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Effect of tibial component alignment on knee kinematics and ligament tension in medial unicompartmental knee arthroplasty

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Objectives

Unicompartmental knee arthroplasty (UKA) is one surgical option for treating symptomatic medial osteoarthritis. Clinical studies have shown the functional benefits of UKA; however, the optimal alignment of the tibial component is still debated. The purpose of this study was to evaluate the effects of tibial coronal and sagittal plane alignment in UKA on knee kinematics and cruciate ligament tension, using a musculoskeletal computer simulation.

Methods

The tibial component was first aligned perpendicular to the mechanical axis of the tibia, with a 7° posterior slope (basic model). Subsequently, coronal and sagittal plane alignments were changed in a simulation programme. Kinematics and cruciate ligament tensions were simulated during weight-bearing deep knee bend and gait motions. Translation was defined as the distance between the most medial and the most lateral femoral positions throughout the cycle.

Results

The femur was positioned more medially relative to the tibia, with increasing varus alignment of the tibial component. Medial/lateral (ML) translation was smallest in the 2° varus model. A greater posterior slope posteriorized the medial condyle and increased anterior cruciate ligament (ACL) tension. ML translation was increased in the > 7° posterior slope model and the 0° model.

Conclusion

The current study suggests that the preferred tibial component alignment is between neutral and 2° varus in the coronal plane, and between 3° and 7° posterior slope in the sagittal plane. Varus > 4° or valgus alignment and excessive posterior slope caused excessive ML translation, which could be related to feelings of instability and could potentially have negative effects on clinical outcomes and implant durability.

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Article focus

- Little is known about the effect of component alignment on kinematics and ligament tension after unicompartmental knee arthroplasty (UKA).
- A computer simulation programme was used to analyze the kinematics and cruciate ligament tension in differently aligned tibial component models.

Key messages

- Slight varus or neutral alignment of tibial component is preferable for minimal translation in the medial/lateral (ML) direction.
- Excessive posterior slope should be avoided due to greater tension of the anterior cruciate ligament (ACL) and excessive translation in the ML direction.

Strengths and limitations

- A computer simulation programme was able to estimate the kinematics and ligament tension for various conditions.
- The study was restricted to only one implant design and two activities.

Introduction

Unicompartmental knee arthroplasty (UKA) is one surgical option for treating symptomatic isolated medial osteoarthritis (OA). Theoretically, UKA has some advantages over total knee arthroplasty (TKA). It is less invasive,[1] promotes rapid recovery,[2] and produces superior functional results.[3] UKA also preserves native structures, including the cruciate ligaments and the patellofemoral joint, which enables UKA to more closely resemble natural knee kinematics.[4] However, despite current progress in the UKA surgical procedure or implant design, previous studies have shown that UKA survival does not match that of TKA.[3,5,6] Further improvements are desired before UKA can be considered as a treatment option more widely.

Component position and alignment are thought to be crucial for better clinical outcomes and durability.[7-12] In addition, the cruciate ligaments, particularly the anterior cruciate ligament (ACL), are also important to the success of UKA. An ACL-deficient knee is thought to be a contraindication due to a higher failure rate.[13,14] In terms of implant survival in a clinical setting, previous studies showed that valgus alignment in the coronal plane and greater posterior slope in the sagittal plane at tibial component should be avoided.[8,9] To investigate the effects of different tibial component alignments, previous studies often utilized finite element analysis (FEA),[15-17] and evaluated the effects on tibial bone stresses in a static condition. So far, no studies have shown the effects of coronal and sagittal plane alignment on ligament tension in a dynamic manner.

Knee kinematics are considered to be related to clinical outcomes after knee prosthesis.[18] Concerning UKA, movement patterns in the anteroposterior (AP) direction are often analyzed and used as parameters. These generally indicate that UKA kinematics closely resemble those of a normal knee.[19] However, the effects of tibial component alignment on knee kinematics and kinetics in UKA are still unclear. In addition, kinematics in the medial/lateral (ML) direction have not been widely analyzed due to limitations of the analysis methods.

In this study, the musculoskeletal knee model (LifeMOD/KneeSIM 2010; LifeModeler Inc., San Clemente, California) was used to clarify the various loading states in dynamic conditions. The purpose of this study was to investigate the effects of different tibial component coronal and sagittal plane alignments on knee kinematics and ligament tension. It was hypothesized that malalignment (relative to preoperative alignment) would adversely affect knee kinematics and ligament tension, which could have a negative effect on clinical outcomes.

Materials and Methods

The musculoskeletal knee model was used for the computer simulation (Fig. 1a). This programme mimics Oxford-rig type setup, commonly used to analyze kinematics of the knee with implant in cadaver specimens.[19] This programme has been used by several researchers in order to investigate the kinematic difference in different implant design or different component positions.[20-23] Regarding tibial component of TKA, detrimental kinematic effects of varus alignment or excessive posterior slope were reported using this programme.[23,24] This programme has been validated in previous reports in patients' fluoroscopic surveillance and cadaveric experiments.[22,24,25] Concerning kinematics, the mean absolute difference of the AP contact positions between this simulation programme and fluoroscopic surveillance was 1.0 mm from 0° to 120° of flexion, during a weight-bearing deep knee bend activity for three patients implanted with TKA.[25]

This simulation programme includes the femorotibial and patellofemoral contact, ACL, posterior cruciate ligament (PCL), medial collateral ligament (MCL), lateral collateral ligament (LCL), elements of the knee capsule, quadriceps muscle and tendon, patellar tendon, and the hamstring muscles as components. A weight-bearing deep knee bend and gait motions were analyzed. In the deep knee bend motion, the knee was flexed from 0° to 150°, and then extended to 0°. In the gait motion, two double knee actions were represented, and the knee was flexed to 15° and 60° during the stance and swing phases, respectively (Fig. 1b). A constant vertical force of 4000 N, equivalent to approximately five times body weight (80 kg), was used.[26-29] The 4000 N load was applied at the hip and loaded onto the knee joint. In this simulation model, the hip joint was modelled as a revolute joint, and it was only allowed to slide vertically. The ankle joint model was allowed to translate freely in the ML direction, and to rotate freely in the direction of flexion and the axial and varus-valgus directions. The simulation was driven using a controlled actuator arrangement, similar to a physical machine. The quadricep and hamstring muscle forces were calculated to induce deep knee bend and gait motions. A closed-loop controller applied tension to these muscles to match the firing at a prescribed flexion angle at each point in the deep knee bend and gait motions; co-contraction between these muscles was defined to coordinate the motion.

The origins of the insertion points and stiffness of the ligaments were based on relevant anatomical studies of normal knees.[30-32] The ACL, PCL, and MCL comprised two bundles in this simulation model. All ligament bundles were modelled as nonlinear springs with material properties obtained from a published report.[33] The stiffness coefficients of the ACL anteromedial (AM) bundle, ACL posterolateral (PL) bundle, PCL anterolateral (AL) bundle, and PCL posteromedial (PM) bundle were all 102 N/mm. The stiffness coefficients were lower for the MCL-anterior and MCL-posterior (both 63 N/mm), and also for the LCL (59 N/mm).[32,34-36] In addition, the joint capsule stiffness coefficients were determined as 1.75 N/mm for the medial and lateral side. The initial strain of each ligament was determined using previous cadaver studies.[36-38]

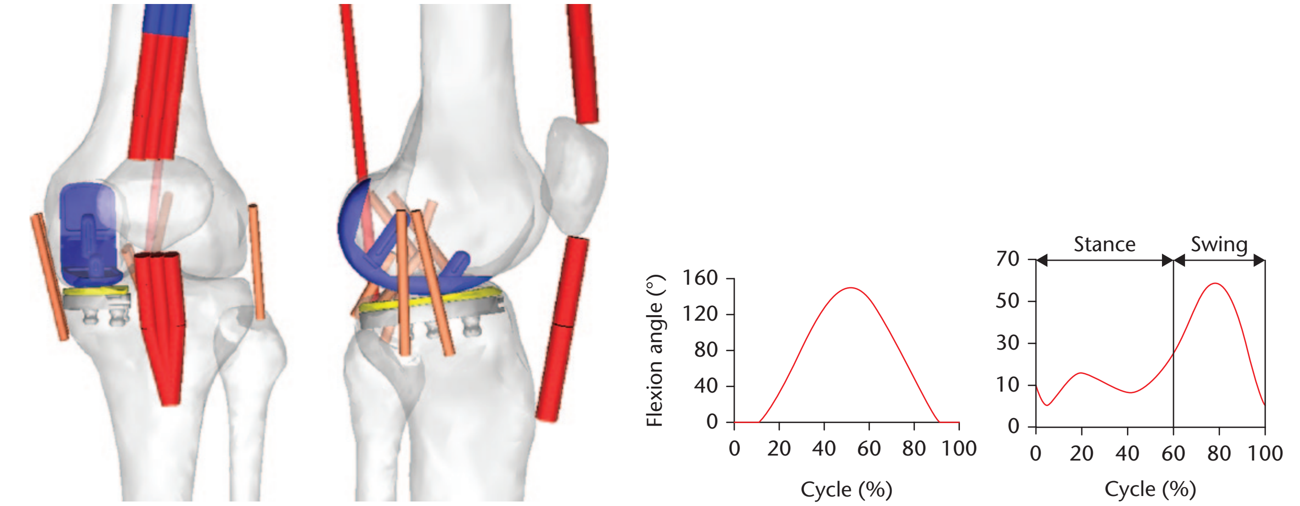


Fig. 1a Fig. 1b Fig. 1c Fig. 1d

Computer simulation model for a) anterior and b) lateral views of the properties. Simulation protocols and range of movement for c) the deep knee bend and d) gait cycles.

The KneeSIM programme used Parasolid geometry for the femoral and tibial components. Parasolid models of a fixed-bearing medial UKA with a multi-radius femoral component and a flat polyethylene insert (TRIBRID; Kyocera, Osaka, Japan) were imported into the simulation model. The femoral component was positioned perpendicular to the mechanical axis of the femur in the coronal plane, and parallel to the transepicondylar axis in the axial plane. The adequate size (size 6) was selected and virtually implanted to keep the level of the joint line on the distal and posterior surface of the femur, without any anterior notching.

The original bone model in the simulation programme had a neutral limb alignment, in which the mechanical axis of the limb passed the centre of the knee joint with 7° of posterior tibial slope. The original bone model (intact knee model) was simulated, and validation was performed by comparing with a previous *in vivo* normal knee kinematic study.[39] In terms of AP position during deep knee bend motion from 0° to 120° of flexion in 10° increments, root mean square errors (RMSE) between our intact knee model and a previous study were 2.3 mm and 1.5 mm at medial and lateral condyles, respectively.

In the basic model, the tibial component was placed in mechanical varus/valgus 0° in the coronal plane, parallel to the AP axis in the axial plane, and with 7° of posterior slope in the sagittal plane. The adequate size (size 4) was determined to avoid overhang of the medial and AP aspects of the tibia. In the proximal/distal direction, the tibial component was implanted to keep the joint line at the centre of the medial tibial plateau, with an 8 mm thickness of polyethylene combined with the tibial component. Subsequently, the coronal alignment was varied from 6° varus to 4° valgus in 2° increments (six different coronal alignment models), based on the centre of the tibial component in the ML direction. The sagittal alignments were also altered from 0° to 11° of posterior slope (six different sagittal alignment models), based on the centre of the tibial component in the AP direction. In all, 11 different alignment models for the tibial component were constructed and analyzed (Fig. 2). Neutral limb alignment was maintained for all models after virtual implantation.

Knee kinematics were analyzed in the ML and AP directions using the tibial component coordinate system. We used flexion facet centres as the reference system of the femur. Flexion facet centres were set as the centres of circles to fit the articular surface of the posterior condyles at sagittal planes.[20] Regarding the AP direction, AP positions of the medial and lateral flexion facet centres were evaluated. Regarding the ML direction, the midpoint between the medial facet centre and lateral facet centre was used as a reference point for the position of femur and evaluated. The anterior and medial directions for the AP and ML positions of the femur, respectively, were denoted as positive in the kinematic analyses. ML translation was defined as the distance between the most medial point and the most lateral point throughout one cycle for both motions (Fig. 3). The tension of each cruciate ligament was calculated as the sum of the forces of the two bundles.

Results

Medial/lateral position. In the coronal models, increased varus alignment medialized the position of femur throughout a cycle of both deep knee bend and gait motions (Fig. 4). The femur at 6° varus model was positioned medially at maximum 5.2 mm and 6.1 mm compared with neutral alignment model during deep knee bend and gait motions, respectively. On the contrary, valgus inclination lateralized the position of the femur. The femur at 4° valgus model was positioned laterally at maximum 4.3 mm

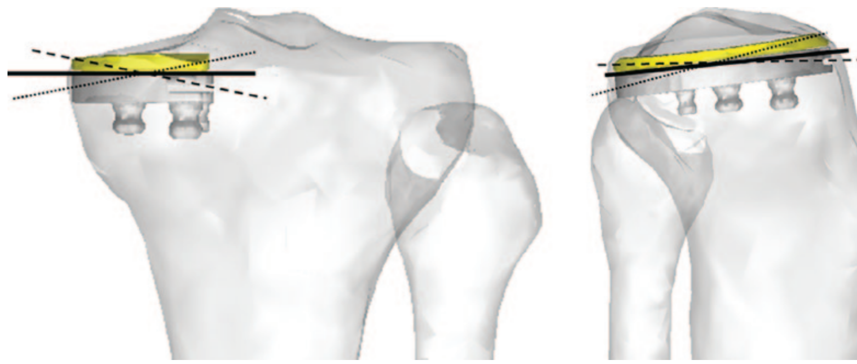


Fig. 2a Fig. 2b

The schema of a) coronal and b) sagittal models analyzed in this study. Coronal model: varus 6° to valgus 4°, posterior slope 7°. Sagittal model: varus/valgus 0° (neutral), posterior slope 0° to 11°.

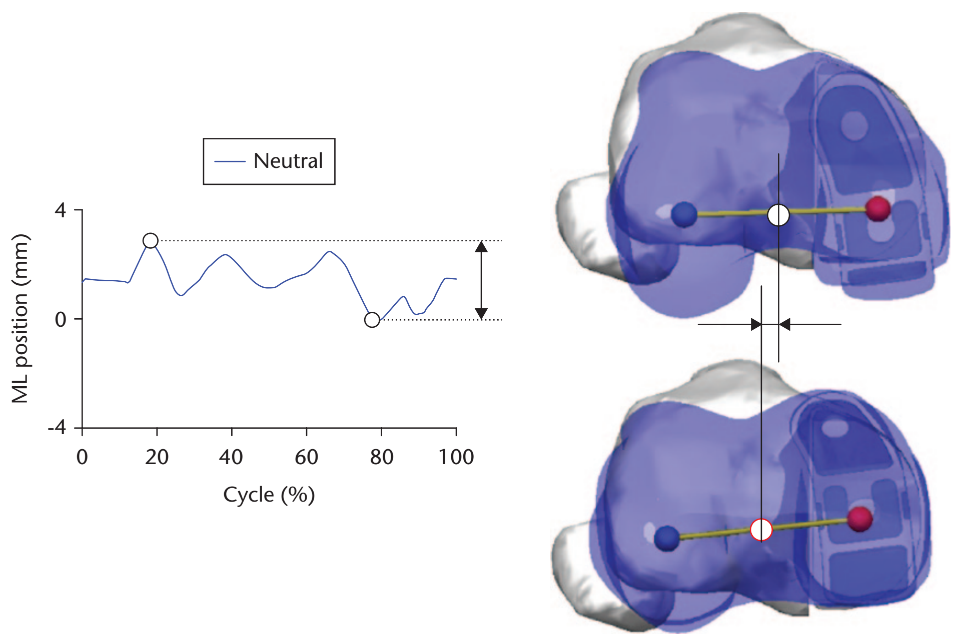


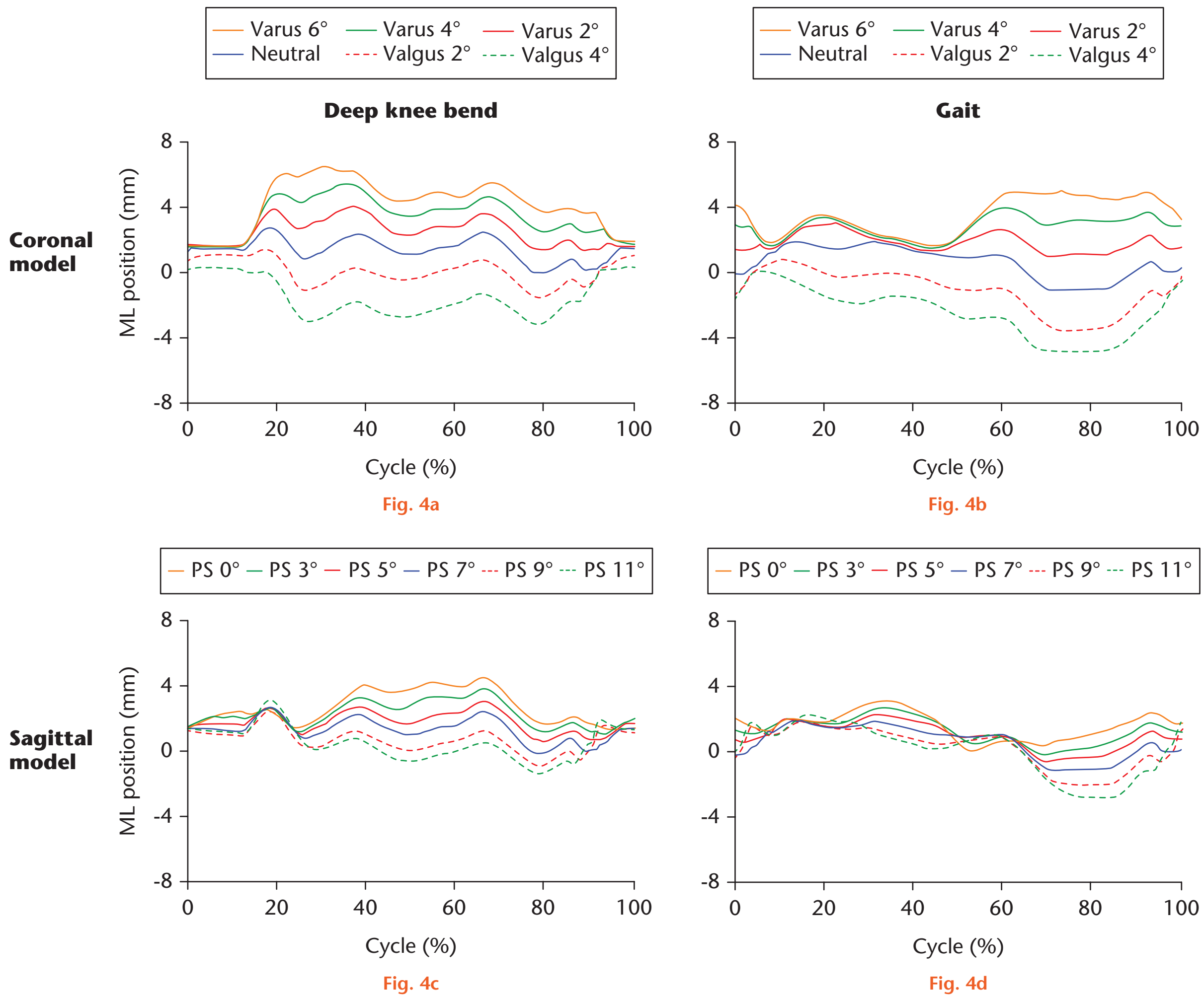
Fig. 3a Fig. 3b

a) Chart showing the evaluation method for medial/lateral (ML) translation for deep knee bend. b) Model showing the evaluation method. Red and blue circles indicate medial and lateral flexion facet centres, respectively, while the white circle indicates the midpoint between both facet centres.

and 4.2 mm compared with neutral alignment model during deep knee bend and gait motions, respectively. In the sagittal models, modification of sagittal alignment had a smaller impact on the ML position.

Medial/lateral translation. The ML translation within one cycle was the smallest for the 2° varus model in the coronal models (2.7 mm and 2.0 mm for deep knee bend and gait motions, respectively) (Fig. 5). The corresponding values increased with both increased varus alignment (4.8 mm and 3.4 mm in the 6° varus model, respectively) and increased valgus alignment (3.5 mm and 4.9 mm in the 4° valgus model, respectively). Regarding the sagittal models, ML translation was increased with a greater posterior slope (> 7°) for deep knee bend and gait motions (4.5 mm and 5.0 mm in the 11° posterior slope model, respectively). At decreased posterior slope (≤ 7°), ML translation was slightly increased at posterior slope 0° model.

Anteroposterior position. In the coronal models, medial and lateral AP positions were similar for both deep knee bend and gait motions (Fig. 6a). In the sagittal models, increased posterior slope posteriorized the medial condyle, whereas there was little effect on the position of the lateral condyle (Fig. 6b). In the 11° posterior slope model, the medial condyle was positioned 4.1 mm and 4.3 mm posteriorly at maximum compared with the 7° model during deep knee bend and gait motions, respectively. In the 0° posterior slope model, the medial condyle was



Medial/lateral (ML) positions of the femur in all models: a) coronal model, deep knee bend; b) coronal model, gait; c) sagittal model, deep knee bend; d) sagittal model, gait. PS, posterior slope.

positioned 6.6 mm and 7.5 mm anteriorly at maximum compared with the 7° model during deep knee bend and gait motions, respectively.

Cruciate ligament tension. In the coronal models, changes in coronal alignment had little impact on ACL and PCL tensions (Fig. 7a). For the sagittal alignment models, ACL tension tended to increase with a greater posterior slope, especially for the gait motion (Fig. 7b). In comparison with the maximum ACL tension with the 7° posterior slope model during the gait motion (222 N), the 11° model and the 0° model showed a 60% increase (356 N) and a 32% decrease (150 N), respectively. PCL tension decreased with a greater posterior slope during both motions. During the deep knee bend motion, the maximum PCL tension showed a slight decrease from 834 N to 776 N when increasing the posterior slope from 7° to 11°, which increased to 899 N in the 0° posterior slope model. During the gait motion, a similar trend was observed when the posterior slope was increased, although the PCL tension was generally smaller than in the deep knee bend motion.

Discussion

In the current study, changes in knee kinematics and ligament tension were analyzed during deep knee bend and gait motions with different coronal and sagittal plane tibial alignments, using computer simulation. Medialization and lateralization of the femoral position was observed in the severe varus and valgus coronal alignment models, respectively. ML translation was small in the neutral and 2° varus models. A posteriorization of the medial condyle and greater ACL tension were observed with increasing posterior slope. Based on the current kinematic and kinetic results, a 2° varus to neutral alignment in the coronal plane and a 3° to 7° posterior slope in the sagittal plane are preferable.

Despite several limitations, computer simulation is a useful tool for kinematic and kinetic analyses. It enables moving patterns to be examined in a dynamic manner, and allows slight differences between conditions (modification of individual parameters) to be explored. Regarding TKA, several computer simulation studies have reported the effect of tibial component alignment on knee

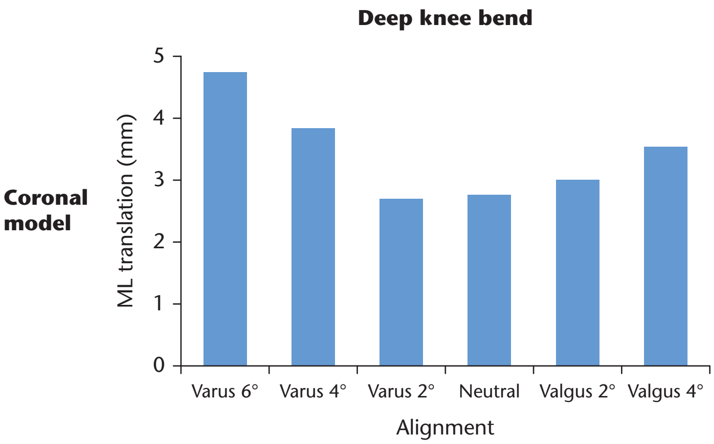


Fig. 5a

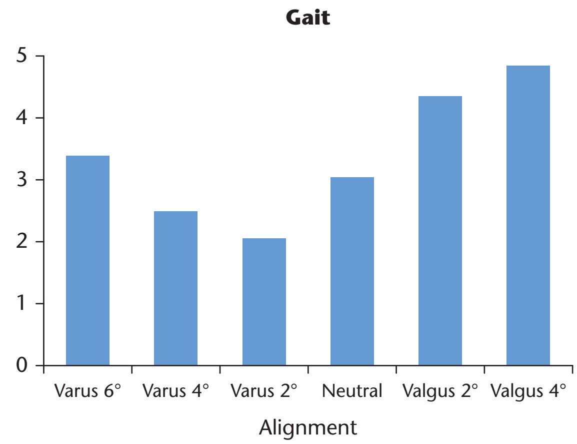


Fig. 5b

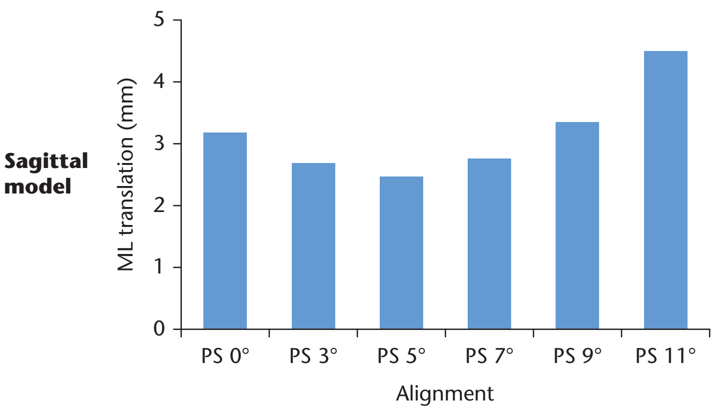


Fig. 5c

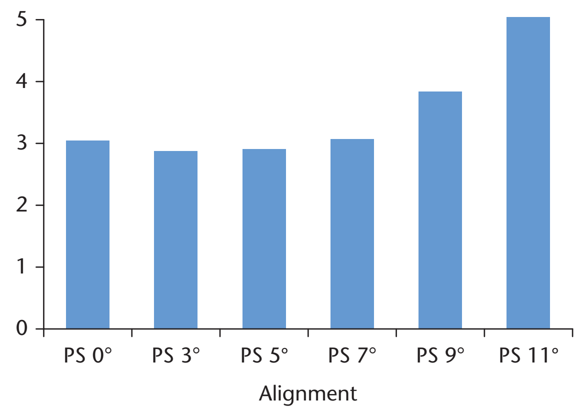


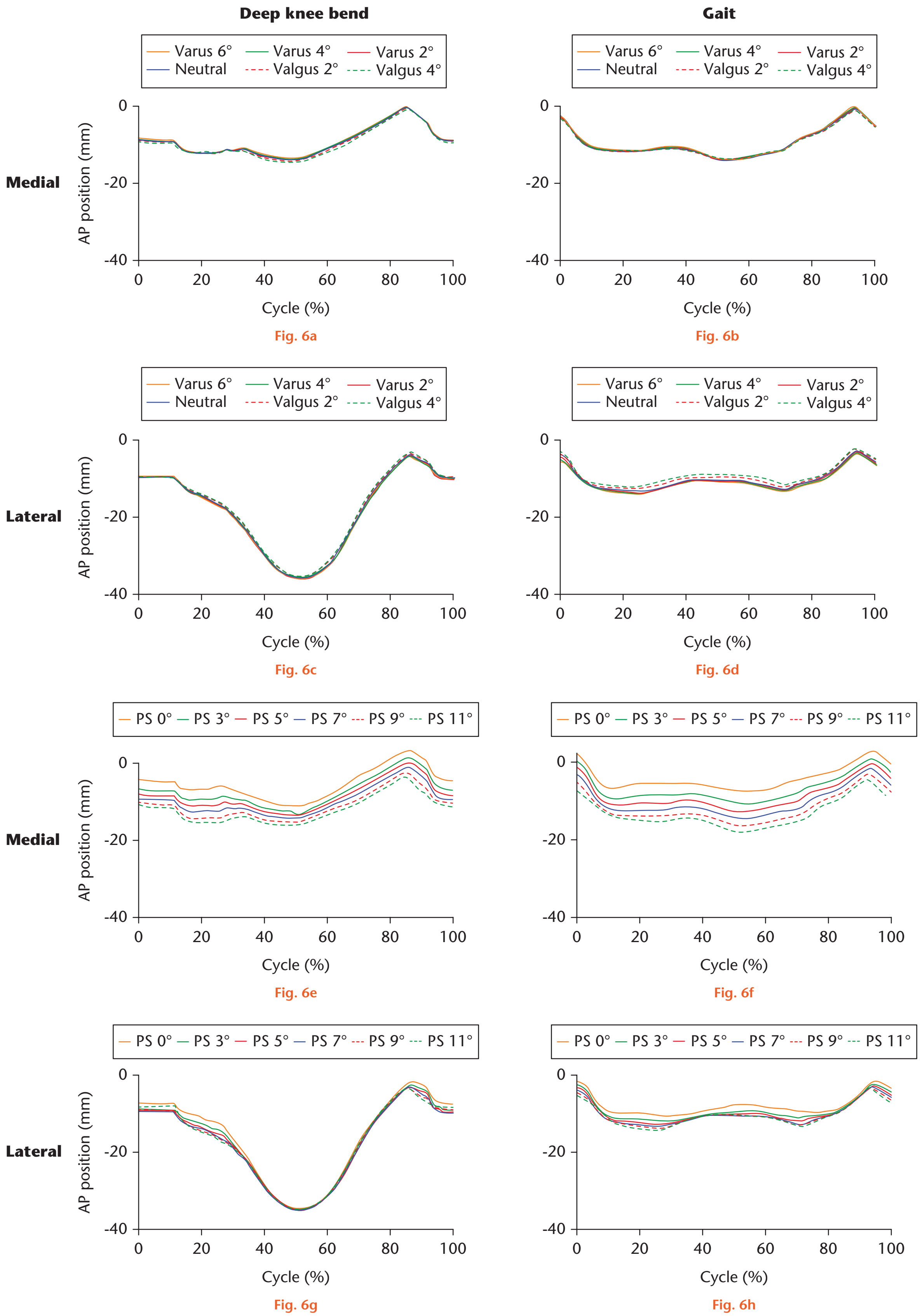
Fig. 5d

Medial/lateral (ML) translation throughout one cycle: a) coronal model, deep knee bend; b) coronal model, gait; c) sagittal model, deep knee bend; d) sagittal model, gait. AR, AL, PS, posterior slope.

kinematics and ligament tension. Varus alignment could lead the ML instability at mid-flexion of knee bending and condylar lift-off when combined with lateral joint laxity.[23] Excessive posterior slope could cause the abnormal anterior sliding of the tibial component at stair-climbing,[24] as well as progressive loosening of the TF joint gap due to a reduction in collateral ligament tension during flexion.[40] However, no computer simulation studies have yet dealt with the effects of component alignment on the kinematics after UKA. Recent advances in surgical procedures, including robotic-assisted surgery, have improved the accuracy of implant positioning.[12,41,42] This enables us to position the tibial component to almost within 2° of the planned position, even in an introduction period.[42] We think that our current study is relevant, and further detailed study using computer simulation for optimal alignment has become increasingly significant, because accurate implant positioning will be easily achieved by the advancement of technology.

Concerning coronal plane alignment, clinical reports described that excessive varus alignment could worsen the survivorship of UKA[8] and increase the risk of loosening.[43] The biomechanical effects were recently analyzed with an FEA model. Inoue et al[16] showed that valgus inclination increased stress concentration on the medial tibial metaphyseal cortices, possibly increasing the risk of medial tibial condylar fractures. Innocenti et al[15] recommended neutral tibial alignment or a slight varus alignment (3°), based on collateral ligament strain and bone and polyethylene insert stress distribution. In the current study, neutral and 2° varus alignments were preferred because of less ML translation within one cycle of the deep knee bend and gait motions. From a kinematic standpoint, this strengthens previous clinical findings and biomechanical theories.

In terms of sagittal plane alignment, previous reports showed that a greater posterior slope could be detrimental to the survivorship of UKA.[8] Hernigou and Deschamps[9] concluded that a tibial implant slope of > 7° should be avoided, because disruptions of the ACL were observed in the group with a greater posterior slope. In the current study, with increasing posterior slope of the tibia, the medial compartment of the femur was positioned posteriorly, and ACL tension was increased. In addition to ACL tension, ML translation was also increased with greater posterior slope (> 7°). Regarding small posterior slope



a) to d) Anteroposterior (AP) positions of the medial and lateral femoral condyles in coronal alignment models: a) medial, deep knee bend; b) medial, gait; c) lateral, deep knee bend; d) lateral, gait. e) to h) AP positions of the medial and lateral femoral condyles in sagittal alignment models: e) medial, deep knee bend; f) medial, gait; g) lateral, deep knee bend; h) lateral, gait. PS, posterior slope.

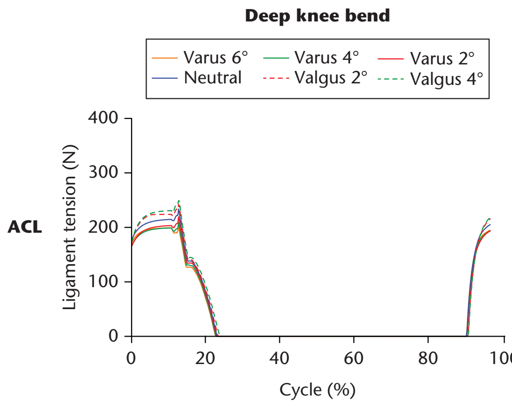


Fig. 7a

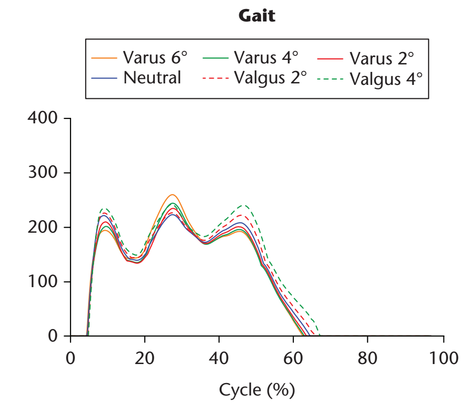


Fig. 7b

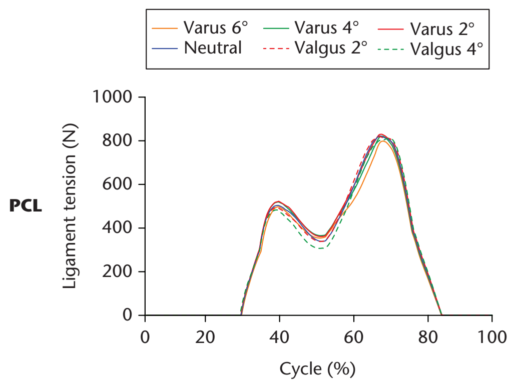


Fig. 7c

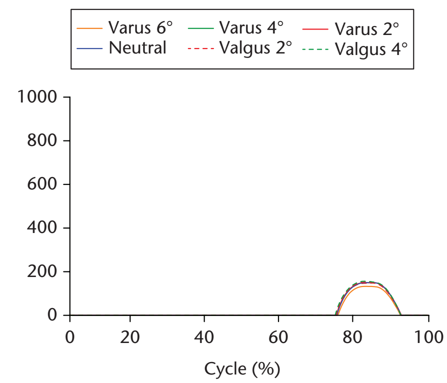


Fig. 7d

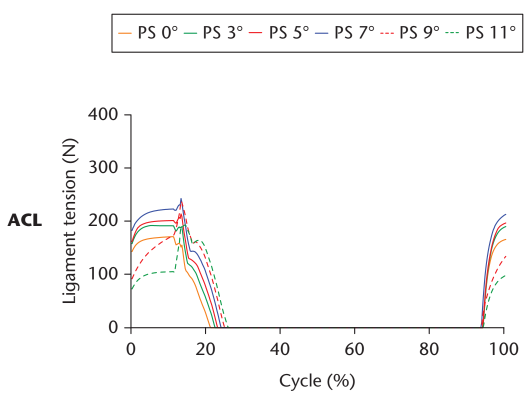


Fig. 7e

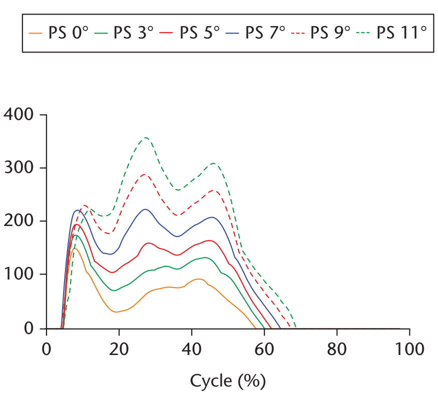


Fig. 7f

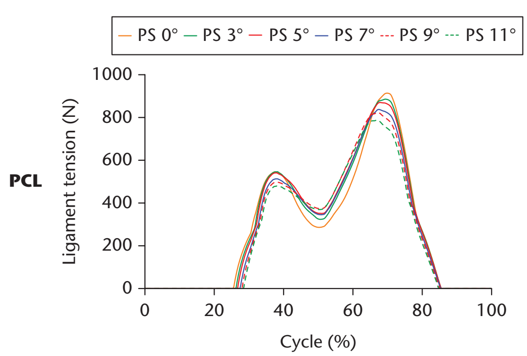


Fig. 7g

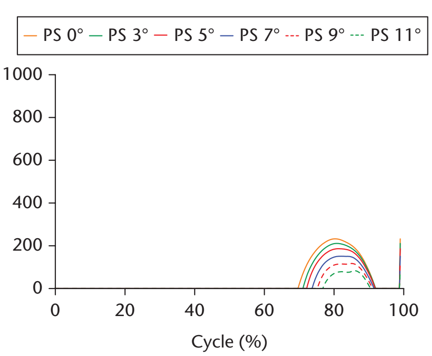


Fig. 7h

a) to d) Tension of the anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL) in coronal alignment models: a) ACL, deep knee bend; b) ACL, gait; c) PCL, deep knee bend; d) PCL, gait. e) to h) Tension of the ACL and PCL in sagittal alignment models: e) ACL, deep knee bend; f) ACL, gait; g) PCL, deep knee bend; h) PCL, gait. PS, posterior slope.

(< 7°), ML translation was slightly increased in the posterior slope 0° model. The results showed, even if not strongly, that a posterior slope of 0° could be suboptimal regarding kinematics. In summary, kinematic results of our models suggested that 3° to 7° of posterior slope were preferable, and that excessive posterior slope (> 7°) should be avoided, which supports the clinical findings.

A novel feature of the current study was the examination of ML translation. Previous methods including single plane fluoroscopic analysis, cadaver studies, and a navigation during surgery had difficulty detecting ML motion. Single-plane fluoroscopy was commonly assessed in the ML direction of the knees.[44] However, the translational error in the position of the ML direction was large because of out-of-plane motion (although it had advantages when evaluating weight-bearing conditions).[45] In cadaver and surgical navigation studies, analyses were performed in non-weight-bearing or much lower weight-bearing conditions compared with full weight acceptance. The computer simulation used in the current study was a useful tool to accurately evaluate the kinematic and kinetic effects of the coronal and sagittal plane alignment of the tibial component. Changes in both coronal plane alignment and sagittal plane alignment affected stability in the ML direction. The significance of ML stability remains to be fully elucidated in the field of UKA. However, thrust in the ML direction could cause a feeling of instability, excessive frictional force, or abnormal tibial stress distribution, possibly leading to poor clinical outcomes or implant failure.

This study had several limitations. First, only one fixed-bearing prosthesis was used in the current study. Several reports describe the difference in knee kinematics between fixed-bearing UKA and mobile-bearing UKA.[46] It is possible that other types of prosthesis could produce different results. Second, the original bone model in the current study had a neutral limb alignment with a 7° posterior slope, and only one bone model was used. Several bone models differing in preoperative alignments are necessary to fully reveal whether it is best to preserve the patient-specific native alignments or not. However, we believed that our model, which has a 7° posterior slope as a native alignment, could be one appropriate model for analysis with medial UKA, because the report showed that the mean preoperative posterior slope of 2031 knees undergoing medial UKA was 6.8° (SD 3.3°).[47] Third, neither statistical processing of the data nor calculation of standard deviations was performed due to setup of this simulation programme. However, we analyzed two motions, and the effect of alignment change was almost consistent, which could strengthen the reliability of the data. Last, this model simplified the properties of the soft tissues (e.g. ligaments and muscles). The simulation model also cannot reproduce all daily activities; however, it is difficult to reproduce the exact *in vivo* mechanical conditions with any of the methods (including mechanical tests and cadaver studies).

In conclusion, regarding tibial component of UKA, slight varus to neutral alignment in the coronal plane and 3° to 7° of posterior slope in the sagittal plane seem to be preferable. Varus (> 4°) or valgus alignment and excessive posterior slope (> 7°) caused the excessive ML translation, which could be related to a feeling of instability and could potentially have negative effects on clinical outcomes and implant durability.

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- K. Sekiguchi: Acquired the data, Wrote the manuscript.
- S. Nakamura: Designed the study, Wrote the manuscript.
- S. Kuriyama: Analyzed the data.
- K. Nishitani: Advised on the study.
- H. Ito: Advised on the study.
- Y. Tanaka: Acquired the data, Advised on the study.
- M. Watanabe: Acquired the data, Advised on the study
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