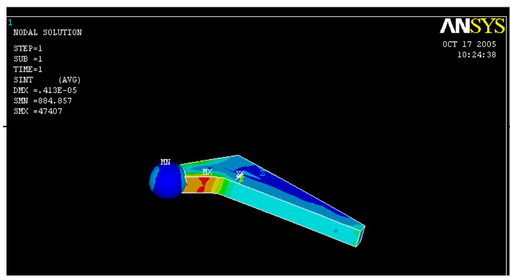

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Vivienne French</td><td>Date: 17/10/04</td></tr>
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FEA Model Report

Job Name: Simplified Un-cemented hip – strong bond on half the contact area

Job Number: 2

Analyst: Vivienne French

Problem Description

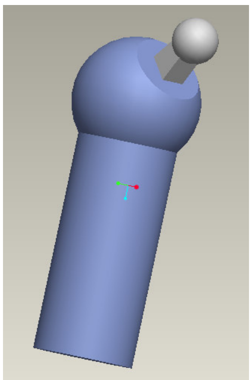
Related Drawings:

This analysis concentrates on a FEA of a simplified model of the un-cemented femoral system. The load on the system will act on the femoral head and will be counter-acted by a constraint in all degrees of freedom on the bottom are of the femur (where it appears to be chopped off). There will be contact pairs created to replicate the bond between the bone and the implant. The first contact pair will be on one side of the implant and will have a COF of 1 and the second contact pair (which select the remanding contact areas) will have a COF of 0.2. This signifies that the half the bond between the bone and the implant has loosened. The properties of the contact pairs will have the material properties for the bone. Both volumes will have there separate material properties (structural, linear, elastic, isotropic):

Implant (Co-Cr-Mo):	Bone:
E=210GPa	E=16.2GPa
V=0.3	v=0.36

Meshing the volumes will be done using the solid brick elements through a free mesh. The results obtained should show the stress intensity throughout the system and the contact pressure between the bone and the element.

The loading in the femoral head will be 40N on 26 separate nodes.



Approach to Analysis

Analysis Type: Structural

Package Used: Ansys v9

Elements Used:

SOLID185

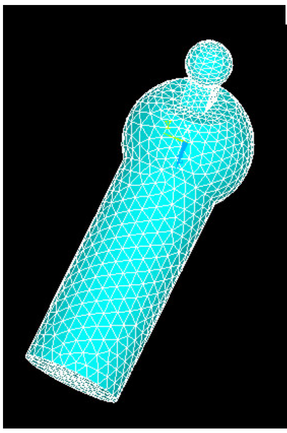
TARGE170 CONTA174

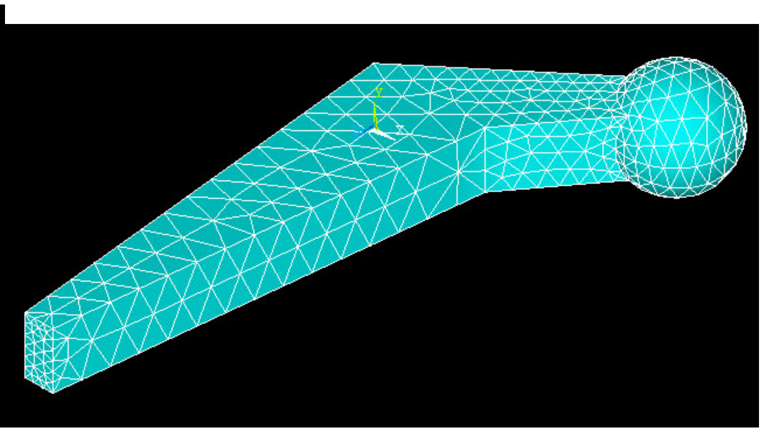
Modelling Assumptions:

- The system is constrained at the base of the bone, where it has been cut off in the solid model
- The bone is isotropic (it is actually anisotropic)
- The contact properties are for that of the bone and the COF of 1 is string and the COF of 0.2 is not strong (loose)
- The load is spread over 26 nodes on the femoral head

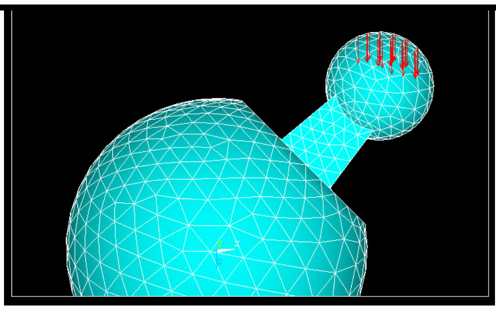
Basic Model (including mesh generation):

The mesh for both components are as follows:





The loading on the system was as follows:



Default Freedom Conditions: ______UX, UY, UZ_____

(e.g. UX,UY for 2D)

Model Summary

Number of Nodes: 3688
Number of Elements: 19332
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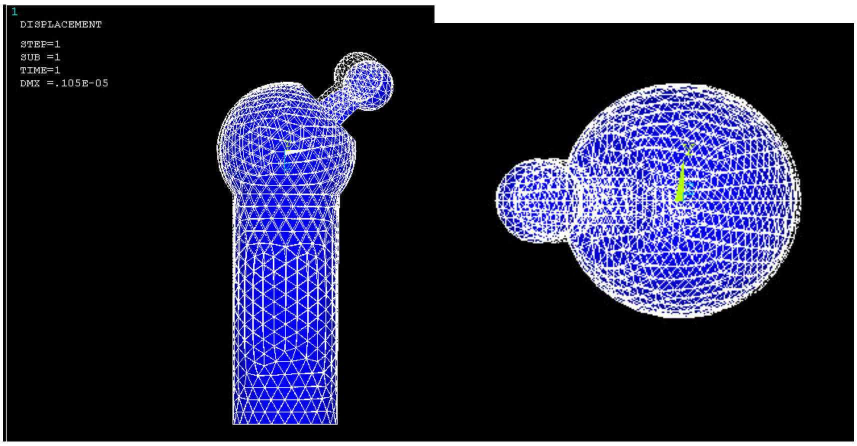
Disk Required: 10.956 Mb
Solution Time (CPU seconds): 3 minutes and 10 seconds
Errors/Warnings

Results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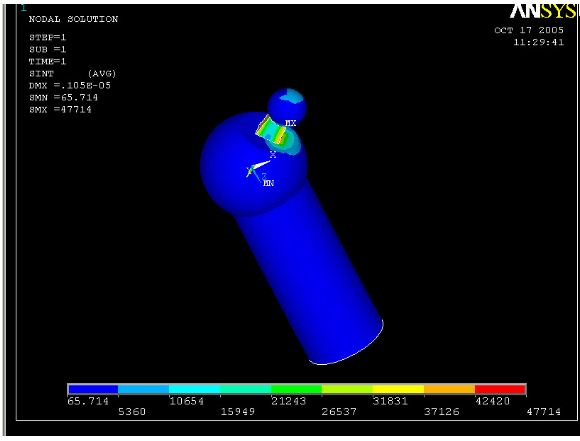
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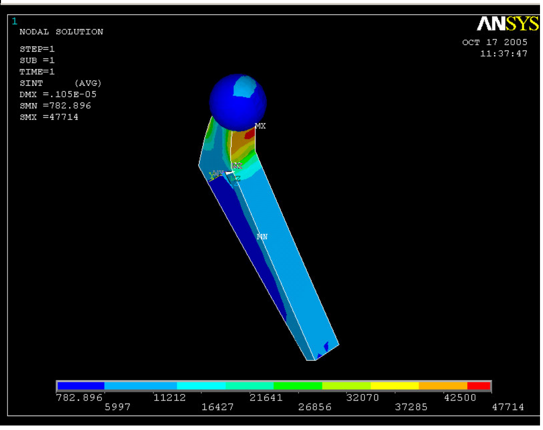
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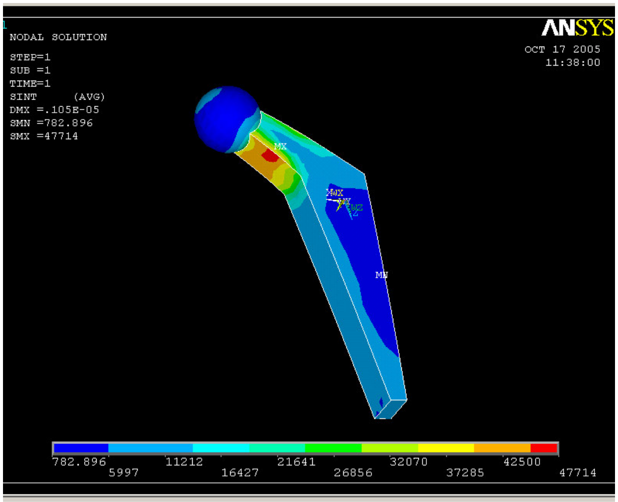
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Action to be taken: **Compare with first analysis**		
	Signed: Vivienne French	**Date: 17/10/05**

FEA Model Report

Job Name: Simplified cemented hip - smaller load

Job Number: 3

Analyst: Vivienne French

Problem Description

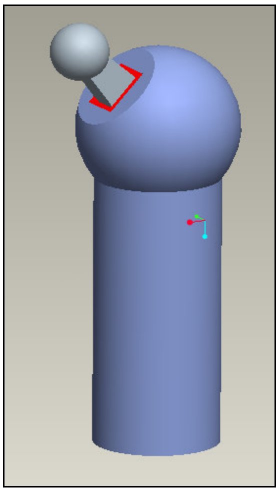
Related Drawings:

This analysis concentrates on a FEA of a simplified model of the cemented femoral system. The load on the system will act on the femoral head and will be counter-acted by a constraint in all degrees of freedom on the bottom are of the femur (where it appears to be chopped off). There will be contact pairs created to replicate the bond for bone and the cement and for the cement and the implant. Both contact pairs will have a COF of 0.95 (string bond). The properties of the contact pair between the bone and the cement will have the material properties for the bone. The contact pair between the cement and the implant will have properties for that of the cement. All volumes will have there separate material properties (structural, linear, elastic, isotropic):

Implant (Co-Cr-Mo):	Bone:	Cement:
E=210GPa	E=16.2Gpa	E=3.6GPa
V=0.3	v=0.36	V=0.3

Meshing the volumes will be done using the solid brick elements through a free mesh. The results obtained should show the stress intensity throughout the system and the contact pressure at all interfaces.

The loading in the femoral head will be 10N on 26 separate nodes.



Approach to Analysis

Analysis Type: Structural

Package Used: Ansys v9

Elements Used:

SOLID185

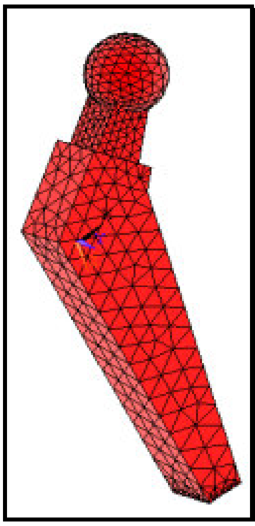
TARGE170   CONTA174

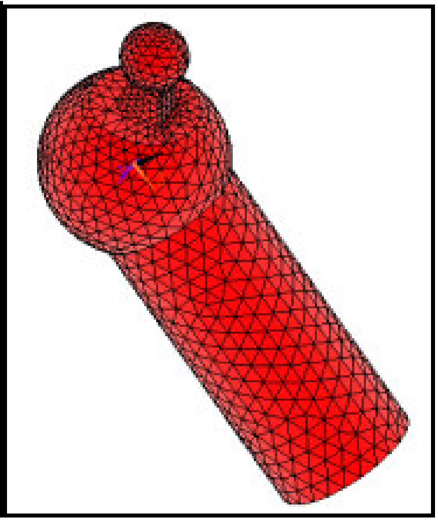
Modelling Assumptions:

- The system is constrained at the base of the bone, where it has been cut off in the solid model
- The bone is isotropic (it is actually anisotropic)
- The load is spread over 26 nodes on the femoral head

Basic Model (including mesh generation):

The mesh for all components are as follows:





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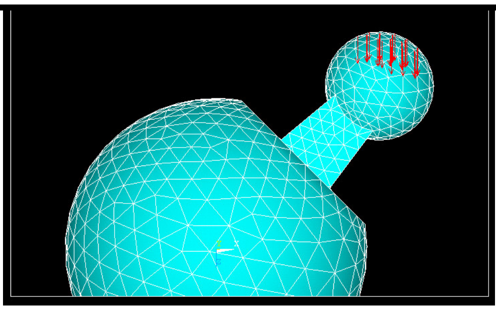


Default Freedom Conditions: ______UX, UY, UZ_____

(e.g. UX,UY for 2D)

Model Summary

Number of Nodes: 6606

Number of Elements: 33609

Number of DOF: 19818

Maximum Half-Bandwidth (or Wavefront):

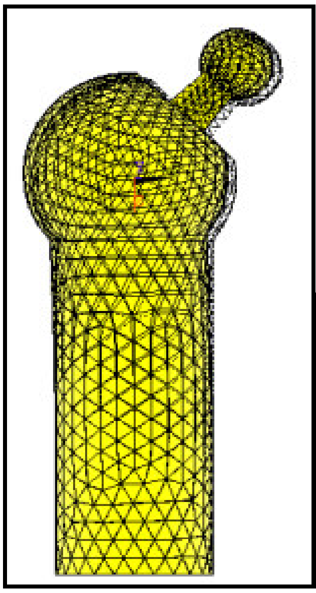
RAM Required: 0.304 Mb

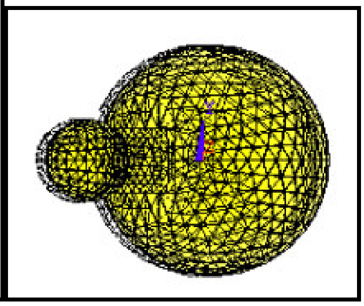
Disk Required: 19.598 Mb

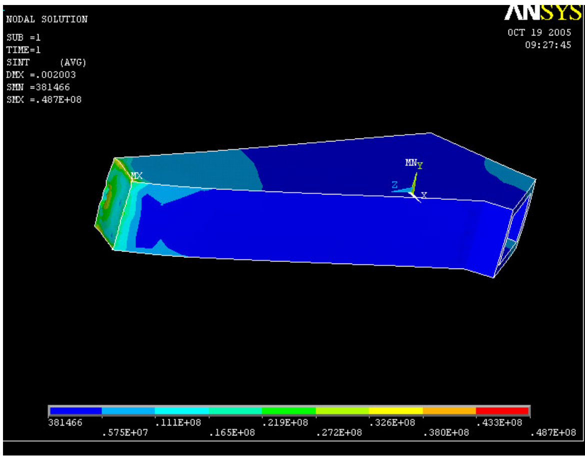
Solution Time (CPU seconds): 5 minutes

Errors/Warnings

Results



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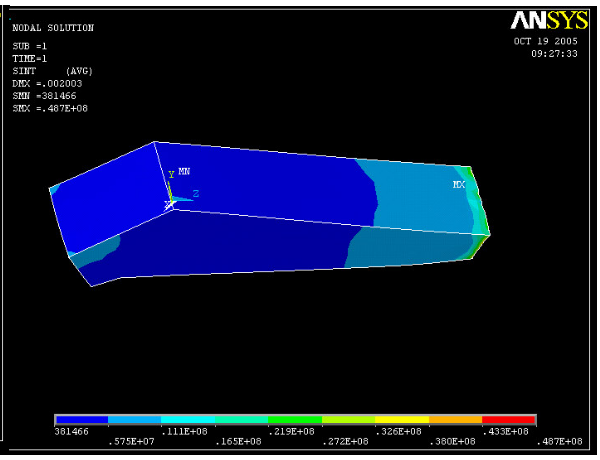
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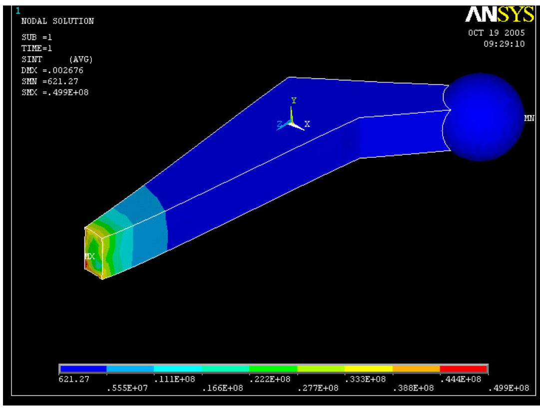
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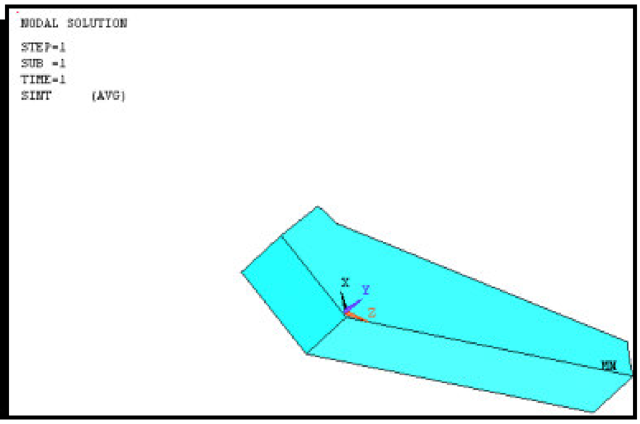
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Compare with next analysis</td></tr>
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Vivienne French</td><td>Date: 19/10/05</td></tr>
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FEA Model Report

Job Name: Simplified cemented hip – larger load

Job Number: 4

Analyst: Vivienne French

Problem Description

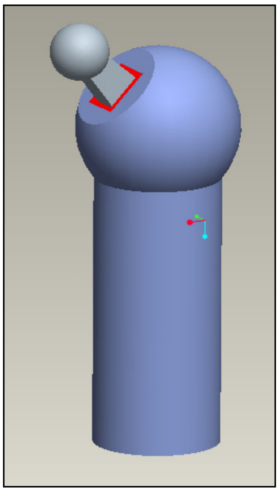
Related Drawings:

This analysis concentrates on a FEA of a simplified model of the cemented femoral system. The load on the system will act on the femoral head and will be counter-acted by a constraint in all degrees of freedom on the bottom are of the femur (where it appears to be chopped off). There will be contact pairs created to replicate the bond for bone and the cement and for the cement and the implant. Both contact pairs will have a COF of 0.95 (string bond). The properties of the contact pair between the bone and the cement will have the material properties for the bone. The contact pair between the cement and the implant will have properties for that of the cement. All volumes will have there separate material properties (structural, linear, elastic, isotropic):

Implant (Co-Cr-Mo):	Bone:	Cement:
E=210GPa	E=16.2Gpa	E=3.6GPa
V=0.3	v=0.36	V=0.3

Meshing the volumes will be done using the solid brick elements through a free mesh. The results obtained should show the stress intensity throughout the system and the contact pressure at all interfaces.

The loading in the femoral head will be 20N on 26 separate nodes.



Approach to Analysis

Analysis Type: Structural

Package Used: Ansys v9

Elements Used:

SOLID185

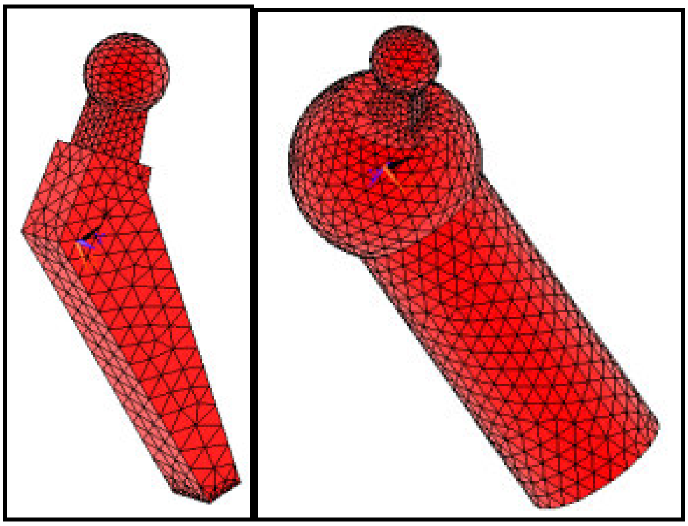
TARGE170   CONTA174

Modelling Assumptions:

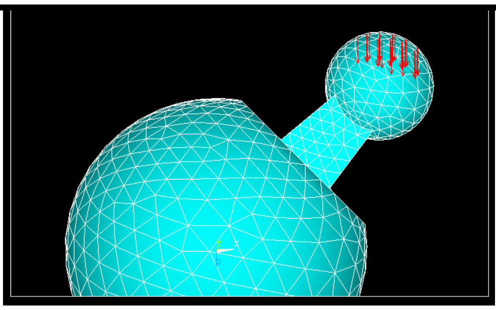
- The system is constrained at the base of the bone, where it has been cut off in the solid model
- The bone is isotropic (it is actually anisotropic)
- The load is spread over 26 nodes on the femoral head

Basic Model (including mesh generation):

The mesh for all components are as follows:



The loading on the system was as follows:



Default Freedom Conditions: ______UX, UY, UZ_____

(e.g. UX,UY for 2D)

Model Summary
Number of Nodes: 6606
Number of Elements: 33740
Number of DOF: 19818
Maximum Half-Bandwidth (or Wavefront):
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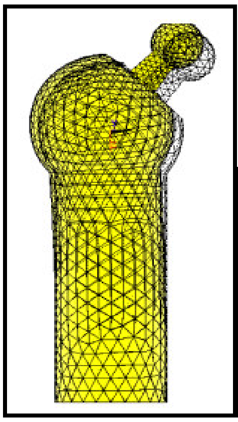
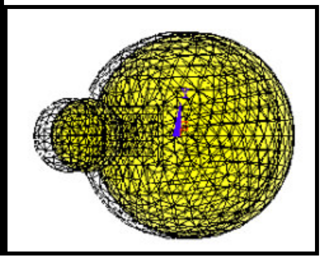
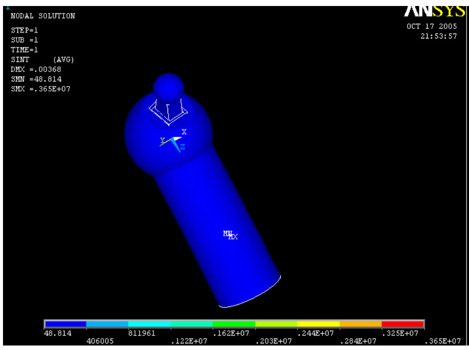
Disk Required: 18.764 Mb
Solution Time (CPU seconds): 3 minutes
Errors/Warnings

Results



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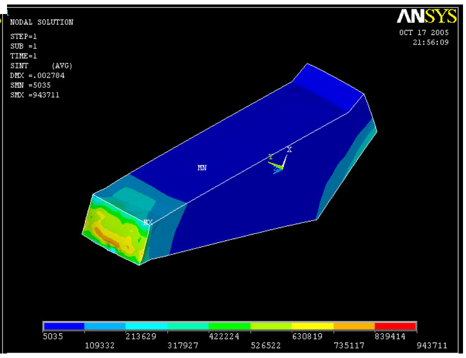
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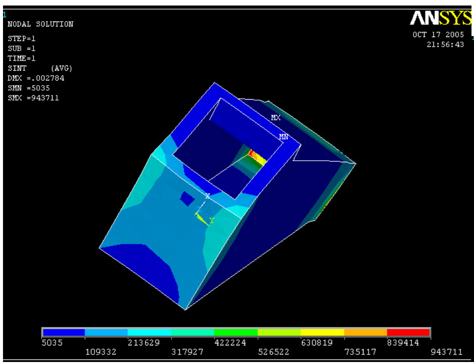
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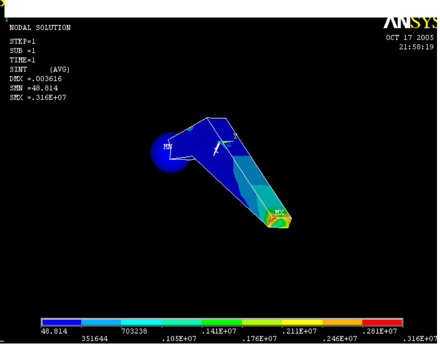
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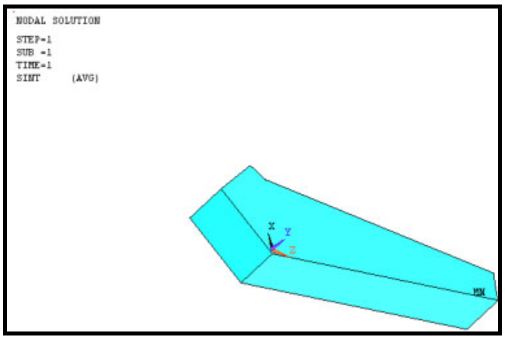
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Action to be taken: **Compare to 1st analysis for the cemented hip (analysis 3)**		
	Signed: Vivienne French	**Date: 17/10/05**

FEA Model Report

Job Name: Simplified cemented hip – smaller load, extra constraints

Job Number: 5

Analyst: Vivienne French

Problem Description

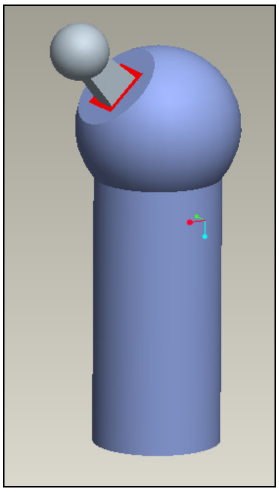
Related Drawings:

This analysis concentrates on a FEA of a simplified model of the cemented femoral system. The load on the system will act on the femoral head and will be counter-acted by a constraint in all degrees of freedom on the bottom are of the femur (where it appears to be chopped off) and an extra constraint on the femoral head. There will be contact pairs created to replicate the bond for bone and the cement and for the cement and the implant. Both contact pairs will have a COF of 0.95 (string bond). The properties of the contact pair between the bone and the cement will have the material properties for the bone. The contact pair between the cement and the implant will have properties for that of the cement. All volumes will have there separate material properties (structural, linear, elastic, isotropic):

Implant (Co-Cr-Mo):	Bone:	Cement:
E=210GPa	E=16.2Gpa	E=3.6GPa
V=0.3	v=0.36	V=0.3

Meshing the volumes will be done using the solid brick elements through a free mesh. The results obtained should show the stress intensity throughout the system and the contact pressure at all interfaces.

The loading in the femoral head will be 10N on 26 separate nodes.



Approach to Analysis

Analysis Type: Structural

Package Used: Ansys v9

Elements Used:

SOLID185

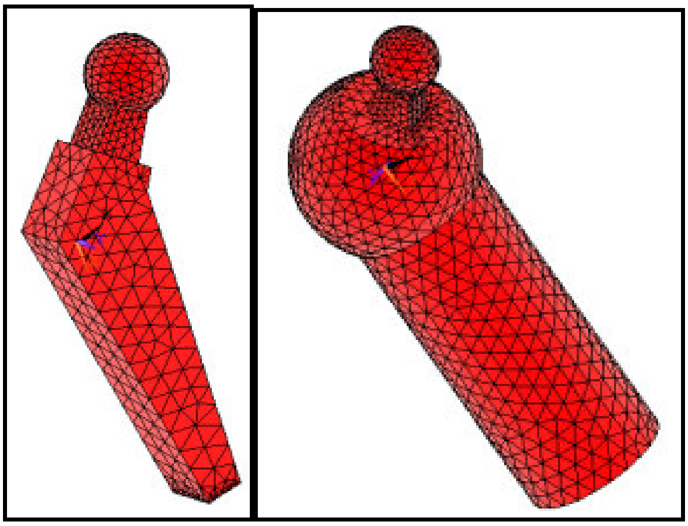
TARGE170   CONTA174

Modelling Assumptions:

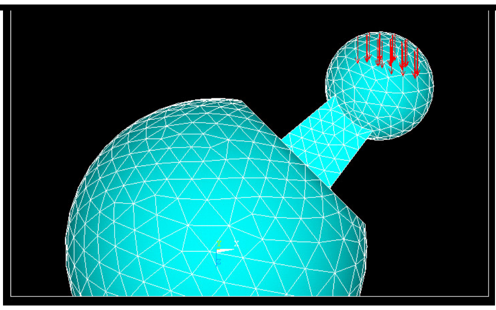
- The system is constrained at the base of the bone, where it has been cut off in the solid model
- The bone is isotropic (it is actually anisotropic)
- The load is spread over 26 nodes on the femoral head

Basic Model (including mesh generation):

The mesh for all components are as follows:



The loading on the system was as follows:



Default Freedom Conditions: ______UX, UY, UZ_____

(e.g. UX,UY for 2D)

Model Summary

Number of Nodes: 6606

Number of Elements: 33740

Number of DOF: 19818

Maximum Half-Bandwidth (or Wavefront):

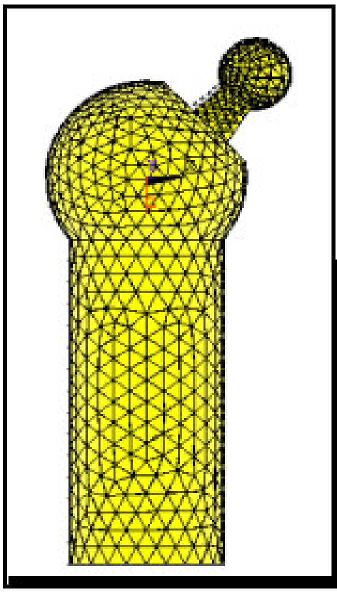
RAM Required: 8.310 Mb

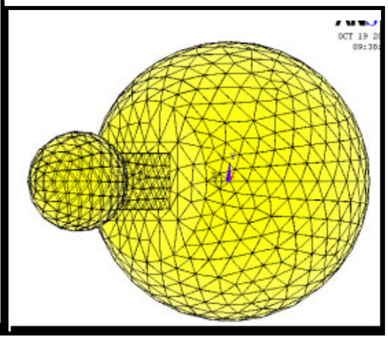
Disk Required: 100.027 Mb

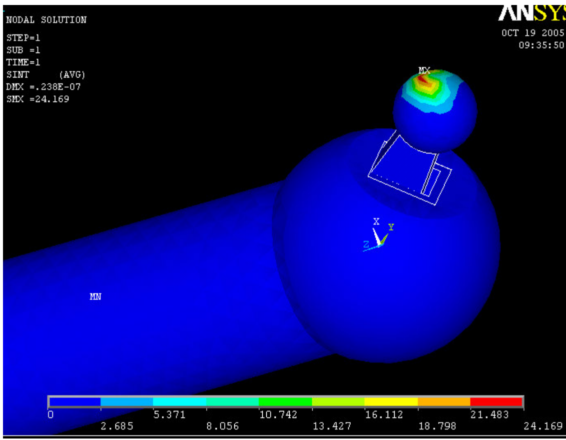
Solution Time (CPU seconds): 3 minutes

Errors/Warnings

Results



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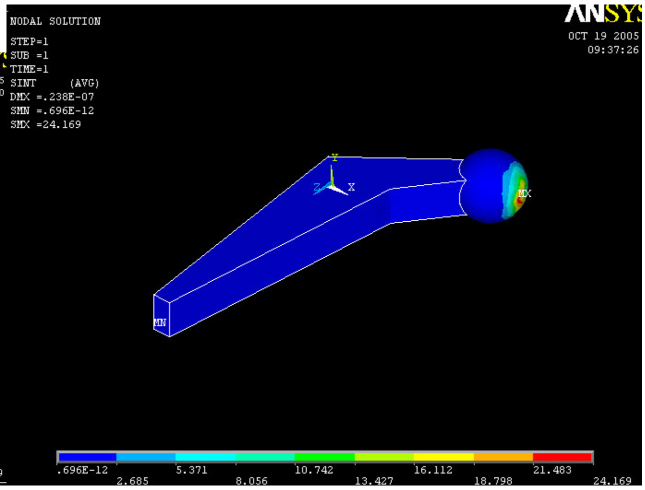
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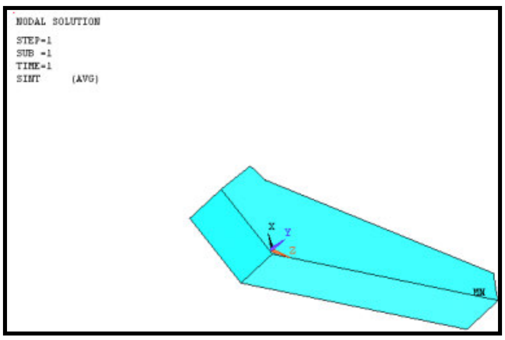
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Compare with 1st analysis for the cemented hip (analysis 3)</td></tr>
<tr><td></td><td>Signed:
Vivienne French</td><td>Date: 19/10/05</td></tr>
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