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Posterior tibial slope and femoral sizing affect posterior cruciate ligament tension in posterior cruciate-retaining total knee arthroplasty

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A B S T R A C T

Background: During cruciate-retaining total knee arthroplasty, surgeons sometimes encounter increased tension of the posterior cruciate ligament. This study investigated the effects of femoral size, posterior tibial slope, and rotational alignment of the femoral and tibial components on forces at the posterior cruciate ligament in cruciate-retaining total knee arthroplasty using a musculoskeletal computer simulation.
Methods: Forces at the posterior cruciate ligament were assessed with the standard femoral component, as well as with 2-mm upsizing and 2-mm downsizing in the anterior–posterior dimension. These forces were also determined with posterior tibial slope angles of 5°, 7°, and 9°, and lastly, were measured in 5° increments when the femoral (tibial) components were positioned from 5° (15°) of internal rotation to 5° (15°) of external rotation.
Findings: Forces at the posterior cruciate ligament increased by up to 718 N with the standard procedure during squatting. The 2-mm downsizing of the femoral component decreased the force at the posterior cruciate ligament by up to 47%. The 2° increment in posterior tibial slope decreased the force at the posterior cruciate ligament by up to 41%. In addition, posterior cruciate ligament tension increased by 11% during internal rotation of the femoral component, and increased by 18% during external rotation of the tibial component.
Interpretation: These findings suggest that accurate sizing and bone preparation are very important to maintain posterior cruciate ligament forces in cruciate-retaining total knee arthroplasty. Care should also be taken regarding malrotation of the femoral and tibial components because this increases posterior cruciate ligament tension.

1. Introduction

Achieving proper soft tissue tension is crucial for controlling knee stability and mobility after total knee arthroplasty (TKA). The posterior cruciate ligament (PCL) plays a very important role in cruciate-retaining TKA (CR-TKA) in terms of both posterior stability and femoral rollback, the latter of which increases the range of motion. A tighter PCL increases joint reactive force on the polyethylene component of the prosthesis (Pagnano et al., 1998), and a previous study reported a rate of flexion tightness of 30% in CR-TKA (Ritter et al., 1988). Alternatively, a looser PCL may result in anterior–posterior instability and knee pain (Morberg et al., 2002; Waslewski et al., 1998).

Partial PCL release is a possible solution to control the tension of the PCL (Ritter et al., 1988). However, this technique has not been investigated thoroughly and has yielded contradictory results, and it may not restore normal kinematics during knee flexion. The appropriate posterior tibial slope (PTS) is also important for maintaining proper PCL tension. Cadaver studies have reported mixed results regarding the effect of the PTS on PCL tension. One study reported that the tension of the PCL measured at a 5° PTS was significantly greater than the tension measured at 8° and 10° (Singerman et al., 1996). Conversely, another study showed that changing the surface slope of the tibial insert from 0° to 15° did not produce any consistent changes in the PCL tension pattern (Incavo et al., 1994). One of the limitations of cadaver studies is that only a small amount of force can be applied to the knee joint and surrounding muscles. Therefore, cadaver models do not reproduce the knee conditions experienced during daily activities such as squatting or deep knee bending.

Another way to balance the PCL is by upsizing or downsizing the femoral component. Some TKA products can downsize the femoral component without additional bone cutting. Although the downsizing of the femoral component can theoretically reduce PCL tension in flexion (Arima et al., 1998; Pagnano et al., 1998), the practical effect on PCL tension remains unclear. Moreover, the rotational alignment of the femoral and tibial components might affect PCL tension, but this has not been evaluated.

The purpose of this study was to use a musculoskeletal computer model to evaluate the effects of PTS, femoral sizing, and rotational

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alignment of the femoral and tibial components on forces at the PCL during weight-bearing deep knee flexion.

2. Methods

The musculoskeletal computer model used in this study is a dynamic program that simulates the knee (LifeMOD/KneeSIM 2010; LifeModeler Inc., San Clemente, CA) (Colwell et al., 2011; Innocenti et al., 2011; Mihalko et al., 2012; Mizu-uchi et al., 2011, 2014). Using this model, we simulated a weight-bearing deep knee bend using an Oxford-type knee rig. This musculoskeletal model included tibiofemoral and patellofemoral contact, the PCL, the lateral collateral ligament (LCL), the medial collateral ligament (MCL), elements of the knee capsule, the quadriceps muscle and tendon, the patellar tendon, and the hamstring muscles. The PCL and MCL comprised one bundle each (Brantigan and Voshell, 1941; Edwards et al., 2007; Harner et al., 1995; Warren et al., 1974). The LCL was considered as a single bundle (LaPrade et al., 2005; Sugita and Amis, 2001). All ligament bundles were modeled as nonlinear springs with material properties obtained from a published report (Blankevoort et al., 1991). The ligaments were simulated as nonlinear force elements with their parabolic and linear equations as follows: If $\varepsilon < 0$, $F(\varepsilon) = 0$; if $0 \le \varepsilon \le 2\varepsilon_1$, $F(\varepsilon) = 0.25\,k\varepsilon^2/0.03$; and if $\varepsilon > 2\varepsilon_1$, $F(\varepsilon) = k\,(\varepsilon - 0.03)$. F is the tension of the ligament, ε is the ligament strain, and k is the stiffness coefficient of each ligament. The linear range threshold was specified as $\varepsilon_1 = 0.03$. The soft tissue elements remained in the same position in all situations tested in this study. The hip joint was modeled as a revolute joint parallel to the flexion axis of the knee and was allowed to slide vertically. The ankle joint was modeled as a combination of several joints that combined to allow free translation in the medial–lateral direction and free rotation in flexion, axial, and varus–valgus directions.

We adjusted the origins of the insertion points, stiffness, and length patterns of each ligament based on the relevant anatomical literature (Edwards et al., 2007; Harner et al., 1995; LaPrade et al., 2005; Liu et al., 2011; Park et al., 2005; Wijdicks et al., 2010). Regarding the insertion points of the PCL, the femoral attachments were on the anterior area of the medial intercondylar wall, and the tibial attachments were on the posterior intercondylar fossa, anteroposteriorly in the order of the anterior–lateral and posterior–medial bundles. The proximal attachment points of the LCL and MCL were the most prominent points of the lateral and medial epicondyles on the femur. The distal attachment points of the LCL and MCL were set as the tip of the fibular head and the midpoint between the tibial attachments of the anterior bundle and the posterior bundle, respectively (Fig. 1). The stiffness coefficients of the anterior–lateral PCL, posterior–medial PCL, LCL, anterior MCL, and posterior MCL were set at 102, 102, 59, 63, and 63 N/mm, respectively, based on reported values (Harner et al., 1995; Robinson et al., 2005; Sugita and Amis, 2001; Wilson et al., 2012). The length and slack of each ligament throughout its range of motion were finely adjusted so that the length patterns of the ligaments were similar to those reported by a previous cadaver study (Fig. 2) (Belvedere et al., 2012).

The KneeSIM program was used to perform analyses using the Parasolid geometry of the femoral and tibial components, as well as the tibial and patellar inserts. Parasolid models of a fixed-bearing CR total knee (NexGen CR-Flex; Zimmer, Warsaw, IN) were imported into the program (Fig. 3).

During the alignment of the components in the coronal plane, the femoral component was set perpendicular to the mechanical axis that connected the center of the knee and the center of the femoral head, and the tibial component was set perpendicular to the mechanical axis that connected the center of the knee and the center of the ankle joint. For sagittal alignment, the femoral component was aligned parallel to the distal anatomical axis of the femur, and the tibial component was aligned parallel to the proximal anatomical axis, using a PTS of 7° as the initial setting. The neutral rotational alignments of the femoral

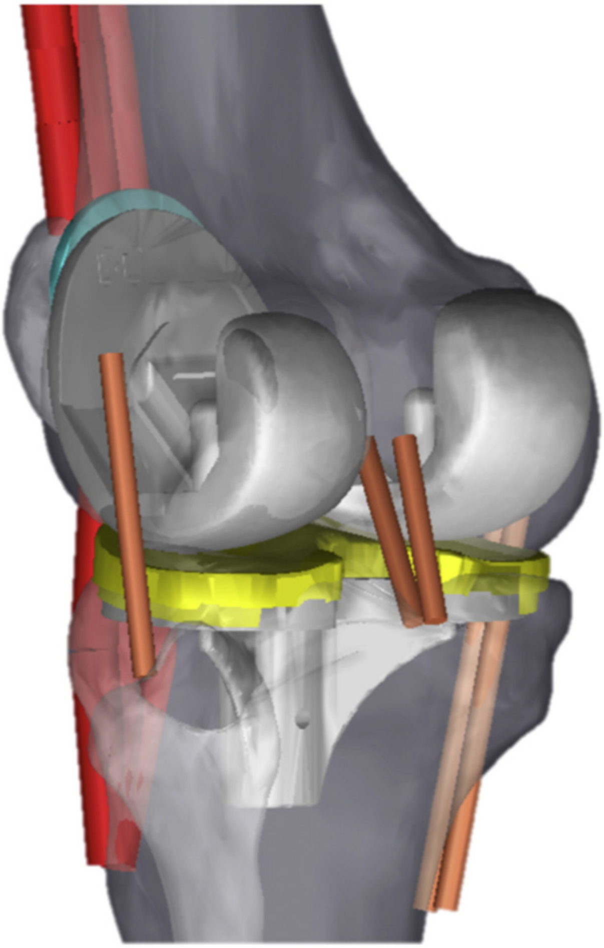

Fig. 1. The schema shows the musculoskeletal computer model of the attachment points of the anterior–lateral and posterior–medial bundles of the posterior cruciate ligament with neutral rotation.

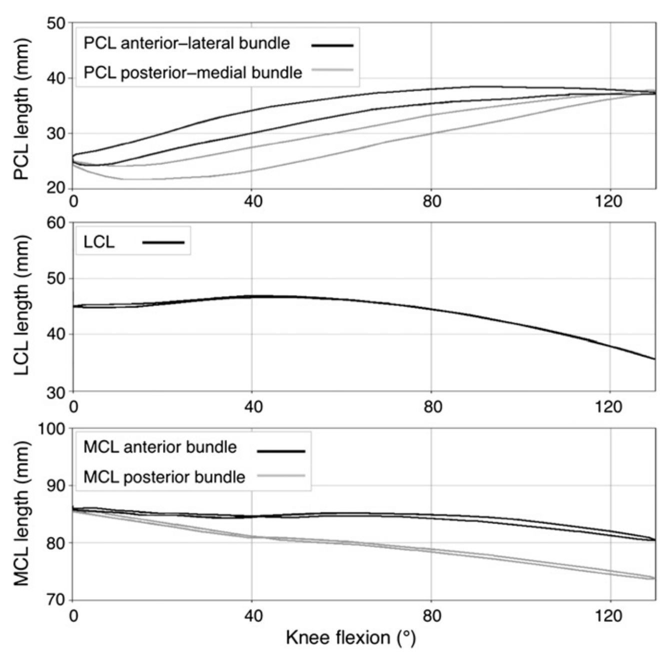

Fig. 2. The graphs show the patterns of change in the translation length (mm) compared with each free length for the posterior cruciate ligament (PCL), lateral collateral ligament (LCL), and medial collateral ligament (MCL) in weight-bearing deep knee flexion (0°–130°–0° flexion).