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Relationship Between Ligament Forces and Contact Forces in Balancing at Total Knee Surgery

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ABSTRACT

Background: Spacer blocks, tensors, or instrumented tibial trials are current methods of balancing the knee during surgery but there are no current techniques for measuring ligament forces. Our goal was to study the relationship between the collateral ligament forces and the condylar contact forces to determine whether there was equivalence.

Methods: A test rig was constructed modeling an artificial knee joint with collateral ligaments. The ligament forces as well as the lateral and medial tibial contact forces were measured during flexion for different positions of the femoral component on the femur, producing a set of forces for the simulated conditions. A regression analysis was used to study the correlation between the ligament and contact forces.

Results: The combined medial and lateral ligament and contact forces showed a linear relation with a correlation coefficient of 0.98. For the medial and lateral sides separately, the correlations were 0.85 and 0.88, respectively, with more than 80% of points within a ±25% deviation from the linear relations. This deviation from the linear correlation is linked to differences in medial-lateral femoral-tibial contact point locations at different flexion angles.

Conclusion: Within balancing accuracies generally achieved at surgery, the collateral ligament forces were linearly correlated to the condylar contact forces. These forces can also be equally correlated to the distraction forces as well as the moments at which condylar liftoff would occur from varus-valgus moments. This indicated a unification of the different balancing parameters, and hence such quantitative methods can be used interchangeably.

It is generally agreed that soft tissue balancing, or ligament balancing, is a necessary part of a total knee procedure for achieving smooth stable motion and avoiding instability. It has been stated that imbalanced conditions of a total knee cause about half of all failures [1]. Several different techniques have been used in balancing, including spacer blocks, laminar spreaders, and tensiometers [2]. The use of spacer blocks at 0° and 90° flexion relies on the feel of the surgeon in determining equal spacing when the blocks are introduced, in assessing whether the block was placed with the knee in neutral resting position with respect to the boundaries of the soft tissue envelope and whether the varus and valgus moments are equal when liftoff occurs. Calibrated distractors can make this process more quantitative by measuring the forces and the size of the gaps between the cut surfaces of the femur and tibia, as well as to measure force and gap distances and to estimate the stiffness of the collateral ligaments [3,4]. One type of

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instrumented balancer measures lateral and medial contact forces at 90° of flexion, in order to set the rotation of the femoral component based on symmetrical values [5]. An instrumented tibial trial measures the contact forces on the lateral and medial condyles throughout flexion where the values are displayed and recorded [1]. Navigation systems are used to measure the varus-valgus laxity envelope throughout the flexion range [6,7]. Currently, there are no surgical techniques available at this time for directly measuring the forces in the ligaments.

Measured imbalances are most often corrected by soft tissue releases such as multiple-puncture needle techniques [8] or by modifications to bone resections [1]. There is no consensus whether the goals of balancing should be based on tibiofemoral gaps, distraction forces, contact forces, ligament forces, varus and valgus moments to cause condylar liftoff, or any combination of the above. It might be assumed that distraction forces will be similar to contact forces across the implant, assuming that the gaps between the cut surfaces of the bone are the same as when the components are in place. However, measuring across bone gaps alone does not necessarily reproduce the relative position of the femur with respect to the tibia due to the missing constraints introduced by the implant components. For the same reason, distraction forces may not be the same as ligament forces. On the other hand, lateral and medial contact forces measured on trial components are likely to be similar to collateral ligament forces, assuming reasonable geometric symmetry between contact points and lines of action of the ligaments in the frontal view. If that is the case, then contact forces will also indicate the varus and valgus moments at liftoff. The actual laxity angles after liftoff however will depend on the relative stiffnesses of lateral and medial structures, which have been shown to be similar in some studies [4,9].

The purpose of this study was to determine whether there is a relationship between the lateral and medial tibiofemoral contact forces and the tensions in the collateral ligaments. If so, this would unify a number of the balancing parameters and allow for more flexibility in the methodologies. To investigate this, a test rig was constructed which simulated a total knee surgery, and where both contact forces and ligament tensions could be varied and measured accurately.

Materials and Methods

Total Knee Model

A test rig was constructed to model an artificial knee joint fixed to a femur and a tibia with collateral ligaments to simulate a total knee, where the contact forces (medial and lateral) and the collateral ligament forces could be simultaneously measured during flexion-extension. A standard, symmetric, posterior-stabilized total knee design (Triathlon, Stryker Orthopedics, Mahwah, NJ) was used for the experiments. Hence, the data can be regarded as being applicable to designs which retain only the collateral ligaments, including posterior-stabilized and ultracongruent types. For the tibial component, an instrumented tibial sensor (VERASENSE for Stryker Triathlon; OrthoSensor Inc, Dania Beach, FL) measured the medial and lateral compartmental forces used in previous studies to study balancing [10–12].

The rig was modeled with 3D printed parts representing the femur and tibia (Fig. 1). Bone resections were made with equal medial, lateral, and posterior cuts consistent with anatomic placement of the femoral component. The tibial resection modeled a typical posterior slope of 5° perpendicular to the anatomic axis in the frontal plane. A motor for driving the flexion and extension of the femur was aligned with the center of rotation of the femoral component and allowed 6 degrees of freedom so the femoral component was unconstrained, guided by the tibia contact surface and the collateral ligaments.

The collateral ligaments were modeled using a woven multifilament polyester fiber (Poly-tape Neoligament; Xiros Ltd, Leeds, UK). This has been clinically used for the reconstruction of ligaments, tendons, and other soft tissues [13]. The artificial ligaments were connected at one end to load cells and to the other end to the femoral attachment points using steel wires. The position of the attachment points on the femur and tibia was obtained from morphological studies of average male knees [14]. On the tibia, the attachment points of the ligaments were modeled using pulleys to guide the steel wires that connected anatomically the Poly-tape ligaments with the femur and tibia (Fig. 2). The Poly-tape ligaments were remote from the femur and the tibia for ease of length adjustment and force measurement. The lateral collateral (LCL) and medial collateral (MCL) forces were measured by calibrated S-type load cells (Phidgets Inc, Alberta, Canada), obtained by a data acquisition system and monitored on a screen during flexion. Ligament pretension forces were adjusted with an accuracy of $\pm$0.5 N.

Tests on knee specimens found that the MCL and LCL had stiffness values of 63 $\pm$ 14 N/mm and 59 $\pm$ 12 N/mm, respectively [9]. On the medial side, the MCL was modeled as 2 fibers that represented the most anterior and posterior fibers of the ligament. The stiffness values were divided between the 2 fibers to model the total stiffness of the MCL. On the lateral side, 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 [15]. Hence,

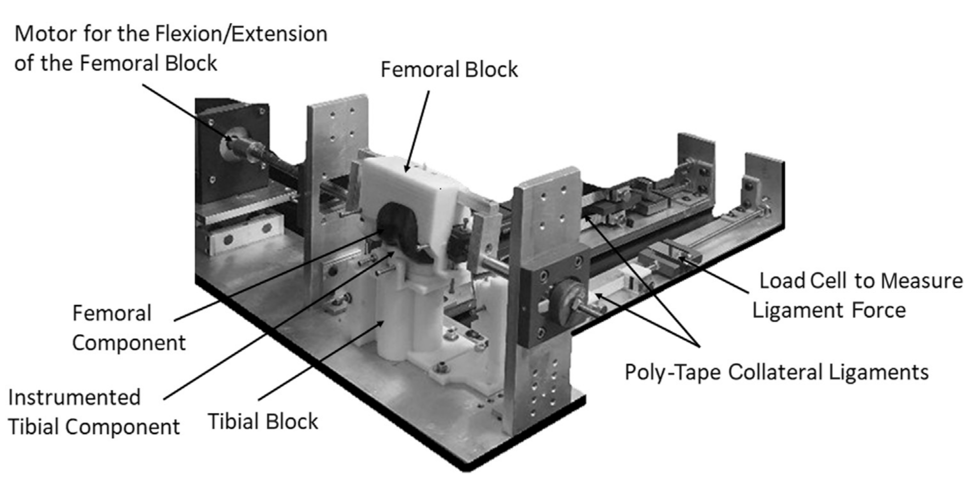

Fig. 1. Isometric view of the rig used to model the total knee and obtain the lateral and medial ligament and condylar forces.

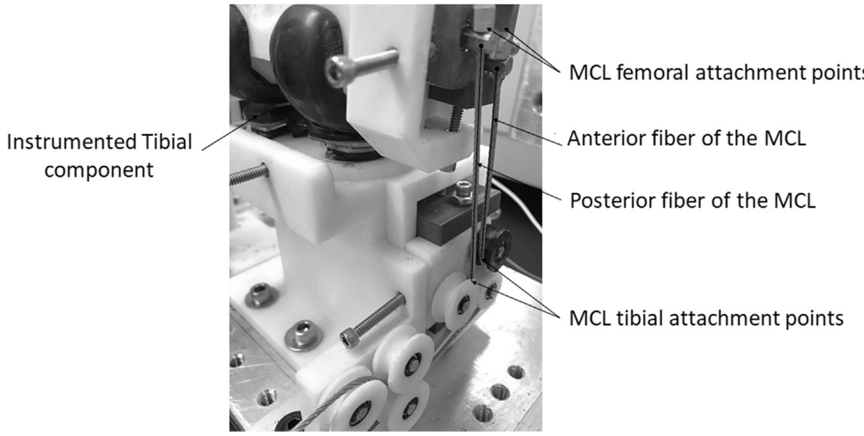

Fig. 2. Femoral and tibial attachment points of the 2 bands of the medial collateral ligament (MCL).

based on the Poly-tape ligament stiffness, the selected length for the LCL was 63 mm while the MCL was represented by two 126-mm-long fibers to simulate the same total stiffness on the medial and lateral sides. The attachment points were secured on adjustable fixtures on the femoral block allowing up to ±3-mm translation in the anterior, posterior, distal, and proximal linear directions. This simulated possible femoral component placement errors during surgery.

Experimental Protocol

A dataset was obtained consisting of multiple values of contact and ligament forces obtained at several flexion angles. Data for incorrect placement of the femoral component were performed by moving linearly the position of the medial and lateral ligament attachment points on the femur to simulate the influence of the placement errors on the center of rotation. For example, Figure 3 shows how proximal femoral component placement error (FEM error) was modeled by moving the medial attachment points on the femur distally corresponding to an excessive cut of the distal femoral condyles. Although not shown in Figure 3, the single lateral ligament attachment point was also moved distally to model proximal FEM error. This situation will cause different ligament lengths and tensions during flexion.

In the normal intact knee, the collateral ligaments have pretension values at all flexion angles. For the tests, the values selected at 0° flexion were 100 N for the LCL and 130 N for the posterior fiber of the MCL. These are close to the mean values measured during a total knee surgery [10]. The MCL was pretensioned more than the LCL to model greater contact forces on the medial side as observed in previous research [10]. The anterior fiber of the MCL was not pretensioned at the starting position because anterior fiber recruitment occurs at higher flexion angles [16]. During flexion, the posterior fiber becomes slack while the anterior fiber is recruited, shifting the ligament force from the posterior to the anterior fibers.

The femoral component placement errors relative to the anatomic placement were anterior, posterior, proximal, and distal. The compartmental and ligament forces were measured at 0°, 30°, 45°, 60°, 90°, and 110° flexion for each of the described femoral placement errors and the reference neutral position. Compartmental forces were measured with an instrumented tibial trial (VERASENSE). This test sequence produced 10 sets of data for each simulated condition, measuring contact and ligament forces, representing realistic placements of the total knee components.

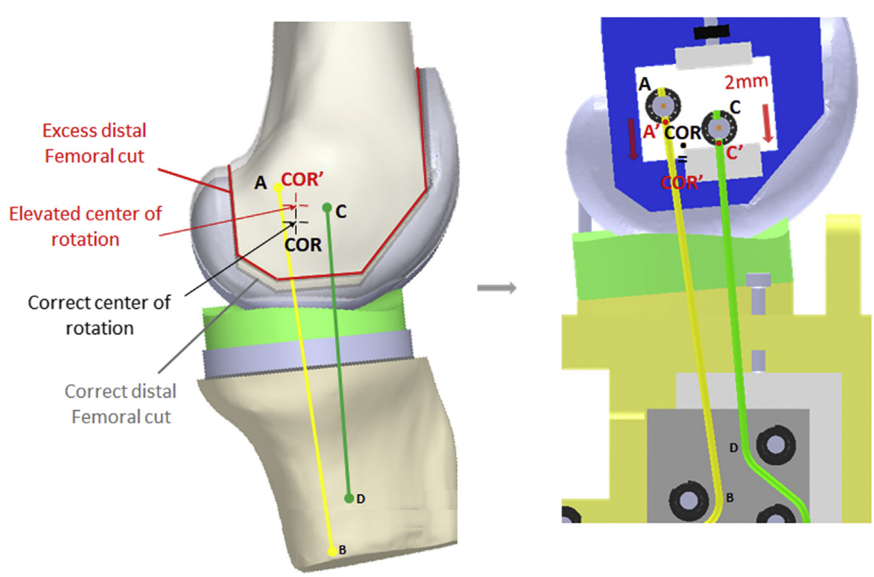

Fig. 3. Representative example of a femoral placement error, excess distal femoral bone cut is simulated by elevating the ligament attachment point. COR (black) = attachment for neutral position. COR (red) = attachment for proximal placement error. COR, center of rotation.

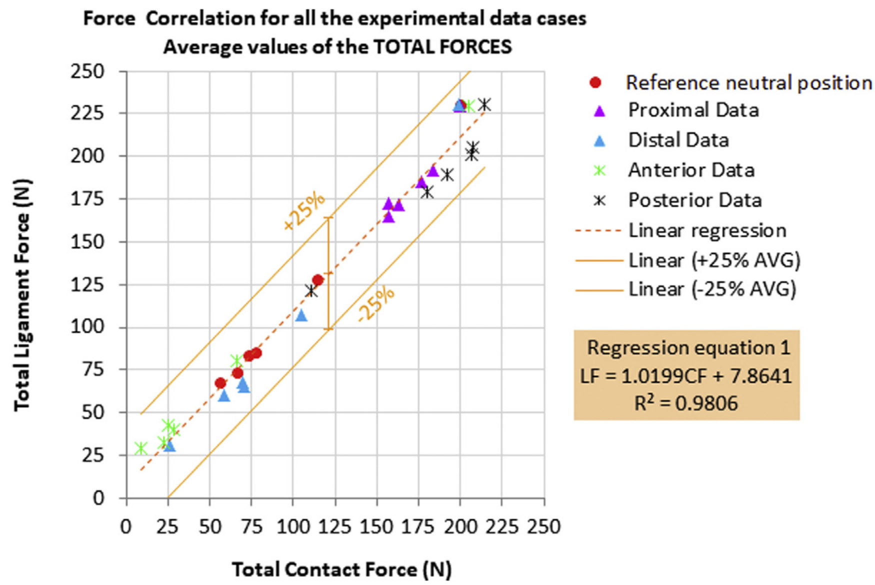

Fig. 4. Total ligament force (medial + lateral) plotted against total contact force. The regression line is shown together with deviations of ±25% from the mean.

Data Analysis

For all of the tests described above, the contact forces were measured by the instrumented tibial trial and averaged from a total of 10 runs. Ligament forces were recorded throughout flexion, and mean values and standard deviations were calculated at the selected flexion angles. Average values of the ligament force vs the contact force measurements for the reference and the simulated FEM errors were plotted for analysis. Linear regression analysis was performed to examine the correlation between the ligament forces and compartmental forces during flexion on the medial and lateral sides individually, as well as total ligament and contact force. On the plots of ligament force vs contact force, recognizing that there is likely to be some inequality, boundary lines were shown representing a ±25% deviation from the mean correlation line to examine the number of individual points inside or outside these boundaries. Correlation linear equations were used to obtain ligament forces as a function of contact forces which can occur at surgery by using instrumented tibial trials.

Results

The first analysis was for the total ligament force (lateral + medial) plotted against the total tibiofemoral contact force. Experimental data from the simulated FEM errors were plotted and fitted with a linear regression equation (Fig. 4). All data points were within the 25% deviation boundary. There was a linear correlation between the ligament and contact forces, independent of the condition simulated represented by the regression equation 1 on Figure 5, where $R^2 = 0.9806$. This shows the accuracy of the data, consistent with a static analysis which would predict equal values between total ligament force and total contact force.

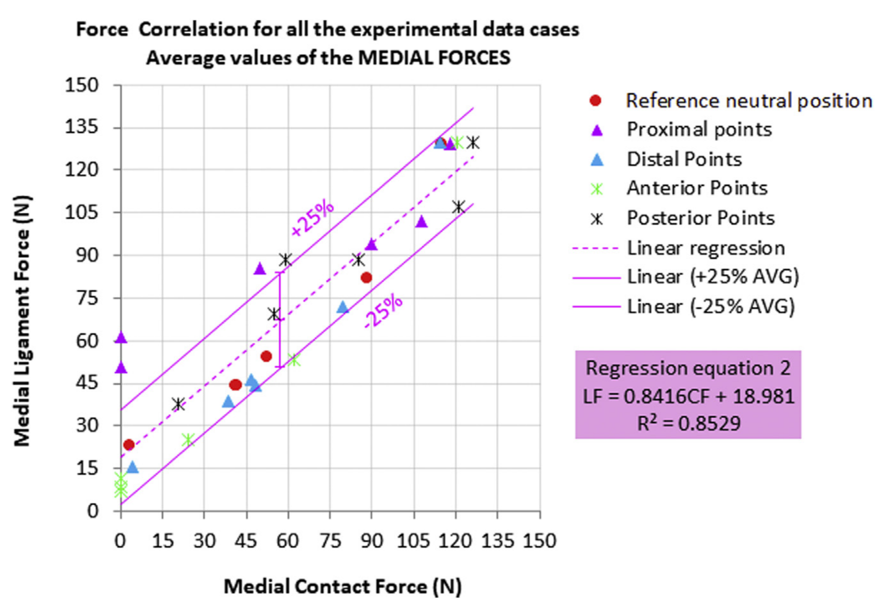

Fig. 5. Medial ligament force plotted against medial contact force. The regression line is shown together with deviations of ±25% from the mean.

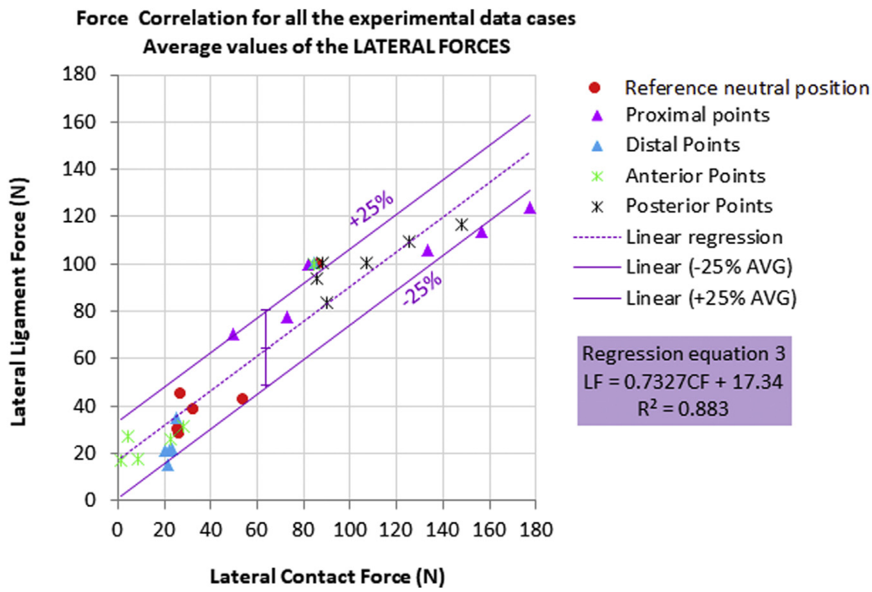

Fig. 6. Lateral ligament force plotted against lateral contact force. The regression line is shown together with deviations of ±25% from the mean.

The data were then plotted between the medial ligament and contact forces (Fig. 5) and between the lateral ligament and contact forces, individually (Fig. 6). Linear regression equations were again calculated. The correlation analysis between medial ligament and contact forces showed that $R^2 = 0.8529$, while 83% of all experimental data points were within the ±25% deviation boundary (Fig. 5). For the lateral side, $R^2 = 0.883$ with 80% of all experimental data points within the ±25% deviation boundary (Fig. 6).

Discussion

In this study, we generated data from a surgical simulation test rig to determine whether there was a correlation between the collateral ligament forces and tibiofemoral contact forces when total knee components were in place. The data analyzed consisted of ligament and contact forces measured on the lateral and medial sides throughout flexion for different placements of the femoral component on the distal femur. Due to the resulting changes in the center of rotation, this produced variations in both ligament forces and contact forces, giving a range of experimental data. Placement errors of 2 mm in different directions were considered sufficient to produce large variations in contact forces, because in previous work it was found that most balancing could be performed with only 2 mm or 2° corrections in ligament lengths or bone cuts [17]. This is consistent with the stiffness values of the collateral ligaments being about 60 N/mm [4,9]. Results showed a high correlation between total ligament and contact forces, with all data points within a ±25% deviation from the regression line. This result for the total forces was expected based on simple static analyses, with the small variations being due to experimental errors. When considering the medial and lateral forces individually, there was still a close linear correlation, although with more scatter of the points, including a few points outside the ±25% boundary deviation lines. The smaller errors were likely due to medial-lateral shifts of the contact points at different flexion angles, changing the lever arms of the forces. This explanation was confirmed in separate tests carried out after the main experiments. However, the larger deviations were possibly due to movement of the center of pressure out of the measurement areas on the condylar surface of the instrumented tibial insert. Other errors evidently occurred when the medial contact forces dropped below zero due to condylar liftoff, affecting the readings of the tibial insert.

It can be concluded that total ligament forces were almost exactly equal to total contact forces for the simulated conditions. On the lateral and medial sides separately, there was a close linear correlation with some variations under certain conditions. To put this in perspective, in total knee surgery, when the goal was equal lateral and medial contact forces, the medial/(medial + lateral) force ratio achieved was between 0.35 and 0.65 in 80% of the cases, while the mean total condyle force was 215 N [11]. This is approximately equivalent to force differences of ±25%. Hence, it can be concluded that lateral ligament force linearly correlates to lateral contact force, and the same for the medial side, with variations from equality being of the same order as mismatches in surgery. One implication of force equality between ligament and contact forces is that if the contact forces were equal, the varus or valgus moments to cause liftoff would be equal. However, it does not mean that for an additional applied moment, the varus and valgus laxity angles will be equal, because the angles will depend on the elongation of the ligaments which will depend on the ligament stiffnesses. Nevertheless as previously stated, the lateral and medial stiffnesses have been measured as being similar, implying that varus and valgus laxity angles may also be similar.

There are certain limitations to the study. The work was not strictly proactive in that the major study originally was for the purpose of determining the effect of femoral component misplacement on the surgical balancing as measured by the lateral and medial contact forces. However, during the experiments, it was realized that the same data could be used to test for a correlation between the ligament and contact forces. The data did cover a realistic range of surgical placements of the femoral component from anatomically correct to ±2-mm errors in all directions. Rotational errors were not considered which could usefully have extended the data. The ligaments were restricted only to the collaterals, although a double band for the medial collateral did simulate the actual behavior where the anterior and posterior bands are tighter or looser depending on the flexion angle. However, it has been pointed out that many other soft tissue structures other than the collaterals can be responsible for imbalance conditions [1]. The obtained data are applicable only to total knees where the cruciate ligaments are resected. The results of this study are representative of a single-radius design knee system, results may vary for dual-radius knee systems, although the data obtained for variations in the position of the femoral component on the femur may mitigate this limitation.

Also the rig models were a simplification of a real knee, due to the technical difficulties of modeling multiple soft tissues. Furthermore, the collateral ligaments themselves are protean in nature, changing their internal structure and stiffness by intertwining of their constituent bundles as a function of flexion. In such a rig, the weight of the limb segments is not accounted for as in a surgical procedure [17] but then the contact forces should more closely approximate the collateral forces. It is also pointed out that the balancing goals are not yet established and may need to be varied for each patient based on their anatomy and deformity. Furthermore, balancing forces should be seen in the context of the muscle and gravity forces which vary with angle and kinematics during function. Hence, surgical balancing is likely to be only one aspect of patient satisfaction, for which further research is needed.

In conclusion, different parameters which can represent balancing are the distraction forces, the condylar contact forces, the varus/valgus moments to cause liftoff, the laxity angles after liftoff, and the ligament forces themselves. It was shown in this study that, allowing for certain factors such as medial-lateral positioning of condyles, and different stiffnesses of collateral ligaments in certain patients, in general all of these parameters are likely to be approximately equivalent. This means that alternative quantitative measurement methods can be used for ligament balancing, with condylar contact forces being a particularly close substitute for ligament forces.

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References

[1] Gustke KA, Golladay GJ, Roche MW, Elson LC, Anderson CR. A targeted approach to ligament balancing using kinetic sensors. J Arthroplasty 2017;32:2127–32.
[2] Mihalko WM, Saleh KJ, Krackrow KA, L.A. Soft-tissue balancing during total knee arthroplasty in the varus knee. J Am Acad Orthop Surg 2009;17:766–74.
[3] D'Lima DD, Colwell CW. Intraoperative measurements and tools to assess stability. J Am Acad Orthop Surg 2017;25:S29–32.
[4] Hasenkamp W, Villard J, Delaloye JR, Arami A, Bertsch A, Jolles BM, et al. Smart instrumentation for determination of ligament stiffness and ligament balance in total knee arthroplasty. Med Eng Phys 2014;36:721–5.
[5] Camarata DA. Soft tissue balance in total knee arthroplasty with a force sensor. Orthop Clin North Am 2014;45:175–84.
[6] Picard F, Deep K, Jenny JY. Current state of the art in total knee arthroplasty computer navigation. Knee Surg Sports Traumatol Arthrosc 2016;24:3565–74.
[7] Hunt NC, Ghosh KM, Blain AP, Athwal KK, Rushton SP, Amis AA, et al. How does laxity after single radius total knee arthroplasty compare with the native knee? J Orthop Res 2014;32:1208–13.
[8] Bellemans J, Vandenneucker H, Van Lauwe J, Victor J. A new surgical technique for medial collateral ligament balancing multiple needle puncturing. J Arthroplasty 2010;25:1151–6.
[9] Wilson WT, Deakin AH, Payne AP, Picard F, Wearing SC. Comparative analysis of the structural properties of the collateral ligaments of the human knee. J Orthop Sports Phys Ther 2012;42:345–51.
[10] Verstraete MA, Meere PA, Salvadore G, Victor J, Walker PS. Contact forces in the tibiofemoral joint from soft tissue tensions: implications to soft tissue balancing in total knee arthroplasty. J Biomech 2017;58:195–202.
[11] Meere PA, Schneider SM, Walker PS. Accuracy of balancing at total knee surgery using an instrumented tibial trial. J Arthroplasty 2016;31:1938–42.
[12] Cho KJ, Seon JK, Jang WY, Park CG, Song EK. Objective quantification of ligament balancing using VERASENSE in measured resection and modified gap balance total knee arthroplasty. BMC Musculoskelet Disord 2018;19.
[13] Chen JM, Xu J, Wang A, Zheng M. Scaffolds for tendon and ligament repair: review of the efficacy of commercial products. Expert Rev Med Devices 2009;6:61–73.
[14] Ozkurt B, Sen T, Cankaya D, Kendir S, Basarir K, Tabak Y. The medial and lateral epicondyle as a reliable landmark for intra-operative joint line determination in revision knee arthroplasty. Bone Joint Res 2016;5:280–6.
[15] Otake N, Chen H, Yao X, Shoumura S. Morphologic study of the lateral and medial collateral ligaments of the human knee. Okajimas Folia Anat Jpn 2007;83:115–22.
[16] Blankevoort L, Huiskes R, de Lange A. Recruitment of knee joint ligaments. J Biomech Eng 1991;113:94–103.
[17] Walker PS, Meere PA, Bell CP. Effects of surgical variables in balancing of total knee replacements using an instrumented tibial trial. Knee 2014;21:156–61.