FINITE ELEMENT SIMULATION OF THE HIP JOINT

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Abstract: Finite element analysis is an established method to assist in the design, materials selection and analysis of the products subjected to different loading conditions before proceeding to the manufacturing stage. It is possible to simulate the joint implant and predict the failure scenario which could experienced in the clinical practice. This paper presents the process of analysing an artificial hip joint subjected to realistic loading conditions, describing materials definitions of bone and the prosthesis and explains the implementation of boundary conditions by applying forces including body weight and muscle load magnitudes. It also identifies instances when improper material selection and loading conditions can lead to inaccurate results.

1. Introduction.

The total number of hip procedures in the UK during 2008 is 71,367, an increase of 3.6% over 2007. Of these, 64,722 are primary and 6,581 (9%) are revision procedures. Indications for surgery for single stage hip revision procedures in 2008 in terms of percentage reported as Aseptic loosening 60%, Lysis 18%, Pain 27%, and Infection 3%. The average age of patients is 66.7 years. Approximately 60% of the patients are female. On average, female patients are older than male patients at the time of their primary hip replacement (68.4 years and 65.8 years respectively) (NJR, 2008).
Using Finite element analysis method it is possible to evaluate and optimise the design of hip joint replacement implant by minimising the weaknesses and stress concentration points so that fewer complications would occur after the operation.

2. Design of 3D Models.

An artificial hip joint consists of two main parts:
1- Femoral stem & Head.
2- Acetabular cup & Liner
In designing the femoral stem there are many points to be considered. The important parameters in the stem design include head diameter, neck diameter, neck length, neck angle, head/ neck ratio, stem length, offset (Figure 1).
The standard femur bone has been used for the FE analysis of hip prosthesis. All curves and details of femur bone are considered including greater and lesser trochanter, head and neck of femur. Femur exhibits a noticeable bow in the anterior–posterior plane.
In modelling the femur-implant joint similar assumptions to those in real surgical process are considered, i.e. the head of femur is first removed, the hollow interior of bone is reamed out and then the prosthesis that is uncoloured and is appropriately designed is placed inside femur.

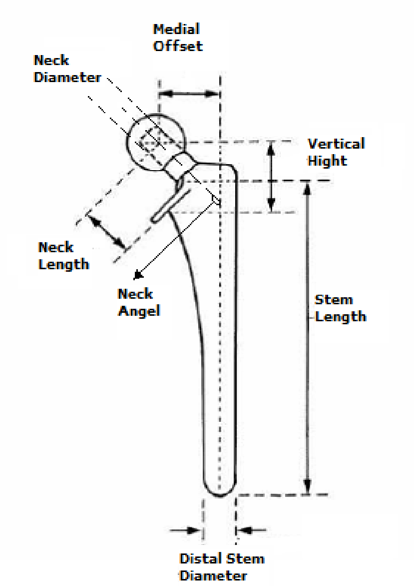

Figure 1. Schematic diagram of a Femoral Stem

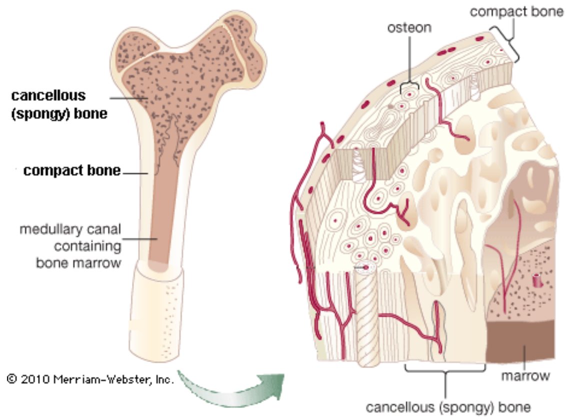

Figure2. Bone tissue consists of two main types:

1-compact bone 2- cancellous bone

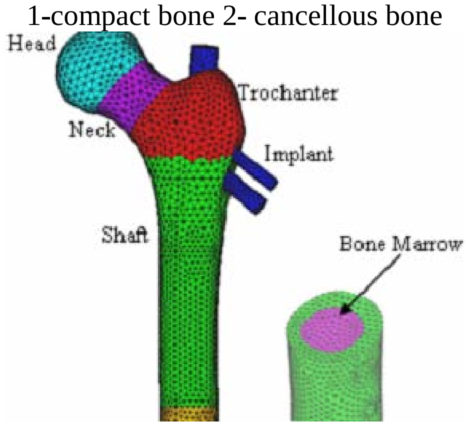

Figure3. Different sections of femur

3. Material Properties.

3.1 Bone Material.

The hip joint consists of two main bones. The femur and pelvis connect together to form the hip joint. The hip joint is a ball and socket joint that helps support the body mass as well as facilitating its movement in many directions.

There are two types of bone tissue: 1-cortical bone or compact bone. 2- cancellous or spongy bone. Cortical bone is denser, harder and stiffer than cancellous bone and it forms an outer shell of bone which supports the whole body. About 80% of the human body weight is attributed to cortical bone. The functional unit of cortical bone is osteon. Compared to cortical bone, cancellous bone is less dense and highly vascular that contains bone marrow where the blood cells produced. It naturally occurs at ends of long bones. The functional unit of cancellous bone is trabecula.

To assign the material properties to the bone, structure of the bone should also be considered. Cortical and cancellous bone has specific mechanical properties. Sowmianarayanan et all, (2006) assigned the material properties to femur as shown in table 1. Items of table consist of the cortical bone in the femoral shaft, cancellous bone in the femoral head, the femoral neck and the trochanteric region. Frictional coefficient of 0.20 is assigned to all the contact elements. Many authors have worked on FEA of femur bone such as Pyburn and Goswani (2004), Nunno and Amabili (2002), Latham and Goswani (2004) and Katoozian and Davy

(2000). The materials properties presented in these papers for the cortical bone eg Poission’s ratio have similar magnitudes while some differences are noticed in material properties of cancellous bone.

Table 1. Material properties of femur

3.2 Prosthesis and cement material.

Author	Material	Young modulus MPa	Poisson’s ratio
Nunno (2002)	Ti–6Al–4V	110,000	0.3
Pyburn (2004)	316L S.S (wrought)	200,000	-
Latham (2004)	Ti6A14V	113,800	-
Katoozian(2000)	cobalt–chromium	200,000	0.3
El’Sheikh(2003)	Ti6Al4V	100,000	0.32

Table2. Material properties of hip prosthesis

The hip joint prosthesis can be of different materials. However hip joint prosthesis is generally produced from some common materials such as cobalt chrome, stainless steel and Titanium alloy. In contrast with cementless hip joint operations, for cement kind of operation, surgeons make use of some sort of adhesives called cement. Prosthesis and cement materials are listed below in two different tables generally according to a number of papers. Overall there are quite the same materials used as cement bone.

Author	Material	Young modulus, MPa	Poisson’s ratio
Nunno (2002)	PMMA mantel	2700	0.35
Pyburn (2004)	PMMA bone cement	2000	-
Latham (2004)	PMMA bone cement	2000	-
Katoozian(2000)	Poly(methyl methacrylate)	2000	0.3
El’Sheikh(2003)	Rostal bone cement (PMMA)	2640	0.4

Table3. Material properties of cement

SI.No	Item	Elastic constant (E), MPa	Poission’s Ratio
1	Head	900	0.29
2	Neck	620	0.29
3	Shaft	17000-14000	0.29
4	Bone Marrow	100	0.29
5	Trochanter	260	0.29

4. Loading and boundary conditions

4.1 Resultant Force

Bergmann et al. (2001) presented a brief calculation of the mechanical loading and function of the hip joint and proximal femur. The average person loaded their hip joint with maximum 238% BW (percent of body weight) when walking at about 4 km/h and with slightly less when standing on one leg. When climbing upstairs the joint contact force recorded 251% BW which is less than 260% BW when going downstairs. Inwards torsion of the implant is probably critical for the stem fixation. On average it was 23% larger when going upstairs than during normal level walking. The inter- and intra-individual variations during stair climbing were large and the highest torque values are 83% larger than during normal walking.

A typical coordinate system for measured hip contact forces is shown in Figure 4. The hip contact force vector $-F$ and its components $-F_x$, $-F_y$, $-F_z$ acts from the pelvis to the implant head and is measured in the femur coordinate system x, y, z.

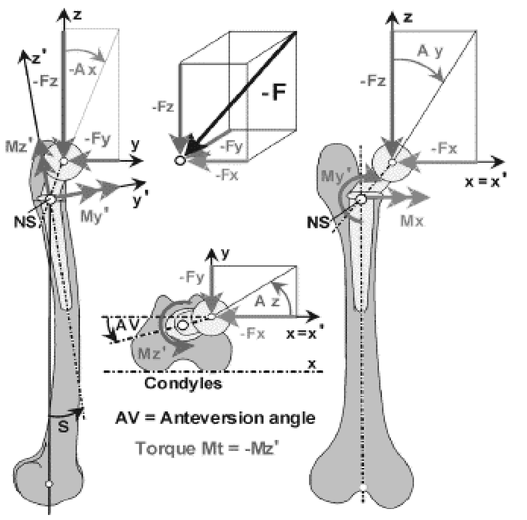

Figure4. Coordinate System at Left Femur (Bergmann et al., 2001)

The magnitude of contact force is denoted as *F* in the text. The axis *z* is parallel to the idealized midline of the femur; *x* is parallel to the dorsal contour of the femoral condyles in the transverse plane. The contact force causes a moment *M* with the components M_x, $M_{y'}$ and $M_{z'}=-M_t$ at the point NS of the implant. A positive torsional moment M_t rotates the implant head inwards. *M* is calculated in the implant system x, y', z'. Both systems deviate by the angle *S*. AV is the anteversion angle of the implant (Bergmann et al., 2001).

One of the major factors to be considered is the loading condition. Some type of loads may have a more significant effect on the design. Biegler et al. (1995) developed a brief FE analysis and calculation of two designs of hip prostheses in one-legged stance and stair climbing configurations. It is shown that torsional loads such as occur during stair climbing contribute to larger amounts of implant micromotion than stance loading does. Contact at the bone-prosthesis interface is more dependent on load type than on implant geometry or surface coating type.

Generally there are various loading conditions calculated and presented in forms of different diagrams based on common real life activities such as slow walking, normal walking, fast walking, up stairs, down stairs, standing up, sitting down, standing on 2-1-2 legs and finally knee bend condition which is shown below in figure 5. Similar diagrams are introduced for moment M.

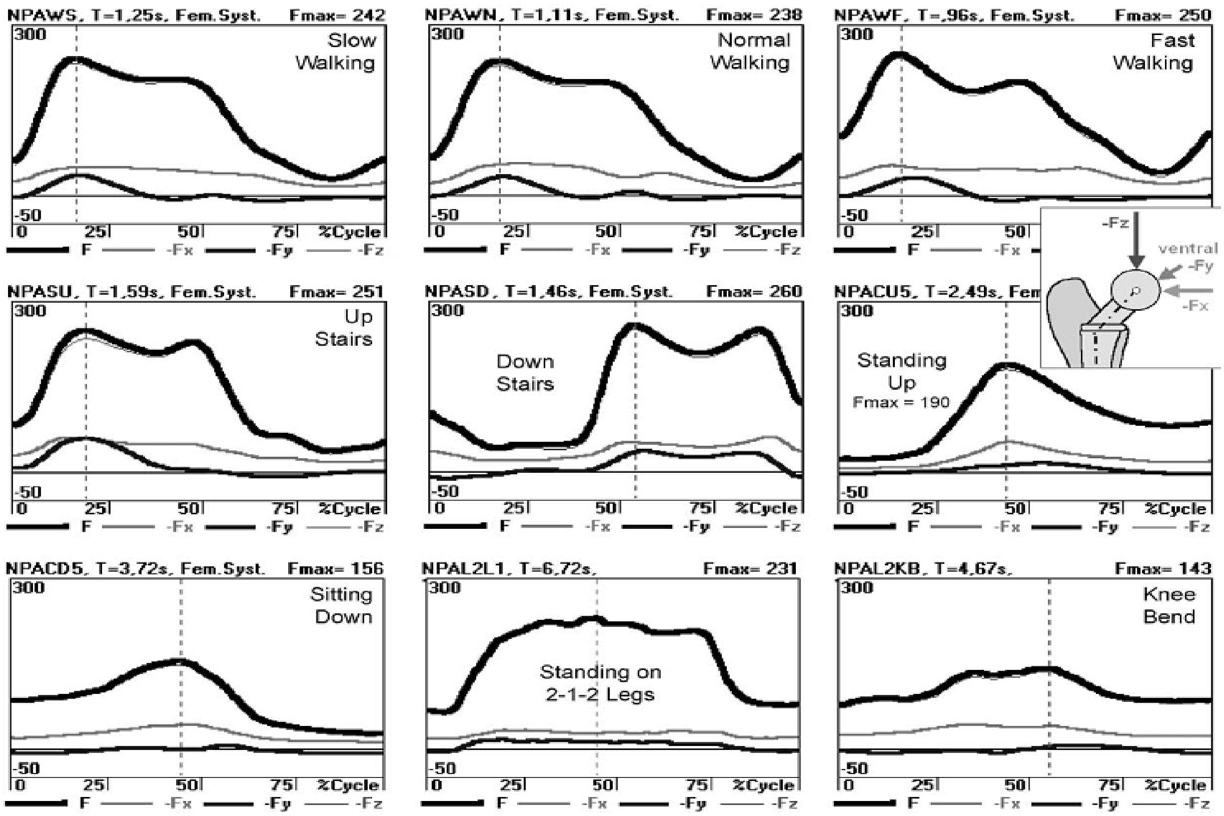

Fig 5. Contact force F of typical patient NPA during nine activities. Contact force F and its components $-F_x$, $-F_y$, $-F_z$. F and $-F_z$ are nearly identical. The scale range is 50–300% BW. Cycle duration and peak force $F_p = F_{max}$ is indicated in diagrams. Bergmann et al. (2001)

4.2 Muscle forces

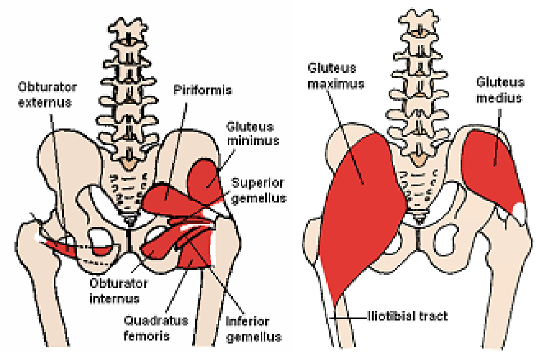

Figure6. The involved muscles with femur: Gluteus medius & Gluteus minimus, ilio-tibial band (Gluteus maximus & tensor fascia latae).

Apart from resultant force applied on the prosthesis, there are few muscles attached to femur that induce extra tension on bone.

At 85% of the gait cycle, a simplified set of active muscles are the abductor muscles, located on the greater trochanter (Gluteus medius and Gluteus minimus), and the ilio-tibial band (Gluteus maximus and tensor fascia latae). El'Sheikh et al (2003). The relative forces are listed in the table4 and shown in figure 7.

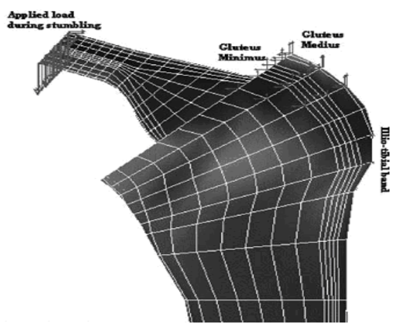

Figure7. Position of applied forces

Component Force (N)	Gluteus Medius	Gluteus Minimus	Ilio-tibial band
F_x	-259	-279	-59
F_y	160	269	-74
F_z	319	134	-58

Table 4. Muscles-forces applied on the femur

Furthermore regarding the muscle forces applied on femur, according to Sowmianarayanan (2006) who also work on finite element analysis of proximal femur nail, the distal end of the femur model, was fully fixed. The various loads due to body weight and various muscles at proximal femur corresponding to Simoes et al. (2000) were considered for the analysis. The applied loads consist of joint reaction force, abductor force, Iliopsoas force and vastas lateral as shown in the table 5 and figure 8.

Generally if any designing steps like: 3d modelling, material selection, boundary conditions or applied forces are not considered properly we will come up with a wrong result. For instance Mathias (1998) has not considered a correct boundary condition for the hip joint prosthesis.

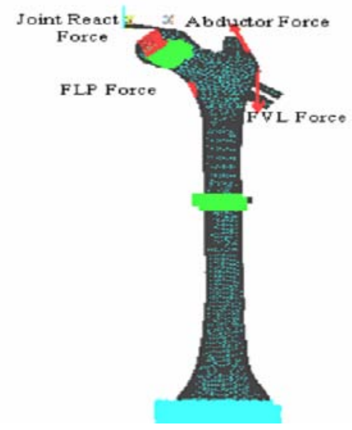

Figure 8. FE model of femur with PFN implant- loads and boundary conditions

SI. NO.	Type of Load	Force N
1	JRF	730
2	Abductors	300
3	Iliopsoas,FVL	188
4	Vastas Laterals, FLP	292

Table 5.Various forces applied on the femur

5. Design optimisation of hip joints.

One may question the reliability of FEA (finite element analysis). In this regard, Stolk et al. (2002) have corroborated that Finite element and experimental models of cemented hip joint reconstructions can produce similar bone and cement strains in pre-clinical tests. They have compared the results of FEA and experimental models. The objective of overall agreement within 10% was achieved, indicating that FE models were successfully validated. Hence the prerequisite for accurately predicting long-term failure has been satisfied.

Many designs have been developed to improve stress, strain, wear and fatigue life of hip implants. To design prosthesis of higher durability the natural processes occurring in bone has to be taken into consideration. Pawlikowski et al. (2003) designed hip joint prosthesis through the acquisition of different steps of CT data, Geometrical modeling of femur, prosthesis design and the numerical analyses of the bone-implant systems helps to finally decide which one of the three designed prostheses is the most appropriate for the patient. Latham and Goswami, (2004) studied the effect of geometric parameters on the development of stress in hip implants.

The parameters include: head diameter, neck diameter, and neck angle. In particular it is shown that as the head diameter increases, the stress at a given location reduces. However, as the surface area from increased head diameter increases, the wear rate also increases. Darwish and Al-Samhan (2009) conducted a parametric study that comprises the parameters affecting the strength of hip-joint cement fixation (offset distance and ball diameter). They recommend offset distance (3-6 mm) and ball sizes (34 and 50 mm) for maximum cement strength. Matsoukas and Kim (2009) performed the design optimisation of a total hip prosthesis for wear reduction. The accumulation of wear debris can lead to osteolysis and the degradation of bone surrounding the implant components. Bennett and Goswami (2008) carried out CAD FEA on six hip stem designs to come up with a hip stem that has a low stress, displacement and wear at a very high fatigue life.

On the effect of different factors on design optimisation, Nicolella et al.**(**2005) investigated the effect of three-dimensional prosthesis shape optimisation on the probabilistic response and failure probability of a cemented hip prosthesis system. It is shown that probability sensitivity factors indicate that the uncertainty in the joint loading, cement strength, and implant–Cement interface strength have the greatest effect on the computed probability of failure.

The main aim of this project is to develop optimum artificial hip joints with new/ improved design features which can address the following requirements:

- To prevent the risk of dislocation in the hip joints
- To be more resistant to damage and failure by suitably adjusting the strength and stiffness in the implant
- To include design features to make it easier for the surgeons to adjust/ tailor make the implant- more surgeon friendly design
- The improved design should potentially remove the risk of further painful experience, by presenting a completely new design of hip joint.

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