Pergamon

J. Biomechanics, Vol. 28, No. 10, pp. 1155–1166, 1995

0021–9290/95 $9.50 + .00

0021–9290(94)00172-3

TRADEOFFS BETWEEN MOTION AND STABILITY IN POSTERIOR SUBSTITUTING KNEE ARTHROPLASTY DESIGN

Scott L. Delp,* Jonathan H. Kocmond* and Steven H. Stern†

*Departments of Biomedical Engineering and Physical Medicine & Rehabilitation, Northwestern University, and Sensory Motor Performance Program, Rehabilitation Institute of Chicago; and †Department of Orthopaedics, Northwestern Memorial Hospital, Chicago, Il, U.S.A.

Abstract—The purpose of this study was to examine how changes in component geometry of posterior substituting knees affect tibiofemoral kinematics and prosthesis stability. Most posterior cruciate ligament substituting prostheses rely on an articulation between a femoral cam and tibial spine to provide anterior–posterior stability of the knee. Failure of this ligament substitution mechanism has resulted in knee dislocations with several different posterior substituting designs. A computer model of a generic posterior substituting prosthesis was altered to analyze the effects of five design parameters (tibial spine height, spine anterior–posterior position, femoral component posterior radius, and femoral cam anterior–posterior and distal–proximal position) on prosthesis stability, tibiofemoral kinematics, and maximum obtainable knee flexion. Prosthesis stability was characterized by a 'dislocation safety factor', defined as the vertical distance from the bottom of the femoral cam to the top of the tibial spine. Computer simulations revealed that posterior substituting knees are most likely to dislocate at maximum knee flexion. Prosthesis stability can be improved by increasing the tibial spine height and moving the femoral cam posteriorly. Our results suggest there is a tradeoff between maximum knee flexion and prosthesis stability. We found that relatively small gains in maximum knee flexion, made through design changes, may cause substantial decreases in prosthesis stability.

Keywords: Knee; Prosthesis; Design; Dislocation; Kinematics.

INTRODUCTION

The modern age in total knee replacements commenced in the early 1970s with the advent of the total condylar prosthesis. The original total condylar prosthesis was a semi-constrained, condylar replacement requiring sacrifice of both cruciate ligaments. Initial results with this prosthesis were encouraging, but there were some important shortcomings. Specifically, posterior tibial subluxation occurred in certain cases, and early reports demonstrated an average of only 90° of knee flexion with this design (Insall *et al.*, 1979, 1983). Theoretically, an intact posterior cruciate ligament would resist posterior tibial subluxation and cause the femur to roll back on the tibia during knee flexion, promoting a greater arc of motion. In an attempt to increase prosthesis stability and range of motion, some knee arthroplasty designers advocated retention of the posterior cruciate ligament. Others promoted a prosthetic design in which a posterior cruciate ligament substituting mechanism was built directly into the components.

The original posterior cruciate substituting design (Insall–Burstein Posterior Stabilized Prosthesis) was introduced in 1978 as a modification of the total condylar design (Insall *et al.*, 1982). The prosthesis was intended to prevent posterior tibial subluxation and increase range of motion. In general, long-term clinical results with implants utilizing the concept of posterior cruciate ligament substitution have been excellent (Aglietti and Buzzi, 1988; Groh *et al.*, 1991; Laskin *et al.*, 1988; Lombardi *et al.*, 1988; Scott and Rubinstein, 1986; Scott *et al.*, 1988; Scuderi and Insall, 1989; Stern and Insall, 1990, 1992; Vince *et al.*, 1988). Thus, the principle of posterior cruciate substitution has now become widespread, and most implant manufacturers offer a posterior cruciate substituting knee prosthesis as one of their component options. The basic substitution mechanism is similar in many designs and is based on interaction between a tibial spine and femoral cam. The spine–cam interaction substitutes for the posterior cruciate ligament by providing anterior–posterior stability to the knee. Spine–cam interaction also causes the femur to roll posteriorly on the tibia (femoral rollback), which potentially increases maximum knee flexion by increasing the knee flexion angle at which the femur impinges on the posterior tibia.

Although the basic principles of posterior cruciate ligament substitution have remained essentially unchanged, the geometry of the spine and the position of the cam vary among manufacturers. As design changes have been implemented, problems that were not originally encountered have occurred. Specifi-

Received in final form 25 November 1994.

Address correspondence to: Scott L. Delp, Sensory Motor Performance Program, Rehabilitation Institute of Chicago (Room 1406), 345 East Superior Street, Chicago, IL 60611, U.S.A.

cally, there have been recent reports of prosthetic knee dislocations with several types of posterior substituting knee designs (Cohen and Constant, 1992; Cohen *et al.*, 1991; Galinat *et al.*, 1988; Gebhard and Kilgus, 1990; Hanssen and Rand, 1988; Lombardi *et al.*, 1991; Sharkey *et al.*, 1992; Striplin and Robinson, 1992). Dislocations of posterior substituting knees occur when the femoral cam translocates anteriorly over the tibial spine (Fig. 1). This results in an acute dislocation with the knee locked in a flexed position.

The factors affecting dislocation of posterior substituting knees remain unclear. It has been suggested that most dislocations occur in flexion and are caused by laxity of the ligaments and other soft tissues when the knee is flexed. However, the design of the tibial spine and the femoral cam, which are intended to provide prosthesis stability, may also affect dislocation.

The purpose of this study was to analyze the effects of five design parameters (tibial spine height, tibial spine anterior–posterior position, femoral condyle posterior radius, and femoral cam anterior–posterior and distal–proximal position) on tibiofemoral kinematics and stability of posterior cruciate ligament substituting components. We specifically aimed to determine how prosthesis stability varied with knee flexion and to analyze the role the design parameters play in increasing or decreasing the risk of prosthetic knee dislocation. A computer model of a generic posterior substituting knee that simulates motion in the sagittal plane was implemented on a computer graphics workstation. The geometry of this standard design was altered to study how design changes affect prosthesis stability, tibiofemoral kinematics, and maximum obtainable knee flexion.

METHODS

The dimensions of the 59 mm Insall–Burstein II prosthesis were used to create a standard design. Based on this geometry, the 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 anterior–posterior placement (Fig. 2A). Spine height was measured from the 7° tibial surface; the anterior–posterior placement of the spine was defined as the distance from the posterior edge of the tibia.

As is standard in the prosthetic industry, the surface of the femoral component was formed from two circles, one that defines the distal surface and another that defines the posterior surface (Fig. 2B). The circles were joined such that the slopes of their tangents were equal at their point of intersection. Each circle was defined by radius and the position of its center. The circle that defines the distal surface has a radius of 51 mm and its center (dc) was placed directly superior to the bottom of the tibial well. The radius of the circle that defines the posterior surface was varied. Once the posterior radius was specified, the position of the center of the posterior circle (pc) was changed to keep the anterior–posterior dimension of the femoral component constant and to assure that the slopes of the tangents of the two circles were equal at the point of intersection. The geometry of the femur was represented as a rectangle extending superiorly from the inside of the posterior aspect of the femoral component. The femoral cam was an ellipse with major and minor axis lengths of 10 and 6 mm. The major axis of the cam was tilted 15° from the horizontal. Cam anterior–posterior and distal–proximal placements were measured from the center of the circle that defines the distal surface (Fig. 2B).

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 that was in contact with the tibia (either the distal or posterior surface). During this phase the tibiofemoral contact point remained constant, essentially assuming that there is no friction between the tibial and femoral components. 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.

The effects of changing tibial spine height, spine position, femoral component posterior radius, and femoral cam anterior–posterior and distal–proximal position were analyzed. Simulations were performed by varying each parameter while the other parameters remained constant. Three simulations were performed for each parameter. One simulation represents the standard knee design. The other two simulations represent a 5 mm increase and a 5 mm decrease in the parameters' standard value. Five millimeter changes were used because they represent reasonable limits on the design of the prosthetic components and clearly demonstrate the effects of each parameter.

Three measures were used to analyze the effects of changes in component geometry: dislocation safety factor (DSF), tibiofemoral contact point, and maximum knee flexion. Dislocation safety factor was used to characterize the stability of each prosthetic knee design. The dislocation safety factor is the distance from the top of the tibial spine to the bottom of the femoral cam (Fig. 3). Negative values of DSF correspond to a situation where the entire femoral cam is positioned superior to the tibial spine. Under these conditions, with a posterior tibial translation, there will be no spine–cam contact to prevent dislocation. In contrast, if DSF is positive, the cam is inferior to the top of the spine. As DSF increases the femoral cam moves inferior with respect to the top of the tibial spine and the knee becomes less likely to dislocate.

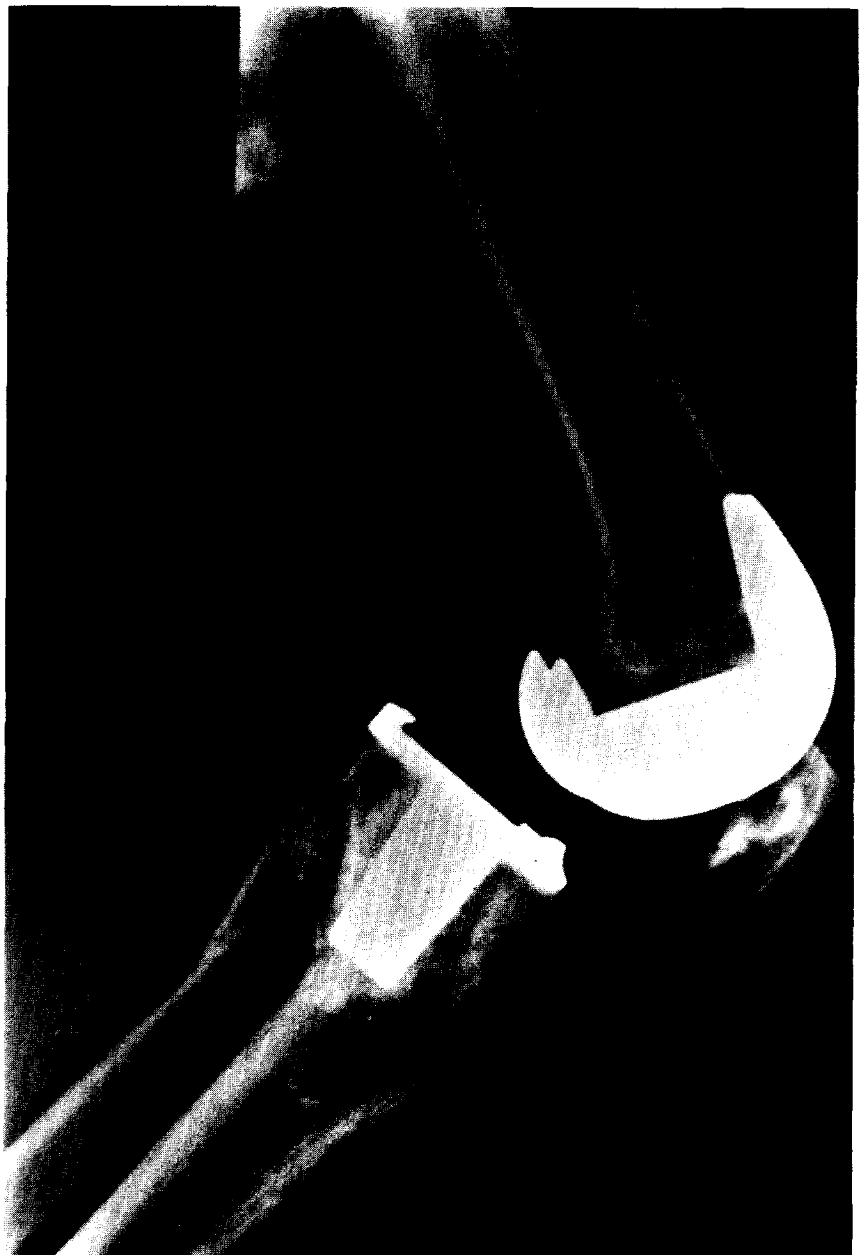

Fig. 1. Radiograph of a dislocated posterior substituting prosthetic knee. Dislocation occurs when the femoral cam translocates anteriorly over the tibial spine.

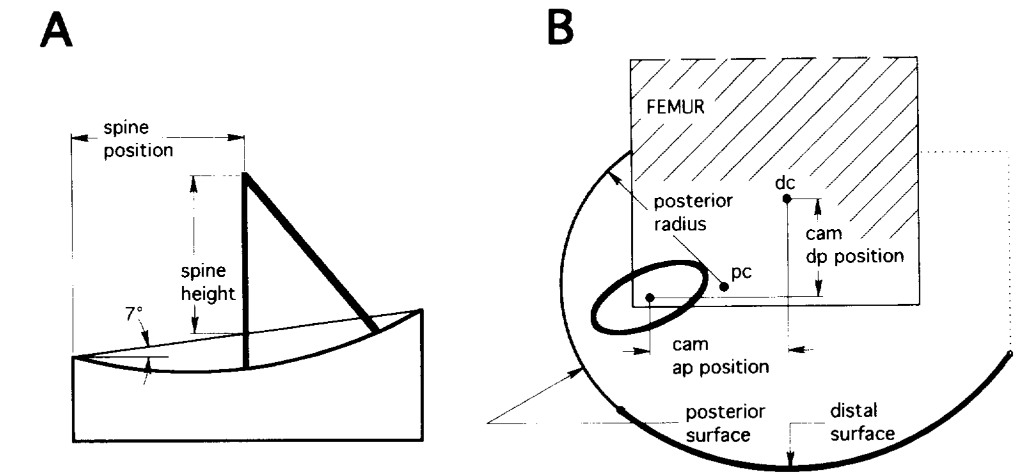

Fig. 2. Geometry of the tibial and femoral components. The tibial surface was sloped 7° posteriorly. Tibial wells were defined by a circle with a radius of 57 mm. Tibial spine height and spine position (A) were varied in this study. The femoral component was formed from two circles that describe distal and posterior surfaces (B). dc is the center of the distal circle; pc is the center of the posterior circle. The distal radius was 51 mm; the posterior radius was varied. The position of the cam in the anterior–posterior (ap) and distal–proximal (dp) directions, measured from the distal center, was also varied.

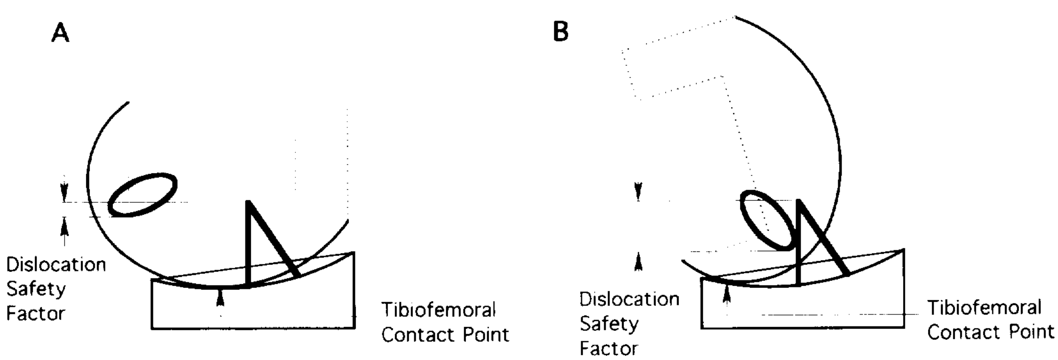

Fig. 3. Definition of dislocation safety factor (DSF) and tibiofemoral contact point with the knee in full extension (A) and with the knee flexed 105° (B).

Thus, DSF is a geometric factor that represents the propensity of the prosthetic knee to dislocate. Both DSF and the tibiofemoral contact point were plotted versus knee flexion angle for each prosthetic knee simulation.

The tibiofemoral contact point is the point where the femoral component contacts the tibial component (Fig. 3). Tibiofemoral contact point is expressed as a percentage of the anterior–posterior tibial width and is given a value from 0 to 100. Zero corresponds to the femoral component contacting the posterior edge of the tibial component and 100 is a contact point on the anterior edge of the tibial component. The standard knee model assumes that the femoral component initially rests on the bottom of the tibial wells. The bottom of the tibial wells occurs 35% anteriorly on the tibia; thus, the initial tibiofemoral contact point is 35. Femoral rollback is the posterior translation of the tibiofemoral contact point. Femoral rollback is defined as the difference between the initial contact point and the final contact point. Since spine–cam interaction causes femoral rollback, the angle at which the tibiofemoral contact point moves posteriorly is the angle at which the spine and cam initially contact.

Maximum knee flexion angle is determined when one of three conditions is met. The first condition is dislocation. In a simulation, dislocation occurs when the tibial spine and femoral cam are in contact and the cam passes over the spine. The second condition occurs when the value of the tibiofemoral contact point reaches 0. This indicates that the femoral component has rolled to the posterior edge of the tibial component and simulated flexion is stopped. Finally, flexion is limited if the femur impinges on the posterior edge of the tibial component.

RESULTS

The variation of dislocation safety factor (DSF) with knee flexion angle is similar for many alterations in component geometry. Specifically, DSF peaks in the middle of the knee flexion range and decreases with either flexion or extension (Figs 4A–8A). Thus, as the knee is flexed or extended from the knee position where DSF peaks, prosthesis stability decreases for all the designs studied here. The knee flexion angle at which DSF peaks and the degree to which DSF changes with knee flexion depend upon the prosthetic component geometry, as discussed in detail below.

Dislocation safety factor (DSF) increases with spine height (Fig. 4A). By definition, changes in spine height are reflected by identical changes in DSF. For example, when spine height is increased 5 mm, the dislocation safety factor increases 5 mm. When the spine height is decreased 5 mm the DSF is negative (i.e. the cam is superior to the top of the spine) for knee flexion angles less than 5°. The knee does not dislocate in this case because the femoral cam and tibial spine are not in contact at this angle of flexion. Spine height has no effect on tibiofemoral contact point or maximum knee flexion for the range of spine heights studied here (Fig. 4B). The contact point remains at 35 (where a contact point of 0 indicates the posterior border of the tibial component and a contact point of 100 indicates the anterior extreme) for flexion angles less than 80°, the angle at which the spine and cam contact. The spine–cam interaction causes the femur to roll back on the tibia; thus, the value of the tibiofemoral contact point decreases. Knee flexion is stopped in all cases because the femur impinges on the posterior tibial component. The maximum knee flexion angle is 125°.

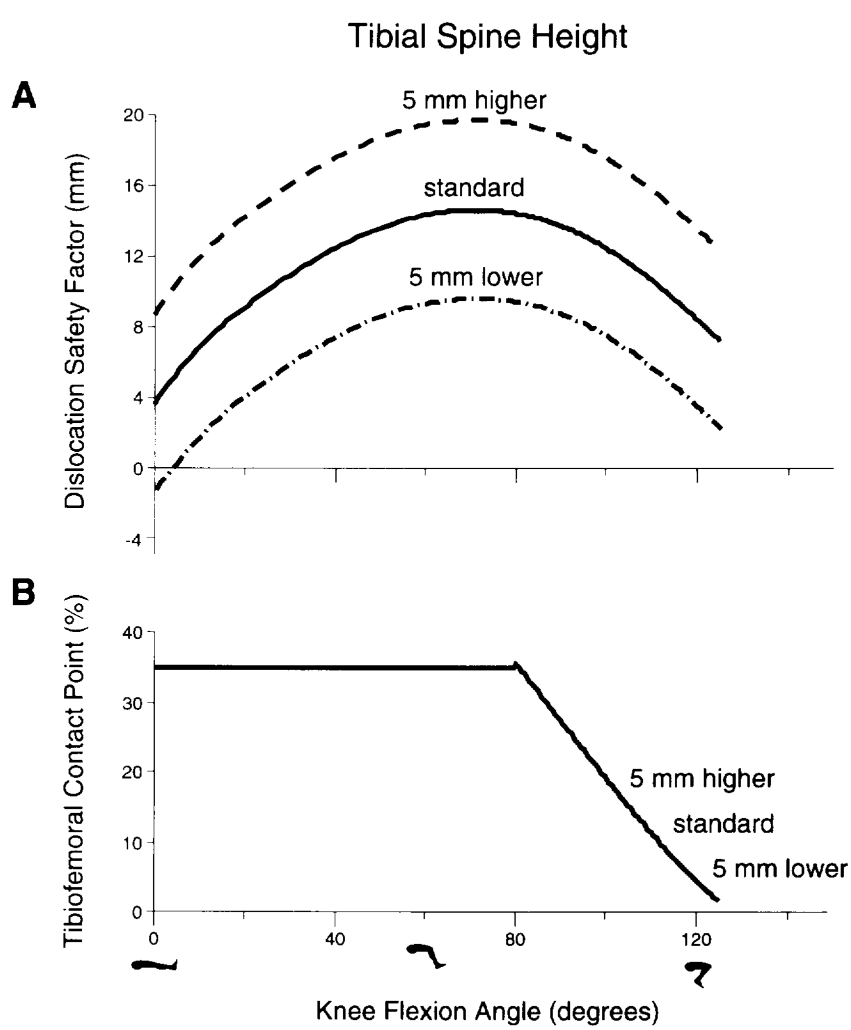

Fig. 4. The effect of spine height on dislocation safety factor and tibiofemoral contact point. Dislocation safety factor and tibiofemoral contact point are plotted vs knee flexion angle for the standard spine height (14 mm; solid curve), and spine heights 5 mm higher than the standard (dashed curve) and 5 mm lower than the standard (dot-dash curve). Note that dislocation safety factor increases with spine height (A). Spine height does not affect tibiofemoral contact point or maximum knee flexion; thus, the curves in (B) overlap.

Dislocation safety factor increases slightly with anterior spine placement (Fig. 5A). The 7° slope of the tibial component causes the top of the spine to be located more superiorly in the tibial coordinate system, which corresponds to a greater DSF. The shape of the tibial well also affects DSF slightly. As the femoral component rolls posteriorly and out of the bottom of the tibial well, the femoral cam is displaced superiorly relative to the tibial spine and the DSF decreases. For example, with the spine moved 5 mm posteriorly from the standard design, the DSF decreases more with knee flexion than in the other simulations for knee flexion angles greater than 75° (cf. the dot-dash curve). This occurs because the femur rolls back and out of the bottom of the tibial wells with a 5 mm posterior displacement of the spine. Spine–cam interaction and, thus, femoral rollback occur at lower knee flexion angles with posterior spine placement (Fig. 5B). With a 5 mm posterior displacement of the spine, the spine and cam first interact (i.e. the tibiofemoral contact point begins to move posteriorly) at 55° knee flexion. In contrast, with a 5 mm anterior displacement of the spine, the spine and cam do not interact until 105° knee flexion. Spine position affects maximum knee flexion. With the spine positioned posteriorly 5 mm, knee flexion is stopped when the femoral component rolls back to the posterior edge of the tibial component at 100° knee flexion. However, when the spine is moved 5 mm anteriorly, spine cam interaction occurs at a greater flexion angle than the standard design and femoral rollback is reduced. This causes the femur to impinge on the tibial component at 111° flexion, which limits maximum flexion.

Decreasing the posterior radius of the femoral condyle increases the knee flexion angle at which maximum DSF is reached (Fig. 6A). After maximum DSF is reached, DSF decreases less with knee flexion for

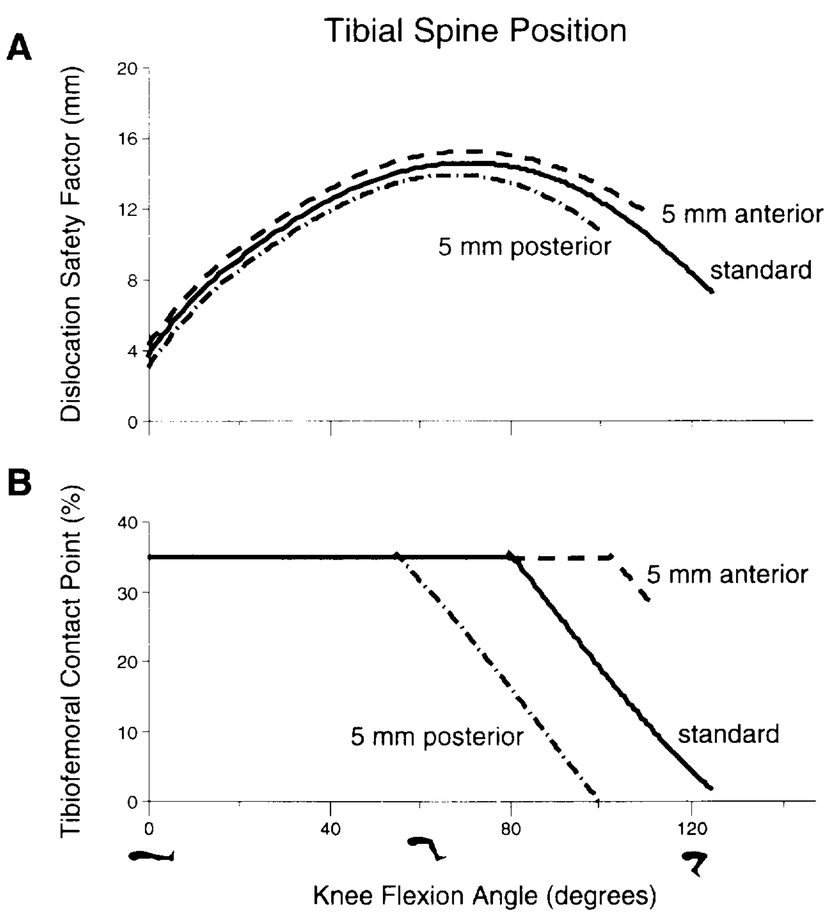

Fig. 5. The effect of spine position on dislocation safety factor and tibiofemoral contact point. Dislocation safety factor and tibiofemoral contact point are plotted vs knee flexion angle for the standard spine position (20 mm anterior to the posterior tibial edge; solid curve), and spine positions 5 mm anterior of the standard (dashed curve) and 5 mm posterior of the standard (dot-dash curve). Dislocation safety factor increases only slightly with anterior spine placement (A). Anterior spine placement causes the spine and cam to interact at greater flexion angles and reduces rollback (B). Maximum knee flexion is reduced when the spine is moved either anteriorly or posteriorly from its standard position.

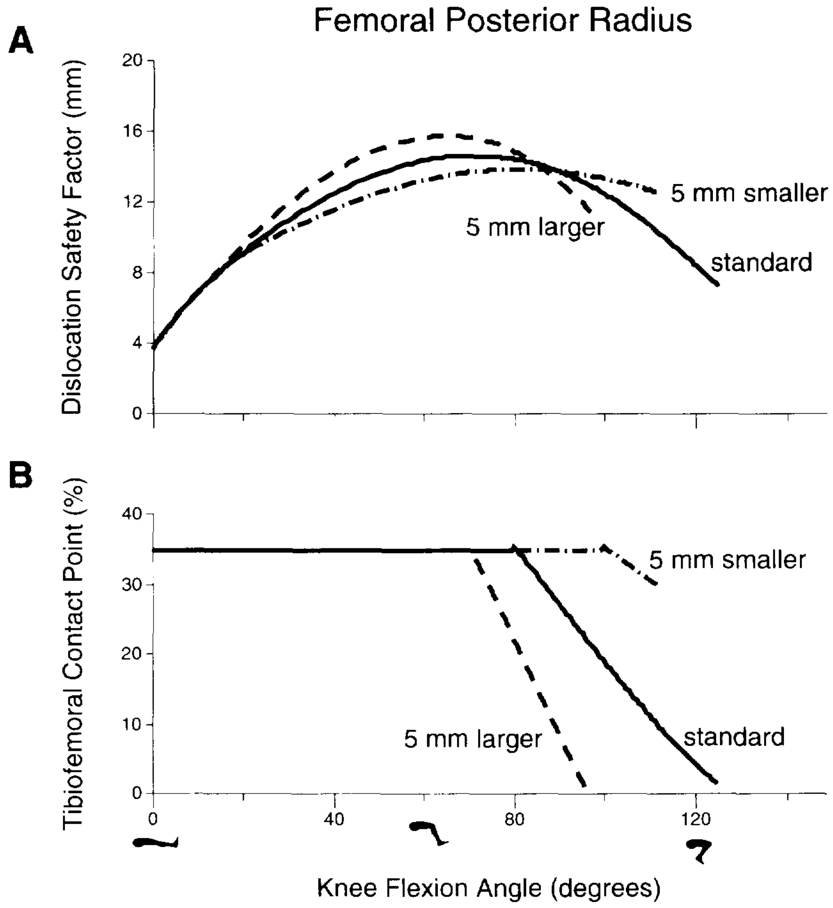

Fig. 6. The effect of the posterior radius of the femoral component on dislocation safety factor and tibiofemoral contact point. Dislocation safety factor and tibiofemoral contact point are plotted vs knee flexion angle for the standard posterior radius (21 mm; solid curve), and posterior radii 5 mm larger than the standard (dashed curve) and 5 mm smaller than normal (dot-dash curve). Note that the knee flexion angle at which the maximum dislocation safety factor is reached increases as the posterior radius decreases (A). With a decreased posterior radius the spine and cam interact at greater knee flexion angles and femoral rollback is reduced (B).

smaller posterior radii. Thus, a smaller posterior radius results in a more stable knee when the knee is flexed. Increasing the posterior radius causes the femur to roll back on the tibia at lower flexion angles (Fig. 6B). When the posterior radius is increased by 5 mm the spine and cam contact at 70° and the femur rolls back to the posterior edge of the tibial component. When the posterior radius is decreased by 5 mm the cam contacts the spine at 102° knee flexion. Maximal knee flexion is strongly influenced by changing the posterior radius. Increasing the posterior radius by 5 mm limits the range of motion to 97°. Decreasing the posterior radius also decreases maximal knee flexion compared with the standard design. When the posterior radius is decreased by 5 mm, femoral rollback is reduced and the femur impinges on the posterior tibial component at 112°.

Dislocation safety factor increases with posterior cam placement for flexion angles greater than 0° (Fig. 7A). With anterior cam placement, the tibiofemoral contact point moves posteriorly less with knee flexion (Fig. 7B). For example, with 5 mm anterior displacement of the cam, the femoral component rolls back from a tibial position of 35 to 15. The standard knee rolls back further (from 35 to 2) and when the cam is moved 5 mm posteriorly the femur rolls back to the posterior edge of the tibial component (tibiofemoral contact point equal to 0). However, anterior cam placement causes the spine and cam to contact at lower knee flexion angles. When the cam is moved 5 mm anteriorly, the spine and cam contact at 74°. For the standard knee and with the cam moved 5 mm posteriorly the spine and cam contact at 80 and 84°, respectively. Maximum flexion is reduced with anterior or posterior cam displacement. When the cam is moved 5 mm posteriorly, the spine–cam interaction causes the femur to roll off the tibia and limit flexion. When the cam is moved 5 mm anteriorly the femur rolls back less on the tibia and flexion is stopped at 117°.

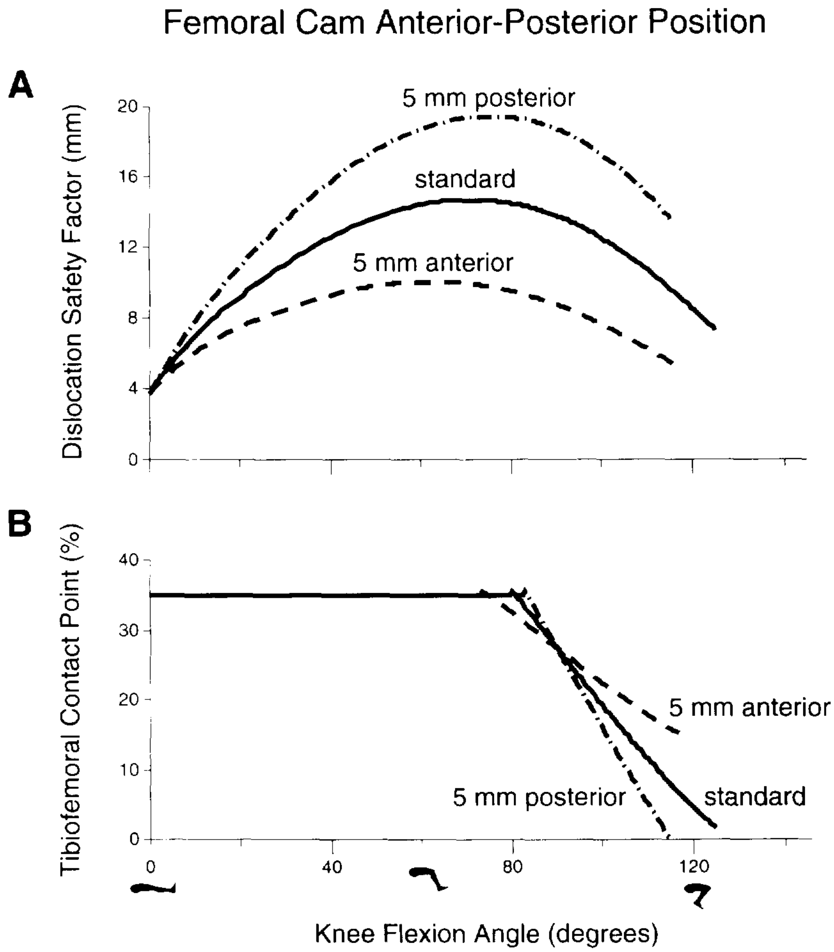

Fig. 7. The effect of anterior–posterior position of the femoral cam on dislocation safety factor and tibiofemoral contact point. Dislocation safety factor and tibiofemoral contact point are plotted vs knee flexion angle for the standard cam position (20 mm posterior to dc; solid curve), and for cam positions 5 mm anterior of the standard (dashed curve) and 5 mm posterior of the standard (dot-dash curve). Dislocation safety factor increases with posterior cam placement (A). The tibiofemoral contact point decreases more with flexion with posterior cam placement (B; see text for details).

The distal–proximal placement of the cam affects the dislocation safety factor significantly (Fig. 8A). The knee flexion angle at which maximal DSF is reached increases with proximal placement of the cam. With a 5 mm distal displacement of the cam, DSF peaks at 55° flexion. For the standard case, and with a 5 mm proximal cam displacement, DSF peaks 70 and 90° knee flexion, respectively. Furthermore, at flexion angles less than 90°, DSF is greater with distal placement of the cam. For flexion angles greater than 90° and at full knee flexion, DSF is greater with proximal placement of the cam. Distal cam placement causes the spine and cam to interact at lower flexion angles (Fig. 8B). When the cam is moved 5 mm distally the spine and cam initially contact at 60° knee flexion, and the femur rolls back to the posterior tibial edge at 101° knee flexion. When the cam is moved 5 mm proximally the spine and cam contact at 103°, femoral rollback is limited, and the femoral bone impinges on the tibial component at 111°. In both cases, maximum flexion decreases compared with the standard knee design.

DISCUSSION

The purpose of this study was to examine the effects of five design parameters of posterior substituting prostheses on tibiofemoral kinematics and prosthesis stability. The simulation results indicate that small changes in component geometry can have substantial effects on prosthesis stability. For example, decreasing the tibial spine height 5 mm from our standard prosthetic design reduces the dislocation safety factor at maximum knee flexion by almost 70% (7.4 mm compared with 2.4 mm). Our finding that prosthesis stability is sensitive to changes in prosthesis geometry is consistent with a recent clinical study, which reports that small design changes altered the dislocation rate of posterior stabilized knees from 0.2 to 2.5% (Lombardi *et al.*, 1993).

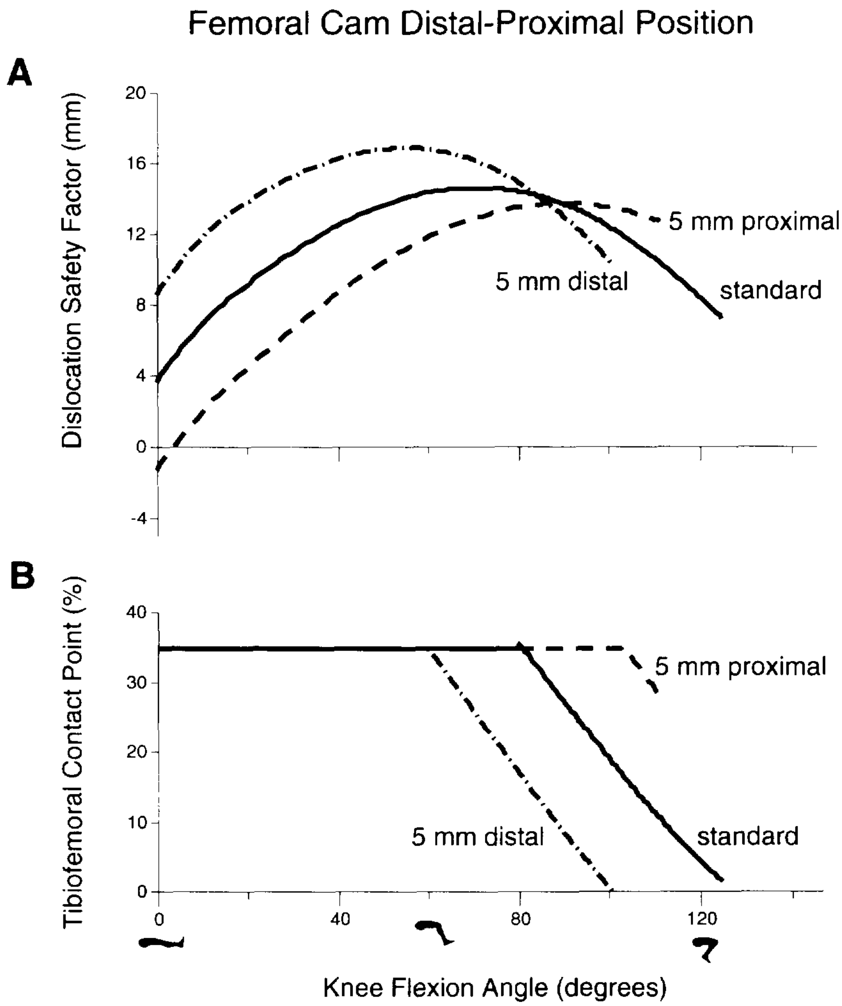

Fig. 8. The effect of the distal–proximal position of the femoral cam on dislocation safety factor and tibiofemoral contact point. Dislocation safety factor and tibiofemoral contact point are plotted vs knee flexion angle for the standard cam position (32 mm distal to dc; solid curve), and for cam positions 5 mm proximal of the standard (dashed curve) and 5 mm distal of the standard (dot-dash curve). The greatest effect on dislocation safety factor is at 0° where DSF increases with distal cam placement (A). However, with proximal cam placement, maximum DSF is reached at greater knee flexion angles. Proximal cam placement also reduces femoral rollback (B).

Before further discussing the implications of our results, the effects of several modeling assumptions should be considered. The computer model used here limits definition and analysis of the knee to the sagittal plane. This constraint is based on previous studies which suggest that the small amount of tibial rotation that occurs in normal knees is largely reduced after knee arthroplasty (Garg and Walker, 1990; Kurosawa *et al.*, 1985). This is partly due to the symmetrically shaped femoral condyles of most knee implants. Dislocations that may occur as a result of motions out of the sagittal plane were not examined in this study.

The current study examined knee motion based solely on component geometry. One should keep in mind that other factors, such as ligamentous laxity, can also play an important role in the dislocation of prosthetic knees. In this study, we concentrated on the effects of component geometry on prosthesis stability. To be consistent with this kinematic analysis, we defined a geometric parameter, the dislocation safety factor, to represent the propensity of the knee to dislocate. Because the dislocation safety factor is defined as the distance between the bottom of the femoral cam and the top of the tibial spine (see Fig. 3), it seems reasonable that knees with a greater dislocation safety factor would be less likely to dislocate. Thus, even though several factors contribute to knee stability, component geometry is important.

This study analyzed the effects of only five design parameters. However, other geometric parameters, such as the size and shape of the femoral cam, may also affect knee motion and stability. Surgical choices, such as the overall size of the prosthesis, the angles at which cuts are made, the alignment of the components, and the extent of soft tissue release, may affect knee motion and stability. The five parameters examined here were chosen based on an initial sensitivity study, which showed that these parameters have substantial effects on knee kinematics and stability, and the observation that these parameters vary among commercially available prostheses.

It should be made clear that knee prosthesis geometry can influence factors other than tibiofemoral kinematics 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 tibiofemoral kinematics and stability only. How changes in component geometry affect other factors, such as component wear, was not addressed in this study.

The variation of dislocation safety factor with knee flexion demonstrated here illustrates an important concept for surgeons employing posterior substituting knee designs. Dislocation safety factor peaks in the mid-portion of the knee flexion range, and decreases with either flexion or extension. This indicates that the knee is susceptible to dislocation at full extension and maximum flexion. However, at full extension the spine and cam are not in contact, and for dislocations to occur the femoral component must translate a significant distance and ride out of the conforming tibial wells. In contrast, in extreme flexion the spine–cam mechanism is engaged and the dislocation safety factor declines with further gains in flexion. Thus, the prosthetic components are most likely to dislocate at maximum flexion. This correlates with the clinical observation that knees with excellent postoperative flexion are the most prone to dislocations (Cohen *et al.*, 1991; Lombardi *et al.*, 1991, 1993; Striplin and Robinson, 1992). Furthermore, since the dislocation safety factor is small at large knee flexion angles, the ligament substituting mechanism is less effective in compensating for excessive soft tissue laxity. This emphasizes the importance that the surgeon balance the ligaments with the knee in flexion.

From a designer's point of view, maximizing the dislocation safety factor at extreme knee flexion is important in terms of reducing the likelihood of dislocation. Our simulations indicate that increasing the tibial spine height and placing the cam posteriorly are effective means to increase the dislocation safety factor. However, there are practical limits on spine height and cam position. For example, if the spine extends too far superiorly into the intercondylar box of the femoral component, it will cause varus–valgus restraint. Most modern cruciate substituting designs specifically strive to avoid varus–valgus restraint. Additional motion constraints caused by large increases in spine height may cause a posterior cruciate substituting arthroplasty to act more like constrained prostheses.

Designers of knee prostheses should also be aware of the tradeoff between knee flexion and prosthesis stability. All of the parameters studied here, except cam anterior–posterior position, have an inverse relationship between femoral rollback and prosthesis stability in flexion (Table 1). Anterior displacement of the femoral cam reduces both the dislocation safety factor at maximum flexion and femoral rollback. Accordingly, if sacrifices in maximum obtainable knee flexion are unacceptable, then placing the femoral cam posteriorly may be helpful, since this increases prosthesis stability and rollback. In this case, care must be taken to ensure that the femoral component does not roll posteriorly off the tibial component. However, most often, relatively small gains in maximum knee flexion made through design changes impose substantial decreases in prothesis stability.

TABLE I

EFFECTS OF THE FIVE PARAMETERS ON DSF* AND FEMORAL ROLLBACK

PARAMETER CHANGE	DSF at MAXIMUM FLEXION	FEMORAL ROLLBACK
Increased Spine Height	↑	NO Δ
Anterior Spine Placement	↑	↓
Decreased Posterior Radius	↑	↓
Anterior Cam Placement	↓	↓
Proximal Cam Placement	↑	↓

* See Fig. 3 for definition of Dislocation Safety Factor (DSF).

Therefore, the spine–cam mechanism in posterior substituting prosthetic knees may be best suited for preventing knee subluxation and dislocation, and not as a mechanism to maximize flexion.

The results presented here raise questions regarding the need to maximize knee flexion using femoral rollback. In our simulations, knee flexion is limited either by impingement of the femur on the posterior tibia or excessive rollback causing the femoral component to roll posteriorly off the tibial tray. Maximum flexion for a prosthetic design is thus theoretically achieved when the femur rolls back to the posterior edge of the tibial component at the same knee flexion angle that the femur impinges upon the posterior tibial component. Rollback allows increased knee flexion, because posterior movement of the tibiofemoral contact provides greater knee flexion before the femur impinges on the posterior tibia. However, even with minimal tibiofemoral rollback, the lowest maximum flexion angle achieved with our knee models was 111° (Fig. 5B). There is evidence that 115° of knee flexion is sufficient for most activities of daily living (Andriacchi *et al.*, 1980; McFadyen and Winter, 1988; Rodosky *et al.*, 1989). The minimum knee flexion (111°) obtained in our simulations does not represent a large decrease from this sufficient level and should be adequate for most arthritic patients. Thus, there may not be a great need to utilize the spine–cam mechanism to maximize knee flexion, especially since doing this generally decreases knee stability.

Acknowledgements—We wish to thank Kevin Greig and Abraham Komattu for their help in the development of the computer model used in this study. This work is supported by NSF Grant #BCS9110741 and the Northwestern Memorial Hospital Orthopaedic Research Fund 125.

REFERENCES

Aglietti, P. and Buzzi, R. (1988) Posterior stabilized total condylar knee replacement. Three to eight years' follow-up of 85 knees. *J. Bone Jt Surg.* **70B,** 211.

Andriacchi, T. P., Andersson, G. B. J., Fermier, B. S., Stern, D. and Galante, J. O. (1980) A study of lower-limb mechanics during stair-climbing. *J. Bone Jt Surg.* **62A,** 749–757.

Cohen, J., Bindelglass, D. and Vince, K. (1991) Dislocation of the spine and cam mechanism of the Insall–Burstein posterior stabilized knee prosthesis—why? *Orthop. Trans.* **15,** 634–635.

Cohen, B. and Constant, C. R. (1992) Subluxation of the posterior stabilized total knee arthroplasy. *J. Arthroplasty* **7,** 161–163.

Galinat, B. J., Vernace, J. V., Booth, R. E. and Rothman, R. H. (1988) Dislocation of the posterior stabilized total knee arthroplasty. A report of two cases. *J. Arthroplasty*, **3,** 363–367.

Garg, A. and Walker, P. S. (1990) Prediction of total knee motion using a three-dimensional computer-graphics model. *J. Biomechanics* **23,** 45–58.

Gebhard, J. S. and Kilgus, D. J. (1990) Dislocation of a posterior stabilized total knee prosthesis. A report of two cases. *Clin. Orthop.* **254,** 225–229.

Groh, G. I., Parker, J., Elliott, J. and Pearl, A. J. (1991) Results of total knee arthroplasty using the posterior stabilized condylar prosthesis. A report of 137 consecutive cases. *Clin. Orthop.* **269,** 58–62.

Hanssen, A. D. and Rand, J. A. (1988) A comparison of primary and revision total knee arthroplasty using the kinematic stabilizer prosthesis. *J. Bone Jt Surg.* **70A,** 491–499.

Insall, J. N., Hood, R. W., Flawn, L. B. and Sullivan, D. J. (1983) The total condylar knee prosthesis in gonarthrosis. A five to nine-year follow-up of the first one hundred consecutive replacements. *J. Bone Jt Surg.* **65A,** 619–628.

Insall, J. N., Lachiewicz, P. F. and Burstein, A. H. (1982) The posterior stabilized condylar prosthesis: a modification of the total condylar design. *J. Bone Jt Surg.* **64A,** 1317–1323.

Insall, J. N., Scott, W. N. and Ranawat, C. S. (1979) The total condylar knee prosthesis. A report of two hundred and twenty cases. *J. Bone Jt Surg.* **61A,** 173–180.

Kurosawa, H., Walker, P. S., Abe, S., Garg, A. and Hunter, T. (1985) Geometry and motion of the knee for implant and orthotic design, *J. Biomechanics* **18,** 487–499.

Laskin, R., Rieger, M., Schob, C. and Turen, C. (1988) The posterior stabilized total knee replacement in the knee with a severe fixed deformity. *Am. J. Knee Surg.* **1,** 199–203.

Lombardi, A. V., Honkala, T. K., Krugel, R., Mallory, T. H. and Vaughn, B. K. (1991) Dislocation following primary posterior stabilized total knee arthroplasty. *Orthop. Trans.* **15,** 716.

Lombardi, A. V., Mallory, T. H., Vaughn, B. K., Krugel, R., Honkala, T. K., Sorcher, M. and Kolczun, M. (1993) Dislocation following primary posterior-stabilized total knee arthroplasty. *J. Arthroplasty* **8,** 633–639.

Lombardi, A. V., Sydney, S. V., Mallory, T. H., Vaughn, B. K. and Kraus, T. J. (1988) Six year survivorship analysis of the Insall–Burstein posterior stabilized knee: a clinical and radiographic evaluation. *Orthop. Trans.* **12,** 711.

McFadyen, B. J. and Winter, D. A. (1988) An integrated biomechanical analysis of normal stair ascent and descent. *J. Biomechanics* **21,** 733–744.

Rodosky, M. W., Andriacchi, T. P. and Andersson, G. B. J. (1989) The influence of chair height on lower limb mechanics during rising. *J. orthop. Res.* **7,** 266–271.

Scott, W. N. and Rubinstein, M. (1986) Posterior stabilized knee arthroplasty: six year experience. *Clin. Orthop.* **205,** 138–145.

Scott, W. N., Rubinstein, M. and Scuderi, G. (1988) Results after knee replacement with a posterior cruciate-substituting prosthesis. *J. Bone Jt Surg.* **70A,** 1163–1173.

Scuderi, G. and Insall, J. N. (1989) The posterior stabilized knee prosthesis. *Orthop. Clin. N. Am.* **20,** 71–78.

Sharkey, P. F., Hozack, W. J., Booth, R. E. Jr, Balderston, R. A. and Rothman, R. H. (1992) Posterior dislocation of total knee arthroplasty. *Clin. Orthop.* **278,** 128–133.

Stern, S. H. and Insall, J. N. (1990) Total knee arthroplasty in obese patients. *J. Bone Jt Surg.* **72A,** 1400–1404.

Stern, S. H. and Insall, J. N. (1992) Posterior stabilized prosthesis: results after follow-up of 9–12 years, *J. Bone Jt Surg.* **74A,** 980–986.

Striplin, D. B. and Robinson, R. P. (1992) Posterior dislocation of the Insall/Burstein II posterior stabilized total knee prosthesis. *Am. J. Knee Surg.* **5,** 79–83.

Vince, K. G., Kelly, M. A. and Insall, J. N. (1988) Posterior stabilized knee prosthesis: follow-up at five to eight years. *Orthop. Trans.* **12,** 157.