The superficial MCL is the primary medial restraint to knee laxity after cruciate-retaining or posterior-stabilised total knee arthroplasty: effects of implant type and partial release.

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Abstract

Purpose Although soft-tissue contribution to stability of the intact knee has been studied in relation to sports injuries, there is little data about their stabilising actions in the replaced knee. The aim of this study was to quantify the contributions of medial soft-tissues to stability following cruciate-retaining (CR) or posterior-stabilised (PS) total knee arthroplasty (TKA).

Methods Using a robotic system, eight cadaveric knees were subjected to ±90N anterior-posterior force, ±5Nm internal-external and ±8Nm varus-valgus torques at various flexion angles. The knees were tested intact and then with CR and PS implants, and successive cuts of the deep and superficial medial collateral ligaments (dMCL, sMCL) and posteromedial capsule (PMC) quantified the percentage contributions of each structure to restraining the applied loads.

Results In implanted knees, the sMCL restrained valgus rotation (62% across flexion angles), anterior-posterior drawer (24% and 10% respectively) and internal-external rotation (22% and 37%). Changing from CR to PS TKA increased the load on the sMCL when resisting valgus loads. The dMCL restrained 11% of external and 13% of valgus rotations, and the PMC was significant at low flexion angles.

Conclusions This work has shown that the sMCL is the primary medial soft-tissue restraint in different planes of laxity in both CR and PS TKA. Medial release in the varus knee should be minimised to maintain a balanced knee. Changing from CR to PS TKA did not show increased stability from the cam-box mechanism, but rather increased loads on the sMCL in the absence of the posterior cruciate ligament.

Keywords: total knee arthroplasty; stability; medial collateral ligament; ligament release; restraint; biomechanics
Introduction

Whilst there have been many studies examining the role of the medial passive soft-tissue structures in the native and injured knee, there remains little information on the restraint that they provide in the presence of a knee replacement [6,27,23,22,12]. In the case of primary total knee arthroplasty (TKA), which is most commonly performed using either a posterior cruciate-retaining (CR) or posterior-stabilised (PS) implant, the onus is on these soft tissues to help provide restraint, but the effect of the altered joint mechanics post-arthroplasty is not known.

The soft-tissues on the medial aspect of the knee have been described as consisting of three distinct layers: the most superficial being a fascial layer, the second layer containing the superficial medial collateral ligament (sMCL), and the deepest layer containing the deep medial collateral ligament (dMCL) and the posteromedial capsule (PMC) [29]. The sMCL has a femoral attachment near the epicondyle [16] and inserts 60-80mm distal to the joint line [24]. The dMCL, previously identified as the mid-third medial capsular ligament [29], lies deep to the posterior fibres of the sMCL as a two-part structure (meniscofemoral and meniscotibial) of the capsule [24]. Posterior to the dMCL, the PMC is formed of a large range of fibres attached to the femur around the base of the adductor tubercle and to the tibia just distal to the joint line posteromedially [29,24].

During TKA surgery, it is general practice for ligaments and soft tissue to be balanced in order to align the knee in extension and flexion. In the case of a varus knee, the medial structures may be overly tight and require judicious release to correct the deformity. However there is a large discrepancy in suggested protocols between studies, with a lack of evidence to support them [13]. Delineation of the contribution of the medial soft-tissues to functional constraint is imperative so as to avoid iatrogenic laxity [3]. Releasing the sMCL may correct varus deformity [31] and relax the lateral structures, but it may adversely affect stability in anterior
drawer or internal-external rotation [6,23]. Similarly, the dMCL may be damaged by tibial bone resection, particularly in a small knee [18], but the role of the dMCL post-arthroplasty has not been measured.

The objective of this study was to determine the contribution of different medial structures in stabilising the implanted knee and identify whether the choice of implant affects this. It was hypothesised that the sMCL would be an important restraint in valgus and internal-external rotation, and that this would be of similar magnitude in both CR and PS knees.

Materials and Methods

Specimen Preparation

Following ethics approval, eight fresh-frozen human cadaver (four female and four male) knee specimens of mean age 79 (range 59–96) were obtained from a tissue bank (five left-sided and three right-sided). None of the knees exhibited more than superficial articular surface changes, misalignment, or fixed flexion. For each specimen, the tibia/fibula and femur were skeletonised 80mm and 110mm from the joint line respectively; within these limits the skin, musculature and soft tissue structures were left intact. The head of the fibula was transfixed to the tibia in an anatomical position using a transcortical bone screw. Any excess proximal femoral bone and distal tibial and fibular bones were removed.

The femur and tibia were fixed in 60mm diameter cylindrical steel pots using polymethyl methacrylate bone cement. The tibia was aligned centrally in the bone pot using a jig with a pointer that located the centre of the tibia as between the tips of the tibial spines [2]. To access the tibial plateau, a midline incision was made to the skin and subcutaneous fat layer, followed by a medial parapatellar arthrotomy [28]. The arthrotomy was opened and resutured at each
stage of the experiment. The femur was cemented into the bone pot whilst in full extension and the posterior condylar axis was aligned parallel to the femoral fixture.

Robotic biomechanical testing system

The cadaveric specimens were tested using a robotic knee joint biomechanical testing system, consisting of a six degree of freedom (DOF) industrial robotic manipulator and robot controller (TX90 and CS8C; Stäubli Ltd, Zürich, Switzerland), and a six axis force/torque sensor (Omega 85; ATI Industrial Automation, Apex NC). The robotic manipulator had a payload of 200N and repeatability of 0.03mm in translation (manufacturer’s specification). The sensor had a force sensing range of 3800N (resolution ±0.43N) for the Z axis and 1900N (resolution ±0.29N) for the X and Y axes, and torque sensing range of 80Nm for Z, X and Y axes (Z resolution ±0.009Nm and X-Y resolution ±0.013Nm). The femoral pot was fixed rigidly in a fixture on the base of the robot, and the tibial pot was attached to the force sensor connected to the end-effector of the manipulator (Fig. 1).

Testing Protocol

The knee was manually flexed 20 times to avoid soft tissue hysteresis, then in the robot the path of passive motion of the intact knee was found by applying a flexion rotation. During the flexion, the robotic system minimised forces and moments acting across the knee. The robotic system recorded the positions during the passive flexion, to determine the starting points for the loaded tests. To allow for variability to attain geometrical full extension in the implanted knees, the starting positions for the loaded tests were 4° (±3.8° standard deviation), then 19°, 49° and 79°. These flexion angles were nominally fixed by the robotic system and confirmed using high-resolution images and ImageJ software (Version 1.49p, National Institutes of Health, Bethesda, MD).
At each flexion position, a ±90N anterior-posterior (AP) force, a ±8Nm varus-valgus (VV) torque and a ±6Nm internal-external (IE) rotational torque were applied. In each situation, the robotic system minimised the loads in the secondary DOF to the primary applied force/torque. These loads were chosen as being comparable to other studies [23,22,9], with 90N being a similar force to that applied by a KT 1000 arthrometer in AP drawer [7]. Each test was repeated three times.

After intact knee data collection, a CR TKA (PFC Sigma; DePuy Synthes Joint Reconstruction, Leeds, UK) was implanted by a consultant orthopaedic surgeon using a medial parapatellar approach. The surgical technique used a standard combination of measured resection and gap balancing performed in full extension and 90° of flexion. The femur was referenced using an intramedullary guide rod set at 5° of valgus. The cutting block was placed against the distal femoral bone in neutral rotation with respect to the epicondylar axis, and a measured 9mm resection was made from the least affected side of the distal femur. Femoral sizing was performed using an anterior down technique. On the tibial side, an intramedullary rod was used with a 3° posterior slope on the tibial block positioned with respect to the tibial anterior prominence. This corresponded to the centre of the tibial tuberosity in our knee specimens. 10mm of bone was resected from the least affected, most superior proximal tibial surface. Gap balancing was confirmed using spacers to achieve a rectangular space both in full extension and flexion after bone resection but before chamfer femoral cuts. The tibial component was cemented to the bone whilst the femoral component was implanted using a press-fit technique. It was known from prior work that press-fitting the femoral component gave secure fixation at the experimental loads [5]. No soft tissue releases were performed and ‘tenting’ of the collateral ligaments was avoided by removing any osteophytes. A stable knee was taken as that which allowed for normal unaided unimpeded patellar tracking, did not exhibit either medial or lateral opening after implant trialling and was confirmed through a passive full range of movement.
from full extension. The knee joint with the CR TKA was then tested using the same procedure as described for the intact knee.

The effect of a medial release with the CR TKA was tested by transecting in sequence the following structures: the dMCL, the sMCL, and the PMC (Table 1). Fibres of the dMCL were cut just distal to the joint line, and remnants of the medial meniscus and its connected tissues were removed. The sMCL was released in two stages: firstly a ‘Whiteside’s release’ was performed using an osteotome passed deep to the anterior fibres and elevating them from their tibial attachment [32,31]. The anterior fibres were easily differentiated in deep flexion because they were more taut than the posterior fibres. The second stage was a transection at the joint line across all the sMCL fibres. The PMC was cut across the fibres attached to the semimembranosus tendon and across other connective tissues just distal to the joint line located posteromedially from the dMCL.

After each transection stage, the measured kinematics from the CR TKA implanted stage were repeated to allow the reduction in restraining force/moment to be measured and thus, using the principle of superposition [25], attribute this to the force/moment restraint which had been offered by each cut soft tissue.

In four out of the eight knees, the CR TKA was replaced prior to the medial release stages with a PFC Sigma PS implant. The PS conversion retained the same tibial tray with a different polyethylene tibial inlay; the resection of the PCL required extra cuts into the femur to fit the PS femoral component with box feature, again cement-free. The knee joint with the implanted PS TKA was then tested using the identical procedure as for the CR TKA.
Data Analysis

Mean peak forces/torques and translations/rotations were calculated using a custom Matlab (Mathworks, Natick, MA) script. After each medial cut, the drop in force/torque required to repeat the kinematics was attributed to the restraint offered by the cut structure as a percentage of the original force/torque value. Statistical analysis was performed in SPSS 22 (IBM SPSS Statistics, version 22, Armonk, NY). Two-way repeated-measures analysis of variance with pairwise comparisons with Bonferroni correction was performed to compare the force/torque contribution (dependent variable) to the medial structure cut at different flexion angles (independent variables), and to compare laxities (dependent) to the knee state (intact, CR TKA, PS TKA) at different flexion angles (independent). To compare the contributions of the sMCL in CR and PS TKAs, a one-way analysis of variance was used. For all analyses, significance level was set at p<0.05. A significant restraint/stabiliser at a given flexion angle was defined as having a mean resisting contribution greater than 10% with p<0.05.

Results

Anterior-Posterior Translation

Under a 90N anterior force, the CR knee was significantly more lax than the intact knee at 19° (p=0.012), 49° (p<0.001) and 79° (p=0.002), whilst the PS knee was significantly more lax than the intact knee at 49° (p=0.005) and 79° (p=0.009). However, a significant difference was not found between the CR and PS knees at any flexion angle (Fig. 2). In all implanted knees, the sMCL was the greatest medial stabiliser to anterior translation (Fig. 3), with a percentage contribution ranging from 18 ± 12% (mean ± standard deviation, p=0.029) at 4° flexion, to 29 ± 11% (p=0.001) at 49°. ‘Whiteside’s release’ of the anterior sMCL fibres reduced the anterior restraint by 10 ± 8% at 49° flexion (p=0.070), however the other medial structures each contributed less than 10% of the anterior restraint.
Under a 90N posterior force, both the CR and PS knees were significantly more lax than the intact knee at 4° (p=0.034, 0.036), 19° (p=0.007, <0.001) and 49° flexion (p=0.037, 0.003) respectively. A significant difference was not found between the CR and PS knees at any flexion angle. On average, the sMCL contributed 10% of the resistance to posterior drawer in implanted knees at all flexion angles (Fig. 3), however this was only found to be significant at 4° (p=0.03). With increasing flexion a trend was observed showing that there was a larger reliance on the sMCL to restraining posterior translation in the CR than the PS implant, however this was not significant.

**Internal-External Rotation**

In response to a 5Nm torque, no significant difference was found in internal rotation laxity between the intact, CR or PS knee at any flexion angle (Fig. 4). The dominant medial restraint to internal rotation in all implanted knees was the sMCL (Fig. 5), with contributions ranging from 17 ± 8% at 4° (p=0.004) to 25 ± 11% at 49° (p=0.003) flexion. ‘Whiteside’s release’, the dMCL and the PMC resisted less than 10% of the internal rotation torque at any flexion angle. When a 5Nm external rotation torque was applied, no significant difference was found in rotational laxity between the intact, CR or PS knee at any flexion angle (Fig. 4). Again, the sMCL offered the greatest medial restraint to rotation in all implanted knees (Fig. 5), with contributions ranging from 33 ± 15% (p=0.003) at 4° to 39 ± 13% (p<0.001) at 49° flexion. ‘Whiteside’s release’ reduced the external restraint by 13 ± 12% at 4° flexion (p=0.104), to 14 ± 12% at 79° flexion (p=0.095), but was not found to be significant. The dMCL had a significant contribution to resisting external rotation of 11 ± 7% at 19° flexion (p=0.028), whereas the PMC peaked at 12 ± 10% at 4° but that was not found to be significant (p=0.056).
Varus-Valgus Rotation

In response to an 8Nm valgus torque, no significant difference in rotational laxity was found between the intact, CR or PS knee at any flexion angle (Fig. 6). In all implanted knees, the sMCL was the primary restraint to valgus rotation at all angles of flexion tested (Fig. 7), with nearly-constant contributions from 59 ± 30% (p<0.001) at 4°, up to 65 ± 14% (p<0.001) at 19° flexion. ‘Whiteside’s release’ reduced the valgus restraint by 16 ± 13% at 4° flexion (p=0.073), to 21 ± 23% at 79° flexion (p=0.186); the change with knee flexion was not found to be significant. The PMC resisted 11 ± 7% of the valgus torque at 4° (p=0.028), which dropped substantially with increasing flexion, and the dMCL restrained 11 ± 6% (p=0.008) and 12 ± 7% (p=0.012) at 4° and 19° respectively. Between the implants a trend was noted: with increasing flexion there was a larger reliance on the sMCL to restrain the valgus torque in the PS than the CR implant: this was found as significant at 19° (p=0.034) and 49° flexion (p=0.011).

In response to an 8Nm varus torque, no significant difference was found in rotational laxity between the intact, CR or PS knee at any flexion angle (Fig. 6). None of the sectioned medial structures resisted the 8Nm varus moment significantly.

Discussion

The most important finding of the study was that the sMCL is the primary medial ligamentous restraint in the implanted knee, demonstrating a consistent role in resisting valgus, internal-external rotations, and anterior translation at all flexion angles examined. Another finding was that no significant difference in laxity was found between the CR and PS TKAs. When comparing the sMCL contributions between the implants, no evidence was found to suggest that the cam-box mechanism of the PS TKA could improve stability of the implanted knee when faced with severe medial ligamentous deficiency; on the contrary, the loss of restraint by
the PCL led to significantly raised restraining actions being imposed onto the sMCL after PS than CR TKA. Therefore surgical release of the sMCL can result in gross laxity not compensated for by the other medial structures or the relatively unconstrained CR or PS implants. Attempts at correction of pre-existent varus through release of the sMCL may inadvertently result in a new combined laxity pattern.

Previous work has investigated the contributions of the medial structures in the native knee [23,22,12,15], but little work exists on the roles of these ligaments after knee replacement [3]. Studies using a robotic system to test CR/PS TKA performance have either focussed on flexion arc comparisons [17,20,33], or have applied AP, IE and VV to implanted knees without investigating soft tissues [21]. Other papers that investigated the function of the MCL and PMC after TKA with surgical releases (such as in operative varus correction) were only able to compare increased laxity and were not able to determine the percentage contribution made by entire structures in restraining motion [26,32]. It is mechanically equivalent to transect the ligaments as a sub-periosteal release, but in this time-zero study healing effects at the insertion points were not of interest.

Previous cadaveric studies on the sMCL in the intact knee have similar findings to the implanted knees in this study, so the mechanics of medial ligamentous stabilisation of the knee were largely preserved after TKA. The changes which have been found include a significant contribution to resisting anterior drawer in an ACL-deficient knee, in internal-external rotation whilst in flexion, and in valgus rotation at all flexion angles [15,23,30]. Along with the sMCL, it was found that the PMC acted as a restraint in valgus rotation when the knee was near full extension, and in valgus, internal and external rotation the contribution of the PMC significantly decreased with increasing flexion. This concurs with the posterior fibres being stretched in extension and slackened with flexion [24]. In some specimens it was difficult to identify the dMCL and its tibial attachments. This observation agrees with the study by Maes
et al. [18], which found that on average 54% of the dMCL tibial attachment area was resected in 33 cadaveric knees after a standard 9mm TKA tibial cut. It was also evident that the dMCL attachment to the medial meniscus [24] was affected by removing the meniscus during implantation. As a comparison to this study, Robinson et al. [23] found no significant increases in IE laxity in intact knees when the dMCL was cut, and a significant increase in valgus laxity only if the sMCL had been cut previously.

This study found increased AP laxity with the CR and PS implants compared with the intact knee. This disagrees with the work by Saeki et al. [26], which only found an increase in AP laxity after TKA at 90° flexion. This may partially be explained by the lack of axial compression loading in the present study: here the knees were loaded in AP to ±90N. Saeki et al. applied a 45N axial load and only a 35N AP force. It is recognised that the concavities on the tibial plateaus of CR/PS implants act to locate the femoral condyles under axial joint compression, and hence provide knee stability [11]. Another explanation may be due to the unconstrained tests in the present study. With the robot being allowed to introduce secondary motion to minimise forces and moments in the other planes of motion, the stability given by the geometry of the articular surface could be overcome and the absence of one or both of the cruciate ligaments in the implanted knees manifested in increased laxity. This study also found no significant difference in VV and IE laxity between the intact and implanted states. This was in agreement with Stoddard et al. [28], who applied 5Nm IE and 3.5Nm torques in addition to 400N extensor loads during active flexion arcs and found no significant difference between intact and implanted state. Other cadaveric studies with loaded quadriceps and hamstrings, which applied VV and IE to the surgeon’s subjective endpoints rather than fixed moments as in this study, also found there to be similar rotatory kinematics between intact and implanted knees [8,14].
When comparing the contributions of the sMCL between the implants, in most cases there were no large differences between the CR and the PS. However, a difference was observed in valgus rotation, when the PS implant had to rely more on the sMCL to offer restraint than the CR: without the PCL, which is a secondary restraint to valgus rotation [10], a larger load was imposed onto the sMCL.

This cadaveric study has measured the contributions of each of the medial passive soft-tissue restraints to tibiofemoral joint laxity in relatively normally-aligned knees, while the surgeon in clinical practice may have to correct limb misalignment. In the varus knee minimal medial release is required to achieve coronal plane symmetry and restore correct mechanical alignment such that the weight bearing axis passes through the centre of the knee [4]. Over-release of any medial (contracted or otherwise) structures risks defunctioning the knee and creating an instability pattern. The ‘Whiteside’s release’ of the anterior fibres caused a 18% loss of valgus restraint on average across flexion angles, equivalent to losing approximately a third of the restraining action of the sMCL. However, that may be regained by healing post-surgery. In valgus alignment, which is more likely to load the MCL during gait, only the anterior portion of the deep medial collateral ligament should be elevated from the tibial bone and only to a depth sufficient to allow for safe tibial bone resection. Such measured release is akin to measured bone resection performed as a standard operating technique.

Limitations of the cadaveric study include being conducted at time-zero and therefore investigation of long-term stability of the implants and ligaments was not possible. However, since ligament balancing is conducted per operatively, the study is able to give an accurate representation of how the medial soft tissues act to restrain knee laxity at the end of this process. With the current set-up it was unfeasible to add axial compression, and the muscles in the femur were not loaded, both of which would contribute to stability of the implanted knees. An example would be loading the semimembranosus tendon, to which the PMC has several
connections [16], although it may be suggested that due to the orientation of the tendon it is unlikely to provide any restraint itself [1]. Additionally the release of the pes anserinus tendons [19] was not assessed as the muscles were not loaded. Thus this study can be considered to have investigated the purely passive mechanics of the ligamentous tissue on the medial side of the knee.

This work has examined the relative contributions of the medial structures to stability of the implanted primary TKA. There is increasing interest in preserving constitutional varus and minimising medial release and this work argues for preservation of the sMCL to afford the surgeon consistent restraint and a balanced knee for the patient, with both CR and PS TKAs.

Acknowledgments

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References


Figure legends

**Figure 1.** Schematic diagram of the robotic knee joint testing system. The arrows detail the forces/torques applied to the knees during the testing protocol. AP = anterior-posterior translation, VV = varus-valgus rotation, and IE = internal-external rotation. X, Y and Z are defined as the axes of the force/torque sensor.

**Figure 2.** Anterior-posterior translation in response to a ± 90N anterior-posterior force. Error bars denote the standard deviation at each flexion angle. * indicates statistical significant translation compared with the intact state (p < 0.05). CR = cruciate-retaining implant, PS = posterior-stabilised implant.

**Figure 3.** Percentage contributions of the deep and superficial medial collateral ligaments (dMCL and sMCL) and posteromedial capsule (PMC) in resisting 90N anterior-posterior force in implanted knees, with 95% confidence intervals. * indicates a statistically significant contribution greater than 10% at the specified flexion angle (p < 0.05). For the sMCL, the contributions are further separated into cruciate-retaining (CR) and posterior-stabilised (PS) implants.

**Figure 4.** Internal-external rotation in response to a ± 5Nm internal-external moment. Error bars denote the standard deviation at each flexion angle. CR = cruciate-retaining implant, PS = posterior-stabilised implant.

**Figure 5.** Percentage contributions of the deep and superficial medial collateral ligaments (dMCL and sMCL) and posteromedial capsule (PMC) in resisting a 5Nm internal-external moment in implanted knees, with 95% confidence intervals. * indicates a statistically significant contribution greater than 10% at the specified flexion angle (p < 0.05). For the
sMCL, the contributions are further separated into cruciate-retaining (CR) and posterior-stabilised (PS) implants.

**Figure 6.** Varus-valgus rotation in response to a $\pm 8\text{Nm}$ varus-valgus moment. Error bars denote the standard deviation at each flexion angle. CR = cruciate-retaining implant, PS = posterior-stabilised implant.

**Figure 7.** Percentage contributions of the deep and superficial medial collateral ligaments (dMCL and sMCL), ‘Whiteside’s release’ of the anterior sMCL fibres, and posteromedial capsule (PMC) in resisting an 8Nm valgus moment in implanted knees, with 95% confidence intervals. * indicates a statistically significant contribution greater than 10% at the specified flexion angle ($p < 0.05$). For the sMCL, the contributions are further separated into cruciate-retaining (CR) and posterior-stabilised (PS) implants, where $l$ indicates a statistically significant difference between the contribution in the CR and PS ($p < 0.05$).
Fig 2
Fig 3

![Graph showing the contribution of anterior/posterior force to anterior/posterior forces at different knee flexion angles.](image-url)

Legend:
- dMCL
- sMCL
- dMCL (CR only)
- sMCL (PS only)
- PMC
Fig 4

![Graph showing knee flexion angle vs external/internal rotation for different conditions.

Key:
- Intact
- CR
- PS

Knee Flexion Angle (°)

External/Internal Rotation (°)
Fig 5

![Graph showing percentage contribution to internal/external moment vs knee flexion angle.](image-url)
Fig 7

![Graph showing % Contribution to Valgus Moment vs Knee Flexion Angle](image)

Legend:
- dMCL
- sMCL
- sMCL (CR only)
- sMCL (PS only)
- White side's sMCL Release
- PMC
Table 1. Outline of the experimental protocol and data obtained.

<table>
<thead>
<tr>
<th>Knee State</th>
<th>Kinematic Test</th>
<th>Data Obtained</th>
</tr>
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<tbody>
<tr>
<td>Intact knee</td>
<td>± 90N AP, ± 8Nm VV, ± 5Nm IE</td>
<td>Kinematics of intact knee (I)</td>
</tr>
<tr>
<td>Implant CR TKA</td>
<td>± 90N AP, ± 8Nm VV, ± 5Nm IE</td>
<td>Kinematics of CR knee (II)</td>
</tr>
<tr>
<td>Implant PS TKA a</td>
<td>± 90N AP, ± 8Nm VV, ± 5Nm IE</td>
<td>Kinematics of PS knee (III)</td>
</tr>
<tr>
<td>Transect dMCL</td>
<td>Repeat kinematics II / III a</td>
<td>Restraining force/moments from dMCL</td>
</tr>
<tr>
<td>Release anterior</td>
<td>Repeat kinematics II / III a</td>
<td>Restraining force/moments from anterior fibres of sMCL</td>
</tr>
<tr>
<td>Fibres of sMCL</td>
<td>Repeat kinematics II / III a</td>
<td>Restraining force/moments from sMCL</td>
</tr>
<tr>
<td>Transect entire sMCL</td>
<td>Repeat kinematics II / III a</td>
<td>Restraining force/moments from PMC</td>
</tr>
</tbody>
</table>

Key to content: AP = anterior-posterior force, VV = varus-valgus torque, IE = internal-external torque, CR TKA = cruciate-retaining total knee arthroplasty, PS TKA = posterior-stabilised total knee arthroplasty, dMCL = deep medial collateral ligament, sMCL = superficial medial ligament, PMC = posterior medial capsule.

The PS TKA was implanted in four of the total eight knees. In these four knees, the following knee states were subject to kinematics III.