Laxity and contact forces of total knee designed for anatomic motion: A cadaveric study

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Abstract

Background: Total knee designs that attempt to reproduce more physiological knee kinematics are gaining attention given their possible improvement in functional outcomes. This study examined if a total knee designed for anatomic motion, where the soft tissue balancing was intended to replicate anatomical tibiofemoral contact forces, can more closely reproduce the laxity of the native knee.

Methods: In an ex-vivo setting, the laxity envelope of the knees from nine lower extremity specimens was measured using a rig that reproduced surgical conditions. The rig allowed application of a constant varus/valgus (V/V) and internal-external (I/E) torque through the range of motion. After testing the native knee, total knee arthroplasty (TKA) was performed using the Journey II bi-cruciate substituting implant. Soft tissue balancing was guided by targeting anatomical compressive forces in the lateral and medial tibiofemoral joints with an instrumented tibial trial. After TKA surgery, the laxity tests were repeated and compared to the native condition.

Results: The TKA knee closely reproduced the coronal laxity of the native knee, except for a difference at 90° of flexion for valgus laxity. Looking at the rotational laxity, the implant constrained the internal rotation relative to the native knee at 45 and 60° of flexion. The forces on the tibial trial for the neutral path of motion showed higher values on the medial side as the knee flexed.

Conclusions: This study suggested that when using an anatomically-designed knee, the soft tissue balancing should also aim for anatomical contact forces, which will result in close to normal laxity patterns.

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1. Introduction

Part of the reason why patients with a total knee arthroplasty (TKA) do not achieve similar functional scores to non-arthritic subjects may be the abnormal kinematics after surgery [1–3]. For example, paradoxical anterior translation of the femur with respect to the tibia can occur, without reproducing a medially pivoting action, while the knee flexes [4,5]. Knee arthroplasties with kinematics deviating from that of normal knees are at risk for reduction in quadriceps efficiency, anterior knee pain, and decreased range of motion (ROM) [6], thus limiting deep knee bending and kneeling activities [7,8].

To overcome these deficiencies, attention has recently been given to total knee designs that aim to reproduce natural knee joint kinematics. An example is the Journey II bi-cruciate substituting (BCS) design, which – in absence of the cruciate ligaments – incorporates an anteroposterior stabilization system that relies on engagement of an asymmetric spine-cam. The design additionally

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aims to reproduce the anatomical geometry of the tibial plateau surfaces, namely concave medially and convex laterally [9]. Therefore, this design can be considered as one of the more anatomically designed implants in today’s market. As a result, it is referred to as an anatomical total knee design in the remainder of this paper. Note that this definition does not necessarily imply full anatomical functionality, i.e. reproduction of the natural knee joint kinematics and laxity. In literature, limited evidence is available in that respect. In an in-vivo fluoroscopy study, Catani et al. and Kuroyanagi et al. have showed that this design philosophy has the potential to restore physiological knee joint motion [10,11]. Other studies showed that this design philosophy can improve the functional outcome after a one-year follow-up [3,12].

The above data is indicative that relying on more anatomic design allows better approximation of normal kinematics and leads to improved outcomes. However, this might also depend upon the surgical ability in restoring the pretension of the ligaments. Soft tissue balancing is an important part of a total knee procedure to achieve an optimal level of laxity by adjusting the soft tissues that surround the knee. To achieve this quantitatively, different instrumented sensors and devices are currently available [13–15]. In this paper, we focused on a particular instrumented tibial trial that provided real-time, intra-operative measurement of the tibio-femoral contact forces on the medial and lateral tibial plateaus.

The main aim of this study was to determine if an anatomically designed TKA, where anatomic tibio-femoral loads were targeted, produced a close reproduction of the laxity of the native knee.

Our first hypothesis was that the varus–valgus and internal–external rotational laxity of a BCS TKA would replicate the laxity of an intact knee.

Our second hypothesis was that through the soft-tissue balancing technique achieved using an instrumented tibial trial, the intraoperative medial and lateral loads during the neutral path of motion of the balanced knee would reproduce anatomic loads.

2. Methods

2.1. Specimens

Nine fresh-frozen lower extremity specimens with an average age of 68 years old (range 55–84, standard deviation (SD) 10.4) and average body mass index (BMI) of 20.3 (range 15.1–25.1, SD 3.8) were tested. Specimens were screened to meet the following inclusion criteria: male subjects, Kellgren–Lawrence score <2 and no lower limb surgery. Each specimen was defrosted for 72 h and then mounted into the test rig. Muscle and soft tissue were left intact. An additional three specimens were used for method development. No institutional review board (IRB) was required for this study.

2.2. Test set-up

A custom mechanical rig that emulated surgical conditions was constructed (Figure 1). The pelvis was constrained in an adjustable frame at the top of the table. After ensuring normal knee and ankle alignment, two pins were drilled into the ilium, fixing...
the pelvis to the table. Another pin was drilled vertically into the femur connecting it to a mobile horizontal arm, preventing axial rotation and abduction of the femur. Hence, the femur was constrained to only flexion about the mediolateral axis of the knee. The ankle was then fixed into an adjustable frame by means of a primary transverse 10 mm rod that passed through the malleoli and a secondary six millimeter pin through the proximal third of the tibia. The ankle frame was mounted on low friction ball bearings, which allowed for medial and lateral translation. At the same time the ankle frame could internally and externally rotate. In addition to this free motion, a known internal/external (I/E) and varus valgus (V/V) moments could be applied to the tibia. The V/V load was applied by means of a weight acting along the medial–lateral rail. The I/E torque was applied by means of a U-joint which transferred a known torque created by a pulley system.

Hence the tibia was allowed five degrees of freedom, the femur just one. The possible directions in which the knee could be moved are shown on the computer-assisted design (CAD) model in Figure 2 as follows:

- **R<sub>1</sub>:** flexion and extension of the femur
- **R<sub>2</sub>:** free rotation around the medial–lateral axis of the tibia (flexion/extension rotation)
- **R<sub>3</sub>:** free rotation around the mechanical axis of the tibia (internal/external rotation)
- **R<sub>4</sub>:** free rotation around the anterior–posterior axis of the tibia (varus/valgus rotation)
- **T<sub>1</sub>:** free translation along the proximal–distal direction
- **T<sub>2</sub>:** free translation along the medio-lateral axis of the tibia

This set-up aimed to replicate intra-operative surgical conditions on the anesthetized patient, therefore no muscle activation was applied.

2.3. Motion and laxity measurements of native knee

After mounting the specimen in the rig, a navigation system (AWD version 5.0, Stryker, Michigan, USA) was installed, which included trackers rigidly attached to the femur and tibia. The system recorded the relative movement of the tibia with respect to the femur. A series of thigh-pull tests were performed to record the neutral path of motion and the laxity for the intact knee. The aim of the thigh-pull test was to flex the knee in a consistent and controlled way: the surgeon lifted the horizontal arm rigidly attached to the femur causing flexion of the knee from 0 to 100° [16]. First, thigh-pull tests were performed to record the neutral path of motion. Then, a load of approximately 24.5 N was applied along the medio-lateral rail, causing a varus or valgus moment of 10 Nm at the knee. The thigh-pull test was subsequently repeated under an externally applied varus resp. valgus moment. The varus (valgus) laxity was then defined as the absolute difference in tibio-femoral angle measured by the navigation system between the varus (valgus) test and the neutral path of motion. After that, an internal/external rotational test was performed. A load of 19.6 N was attached to a wire at the end of the pulley system. The torque was transmitted to the tibia by the U-shape joint and pulley connected to the tibial frame, creating an internal or external rotational moment of two newton meters to the

![Figure 2. CAD model of the rig showing the degrees of freedom of the knee.](image-url)
A thigh-pull test was performed, flexing the knee again from full extension to 100°. The tibiofemoral rotation angles were recorded using the surgical navigation system. The rotational laxity was defined as the difference of the angles between the neutral path of motion and the internal/external laxity test.

2.4. Balancing and contact forces

After testing the native knee, a primary total knee arthroplasty was performed using a BCS implant (Journey II, Smith and Nephew, Memphis, TN, USA). The bone cuts were performed with an optical navigation system using a measured resection technique. A subvastus approach was used. The femoral and tibial alignment was obtained using surgical navigation aiming for neutral mechanical alignment, including a tibial cut perpendicular to the tibial mechanical axis (Stryker, Michigan, USA). Subsequently, trial components were inserted. The tibial liner trials were thereby substituted by sensorized trials (Verasense, Orthosensor, Dania Beach, Florida, USA), which measured the medial and lateral tibiofemoral contact forces and both tibiofemoral contact points. Based on these contact points, tibial rotation was derived, such that no excessive tibiofemoral rotation was present in full extension. After determining the thickness of the trial, based on achieving full extension using the sag test [16], the knee was balanced using an algorithm of specific surgical corrections based on the sensor-derived medial and lateral compartmental loads [17].

Balancing was thereby quantitatively defined using the compartmental load ratio (CLR), defined as the ratio of the medial force over the total forces measured on the tibial plateau \( F_{\text{medial}} / (F_{\text{medial}} + F_{\text{lateral}}) \). A ratio of 0.5 indicates equal lateral and medial forces. This target has traditionally been the target for standard total knee designs [15]. However, the implant design considered in this study more closely represents the native anatomy (e.g. including an oblique joint line). Consequently, the target was modified to allow for higher forces on the medial side compared to the lateral, particularly in high flexion. The target loads thereby more closely match the condition of the intact knee, associated with rollback of the lateral femoral condyle in flexion and relaxation of the lateral collateral ligament [18]. Relatively minor surgical corrections were needed, attributed to the normal non-arthritic status of the specimen. Details regarding the surgical corrections performed on each knee are given in Appendix A.

After balancing the knee, the parapatellar arthrotomy was sutured and the unloaded thigh-pull tests were performed, followed by the laxity tests (varus/valgus and internal/external rotation).

2.5. Statistical analysis

Descriptive statistics including mean, SD and range are presented for continuous variables. The Shapiro–Wilk test for normality was applied to the variables in order to verify normality, which was the case for all data presented in this paper. The laxity differences were subsequently assessed using a two-sided paired t-test.

3. Results

3.1. Laxity measurements of native and TKA knee

The coronal and rotational laxities were averaged for the nine specimens and plotted as a function of the flexion angle (Figure 3). All the specimens reached at least 90° of flexion. Varus angles are represented with a positive sign, while valgus with a negative. The

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**Figure 3.** Coronal (A) and rotational (B) laxity measurements averaged for the nine specimens tested. The shaded areas surrounding the solid lines represent the standard deviation. Significant differences indicated by symbol.
average varus laxity of the TKA knee matched the laxity of the intact knee for the whole range of motion. On the other hand, the average valgus laxity of the TKA matched the intact knee laxity up to 45°, then started to increase, reaching a significant difference ($p < 0.05$) of four degrees at 90° of flexion (Figure 3A). The internal rotational laxity of the BCS TKA knee was lower than the laxity of the native knee for the whole range of motion, with significant differences at 45 and 60° of flexion. The average of the external rotational laxity of the BCS TKA implant was higher than that of the native knee (Figure 3B) but no significant difference was found between the two cases.

3.2. Contact force and CLR

The compartmental loads (medial and lateral) and the CLR averaged for nine knee specimens are plotted as a function of the flexion angle (Figure 4). The compartmental loads are the loads recorded during the thigh-pull test for the neutral path of motion (no external load applied). The lateral load decreased as the knee flexed, while the opposite was true for the medial side. The medial load showed a higher standard deviation at higher degrees of flexion, while the standard deviation of the lateral load decreased as the flexion angle increased. The trend of the compartmental loads was reflected in the CLR during neutral path of motion. The CLR started with a value of 0.41 at full extension and then slowly increased until reaching 0.80 at 90° of flexion. The CLR increase during the range of motion corresponded to the transmission of the majority of the load through the medial side after 30° of flexion.

4. Discussion

This study focusses on the ability to replicate the laxity of the native knee after total knee arthroplasty. Therefore, a specific knee implant design was considered, that replicates the intact knee anatomy in various ways, notwithstanding both cruciate ligaments are replaced by a post and cam mechanism. In a cadaveric setting, we have demonstrated that by targeting anatomical tibiofemoral contact force levels in the medial and lateral compartments during surgery, the coronal and rotational laxities approximate, if not match, the laxity of that particular knee before TKA surgery.

Instability remains a primary cause of revision surgery [19–21]. In an attempt to restore the stability of the intact knee, we hypothesized that implant design and the knee specific soft tissue balancing at surgery are interrelated concepts and mutually influence the laxity envelope of the replaced knee. On the one hand, we implemented a total knee design that, through various design features, aims to mimic the intact knee anatomy. On the other hand, to handle soft tissue balancing during surgery, our surgical approach relied on the use of instrumented tibial inserts. These sensors allowed targeting medial and lateral compartmental load levels representative for the intact knee during arthroplasty surgery [18].

Looking at the coronal laxity of the intact knee, we found that the varus laxity was slightly higher than the valgus laxity at a given flexion angle. This confirms earlier publications on both in-vivo and in-vitro studies [22–24]. After TKA surgery, and considering the intraarticular loads, we were able to replicate this laxity signature remarkably well. However, from 45° of flexion onwards, a slightly increased valgus laxity was observed. This difference became statistically significant at 90° of flexion. To appreciate this difference, the primary stabilizers for coronal stability should be considered, namely the medial and lateral collateral ligaments [25]. The increased valgus laxity is hence linked to a decreased stiffness of the medial collateral ligament compared to the intact knee at these flexion angles. The ligament's stress–strain properties are characterized by a strongly non-linear relationship, with an increasing stiffness as the tension in the ligament increases [26]. Therefore, the low stiffness of the medial collateral suggests that the pretension in the
medial collateral ligament was too low. Considering the design of the instrumented tibial trials and previous work on the relationship between laxity and intraarticular loads, it is fair to assume that the pretension in these collaterals is directly assessed using these sensors [27]. Hence, the increased valgus laxity likely indicates that the load in the medial compartment was still too low to replicate the laxity signature of the intact knee, despite the unequal CLR that was targeted during soft tissue balancing.

Looking at the rotational laxity, the obtained values for the intact knee are again comparable to previous publications [24, 28]. However, when the prosthesis is implanted, there is a significant decrease of the internal rotational laxity during mid-flexion (45–60°). For the intact knee, the cruciate ligaments are the primary stabilizers for rotational stability [25]. Both are however sacrificed during TKA surgery. Therefore, the reason for this discrepancy is believed to be attributed to either the tension in the secondary stabilizing structures (the collateral ligaments) or the implant design. The former appears unlikely, given the excellent agreement in coronal stability between the intact and TKA knees in this flexion range. Hence, the implant design is deemed responsible for this reduced internal rotation. The implant seemed to constrain the tibial internal rotation (external rotation of the femur) and allowed for slightly more tibial external rotation compared to the intact knee. This phenomenon is attributed to the asymmetric post-cam mechanism of this particular design, whose aim is to engage the femoral component during flexion (at about 50–60°) [9]. However, this cam may in principle also prevent tibial internal rotation at those specific degrees of flexion.

With a minimal number of surgical corrections, as detailed in Appendix A, the knees tested during our test series indicated that the majority of the load was transferred through the medial compartment after TKA surgery. This was particularly the case at higher flexion angles, with an average CLR of 0.80 at 90° of flexion. In contrast, the medial and lateral loads were almost identical near full extension (CLR of 0.4–0.5). These load levels resemble the distribution observed for intact knees, as demonstrated by earlier work from our group [18]. This resemblance is not surprising given the various anatomical design features of the considered implant system, e.g. difference in tibial height between the medial and lateral compartments resulting in a joint line obliquity, and the low articular nature of the tested knees.

The clinical implications of this study focus on preventing knee instability and minimizing the risk for requiring post-operative manipulation under anesthesia. We have demonstrated that restoring knee stability to levels observed in intact knees is possible, when taking into account the interplay between implant design and soft tissue balancing. The target load levels can thus be linked to the implant design. For our study, considering the anatomically designed implant, the stability of the intact knee is restored after TKA surgery by targeting lower lateral than medial compartmental loads; a situation that closely resembles the intact knee [18]. This contrasts with previous studies that aimed for equal loads in the medial and lateral compartments [17, 29]. Nevertheless, using these equal target loads appeared successful in these papers, given the excellent patient reported outcomes. This discrepancy in target loads is therefore accepted as the implants considered in those studies had a geometric symmetrical medial and lateral tibial plateau that may hence benefit from a more symmetrical (CLR = 0.5) load distribution.

A limitation of this study is that the specimens were close to normal without arthritic changes, and hence not reflecting any soft tissue balancing which might be required under actual surgical conditions. At the same time, this allowed us to assess the load patterns without the need for numerous, complex soft tissue releases. Furthermore, although we determined that the laxity and contact force patterns after implantation were similar to those in the native knee, this does not in itself imply the reproduction of this behavior on human subjects. This would need to be verified with an appropriate clinical study where this paper can be the base of.

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Appendix A

In this paper, surgical corrections are used to indicate surgical adjustments to achieve an equal load balance at the knee. Two major categories are identified based on earlier work, namely soft tissue adjustments and bony adjustments [16]. The soft tissue adjustments consisted of subtle pie-crusting of the listed ligaments, though it is acknowledged that these were not exactly quantified. Table number 1 provides an overview of the surgical corrections performed in each experimentally assessed knee.

<table>
<thead>
<tr>
<th>Specimen #</th>
<th>Surgical corrections</th>
</tr>
</thead>
<tbody>
<tr>
<td>4</td>
<td>None</td>
</tr>
<tr>
<td>5</td>
<td>Bone cut on the distal femur</td>
</tr>
<tr>
<td>6</td>
<td>None</td>
</tr>
<tr>
<td>7</td>
<td>None</td>
</tr>
<tr>
<td>8</td>
<td>Soft tissue release of popliteus</td>
</tr>
<tr>
<td>9</td>
<td>Soft tissue release of the anterior part of the medial collateral ligament</td>
</tr>
<tr>
<td>10</td>
<td>Soft tissue release of the medial collateral ligament</td>
</tr>
<tr>
<td>11</td>
<td>Soft tissue release of lateral collateral ligament</td>
</tr>
<tr>
<td>12</td>
<td>None</td>
</tr>
</tbody>
</table>

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