Stability and Range of Motion of Insall-Burstein Condylar Prostheses

A Computer Simulation Study

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Abstract: The Insall-Burstein Posterior Stabilized Prosthesis (Zimmer, Warsaw, IN) uses an articulation between a femoral cam and tibial spine to provide anteroposterior stability to the knee. Dislocation can occur if the femoral cam translocates anteriorly and over the tibial spine. A computer model was used to examine the effects of design changes made between the Insall-Burstein I (IB I), Insall-Burstein II (IB II), and revised Insall-Burstein II (IB II^R) knees. The effects of these design changes were determined from their influence on knee stability and maximum obtainable knee flexion. Knee stability was characterized by a *dislocation safety factor,* defined as the vertical distance from the top of the tibial spine to the bottom of the femoral cam. Our analysis showed that the dislocation safety factor is greatest at approximately 70° of knee flexion for all IB knees. As knee flexion is increased from this angle, the dislocation safety factor decreases, reducing knee stability. The simulations highlighted a trade-off between improving knee flexion and improving knee stability. The geometry of the IB II knee allowed greater knee flexion. The maximum flexion achieved with the IB II knee was 125° compared with 115° and 117° for the IB I and IB II^R knees, respectively. However, the simulations indicate that the IB I and IB II^R knees are less likely to dislocate because they have greater dislocation safety factors than the IB II knees. **Key words:** knee, arthroplasty, dislocation, biomechanics, computer simulation, prosthetic design.

The Insall-Burstein Posterior Stabilized Prosthesis (Zimmer, Warsaw, IN) has provided excellent long-term knee arthroplasty results using the concept of posterior cruciate-ligament substitution.[1-11] The substitution mechanism depends on the interaction between the tibial spine and femoral cam. This spine-cam interaction substitutes for the posterior cruciate ligament by providing anteroposterior stability to the knee. Spine-cam interaction also causes the femur to roll posteriorly on the tibia, which increases maximum knee flexion. Although this basic principle of posterior ligament substitution has remained unchanged,[3,6,12-16] modifications to the design of these knees have been made. Design modifications have included changes in the component geometries, the addition of metal backing to the tibial component, and the advent of modular components.

Although these modifications have generally improved the results obtained with these knees, changes in component geometry may also produce unpredictable or unwanted effects. For instance, analysis of several clinical studies suggests that differences in prosthetic knee dislocation rates with

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the Insall-Burstein knees[17–22] may be caused, in part, by the differences in the geometries of the Insall-Burstein I (IB I), the Insall-Burstein II (IB II), and the revised Insall-Burstein II (IB II^R) knees. Dislocation of the posterior stabilized knee occurs when the femoral cam translocates anteriorly and over the tibial spine. This results in an acute dislocation of the prosthesis, with the patient's knee locked in a flexed position. In most instances, these dislocations cannot be reduced without formal sedation and treatment. Galinat et al. reported only 2 dislocations in 832 surgeries using the IB I knee.[19] Cohen et al. reported no dislocations in 105 IB I knee arthroplasties, but 5 dislocations occurred in 100 IB II knee arthroplasties.[18] Similarly, Striplin and Robinson did not observe any dislocations in their 428 IB I arthroplasties, yet reported 3 dislocations (in 2 knees) in 240 IB II knee arthroplasties.[22] Cohen and Constant reported similar problems, observing 5 dislocations (in 2 knees) in the 437 knees they replaced with the IB II knee.[17] A recent report from Lombardi et al. showed that the dislocation rate increased from 0.2 to 2.5% when the IB II was introduced.[21] Overall, the reported incidence of dislocation was greater with the IB II components than with the IB I components (Table 1). Because of this problem with the IB II knee, modifications to the spine geometry were made, which seem to have decreased dislocations.[21,23]

The purpose of this study was to examine how the design changes made to the IB knees affected the anteroposterior stability and maximum flexion of these knees. In this way, we hoped to gain a more thorough understanding of how component geometry influences dislocation of posterior-stabilized knees.

Materials and Methods

A computer graphics model of an IB knee that simulates motion in the sagittal plane was implemented on a Silicon Graphics (Mountain View, CA) workstation. The dimensions of the 59-mm IB II prosthesis were used to model the knee based on drawings provided by the manufacturer. The geometry of this IB II knee model was then altered to represent the IB I and IB II^R knees. Motion was simulated and the differences in knee stability and maximum obtainable knee flexion were analyzed.

To simulate the IB II knee, tibial wells were defined by a circle with a radius of 57 mm. The geometry of the tibial spine was defined by two parameters: spine height and anteroposterior placement (Fig. 1A). The spine height was 14.5 mm from the 7° tibial surface, and the spine was placed 20 mm anterior from the posterior tibial edge, based on dimensions provided by the manufacturer.

The distal and posterior surfaces of the femoral component were represented by two circles (Fig. 1B). The circles were joined such that the slopes of their tangents were equal at their point of intersection. Each circle was defined by its radius and the position of its center. The circle that defines the distal surface was given a radius of 51 mm, and its center was placed directly superior to the bottom of the tibial well. In the IB II knee model, the radius of the circle that defines the posterior surface was 21 mm. The position of the center of the posterior circle was defined by the constraint that the slopes of the tangents of the two surface circles must be equal at the point of intersection. The femur was modeled by extending it superiorly from the inside of the posterior femoral aspect. The femoral cam was shaped as an ellipse with major and minor axis lengths of 10 and 6 mm. The major axis of the cam was tilted 15° with respect to horizontal. The cam was placed 32 mm distal and 20 mm posterior from the distal center.

Differences between the IB II knee design and the IB I and IB II^R designs, as discussed by Striplin and Robinson,[22] were implemented using the computer model. The IB I has a smaller height of the poste-

Table 1. Clinical Dislocations Reported With the Insall-Burstein I and II Knees

Author	IB I Knee: Incidence of Dislocations	IB I Knee: Dislocation Rate (%)	IB II Knee: Incidence of Dislocations	IB II Knee: Dislocation Rate (%)
Galinat et al., 1988[19]	2 in 832	0.2	—	—
Cohen et al., 1991[18]	0 in 105	0.0	5 in 100	5.0
Striplin and Robinson, 1992[22]	0 in 428	0.0	2 in 240	0.8
Cohen and Constant, 1992[17]	—	—	2 in 437	0.5
Lombardi et al., 1993[20]	4 in 1,978	0.2	10 in 398	2.5
Total	6 in 3,343	0.2	19 in 1,175	1.6

IB, Insall-Burstein.

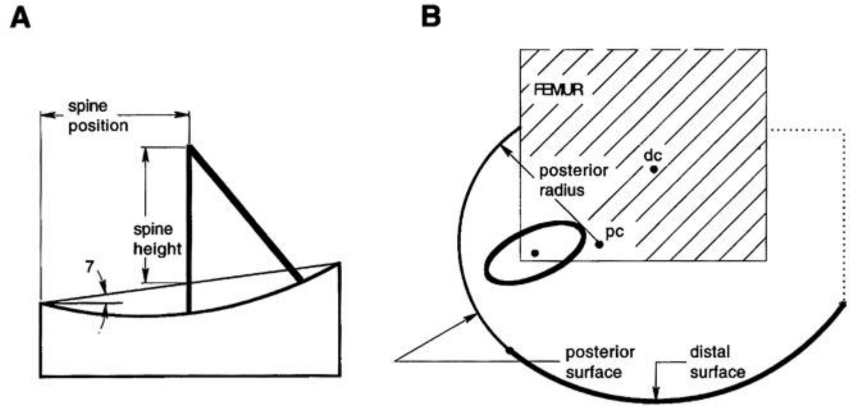

Fig. 1. Geometry of the tibial and femoral model components. (A) The tibial surface was sloped 7°. Tibial spine height was 14.5 mm and the spine position was 20 mm with the Insall-Burstein (IB) I and IB II knees. The spine height and position were changed to 16.5 mm and 22 mm, respectively, to represent the IB II^R knee. (B) The femoral component was formed from two circles that describe distal and posterior surfaces. The posterior radius was 18 mm with the IB I knee model and 21 mm with the IB II and IB II^R knee models. dc, the center of the distal circle; pc, the center of the posterior circle.

rior femoral flange than the IB II (see Fig. 3 of Striplin and Robinson[22]). We simulated this change by reducing the height of the posterior femoral flange of the IB II model to represent the IB I. This change was implemented by decreasing the posterior radius of the femoral component to 18 mm (a 3-mm decrease) in the IB I knee model. A model of the IB II^R was made by moving the tibial spine of the IB II model 2 mm anteriorly and making it 2 mm higher, as detailed by others.[18,20,22] All other dimensions remained constant in our simulations.

Knee flexion was simulated in two phases. The first phase consisted of a pure rotation of the femur about the center of curvature of the femoral surface in contact with the tibia (either the distal or posterior surface). During this phase, the tibiofemoral contact point remained constant. The second phase began when the femoral cam contacted the tibial spine. The motion of the femoral component during the second phase consisted of a combination of rolling and sliding that caused the femur to move posteriorly with respect to the tibial component. This motion was simulated by first rotating the femoral component and then translating it such that the spine and cam surfaces, as well as the tibial and femoral articulating surfaces, remained in contact.

Two measures were used to analyze the effects of changes in component geometry: dislocation safety factor (DSF) and maximum knee flexion. Dislocation safety factor was used to characterize the stability of each prosthetic knee design. The DSF was calculated as the distance from the top of the tibial spine to the bottom of the femoral cam (Fig. 2). When the DSF is positive, the cam is inferior to the top of the spine. As DSF increases, the femoral cam moves inferiorly with respect to the top of the tibial spine and the knee becomes less likely to dislocate. Thus,

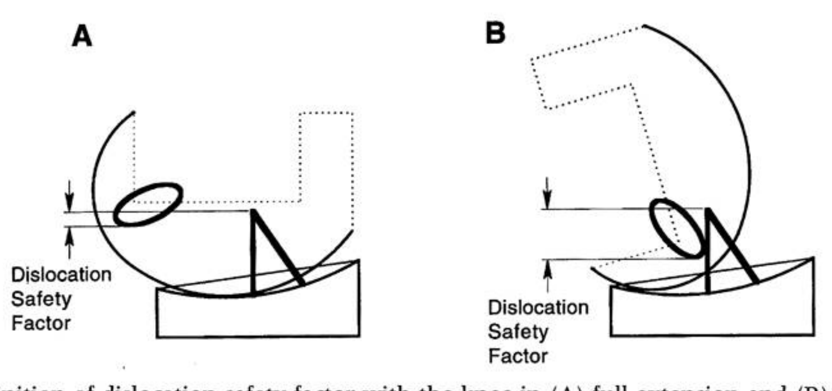

Fig. 2. Definition of dislocation safety factor with the knee in (A) full extension and (B) flexed 105°.

DSF is a geometric factor that represents the propensity of the prosthetic knee to dislocate. Dislocation safety factor was plotted versus knee flexion angle for each prosthetic knee simulation. The maximum knee flexion angle was determined as the flexion angle at which the femur impinged on the posterior lip of the tibial component.

Results

The variation of the DSF with knee flexion angle is similar for all of the IB knees studied (Fig. 3). The DSF is greatest at approximately 70° of knee flexion; as the knee is flexed or extended from this midrange value, DSF decreases, resulting in decreased knee stability. Thus, IB knees are most stable in midflexion and are more likely to dislocate at full extension and maximum flexion.

The IB I knee was changed (to the IB II) in an attempt to increase maximum knee flexion by increasing the height of the posterior femoral flange. Our results indicate that maximum flexion increased from 115° with the IB I to 125° with the IB II (Fig. 3, arrows); however, the geometry of the IB II also decreased DSF at maximum knee flexion. In our simulations, DSF at maximum flexion decreased from 11 mm with the IB I to 7 mm with the IB II. In other words, at maximum flexion, the bottom of the femoral cam is only 7 mm inferior to the top of the tibial spine with the IB II knee model. This occurs because DSF decreases more with flexion past 70° with the geometry of the IB II (ie, the slope of the dashed curve is steeper in Fig. 3). Also, because the IB II allows greater knee flexion and DSF decreases with flexion, DSF is lower for the IB II. This may explain the greater incidence of dislocations that have been reported with the IB II.

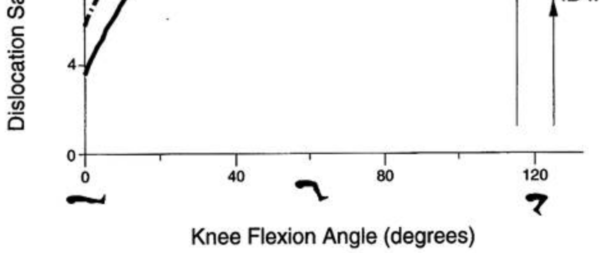

Fig. 3. Dislocation safety factor (DSF) versus knee flexion angle for the Insall-Burstein (IB) I (solid curve), the IB II (dashed curve), and the IB II^R knee (dot-dash curve). Note that the IB II knee provides greater knee flexion than the IB I knee (see arrows), but has a lower DSF at maximum flexion. The IB II^R increases DSF, but decreases maximum knee flexion when compared to the IB II.

The IB II was revised (to the IB II^R) to improve stability by increasing the height of the tibial spine 2 mm and moving the spine 2 mm anteriorly. Our simulations show that the DSF at maximum knee flexion increased from 7 mm with the IB II model to 12 mm with the IB II^R model. Because DSF is defined as the distance from the top of the spine to the bottom of the cam, increasing spine height 2 mm increased DSF 2 mm (note that the dot-dash curve is 2 mm higher than the dashed curve throughout the range of motion in Fig. 3). The anterior placement of the spine caused a decrease in maximum knee flexion from 125° to 117°. Moving the spine anteriorly caused the cam to contact the spine at a greater knee flexion angle, decreasing femoral rollback. Decreasing femoral rollback limited knee flexion by causing the femur to impinge on the posterior tibial component at a smaller knee flexion angle. Limiting knee flexion has the effect of increasing DSF at maximum flexion.

Discussion

Before discussing the implications of these results, the effects of several modeling assumptions should be considered. First, the model limits definition and analysis of the knee to the sagittal plane. This constraint is based on previous studies that suggest that the small amount of tibial rotation that occurs in normal knees is largely reduced after knee arthroplasty.[24] This is partly due to the symmetrically shaped femoral condyles of the posterior-stabilized knee.[25] Dislocations that may occur as a result of motions out of the sagittal plane were not examined in this study. Second, although external forces, or forces generated by muscles and ligaments, can affect knee motion, this study examined knee motion based solely on component geometry. To be consistent with this kinematic analysis, a geometric parameter, the DSF, was used to represent the propensity of the knee to dislocate. Since DSF is defined as the distance between the top of the tibial spine and the bottom of the femoral cam (Fig. 2), it seems reasonable that knees with a greater DSF would be less likely to dislocate. Finally, it should be made clear that knee prosthesis geometry can influence factors other than knee range of motion and stability. For instance, decreasing the radius of curvature of the posterior femoral condyle may decrease the contact area of the femoral and tibial components and promote component wear. Our study focuses on the effects of changes in component geometry on knee kinematics and stability only. How changes in component geometry affect other factors, such as component wear, is also very important, but was not considered here.

The results of our simulations have several important implications. First, small changes in component geometry can have substantial and unpredictable effects on knee stability. For instance, a 3-mm decrease in the posterior radius of the femoral component of our model increased maximum knee flexion 10°, as expected, but unexpectedly decreased DSF at maximum flexion by 4 mm. Also, making small changes to the geometry of the IB II to create the IB II^R had substantial effects on the stability and range of motion of the knee. The tibial spine of the IB II was moved 2 mm anteriorly and its height was increased 2 mm. This decreased maximum flexion of the knee 8° and increased DSF 5 mm (71%) at maximum flexion.

In terms of component geometry, there is a trade-off between increasing knee flexion and improving knee stability. The original IB I knee was changed into the IB II knee to increase flexion of the knee. The changes made to increase flexion decreased the DSF at maximum knee flexion. This corresponds with clinical reports showing more dislocations with the IB II than with the IB I.[17–19,21,22] The IB II^R knee increased stability, but decreased maximum obtainable flexion. This corresponds to clinical reports showing a decreased dislocation rate with the IB II^R [21] and highlights the trade-off between flexion and stability.

There is evidence that 115° of knee flexion is sufficient for most activities of daily living.[26–28] Thus, there may not be a great need to use the spine-cam mechanism to maximize knee flexion, especially because doing this generally decreases knee stability. This suggests that the spine-cam mechanism may be best suited for preventing knee subluxation and dislocation rather than maximizing knee flexion.

The shape of the DSF-versus-knee angle curves illustrates an important concept. The DSF peaks in the midportion of the knee flexion range and decreases with flexion beyond this point. Thus, all knees become inherently less stable as flexion increases past approximately 70°. The slope of the DSF curve at maximum flexion determines how much the femoral cam moves up the tibial spine near the end of the range of motion. If DSF decreases rapidly with flexion (a steep slope), then the knee becomes even more likely to dislocate as knee flexion increases or with forced hyperflexion. Indeed, previous clinical studies have reported that knees in which excellent postoperative flexion is obtained are more likely to dislocate.[18,20–22]

It has been suggested that several factors work in conjunction to cause a prosthetic knee to dislocate.[18,22] These factors include (1) component geometries that allows the femoral cam to ride far up the tibial spine, (2) knees that have exceptional flexion, (3) knees that have ligamentous laxity, and (4) a large posteriorly directed load applied to the tibia. Thus, although component geometry is only one of several factors that can contribute to a dislocation, it is critically important in situations of extreme flexion, ligamentous laxity, or large external loads.

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