# Kinematic alignment produces near-normal knee motion but increases contact stress after total knee arthroplasty: A case study on a single implant design

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**Abstract**

*Background*: Kinematically aligned total knee arthroplasty (TKA) is of increasing interest because this method might improve postoperative patient satisfaction. In kinematic alignment the femoral component is implanted in a slightly more valgus and internally rotated position, and the tibial component is implanted in a slightly more varus and internally rotated position, than in mechanical alignment. However, the biomechanics of kinematically aligned TKA remain largely unknown. The aim of this study was to compare the kinematics and contact stresses of mechanically and kinematically aligned TKAs.

*Methods:* A musculoskeletal computer simulation was used to determine the effects of mechanically or kinematically aligned TKA. Knee kinematics were examined for mechanically aligned, kinematically aligned, and kinematically aligned outlier models. Patellofemoral and tibiofemoral contact forces were measured using finite element analysis.

*Results*: Greater femoral rollback and more external rotation of the femoral component were observed with kinematically aligned TKA than mechanically aligned TKA. However, patellofemoral and tibiofemoral contact stresses were increased in kinematically aligned TKA.

*Conclusions*: These findings suggest that kinematically aligned TKA produces near-normal knee kinematics, but that concerns for long-term outcome might arise because of high contact stresses.

Keywords: Total knee arthroplasty, Kinematically aligned TKA, Knee kinematics, Finite element analysis

## 1. Introduction

Total knee arthroplasty (TKA) is a well-established procedure for improving pain and restoring function in patients with arthritic knees. The postoperative alignment of the knee affects the longevity of the implant and postoperative knee function [1-3]. The traditional “mechanical alignment” method, which involves a cut perpendicular to the mechanical axes of the femur and tibia, is a commonly used technique; however, this method does not always result in high patient satisfaction after TKA [4,5]. Thus, there is a need for new or improved TKA techniques that provide better functional results and greater postoperative patient satisfaction.

Howell et al. recently proposed a technique called the “kinematically aligned” TKA [6,7]. This method strives to reproduce near-normal knee function by restoring premorbid joint levels and angles during TKA. To do this, the femoral component of the implant is placed in a slightly more valgus and internally rotated position, and the tibial component is placed in a slightly more varus and internally rotated position, compared with the placement of the implants in a mechanically aligned TKA [6,8-10]. Recently, a randomized controlled study has shown that kinematically aligned TKA resulted in better pain relief, postoperative function, and range of motion than mechanically aligned TKA [11]. However, a potentially serious complication of kinematically aligned TKA is that the varus alignment of the tibial component might lead to higher stresses on the tibial insert. Also, the internal rotation of the femoral and tibial components might affect the contact stresses on the patellofemoral joint. In spite of these concerns, extensive biomechanical analyses of knees that have undergone kinematically aligned TKA have not been performed.

Recent advances in computer technology have allowed detailed analyses of the human knee [12-17]. A computational kinematic knee simulator provides a simulation of continuous implant translation and contact force during daily activities such as walking and deep knee flexion, and the accuracy of knee simulation has been validated [12,16-20]. Thus, computer simulation is a useful tool for examining the factors, including surgical techniques and implant orientation and design, which may influence the kinematic function of the knee.

The purpose of this study was to compare the kinematic outcomes of mechanically and kinematically aligned TKAs using a computational knee simulator. We hypothesized that these two methods would result in different kinematic patterns. In addition, we evaluated the contact stresses resulting from these two methods using finite element analysis. We hypothesized that the stresses in the patellofemoral and tibiofemoral joints would be greater after kinematically aligned TKA than after mechanically aligned TKA.

## 2. Materials and Methods

A musculoskeletal computer simulation was used to evaluate the results of the different alignment techniques. This musculoskeletal computer model provided a dynamic simulation of the knee (LifeMOD/KneeSIM 2010; LifeModeler Inc., San Clemente, CA, USA). The model included tibiofemoral and patellofemoral contact, LCL, MCL, posterior cruciate ligament (PCL), elements of the knee capsule, quadriceps muscle and tendon, patellar tendon, and hamstring muscles. The LCL was considered to be a single fiber bundle, and the MCL was considered to consist of anterior and posterior bundles [21-24]. All ligament bundles were modeled as nonlinear springs with material properties obtained from a published report [25]. We first adjusted the insertion point of each ligament, and next determined that the stiffnesses and lengths of the ligaments at each flexion angle were similar to those reported in the literature [22,26-29]. The proximal attachment points of the LCL and MCL were defined as the most prominent points of the femoral epicondyles. The distal attachment points of the LCL and MCL were defined as the tip of the fibular head and the midpoint between the tibial attachments of the anterior and posterior bundles, respectively. The PCL comprised two bundles [30,31]; its femoral attachments were defined as the anterior area of the medial intercondylar wall, and its tibial attachments were defined as the posterior intercondylar fossa, with the anterior-lateral bundle anterior to the posterior-medial bundle. The stiffness coefficients of the LCL, MCL-anterior, MCL-posterior, and PCL were defined based on reported values [23,31,32]. Finally, we adjusted the attachment points of each ligament, and their slack during weight-bearing deep knee flexion, so that their lengths were similar to those reported in a previous cadaver study [29].

The KneeSIM program uses the implant geometry to analyze the performance of the femoral, tibial, and patellar components, as well as the polyethylene inserts, under a variety of conditions. We have previously reported the kinematics and kinetics of the knee implants using this computer simulation [33,34]. In the present study, the model parameters for a fixed-bearing, cruciate-retaining, total left knee (NexGen CR-Flex; Zimmer, Warsaw, IN, USA) were imported into the program, and tested during a simulated weight-bearing deep knee bend using an Oxford-type knee rig as described previously [33]. The femoral component of the implant had a multi-radius, asymmetrical condyle design, while the design of the tibial insert included a low anterior lip and symmetrical condyles. Figure 1 shows the structure of the KneeSIM model. During movement, the hip joint was allowed to flex and extend and to slide vertically, while the ankle joint was allowed free translation in the medial-lateral direction and free varus-valgus and axial rotation.

Previous studies have reported that peak tibiofemoral contact force in normal or TKA patients during a squat motion increased by up to 4–6 times body weight [13,35,36]. The model parameters were set so that the constant vertical force was converted to a 4,000N load on the bicondylar joint of the knee, which corresponds to approximately 5 times a body weight of 80 kg. This force was applied at the hip and its active driving elements were the forces of the quadriceps and hamstring muscles. The simulation was driven by a controlled actuator arrangement similar to a physical machine, such as an Oxford-type knee rig. A closed-loop controller applied tension to the quadriceps and hamstrings to match firing to a prescribed flexion angle at each point, and cocontraction between these muscles was defined. The models were subjected to a 4.5 sec cycle for a squat motion (0°–130° flexion).

The mechanical axis of the femur was defined as the line from the center of the femoral head to the center of the knee joint. The mechanical axis of the tibia was defined as the line extending from the center of the tibiofemoral joint to the center of the talocrural joint. The femoral component was aligned perpendicular to the mechanical axis of the femur in the coronal plane and parallel to the distal anatomical axis of the femur in the sagittal plane. The tibial component was aligned perpendicular to the mechanical axis of the tibia with 7° of posterior tibial slope. The rotational alignments of the femoral and tibial components were determined based on the femoral epicondylar axis and the tibial anteroposterior (AP) axis, respectively [37]. When the implants were positioned in this manner they were defined as being in neutral alignment.

In the current study, three different alignment models were examined: (1) femoral and tibial component in neutral alignment (mechanical alignment model); (2) femoral component with 3° valgus and 3° internal rotation, and tibial component with 3° varus and 3° internal rotation (kinematic alignment 3° model); and (3) femoral component with 5° valgus and 5° internal rotation, and tibial component with 5° varus and 5° internal rotation (kinematic alignment 5° model: outlier model).

### *2.1. Kinematic analysis*

All kinematic measurements were performed at 0°, 30°, 45°, 60°, 90° and 120° of knee flexion. The medial and lateral centers of the femoral condyles were used as geometric reference points, as previously described [19]. The axis of the femoral component was defined as the line connecting the medial and lateral reference points. For the tibiofemoral joint, the AP positions of the medial and lateral reference points were measured using the coordinate system of the tibial component. The axial rotations of the femoral and patellar components were determined relative to the tibial component. For the patellofemoral joint, patellar shift indicated the position of the patella relative to the tibial component. Patellar tilt was defined as the angle of the patella relative to the femoral component, which was defined as positive if the patellar component was externally rotated relative to the femoral component.

### *2.2. Finite element (FE) analysis*

Patellofemoral and tibiofemoral contact forces were measured under the same test conditions. The position of the components, and the magnitude and direction of each force, computed by KneeSIM at 30°, 60° and 90° of knee flexion, were used in the finite element (FE) analysis. Contact stresses on the patellar component and on the tibial insert against the femoral component interfaces were calculated using three-dimensional FE analysis. FE simulations were performed using ANSYS Workbench ver. 12.0.1 (ANSYS, Inc., Canonsburg, PA, USA). The femoral component and tibial insert were both modeled as rigid bodies. The Young's modulus of the femoral component was set at 240 GPa, which is consistent with data for Co–Cr–Mo alloy femoral components. The tibial insert and patellar polyethylene component were modeled as nonlinear elastoplastic materials, as described in a previous study [38]. The mesh of the femoral component and the tibial insert were generated based on 10 node quadratic tetrahedral elements sized at 0.8 mm. The generated mesh contained a total of 687152 nodes as a result of 434348 total elements. The mesh of the patellar component was generated based on 20 node quadratic hexahedral elements sized at 0.8 mm. The generated mesh contained a total of 793803 nodes as a result of 430318 total elements. The meshed model is shown Figure1 (right). Contact was considered to occur when the perpendicular distance between the surfaces of the femoral component and the tibial insert was < 0.3mm.

## 3. Results

The AP positions of the medial and lateral femoral reference points from 0° to 120° of knee flexion are shown in Fig. 2. All three models exhibited anterior translation of the femoral component relative to the tibia during the early flexion phase, and then posterior translation as flexion increased. The anterior translation from 0° to 30° of flexion was similar bilaterally in all three models. The lateral posterior translation from 0° to 120° of knee flexion was greater in the kinematic alignment models than in the mechanical alignment model (-10.6, -12.0, and -12.8 mm in the mechanical alignment, kinematic alignment 3°, and kinematic alignment 5° models, respectively). However, the corresponding values on the medial side were smaller in the kinematic alignment models than in the mechanical alignment model (-12.6, -11.0, and -10.6 mm in the mechanical alignment, kinematic alignment 3°, and kinematic alignment 5° models, respectively). A normal axial rotation pattern was observed in the kinematic alignment models, especially in the 5° rotation model, from 0° to 120° of knee flexion. In the mechanical alignment model, however, the femoral component rotated internally from 0° to 120° of knee flexion, showing a reverse rotation pattern.

The effects of kinematically aligned TKA on the patellofemoral joint are shown in Fig. 3. Patellar maltracking was observed during early flexion in the kinematic alignment models. The patellar component shifted laterally in the kinematic alignment models during the early flexion phase; this lateral shift was gradually reduced with increasing knee flexion (Fig. 3). Similar patellar tracking was observed from mid- to full-flexion in all three models. In the kinematic alignment models, the patella tilted more externally relative to the tibial component at 0° and 30° compared with the mechanical alignment model, whereas similar tilts occurred during mid to deep flexion in all three models (Fig. 3). The patellar tilt was considerably greater at 0° and 30° of knee flexion in the kinematic alignment 5° rotation model than in the other models.

Finite element analyses of the patellofemoral joint are shown in Fig. 4. At the lateral side, the maximum peak contact stress in the kinematic alignment 5° rotation model was 88 Mpa at 30° of knee flexion, which was 2.7 times greater than the maximum peak contact stress in the mechanical alignment model. Similarly, the corresponding value at 60° of knee flexion in the 5° rotation model was 1.3 times greater than in the mechanical alignment model. During deep flexion, the peak contact stresses in all three models were similar. At the medial side, the maximum peak contact stress was similar in the 5° rotation model and mechanical alignment model.

Finite element analyses of the tibiofemoral joint are shown in Fig. 5. The peak contact stresses in the kinematic alignment models were greater than in the mechanical alignment model at all flexion angles. In the kinematic alignment models, the peak contact stresses on both sides tended to increase with greater varus tilt of the tibial component. The peak contact stresses at 30° and 60° of knee flexion in the 5° rotation kinematic alignment model were up to twice as large as in the mechanical alignment model.

## 4. Discussion

Kinematically aligned TKA strives to replicate the premorbid joint line and morphology. Several studies have reported that the kinematic alignment method results in more near-normal knee kinematics than mechanical alignment, and that patients experience this motion as natural [6,39,40]. A randomized controlled study has shown that kinematically aligned TKA achieves better flexion and higher clinical outcome scores than mechanically aligned TKA [11]. However, the biomechanical advantages and disadvantages of kinematically aligned TKA remain unclear. In the current study, we investigated the kinematics and kinetics of the knee after kinematically aligned TKA and compared these with the results achieved using a mechanically aligned model. In our computer simulation, mechanically aligned TKA resulted in internal rotation of the femoral component relative to the tibia, which is consistent with previous findings [14]. In contrast, kinematically aligned TKA achieved near-normal knee kinematics, including greater rollback of the lateral femoral condyle and external rotation of the femoral component relative to the tibia. The results of the current study suggest that restoring the joint line to close to its normal position can reproduce near-normal joint kinematics. Thus, the better clinical results of kinematically aligned TKA found in previous studies might be associated with the reproduction of more normal knee kinematics [7,11].

In the kinematic alignment model, the internal rotation of the femoral and tibial components resulted in a lateral shift and tilt of the patellar component during early knee flexion and also increased patellofemoral contact stresses, which were up to 267% greater in the 5° rotation model than in the mechanically aligned model at 30° of knee flexion. Although the 10-year longevity after TKA using any endpoint is over 95% [41-43], patellofemoral complications are still one of the most common problems leading to revision TKA [44]. Among the patellofemoral complications, patellar maltracking is a major problem that causes subluxation with increased polyethylene wear, and previous studies have shown that internally rotated femoral and tibial components cause patellar maltracking [45]. Most current implants are designed to be aligned perpendicular to the Whiteside line, which is determined by patellar groove anatomy, or parallel to the epicondylar axis. With currently used implants, a near-normal patellar groove can be replicated only when the femoral component is aligned with the Whiteside line or the transepicondylar axis. Therefore, kinematically aligned TKA using a conventional implant may increase the risk of patellofemoral joint complications.

The varus tilt of the tibial component is another important issue in kinematically aligned TKA. In this study, greater varus tilt of the tibial component was correlated with greater tibiofemoral contact stress. These findings suggest that the varus tilt of the tibial component might cause more polyethylene wear and component loosening, even if the overall alignment is neutral. Srivastava et al. recently examined 16 modern tibia inserts retrieved during revision surgery and found that varus alignment of the tibial component was associated with increased medial and total compartment wear, even when overall limb alignment was almost ideal [46]. They concluded that varus tibial malalignment of as low as 3° may result in accelerated wear. Ritter et al. suggested that correction of the alignment of the femoral component to compensate for a varus tibial component increases the risk of implant failure [47]. Therefore, the kinematically aligned TKA with a varus aligned tibial component may increase the risk of polyethylene wear and component loosening.

It is well known that overall leg alignment affects the longevity of the TKA. Kinematically aligned TKA does not aim for neutral alignment, but tries to restore premorbid alignment; therefore, concerns remain for the "constitutional varus" knee. Bellemans et al. showed that the incidence of natural limb alignment of 3° varus or more is approximately 32% in men and 17% in woman, and this is defined as constitutional varus [48]. For the constitutional varus knee, use of the kinematic alignment method could result in a postoperative alignment of more than 3 degrees varus because kinematic alignment restores the premorbid joint alignment. Fang et al. found that overall varus alignment was associated with a 6.9 times greater risk of medial tibial collapse compared with overall proper alignment [1] and concluded that the ideal coronal alignment to achieve the best TKA survival is 2.4° to 7.2° valgus. This finding suggests the importance of overall neutral limb alignment for implant longevity.

This study had several limitations. First, the TKA system used in this study had a multiple radius femoral component, so a single radius femoral component might move differently. However, the effects of a varus aligned tibial component on the tibiofemoral joint, and the effects of an internally rotated femoral component on the patellofemoral joint, would not significantly differ for single or multiple radius components. Second, a computational model cannot reproduce all inherent soft tissue conditions. However, it is also difficult to reproduce the exact *in vivo* mechanical loading in cadaver studies. In addition, fluoroscopic studies do not allow comparison of the motion of different alignments in the same individual, whereas a computational model does permit this comparison. Recently, Mihalko et al. clearly showed that a computational model with varying implant positioning gave results comparable to those from TKA fluorokinematic data [14]. Morra et al. also showed that the patterns of damage to tibial inserts predicted using computational finite element analysis correlated with physical measurements of contact area and stress, laboratory wear simulations, and damage patterns found after clinical retrievals [20]. Hence, a computational model is suitable for comparison of different alignment techniques in dynamic conditions. Third, in the current study, the PCL stiffness was created without any release, and this was applied to any model set-ups. Thus, we cannot assess the influence of PCL balancing. Finally, the knee kinematics in the current study were analyzed with reference to the tibial component. The results might have been different if the tibia itself had instead been used as the reference.

In conclusion, kinematically aligned TKA achieves sufficient femoral rollback and external rotation of the femoral component. These results suggest that kinematically aligned TKA results in close-to-more normal knee kinematics, providing better clinical results than mechanical alignment TKA. However, contact stresses on the patellofemoral and tibiofemoral joints are increased considerably after kinematically aligned TKA. This might result in reduced implant longevity if the current prostheses commonly used are implanted with the kinematically aligned method.

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## Figure Legends

Fig. 1. The insertion point of each ligament, the boundary conditions, and the KneeSIM (left) and finite element (right) models. The patellofemoral force was calculated as a single force on the patellar component, and the tibiofemoral forces were calculated as medial and lateral forces on the tibial insert.

Fig. 2. Images showing the anteroposterior positions and angles of the femoral reference axis at 0°, 30°, 45°, 60°, 90°, and 120° of knee flexion in the three models.

Fig. 3. Diagrams showing the patellar shifts and tilts at 0° and 30° flexion in the three models. (A) Patellar shift in the axial plane was defined as the distance between the mediolateral center of the femoral and patellar components.

Fig. 4. Images showing the peak contact stress on the patellofemoral joint in the three models. Colors of the contact areas indicate the degree of peak contact stress.

Fig. 5. Images showing the peak contact stress on the tibiofemoral joint in the three models. Colors of the contact areas indicate the degree of peak contact stress.

**Fig. 1**

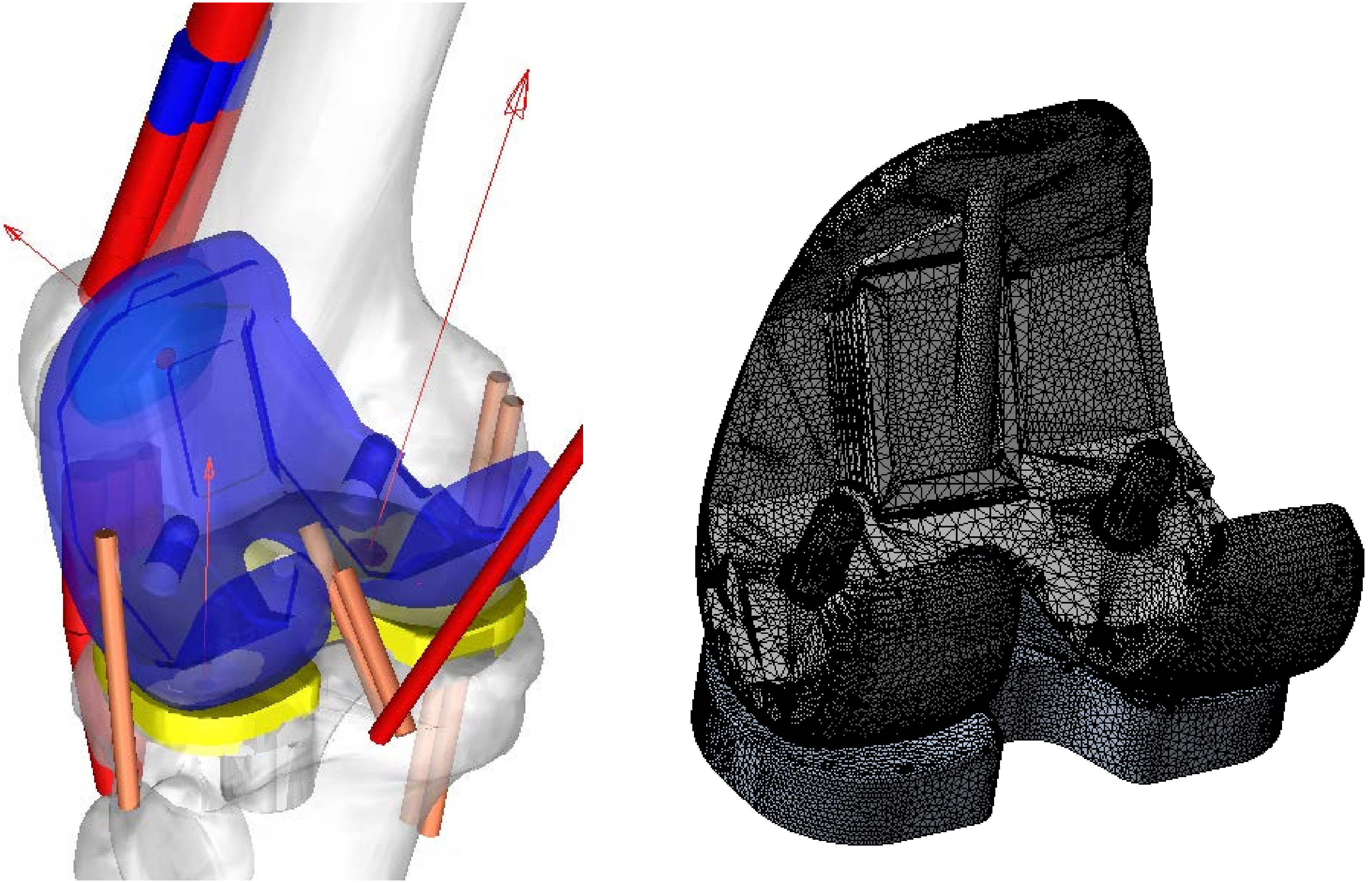
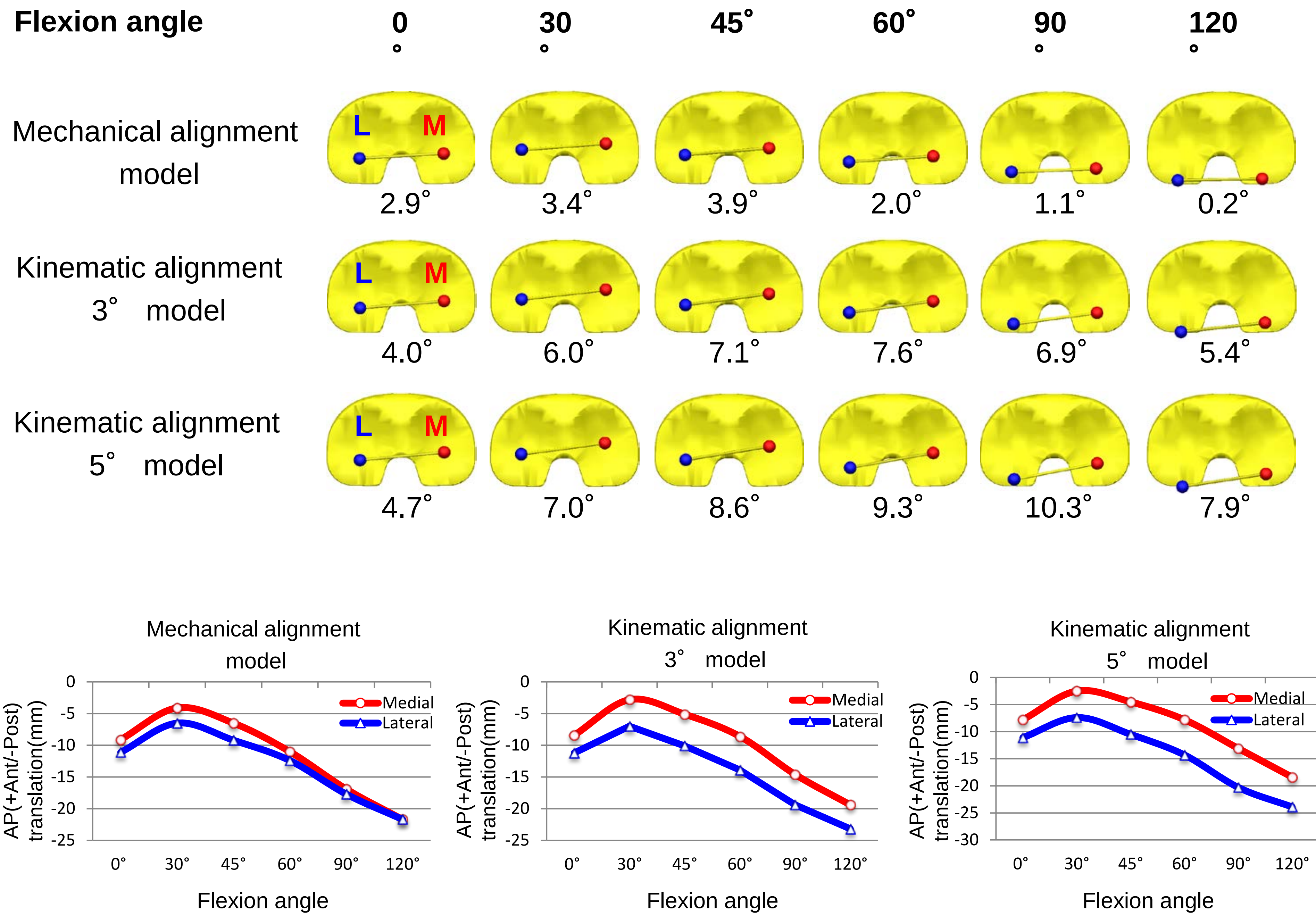
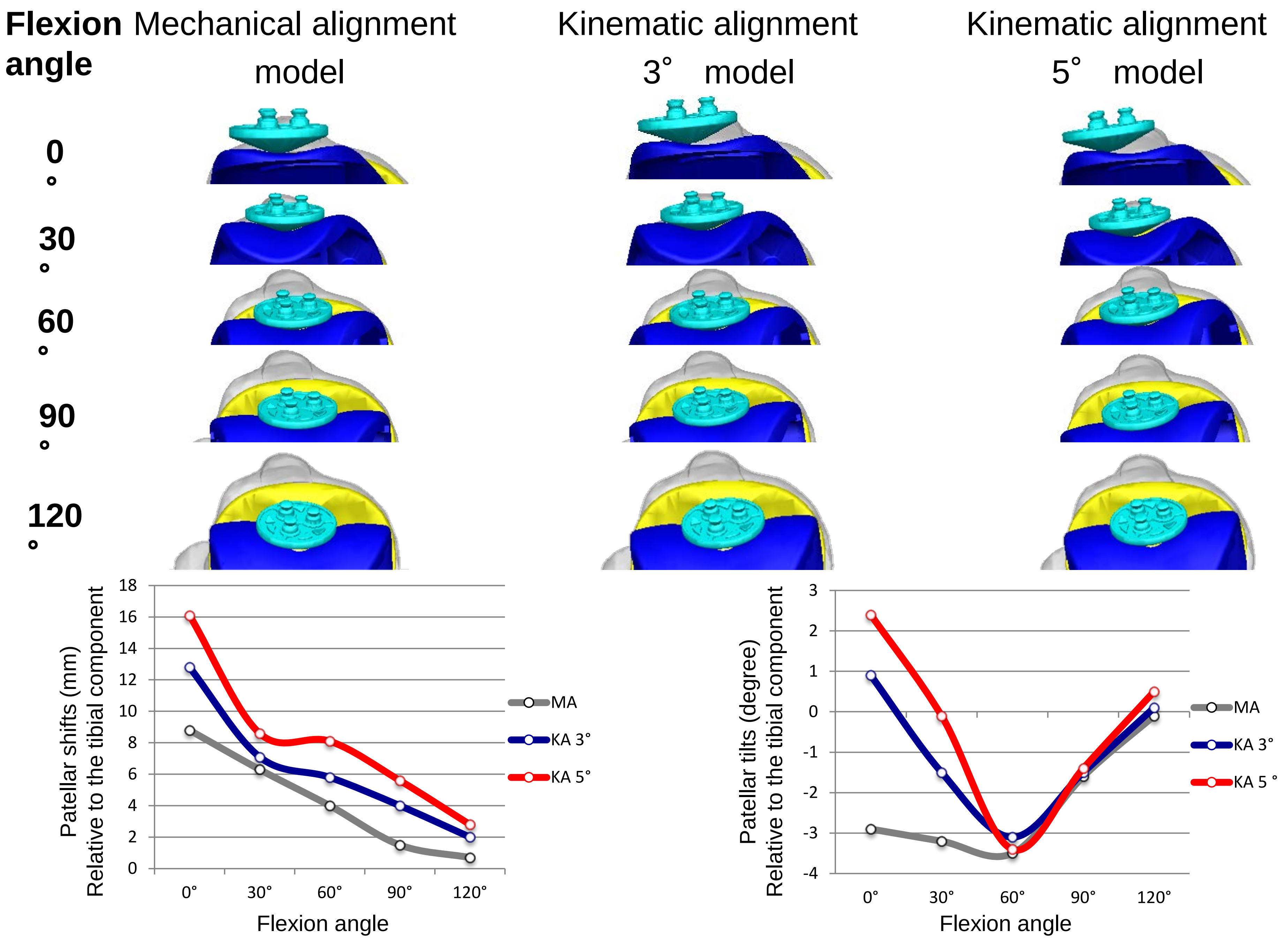


Fig. 2



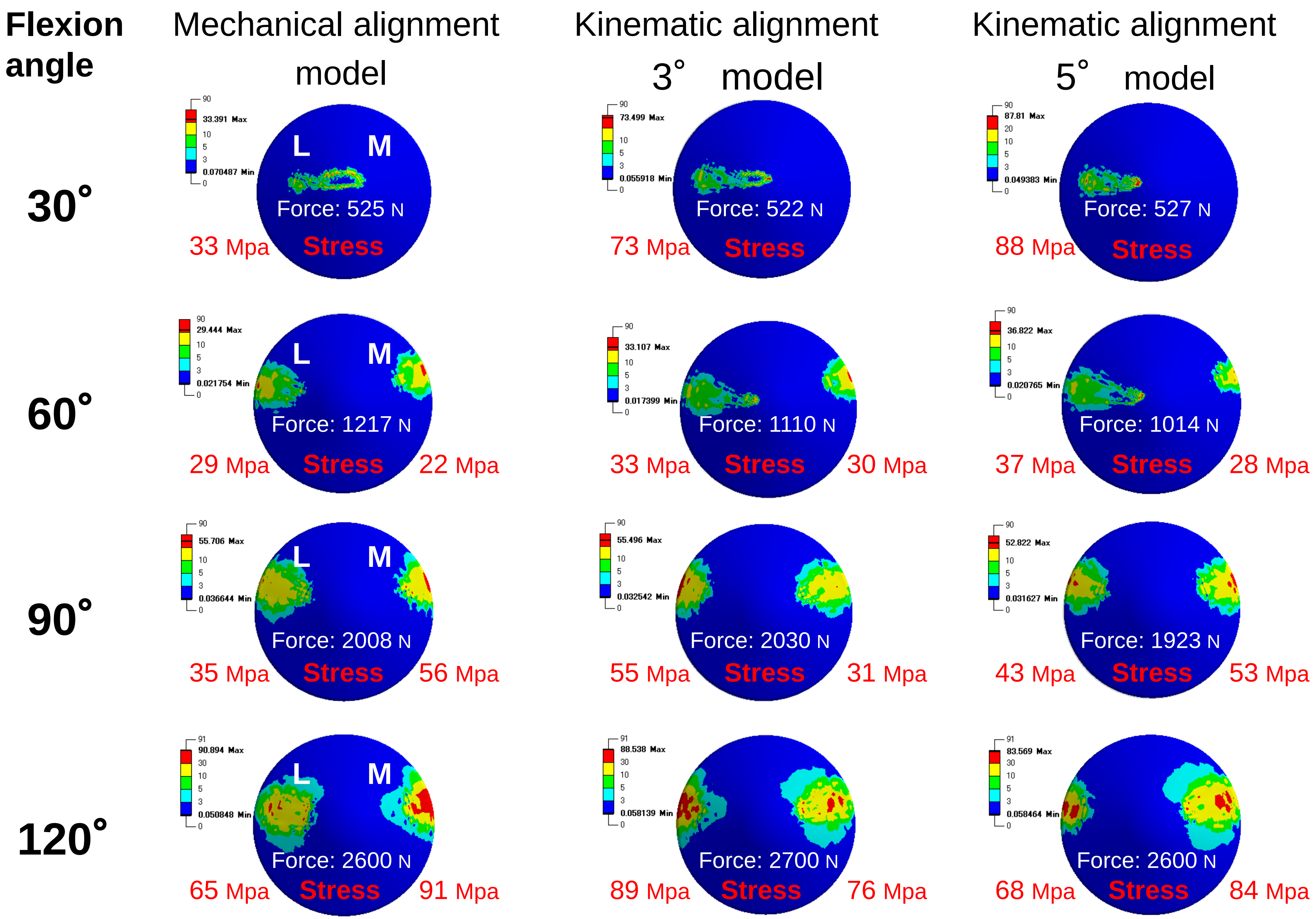
Flexion angle
0°
30°
45°
60°
90°
120°
Mechanical alignment model
L
M
2.9°
3.4°
3.9°
2.0°
1.1°
0.2°
Kinematic alignment 3° model
L
M
4.0°
6.0°
7.1°
7.6°
6.9°
5.4°
Kinematic alignment 5° model
L
M
4.7°
7.0°
8.6°
9.3°
10.3°
7.9°
Mechanical alignment model
Kinematic alignment 3° model
Kinematic alignment 5° model
AP(+Ant/-Post) translation(mm)
0
-5
-10
-15
-20
-25
-30
Medial
Lateral
0°
30°
45°
60°
90°
120°
Flexion angle

Fig. 3



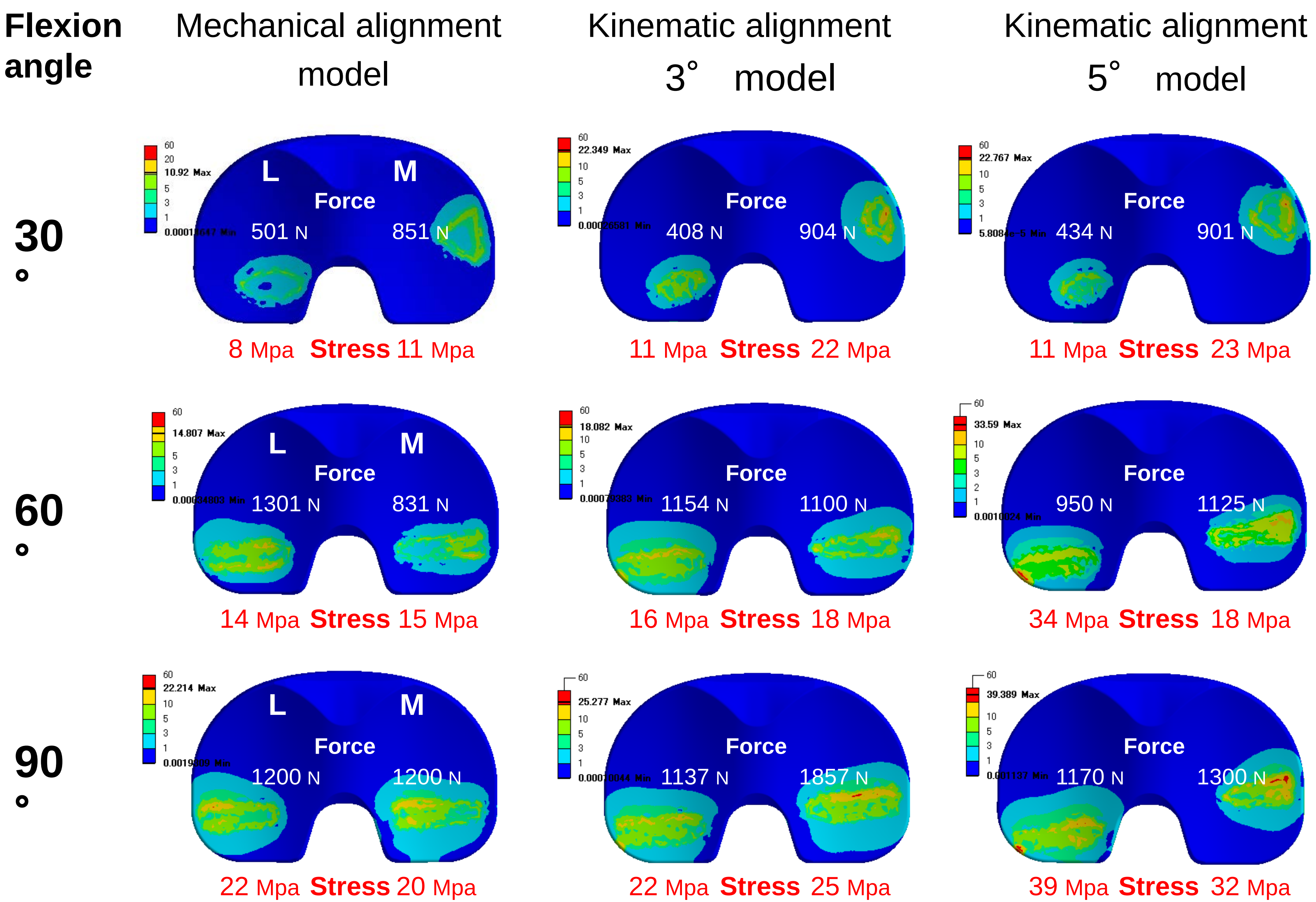
Flexion angle
Mechanical alignment model
Kinematic alignment 3° model
Kinematic alignment 5° model
0°
30°
60°
90°
120°
Patellar shifts (mm)
Relative to the tibial component
18
16
14
12
10
8
6
4
2
0
0°
30°
60°
90°
120°
Flexion angle
MA
KA 3°
KA 5°
Patellar tilts (degree)
Relative to the tibial component
3
2
1
0
-1
-2
-3
-4
0°
30°
60°
90°
120°
Flexion angle
MA
KA 3°
KA 5 °

Fig. 4



Flexion angle
Mechanical alignment model
Kinematic alignment 3° model
Kinematic alignment 5° model
30°
60°
90°
120°
L
M
Force: 525 N
Force: 522 N
Force: 527 N
Force: 1217 N
Force: 1110 N
Force: 1014 N
Force: 2008 N
Force: 2030 N
Force: 1923 N
Force: 2600 N
Force: 2700 N
Force: 2600 N
Stress
33 Mpa
73 Mpa
88 Mpa
29 Mpa
22 Mpa
33 Mpa
30 Mpa
37 Mpa
28 Mpa
35 Mpa
56 Mpa
55 Mpa
31 Mpa
43 Mpa
53 Mpa
65 Mpa
91 Mpa
89 Mpa
76 Mpa
68 Mpa
84 Mpa

Fig. 5



Flexion angle
Mechanical alignment model
Kinematic alignment 3° model
Kinematic alignment 5° model
30°
L
M
Force
501 N
851 N
8 Mpa Stress 11 Mpa
Force
408 N
904 N
11 Mpa Stress 22 Mpa
Force
434 N
901 N
11 Mpa Stress 23 Mpa
60°
L
M
Force
1301 N
831 N
14 Mpa Stress 15 Mpa
Force
1154 N
1100 N
16 Mpa Stress 18 Mpa
Force
950 N
1125 N
34 Mpa Stress 18 Mpa
90°
L
M
Force
1200 N
1200 N
22 Mpa Stress 20 Mpa
Force
1137 N
1857 N
22 Mpa Stress 25 Mpa
Force
1170 N
1300 N
39 Mpa Stress 32 Mpa