INTRODUCTION

A 3D finite element model has been developed to study the mechanics of total knee replacements (TKRs). In particular, this study looked at the effects of certain design parameters on the stresses found in the meniscal insert component of a TKR. A primary reason for the failure of TKRs is failure of the meniscal insert. The meniscal insert is made from Ultra High Molecular Weight Polyethylene (UHMWPE) and the femoral component is commonly made from a Cobalt Chromium alloy. Failure therefore tends to occur to the meniscal insert because the femoral component is much stiffer, in the range of 100-200 times, and acts as a rigid indentor (Bartel 1985) as it articulates, rolls and slides with the meniscal insert under cyclic compressive forces.

The most significant and detrimental failure that occurs to the meniscal insert is delamination. Delamination is a process where micro cracks are formed under the surface, which propagate under the cyclic loads leading to sheets of the material laminating away (Walker 1993). If this type of failure occurs within the first four years of implantation then it is most likely due to imperfections in the material from the manufacturing processes. If the failure occurs after this time then it is generally a result of fatigue (Blunn 1997). To increase the longevity of a TKR the stresses that cause delamination must be reduced.

There are four types of stress that play a significant role in the initiation, propagation and eventual failure in the UHMWPE. Equivalent stresses and contact stresses initiate the micro cracks below and at the surfaces respectively (Blunn 1991). The tensile stresses and shear stresses then assist in the propagation of these cracks (Bartel 1986). By using Finite Element Analysis it is possible to see where these stresses are occurring and their extent.

Finite element analysis (FEA) is a powerful tool that can be used to assist in the design of TKRs. Two methods of analysis were used in this study, static and pseudo-dynamic, depending on the parameter being studied. Up until recently, most finite element modelling of TKRs has been statically analysed (Bartel 1995, Mottershead 1996). Although this is valid for looking at simple parameters such as meniscal thickness it is very limited at exploring more complex concerns such as misalignment, which can also influence the kinematics of the TKR. For these parameters a pseudo-dynamic model based on the walking gait cycle was used.

METHODS

A 3D, non-linear, finite element model was developed, solved and analysed with in the finite element software ANSYS 7.0. The geometry of the TKR used was based on the regular sized prosthesis (size 3) in the Active Knee System by ASDM, Australia. The geometry of the meniscal insert component was simplified for easier meshing within ANSYS, however the integrity of the design for the purpose of this study was not sacrificed. It was map-meshed using 20 node hexahedral elements with multi-linear elastic material properties for GUR 1050 UHMWPE. The femoral component was simplified by only modelling the outer contacting surface and was meshed as a rigid surface. This was considered to be adequate as the stiffness of chromium cobalt alloy is significantly greater than that of UHMWPE. Contact was modelled between the two articulating surfaces using the ANSYS contact capabilities. The inferior surface of the meniscal component was modelled to be rigid and controlled by a master node on the surface. A pilot node similarly controlled the femoral component surface. Figure 1 shows a comparison between the real TKR and the finite element model.

Two different types of analysis were used to look at the effect of different parameters. Firstly, a static analysis was used to look at the influence of the meniscal insert thickness and patient body weight on the stresses in the meniscal insert. The femoral and meniscal components were aligned in the designed neutral position at 0 Degrees flexion, stance position. The meniscal insert was constrained and the femoral component was loaded through the pilot node. The model was loaded with an axial force of 3 times the body weight of an 80kg person, the approximate mass of an 180cm tall adult male (Diem 1970).

Four different sizes of minimal meniscal thickness were analysed; 4mm, 6mm, 9.5mm, and 13.5mm. The three larger sizes represented the minimum, middle, and maximum thicknesses of the Active Knee System. A thickness of 4mm was chosen to demonstrate the effects of smaller thicknesses. The 6mm meniscal component was then used for all other analyses. A range of masses were analysed from 80kg to 150kg at increments of 10kg.

A pseudo-dynamic model was then used to analyse the effects of friction, cross-sizing and surgical misalignment. The model set-up was based on the specifications for wear
Simulating machines found in the ISO draft standard for wear testing of total knee joint prostheses with load control, ISO 14243-1.4. Four loading curves, controlling flexion-extension angle, axial force, anterior-posterior force, and medial-lateral rotation torque, were then applied to the model.

The rotational displacement was applied to the pilot node of the femoral component to control the flexion-extension angle of the knee joint. The femoral component was fully constrained in all other degrees of freedom.

The axial force was applied to the master node on the inferior surface of the meniscal insert. The master node was at an offset equivalent to 0.07*width of the meniscal insert medially from the centre line. This offset was to allow for the unequal loading between the medial and lateral condyles found in the normal knee joint.

The anterior-posterior force was also applied to the master node with positive and negative values representing the forces in the anterior and posterior directions respectively. The internal-external moment was again applied to the master node with positive and negative values representing the moments in the medial and lateral directions respectively.

No force was applied to represent the medial-lateral translation of the knee joint as it was considered sufficiently small that it could be neglected (Walker 1997), however the meniscal component was left unconstrained for this degree of freedom. The meniscal component was also unconstrained in the varus-valgus rotation degree of freedom to allow for the effect of the axial force being applied slightly more to the medial side. It was however, constrained so as not to allow rotation of flexion-extension as this degree of freedom has been accounted for by the femoral component.

RESULTS AND DISCUSSION

Varying the meniscal insert thickness produced expected results. As the meniscal insert thickness is increased the control the anterior-posterior translation with one end attached to the master node of the meniscal component, the other end constrained in the AP direction and with a spring constant of 30N/mm. A torsional spring was used to control the medial-lateral rotation. Similarly, one end was attached to the master node, the other constrained for ML rotation, and with a spring constant of 0.6Nm/degree. Figure 2 is a representation of the model set up for the pseudo-dynamic analysis.

An average walking gait cycle, described in the ISO draft standard, was applied to the model over 1 second. This data was based on experimental work that combines ground reaction force plate data and internal muscle forces (Morrison 1970). The graphs representing these input values are in Figure 3.

The published values for the coefficient of friction of cobalt chromium alloy on UHMWPE vary greatly. For this study three values for the coefficient of friction were compared; 0.01, 0.1, and 0.16, representing the variation in published values (McKellop 1981). Cross sizing is a feature available in the Active Knee System range. It offers the option to cross a smaller meniscal insert with a larger femoral component to allow for patients with varying bone morphology (Palmer 2001). The effect of crossing a regular sized (size 3) meniscal component with a large (size 4) femoral component was analysed. Surgical misalignment is a realistic concern that can occur during the implant operation. For this study we looked at the influence of a 5 Degree surgical misalignment and compared this to a correctly positioned prosthesis.

The maximum equivalent stress (Von Mises Criterion) was used as a comparison indicator for all analyses. This stress criterion was chosen because it is an indicator for where the delamination process begins.
equivalent stress decreased, as shown in Figure 4. Essentially, as the component becomes thinner the material becomes stiffer producing an increase in stress. The stress found in the 4mm meniscal component was significantly higher, which suggested that thicknesses below 6mm may considerably decrease the longevity of a component. The concern with increasing meniscal insert thickness excessively is maintaining anatomically correct joint space. This has been achieved by sacrificing proximal tibial bone. Too much bone loss is undesirable; therefore there must be a balance between meniscal thickness, joint space and bone loss to achieve the best possible outcome for the patient.

Figure 4: The influence of varying meniscal component thickness

As the lowest minimum meniscal thickness found in the Active Knee System is 6mm and would consequently produce the highest stresses, this was the size used for the remaining analyses.

The second part of the static study looked at the effects of various body masses on the stress found in the meniscal insert. The results, found in figure 5a, showed that although there was a steady increase in stress up until 130kg, there was a significant rise in stress beyond this mass. These results raised a few questions. The first concern was whether this phenomenon would be more significant in the lower size range. A similar analysis was consequently conducted for the size one combination with masses varying from 60kg to 85kg in 5kg increments. The results showed that there was a small rise in stress after 60kg and then another significant rise after 75kg. These results show that patient body mass can have a considerable influence on the longevity of a TKR. There is a possible need here for more information to be made available to the surgeon about safe implantation environments, including increasing the thickness of the meniscal insert for patients who are considered over weight.

Figure 5: The effect of body mass for (a) Size 3 prosthesis (b) Size 1 prosthesis

The coefficient of friction values used in the pseudo-dynamic analysis were 0.03, 0.1, and 0.16. The results for each value over the gait cycle are shown in Figure 5. There is only a small amount of variation between the lowest and highest values. This suggested that for this range of values the coefficient of friction is not a significant parameter. For the remaining models a coefficient of friction of 0.1 was used.

Figure 5: The effect of varying the coefficient of friction
The effect of cross sizing a size 3 meniscal component with a size 4 femoral component was then analysed and compared with a normal size 3 combination. Figure 6 shows a small increase of 12% in the equivalent stress with a maximum stress of 10.5MPa. The increase was not considered to be significant and can confirm the safe and affective use of cross sizing.

The final parameter was the effect of a 5 Degree surgical misalignment. To do this the femoral component was rotated 5 degrees from its neutral position and the gait walking cycle applied. The results in Figure 7 show that there was a significant rise in stress in comparison to the correctly surgical aligned model. These high peaks of stress coincided with the initial full weight bearing and the toe-off points in the gait cycle. With such a significant rise in stress the longevity of the TKR would be greatly reduced. These results promoted the need for better implantation procedures to reduce the amount of surgical misalignment occurring.

**SUMMARY**

Through finite element modelling and analysis it has been possible to observe the effects of different variables and the extent of those effects on UHMWPE stresses. This information can lead to better design and increased longevity of a TKR. It was shown that the patient bodyweight could have a significant influence on the results of TKR surgery, leading to the need for better implant information for size selection. The most important result is the considerable increase of stress due to surgical misalignment, which highlights the need for manufacturers to develop accurate fixture and cutting guides for TKR implantation.

**REFERENCES**