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Effects of femoral component placement on the balancing of a total knee at surgery

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ABSTRACT

Misalignment and soft-tissue imbalance in total knee arthroplasty (TKA) can cause discomfort, pain, inadequate motion and instability that may require revision surgery. Balancing can be defined as equal collateral ligament tensions or equal medial and lateral compartmental forces during the flexion range. Our goal was to study the effects on balancing of linear femoral component misplacements (proximal, distal, anterior, posterior); and different component rotations in mechanical alignment compared to kinematic alignment throughout the flexion path. A test rig was constructed such that the position of a standard femoral component could be adjusted to simulate the linear and rotational positions. With the knee in neutral reference values of the collateral tensions were adjusted to give anatomic contact force patterns, measured with an instrumented tibial trial. The deviations in the forces for each femoral component position were then determined. Compartmental forces were significantly influenced by 2 mm linear errors in the femoral component placement. However, the errors were least for a distal error, equivalent to undercutting the distal femur. The largest errors mainly increase the lateral condyle force, occurred for proximal and posterior component errors. There were only small contact force differences between kinematic and mechanical alignment. Based on these results, surgeons should avoid overcutting the distal femur and undercutting the posterior femur. However, the 2–3 degrees varus slope of the joint line as in kinematic alignment did not have much effect on balancing, so mechanical or kinematic alignment were equivalent.

1. Introduction

During passive flexion of the normal knee, the geometry of the joint surfaces, menisci, and ligaments allow for smooth and stable motion, but with laxity in all directions due to the elasticity of the structures. However, the laxity has boundaries due to tightening of the soft tissues such that adequate stability is maintained (Markolf et al., 1978). The ideal goal of TKA is to restore the same laxity and stability as in the normal knee which requires that the bearing surfaces of the femoral component replicate those of the knee it is resurfacing. This is difficult due to the geometric differences between individuals and the surgical precision required (Ewing et al., 2016). Another deficiency is that most implants used today sacrifice one or both cruciate ligaments and the stiffness of the bearing surfaces is higher compared to the articular cartilage (Liu et al., 2010).

After insertion of total knee components there is usually some imbalance of the soft tissues as indicated by different laxities (Ewing et al., 2016). Balancing is obtained in different ways, such as using spacer blocks or distractors, or instrumented trial components to measure the contact forces on the condyles (D'Lima and Colwell, 2017). Adjustment of tight soft tissues is achieved by releasing fibers from bony attachments, making small punctures in the body of the ligament, or by adjusting bone cuts (Gustke et al., 2017). Palpitation guided by information provided by any measurement device is used to detect tight fibers. Normal ligament length patterns and pre-tensions during flexion should produce normal contact forces on the lateral and medial condyles (Hosseini et al., 2015; Verstraete et al., 2017). A misplaced femoral

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component would change the length patterns of the collateral ligaments, altering the tensions and forces. This was confirmed in cadaveric studies where major changes in the contact forces occurred for only 2 mm or 2° changes in position of the components (Walker et al., 2014).

Variations in tensions and forces can also result from different surgical alignment techniques. Kinematic alignment aims to restore the natural articulating surface by aligning the distal and posterior femoral joint line of the femoral component to the transverse axis of the femur while in mechanical alignment the resection amounts on the lateral and medial condyles of the femur are unequal both distally and posteriorly (Schiraldi et al., 2016).

The purpose of the study was to determine the effects on contact forces, and therefore balancing, that result from misplacement of the femoral component or from different alignment techniques. We modeled placement errors by varying the position of the femoral component by 2 mm while bone cuts were varied to model kinematic and mechanical alignment. The data could then be used as guidelines for surgeons in obtaining a balanced knee and improving kinematics.

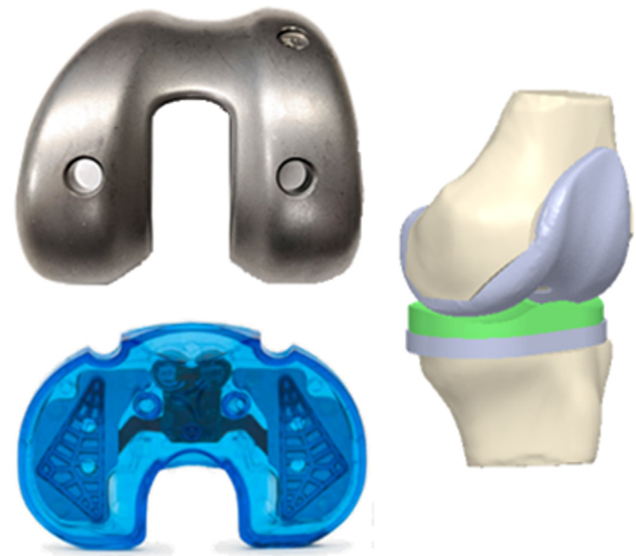

Fig. 1. Total knee Stryker Triathlon®, Verasense™ tibial sensor and assembled Total Knee System.

2. Materials and methods

A test rig was built to measure the changes in compartmental and ligament forces on a total knee during flexion/extension. The ligaments modeled were the lateral collateral (LCL) and medial collateral (MCL). A standard symmetric total knee design (Triathlon, Stryker Orthopedics, Mahwah, NJ, US) was chosen for the experiments (Fig. 1). For the tibial component, an instrumented tibial sensor (Verasense for Stryker Triathlon, Orthosensor Inc., Dania Beach, FL, US) previously employed to study balancing (Salvadore et al., 2018; Meere et al., 2016; Cho et al., 2018), was used to measure the medial and lateral compartmental forces. The geometry of the tibial sensor bearing surface reproduced that of the Triathlon®.

2.1. Design of the rig

An innovative test rig was designed to assess balance and imbalance conditions in the knee, evaluating the compartmental loads during flexion (Fig. 2). The design specifications of the rig were:

- Femur allowed six (6) degrees of freedom (DOF) relative to the tibia.
- Range of flexion: 0–110°.
- Typical symmetric total knee design for the femoral and tibial components.
- Adjustability of the ligament pre-tension load, stiffness, and attachment points on the femur.
- Ability to use two medial collateral ligament fiber bundles.
- Monitoring ligament and contact forces during flexion.

Flexion was driven by a stepper motor (Nema 34 Hybrid Stepper Motor). The motor was mounted on a linear guide that allowed for ±3 mm of medial/lateral translation. The motor shaft was coupled with a double-universal joint that connected to the femoral block through a rotational lever. The coupling allowed unconstrained proximal/distal, anterior/posterior translation and internal/external and varus/valgus rotation of the femoral block up to ±5°. Stops were provided to prevent excessive posterior displacements of the femoral condyles, consistent with the motion measured on the Triathlon knee using fluoroscopy (Grieco et al., 2016).

2.2. Total knee model

Blocks were designed to represent the femur and tibia, for mounting the femoral and tibial components. The femur block was designed to align the TKA femoral component to represent kinematic alignment, with equal medial and lateral cuts both distally and posteriorly. The rotational axis of the single-radii femoral component was aligned to the center of the motor shaft (Gomez-Barrena et al., 2010; Cook et al., 2012). The tibia block was designed to model a resected tibia with a 5° slope typical of a posterior cruciate resection implant. The fibula was modeled separately to provide the distal anatomic location of the LCL attachment point.

2.3. Ligament attachment points

The variation of the pre-tension forces and stiffness of the collateral ligaments was achieved with adjustable fixtures at the lateral and the medial sides of the femur, allowing easy linear translation of the attachment points (Fig. 3). The ligaments were connected to these fixtures using steel wires. A clamp system was used to fix the material that modeled the ligaments. Load cells were attached to the clamps for ligament force monitoring. The adjustable parts where the ligaments were attached to the femoral and tibial blocks allowed up to ±3 mm translation in the anterior, posterior, distal and proximal linear directions. This simulated femoral component placement errors during surgery. The normal position of the attachment points on the femur, tibia and fibula were obtained from morphological studies of average male knees (Ozkurt et al., 2016).

The MCL was modeled as two fibers that represented the most anterior and posterior fibers of the ligament. The stiffness values were divided between the two fibers to model the total stiffness of the MCL. The LCL was considered as one single fiber. The attachment point was represented as the centroid of the anatomic attachment point and the length of the fiber corresponded to morphological study values (Otake et al., 2007).

2.4. Modeling of the ligaments

Previous studies have used elastomeric bands to model ligaments, due to the non-linear stress-strain behavior of the material

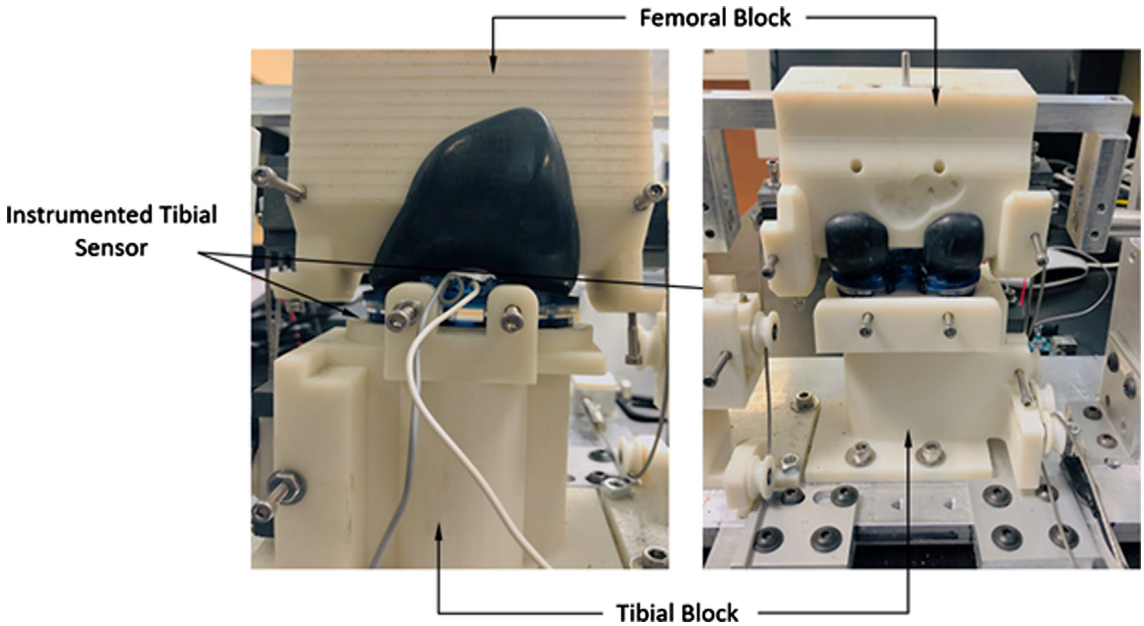

Fig. 2. Anterior (left) and posterior (right) views of the assembly of the femoral component and the tibial sensor component on the femoral and tibial blocks. The blocks include the collateral ligament attachment fixtures.

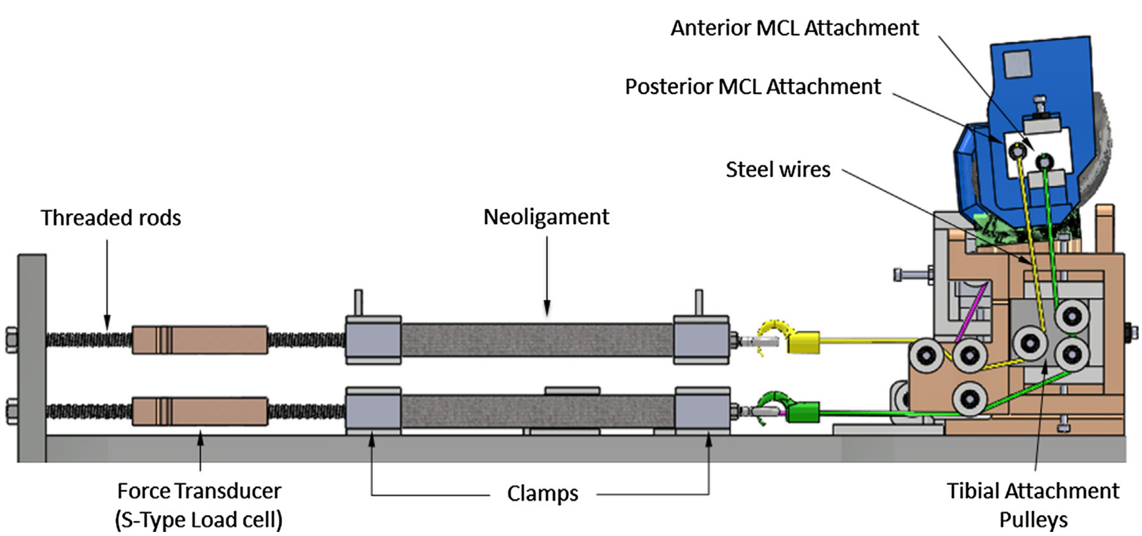

Fig. 3. Side view of the test rig depicting the medial collateral ligament attachments and clamp system.

(Lowry et al., 2016). The material selected to model the collateral ligaments was polyethylene terephthalate manufactured as woven multifilament polyester fibers, named Neoligaments (Xiros Ltd., Leeds, UK). This has been clinically used for the reconstruction of ligaments, tendons and other soft tissues (Chen et al., 2009). To determine the dimensions to model the collateral ligaments, tensile tests were conducted on a calibrated material testing machine (MTS Systems Corporation, MN, US), the samples being 63 mm in length and 15 mm width. The average ultimate strength value was 603 N. The Young's modulus (E) and stiffness (K) were calculated in the force range between 100 and 300 N. The average values were E = 549 N/mm^2 and K = 65.4 N/mm. Cadaveric studies have reported MCL and LCL stiffness values of 63 ± 14 N/mm and 59 ± 12 N/mm, respectively (Wilson et al., 2012). Hence, the LCL length used was 63 mm while the MCL was represented by two 126 mm long fibers to generate the same total stiffness on the medial and lateral sides.

A clamp system allowed for the adjustment of ligament stiffness by varying their lengths. One clamp was connected to the metal wires, while the other clamp fastened to a load cell. Pretensioning of the ligaments was accomplished by threaded rods connected to the load cells that connected to the frame of the rig. Three S-type load cells (Phidgets Inc., Alberta, Canada) were used to monitor the force on each ligament fiber. The calibration was performed using the MTS testing machine. Each load cell was assigned the corresponding gain value obtained from a linear regression analysis from data sets.

2.5. Hardware and software

Ligament forces measured by the load cells were amplified by a Phidget Bridge (Phidgets Inc., Alberta, Canada) connected to a development board (Raspberry Pi Foundation, Cambridge, UK) that monitored and recorded the ligament force data. To control the position of the stepper motor, a motor driver DM860T(I) (Leadshine Technology Co., Shenzhen, China) was connected to an Arduino board. Compartmental forces were measured at the following flexion angles: 0°, 30°, 45°, 60°, 90° and 110°.

2.6. Experimental protocol

To determine the effect on balancing of placement errors of the femoral component, a reference starting position was required. The reference positions of the femoral MCL and LCL attachment points were first determined from literature values (Ozkurt et al., 2016). During flexion tests, the attachment points were varied within the standard deviations mentioned in this study, targeting compartmental loads observed for the intact knee (Verstraete et al., 2017; Salvadore et al., 2018). Systematic modifications of the initial pre-tensions of the two fibers of the MCL were also required.

The pre-tension values at 0° flexion selected for the tests were 100 N for the LCL and 130 N for the posterior fiber of the MCL. The MCL was pre-tensioned more than the LCL to model greater contact forces on the medial side as observed in previous research (Verstraete et al., 2017). The anterior fiber of the MCL was not pre-tensioned at the starting position since anterior fiber recruitment occurs at higher flexion angles (Blankevoort et al., 1991). Therefore, during flexion the posterior fiber becomes slack while the anterior fiber is recruited, shifting the ligament force from the posterior to the anterior fibers. Ligament forces were recorded throughout flexion and, simultaneously, medial and lateral compartmental forces were measured by the Verasense sensor.

2.7. Modeling femoral component placement error

Incorrect linear placement of the femoral component was evaluated by moving the attachment points on the femur (Fig. 4). The acronym FEM error is designated to describe incorrect linear placement of the femoral component in the anterior, posterior, proximal and distal directions. The rationale for moving the ligament attachment points to model each of the FEM errors relates to the influence on the center of rotation (COR). Incorrect placement of the femoral component moves the kinematic COR relative to the ligament attachment points on the femur. When femur bone cuts deviate from the ideal, the error affects the COR of the femur creating a mismatch with the COR' of the implant. For example, in Fig. 5 distal FEM error is modeled by moving the medial and lateral ligament attachment points on the femur proximally.

Based on feedback obtained from surgeons, errors up to ±3 mm linearly could result from a combination of misplacement and mismatch of component geometry with anatomic geometry. A value of 2 mm error was chosen for our experiments. After the ligament attachment points were changed to simulate an error in femoral component placement, the pre-tensions in the collateral ligaments were checked and if necessary reset to the reference values at 0° flexion. This is equivalent to the surgeon choosing the most suitable tibial thickness and carrying out minor ligament releases or changes in bone cuts, to achieve balancing at 0° flexion. The effects of the FEM error on the contact forces during flexion were then measured.

2.8. Modeling mechanical alignment

In mechanical alignment, the femoral component is aligned perpendicular to the mechanical axis. Compared to the kinematic alignment, there is 2–3° of valgus and external rotation of 2–3° which results in larger bone resections on the medial condyle than the lateral both distally and posteriorly (Schiraldi et al., 2016). Three different combinations were tested. First 12 mm medial condyle resection paired with 10 mm on the lateral; second, medial

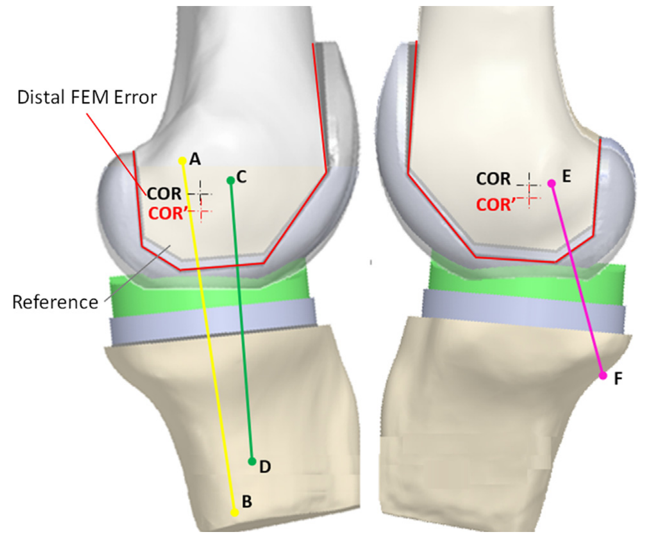

Fig. 4. Center of rotation (COR) of the femoral component and the adjusted center of rotation (COR′) resulting from distal FEM error with respect to the attachment points. Points A and C are the femoral attachment points for the posterior and anterior medial collateral ligament fibers (MCL), respectively. B and D are the attachment points on the tibia for the posterior and anterior MCL fibers, respectively. E is the femoral attachment point for the lateral collateral ligament (LCL) fiber and F is the attachment point on the fibula.

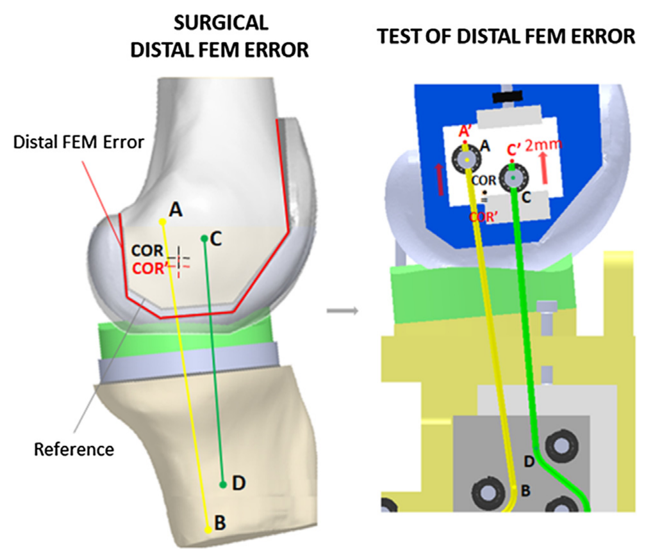

Fig. 5. Distal FEM error, outlined in red, was modeled on the test rig by linearly moving the ligament attachment points (A and C) on the femoral block 2 mm in the proximal direction. Shifts to the center of rotation (COR) and the ligament attachment points are labeled as COR′, A′ and C′. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article.)

11 mm and lateral 9 mm; third, medial 10 mm and lateral 8 mm. These variations were obtained by moving appropriate femoral ligament attachment points. As for changes in femoral component position, the initial ligament forces at 0° flexion were set to the reference values. This would account for any changes in the tibial attachment points due to mechanical alignment.

2.9. Data analysis

For all of the tests described above, the compartmental forces were measured by the tibial sensor and averaged from a total of ten runs. The results were plotted with standard deviations. For each of the FEM errors, the force values were compared to the reference values, to demonstrate changes in force due to the FEM error.

3. Results

The following group of tests refer to the placement errors of the femoral component (Figs. 6 and 7). For distal FEM error, there were only small changes in the contact forces except for a decrease in the lateral force in higher flexion. For proximal FEM error, the main change was a large increase in the lateral force after 30° flexion. For anterior FEM error, both the medial and lateral forces decreased relative to the reference. However, the reduction on the medial side was the most substantial in higher flexion. For the posterior FEM error, the medial forces were slightly higher throughout flexion, but the lateral forces increased substantially after 30° flexion. The data above was replotted to make an easy comparison of the forces on the medial and lateral condyles for the four different placement errors, in comparison to the reference forces (Fig. 7). For example, it can be seen that the distal error made minimal difference to the forces, anterior error made significant differences medially, while posterior and proximal errors causes large force increases on the lateral side. Table 1 depicts the major changes in contact force for each FEM error compared to the reference.

The results of the second group of tests of the effect of mechanical alignment compared with kinematic alignment are shown (Fig. 8). Medial and lateral compartmental forces for mechanical alignment behaved similarly to the kinematic alignment where both decreased throughout flexion. For the medial side, all three resection combinations gave similar force values. For the lateral side, the 12/10 mm combination gave the best results, but cutting less bone reduced the forces to almost zero after about 30° flexion.

4. Discussion

In this study, we examined how errors in femoral component placement and alignment techniques of the femoral component affect the tibiofemoral contact forces during flexion. Imbalance during surgery indicated by changes in tibiofemoral contact forces, can be associated to each FEM error and provide feedback that would indicate potential correction strategies. The effects on forces observed with 2 mm placement errors were large in most cases, except for distal FEM error. The effect of mechanical alignment, relative to kinematic alignment, on the tibiofemoral contact forces was limited mainly to changes on the lateral side. If the resections on the medial femoral condyles are 11 mm or less, the forces on the lateral compartment decreased to about zero after 30° flexion. However, for all the resections tested, forces on the medial compartment were little changed. This result is consistent with the results for linear placement error of the femoral component described above, smaller condyle resections being analogous to distal component placement error, having the least effect.

In interpreting the results, a major factor is that small dimensional differences of the collateral ligament length cause large changes in tension due to its high stiffness (Wilson et al., 2012; Walker et al., 2014). The eccentricity of ligament attachment points relative to the center of rotation of the femoral component resulted in large changes in ligament extensions relative to the baseline condition due to femoral component misplacement (Figs. 9 and 10). A thinner or thicker tibial component would be

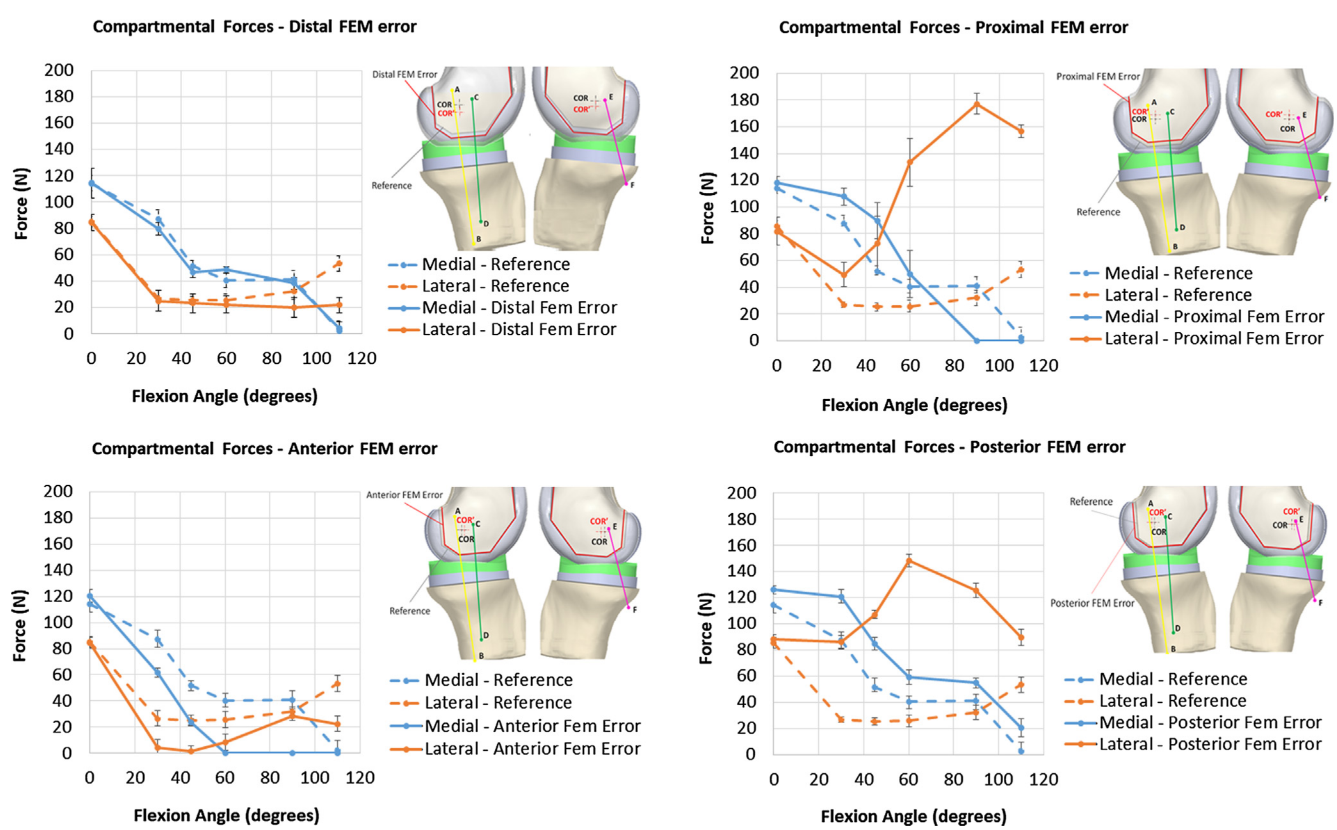

Fig. 6. Medial and lateral compartmental forces for the distal, proximal, anterior and posterior FEM errors with respect to the reference position.

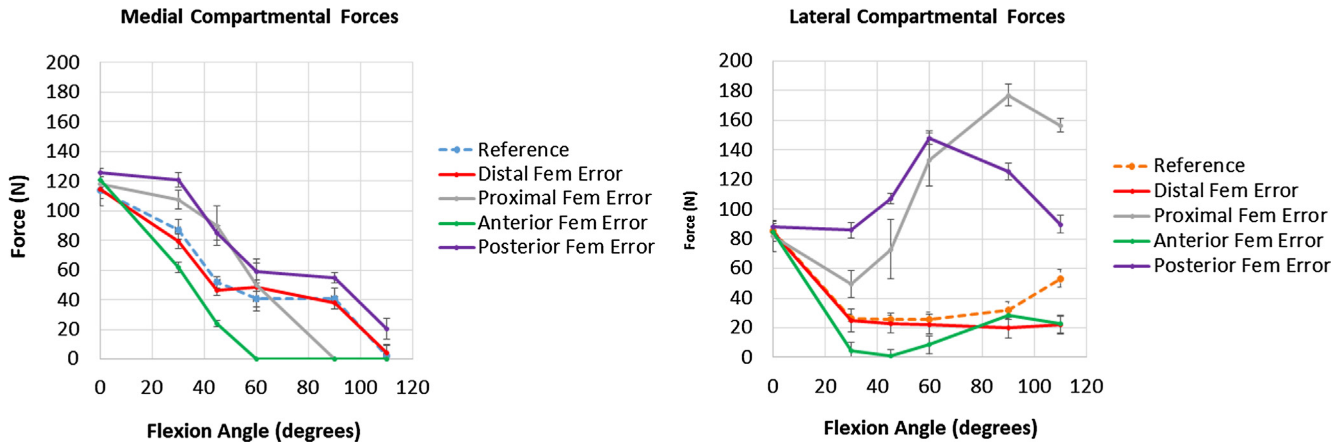

Fig. 7. Medial (left) and lateral (right) compartmental forces for the femoral placement errors: distal, proximal, anterior, and posterior.

Table 1
Maximum changes of compartmental forces compared with the reference values, for the different femoral component placement errors.

FEM error	Flexion angle	Medial	SD	Flexion angle	Lateral	SD
Distal	60°	+8N	±2.5	110°	−31.6N	±6.1
Proximal	90°	−41N	±7	90°	+145N	±7.5
Anterior	60°	−40.5N	±5.2	45°	−24N	±4.2
Posterior	30°	+33.5N	±5	60°	+122.4N	±4.7
Mechanical Alignment (MC10 mm – LC8 mm)	60°	+12.9N	±2.1	110°	−35.2N	±2.6

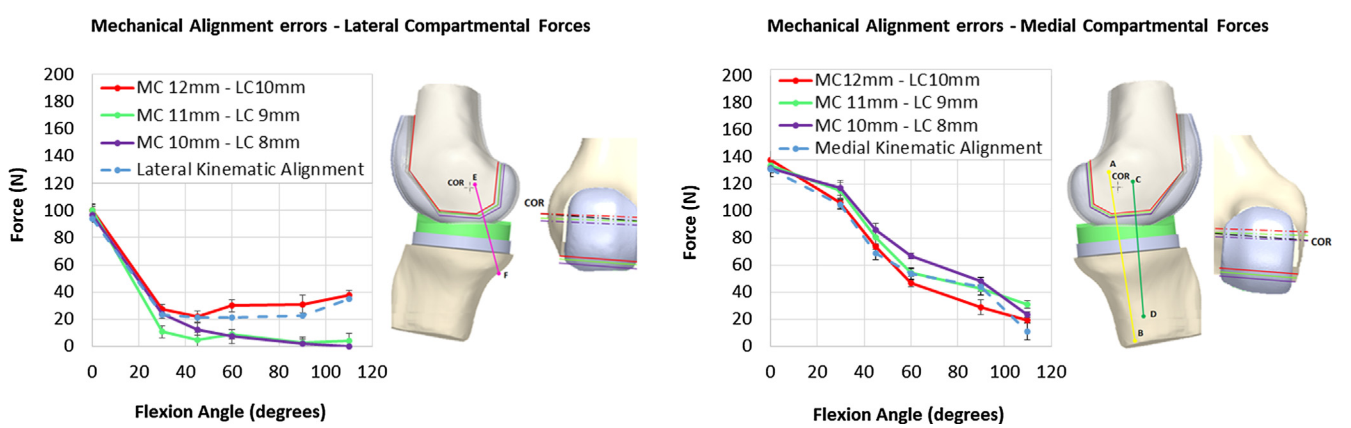

Fig. 8. Medial (left) and lateral (right) compartmental forces for the mechanical alignment variations. The forces are shown relative to the reference forces for kinematic alignment.

expected to change the contact forces by a uniform amount throughout flexion.

Up until now, the surgical goal has been for the forces to be equal throughout flexion on the lateral and medial sides, although it is generally recognized that in full extension the forces elevate due to tightness in collateral and posterior soft tissues. In a measured resection technique, such as represented by the kinematic alignment simulated in our study, unequal forces would occur as observed in previous studies (Meere et al., 2016; Gustke et al., 2017). In those studies, ligament releases were carried out to equalize the forces, usually on the medial side. However, for symmetric types of total knee as used in this study, equal balancing may be justified and easier for the surgeon to attain. The gap balancing technique where the bone cuts are made so that the forces on the condyles are symmetric and equal at 0° and 90° flexion, was not studied in our experiments (Churchill et al., 2018). Such a technique is particularly applicable when there are large deformities, which are more amenable to correction by bone cuts as well as with ligament releases.

The ideal situation, whatever technique is used, is that balancing is measured during the whole flexion range. Imbalances can be detected for example at three positions of 10°, 45°, and 90° (Gustke et al., 2017). Previous studies have provided guidelines to translate force data into the steps needed to achieve balancing by additional bone resections or pie-crusting of the ligaments (Meere et al., 2016; Gustke et al., 2017).

It has been found that kinematic alignment results in a reduced need for ligament balancing following bony resection, earlier recovery and improved functional results in the short term (Riviere et al., 2017). These results may also relate to the overall limb alignment a factor we did not account for in our study. It was pointed out that even though overall limb alignment can be the same after both kinematic or mechanical alignment techniques, the normal coronal medial tilt of the joint line is

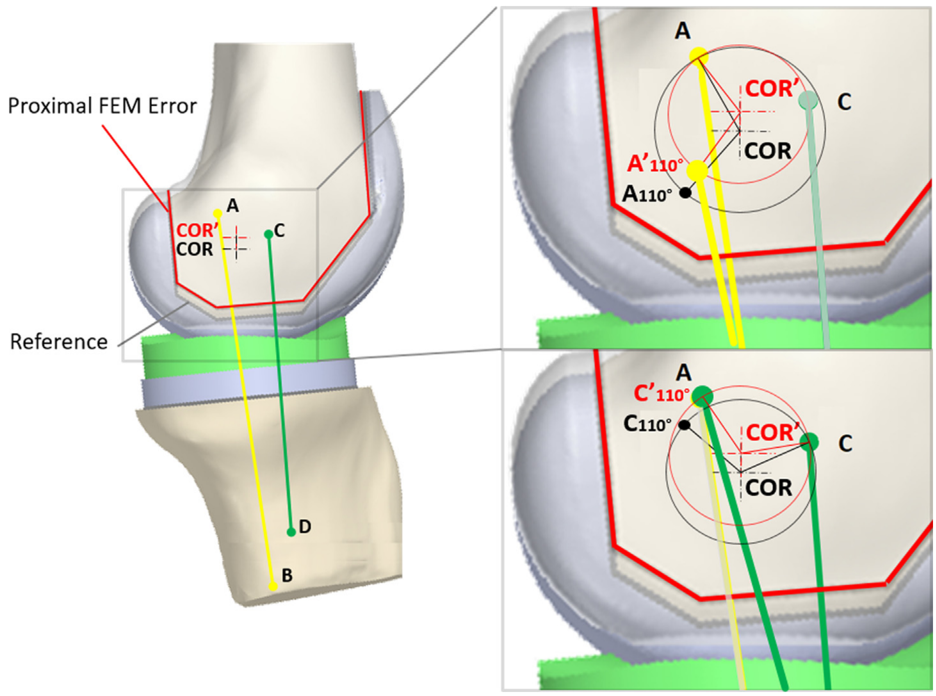

Fig. 9. Proximal FEM error results in the proximal shift of the center of rotation (COR) to COR'. During the flexion range, changes in the rotational path of the medial attachment points from the reference affect the ligament lengths. This is shown by the shift in the location of the ligament attachment points at 110° flexion in the reference, A_{100}° and C_{100}°, to the proximal FEM error , A'_{100}° and C'_{100}°.

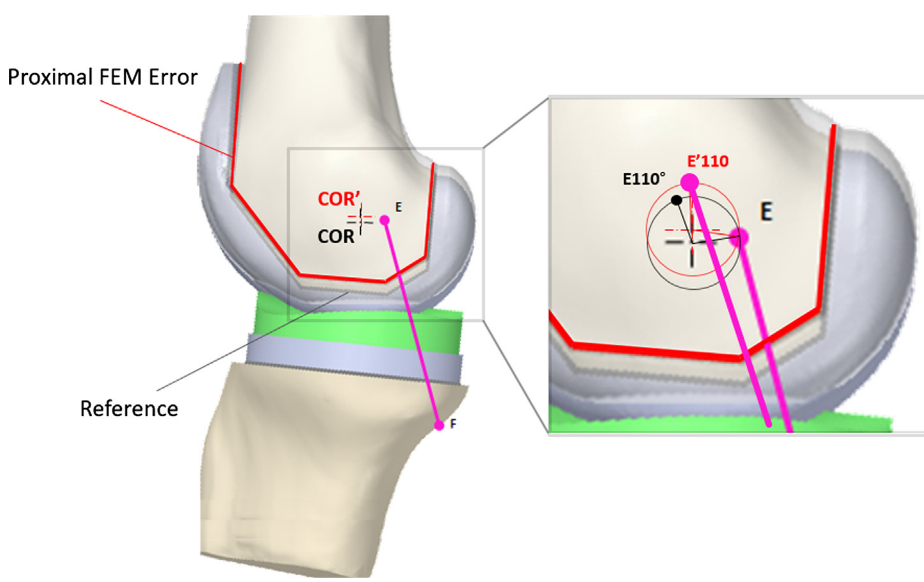

Fig. 10. Proximal FEM error also changes the rotational path of the lateral attachment points thus affecting the ligament lengths throughout flexion. This is shown by the shift in the ligament attachment point at 100° flexion in the reference, E_{100}°, to the proximal FEM error, E'_{100}°.

maintained in kinematic alignment and hence the ligament lengths are likely to be more anatomic. Consequently, imbalance would be less likely and the need for ligament balancing procedures reduced (Schiraldi et al., 2016). Nevertheless, the contact forces did not change substantially after mechanical alignment.

There are certain limitations to our study. The rig models only the collateral ligaments, however, there are several ligamentous structures in the knee that may also influence contact forces. The medial collateral ligament was modeled only using two fibers, whereas in reality it is a fan-shaped ligament with a broad attachment on the medial femoral condyle (Wilson et al., 2012). The MCL has been modeled by multiple fibers using an embedded computer model (Willing and Walker, 2018). The deep fibers of the MCL were not represented since they are largely resected in knee arthroplasty due to the tibial bone resection. The posterior cruciate (PCL) was not modeled in our test rig, although it is usually retained for the Triathlon Knee. However, the main function of the PCL is in high flexion and in controlling axial rotation (Logterman et al., 2018; Race and Amis, 1994). The load-elongation curve of the Neoligaments had a very similar stiffness to that of an anatomic collateral ligament, but did not show a toe region at low forces (Wilson et al., 2012). This would cause forces below 50 N to decrease more rapidly compared to an anatomic ligament. The results of this study are representative of a single radius design, results may change for other designs with differing radii.

The experiments highlighted important phenomena that can help the surgeon focus on the goals of balancing and how to achieve it. While certain corrections may be obvious, such as a tight lateral or medial side throughout flexion, subtleties of imbalance can only be dealt with at surgery by examining the tightness of various soft tissues by probing, rather than by a reliable predictive method which would be difficult to formulate. These authors suggested a methodology where attachment points and stiffnesses are measured at surgery and plugged into equations (Ewing et al., 2016).

Constant measurement of the contact forces across the condyles is a major advantage for determining the status of the balancing during flexion. Our reference values of lateral and medial contact forces were taken to simulate the anatomical values, rather than making them equal at all flexion angles. This is mainly a consequence of attempting to simulate kinematic placement of the knee, which arguably can be considered an ideal situation.

The main conclusion from this study is the importance of the accurate placement of the femoral component to achieve a well-balanced knee, which requires careful sizing of the component and precise bone resections to remove the exact amounts from the distal and posterior condyles. A second conclusion is that inadequate bone resections in mechanical alignment technique can produce excessive lateral looseness in flexion, although for normal bone resection values, there is little difference between mechanical and kinematic alignments.

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Conflict of interest

One of the authors (PSW) receives royalties from Zimmer Biomet and Stryker for total knee designs and owns shares in Genovel through NYUMC Department of Technology Transfer. Another author (MV) is an employee of Orthosensor Inc.

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