In vitro Kinematic evaluation of Knee prosthesis

A. Pérez\textsuperscript{1}, M. Salmerón\textsuperscript{1}, X. Sancho\textsuperscript{1}, M. Comín\textsuperscript{2}, J. Prat\textsuperscript{2}, J.L. Peris\textsuperscript{2}

\textsuperscript{1}Área de Ingeniería Mecánica, Dpto. de Tecnologia, Universitat Jaume I, 12071 Castellón, Spain. 
\textsuperscript{2}Instituto de Biomecánica de Valencia, Parc Tecnològic, 46980 Paterna (Valencia), Spain

Introduction

Normal kinematics of the natural joint is expected to be restored when a knee joint replacement is performed. Different prosthesis models exist, with different features. In this work, the kinematic behaviour of several models is analysed by means of the measure of the rotations and translations of a fresh human knee. All the prostheses are implanted using the same surgical procedure to avoid the influence of the different position after the implant.

Material and Methods

A post-morten human left knee joint specimen with no pathology, resected approximately 15cm above and below from the joint center, was used to implant three different prosthesis models: posterior cruciate ligament retaining (PCLR), posterior stabilized (PS) and meniscal bearing (MB) models.

First the natural knee was mounted on a specially designed testing rig (Fig. 1). The apparatus maintained the femur in horizontal fixed position, allowing the tibia to rotate freely with respect to the femur about the knee. In the starting position the knee was put in full extension thanks to an extending load obtained from a cord-pulley-mass system actuating over a plate at the free end of the tibia. A small compressive and flexing load acting in the same plate through a cord is used to assure bone contact in the joint. Different equilibrium positions in the range of knee motion from extension to flexion were reached changing the compressive and flexing load and maintaining the extending load. A total of 9 positions from full extension to near 100° of flexion were measured.

![Figure 1: Testing rig](image)

The equilibrium positions were measured with an electromagnetic system (Fastrak, Polhemus Inc.). The system consists of a reference transmitter, two sensors and a digitising stylus (pen). The transmitter was fixed to the frame of the testing rig and the two sensors were screwed one to the femur and the other to the tibia. The Fastrak gives the position and orientation of the sensors axes and stylus axis with respect to those of the transmitter.

The magnitudes of the forces applied to the tibia were measured with loading cells and their directions were obtained from two measurements with the stylus over two points in the cords. Two direct coordinate systems, one for the tibia (T) an other for the femur (F), were defined to be coincident in full extension. The origin of these two coordinate systems in full extension was the mid
point between the two epicondyle protusions in the femur and their axis point toward the approximate anatomical directions: distal to proximal (x), posterior to anterior (y), medial to lateral (z).

For each of the 9 equilibrium positions the Fastrak recorded the positions and orientations of the sensors and the different stylus positions. From these data, the information of position and orientation of the tibia coordinate system (T) with respect to the femur one (F), and the magnitude and direction of the forces over the tibia could be obtained. The Euler angles between T (moving) and F (fixed) axes with the rotation order z-y-x, gives the flexion/extension, adduction/abduction and external/internal rotation angles, respectively. The relative position between the origin of T and that of F gives the medial/lateral, posterior/anterior and distal/proximal displacements.

In a second step, the natural knee was removed from the loading apparatus and the PCLR prosthesis model (fig. 2) was implanted in the knee following the surgical protocol given by the manufacturer. After the implant, the knee was mounted in the test rig by fixing the femur in the same way that for the natural knee. Then, the same protocol described before for the natural knee was repeated. This same procedure was repeated for the PS (fig. 3) and MB (fig. 4) prosthesis models.

**Results and Discussion**

The external net moment and force applied to the tibia were about 3Nm and 30 N, respectively, for all flexion angles, differing their directions less than 4 and 5 deg, respectively, among prostheses. The external moment had an adduction component but no significant component in the direction of internal/external rotation. The external net force was compressive, nearly normal to the bone contact with no significant component in the medial-lateral direction. No important differences were observed in the pattern of variation of the net force and moment with flexion angle among prostheses. From these results it can be assured that the differences in the equilibrium positions among prostheses are due to the geometry differences and not to the external forces applied.

The measured values for the internal/external (IE) rotation and the antero/posterior (AP) displacement of the tibia as a function of the flexion angle are shown in figures 5 and 6, respectively.
As it was expected, tibia rotates internally with flexion (Fig. 5). The measured values of internal rotation are in accordance with the data obtained by other authors (Kurosawa et al., 1985, Blankevoort et al., 1990, Feikes, 1999). A value ranging from 12° to 20° of internal rotation for 90° of flexion is obtained for the different prostheses. The rotation pattern for the three prostheses is similar and similar to that of the natural knee. However the rotation for the MB one is higher as it was expected from its design. The AP displacement has the same tendency for all the models. An anterior displacement is obtained for 90° of flexion ranging from 0.2 to 0.8 cm. Exact comparison between these data and those published by other authors can not be done because they depend on the chosen reference point. In our case the reference point is the center of the T coordinate system which is in the mid point between the epicondyles in full extension. This point moves anteriorly with flexion. The anterior displacement is lower for both the MB and PS models than for the natural knee and PCLR model. The central post introduced in both PS and MB models when the PCL is resected seems not to reproduce the natural knee behaviour correctly. Other result observed from the experiment was an increment in the adduction angle with the prostheses respect to that of the natural knee. For 90° of flexion an increment of 4° of adduction was measured for the PCLR and MB models and 2° for the PS model.

Conclusions

It can be concluded that
1. The internal rotation achieved with the PS prosthesis model and the posterior displacement of the PCLR model are quite similar to those of the natural knee.
2. The rotation of the MB model seems to be higher than that required for the normal knee kinematics, and the anterior displacement of both MB and PS seems to be lower.
3. The prostheses models increase the observed coupled adduction with flexion movement.
4. The changes performed on the basic model (PCLR) to get a valid design when PCL is resected (PS and MB prostheses) do not achieve the expected results.

References