The Influence of Muscle Load on Tibiofemoral Knee Kinematics

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ABSTRACT: A comparative kinematics study was conducted on six cadaver limbs, comparing tibiofemoral kinematics in five conditions: unloaded, under a constant 130 N ankle load with a variable quadriceps load, with and without a simultaneous constant 50 N medial and lateral hamstrings load. Kinematics were described as translation of the projected centers of the medial (MFT) and lateral femoral condyles (LFT) in the horizontal plane of the tibia, and tibial axial rotation (TR) as a function of flexion angle. In passive conditions, the tibia rotated internally with increasing flexion to an average of -16° (range: $-12/-20^\circ$, SD = 3.0°). Between 0 and 40° flexion, the medial condyle translated forwards 4 mm (range: 0.8/5.5 mm, SD = 2.5 mm), followed by a gradual posterior translation, totaling -9 mm (range: $-5.8/-18.5$ mm, SD = 4.9 mm) between 40–140° flexion. The lateral femoral condyle translated posteriorly with increasing flexion completing -25 mm (range: -22.6 to -28.2 mm, SD = 2.5 mm). Dynamic, loaded measurements simulating a deep knee bend were carried out in a knee rig. Under a fixed ankle load of 130 N and variable quadriceps loading, tibial rotation was inverted, mean TR = 4.7° (range: $-3.3^\circ/11.8^\circ$ SD = 5.4°), MFT = -0.5 mm (range: = $-4.3/2.4$ mm, SD = 2.4 mm), LFT = 3.3 mm (range: = $-3.6/10.6$ mm, SD = 5.1 mm). Compared to the passive condition, all these excursions were significantly different ($p \leq 0.015$). Adding medial and lateral hamstrings force of 50 N each reduced TR, MFT, and LFT significantly compared to the passive condition. In general, loading the knee with hamstrings and quadriceps reduces rotation and translation compared to the passive condition. Lateral hamstring action is more influential on knee kinematics than medial hamstrings action. J Orthop Res 28:419–428, 2010

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Knee kinematics are complex and intriguing and have been studied extensively.[1] The deduced model describes posterior translation of the femoral condyles relative to the tibia with increasing flexion. This translation is greater on the lateral than on the medial side, leading to relative internal tibial rotation. However, different methods reveal different patterns, and existing literature is not unanimous in describing "normal" knee kinematics. Differences can be attributed to intrinsic and extrinsic factors. Intrinsic factors relate to the interindividual differences. Most studies included small numbers of specimens, patients, or volunteers, so bias cannot be excluded, even with a normal distribution of anatomic features or kinematic patterns. Extrinsic factors include the experimental setup with differences in quadriceps force, hamstrings cocontraction, triceps surae cocontraction, loads, and mechanical constraints imposed on the joint. Also, the mathematical model used for describing knee kinematics will influence the results.[2–5]

Studies describing passive or unloaded knee kinematics[6–12] should be differentiated from studies describing loaded kinematics. Loaded in vitro experiments typically use load frames, knee simulators, or robots where controlled loads are applied to the joint.[13–25] These experiments can use a feedback loop between ankle load and applied quadriceps force, but are arbitrary when it comes to cocontraction of important muscle groups. Loads exerted by cocontraction of hamstrings and triceps surae remain uncertain, which is reflected in the large range of applied loads use in kinematic experiments.[14,16,20,21,24,25]

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Loaded in vivo research uses MRI imaging,[10,26,27] roentgen stereophotogrammetric analysis,[28] and 2D fluoroscopy with shape matching techniques, based on CT models or combined CT/MRI models.[29–32] These studies are limited due to unknown loading conditions and variation in the activity performed. In contrast, they better simulate normal activities with correct tuning of cocontractions as the subject adapts to external forces imposed by the task.

The aim of this study was to isolate the impact of muscular action on knee kinematics by keeping all other variables constant. In vitro knee kinematics are described in passive conditions, with variable quadriceps loads, with and without hamstrings loads on the medial and on the lateral side. The null hypothesis was that muscle action does not change knee kinematics.

MATERIALS AND METHODS

Six unmatched cadaver limbs (three male, three female; ages: 78–87 years) were disarticulated at the hip and frozen at -20°C. Optical reflective markers were rigidly attached to the femur, tibia, and patella. Volumetric CT scans on a 64-row multidetector computed tomography (MDCT) scanner (General Electric Lightspeed VCT, Milwaukee, WI, USA) were performed on the disarticulated specimens. All CT data were loaded in a 3D visualization software system (Mimics 11.02 and its MedCAD module, Materialise, Haasrode, Belgium) for analysis. After creating a bone surface reconstruction mask, the markers and relevant landmarks for describing alignment and kinematics were marked on the tibial and femoral scans.[33] Prior to the experiment, the specimens were thawed. The hip was amputated 32 cm cranial and the foot 28 cm caudal to the knee joint line. The femur and tibia were rigidly fixed with PMMA in containers. The quadriceps tendon was dissected and looped around a metal bar, 7 cm proximal from its attachment to the patella and securely fixed with Ticron® no. 5 and Mersilene® tape. The biceps tendon was dissected and attached to a Ticron® no. 5 suture. The semimembranosus and semitendinosus tendons were prepared in a similar fashion.

A dynamic knee simulator system, based on the Oxford Rig, was designed to simulate and record motions and loads during squatting. The hip joint can move up and down and can flex and extend. The ankle joint can move mediolaterally and has all three rotational degrees of freedom (flexion–extension, internal–external rotation, ab–adduction). Thus, the knee has all six degrees of freedom. Only knee flexion angle is controlled by programming the hip position as a function of time: upward movement extends the knee, downward induces flexion. Translations and rotations along these axes are governed by the geometry and anatomy of the joint. One actuator simulates the quadriceps muscles, and a second produces vertical hip motion. The quadriceps actuator is positioned on the upper leg in a way that reproduces its anatomical location and thus its moment arm with respect to the knee. Two constant force springs ($50\ N \pm 5\ N$ each) load the hamstrings on the lateral and medial side of the tibia. They are fixed to the metal frame representing the pelvis to reproduce their biarticular function. Their position is such that their moment arms are similar to in vivo. Sensors placed in line with the actuators detect the quadriceps force, the ankle forces and moments, and hip height relative to the ankle. A real-time data acquisition and closed feedback system (Labview, National Institute, Denton, TX, USA) was used to perform a squat with a certain hip velocity, while simultaneously applying a quadriceps force to induce a vertical ankle force (defined as a function of time). The hip actuator is controlled by error feedback from the hip position sensor using a proportional-integral-derivative (PID) controller. If the hip position is too high with respect to the programmed position (i.e., the hip is lagging behind), the actuator is instructed to speed up and vice versa. Likewise, the quadriceps actuator is controlled by error feedback from a six-axis ankle load cell under the ankle using a similar PID system. Both control loops are nearly independent. Based on repeated simulated squats, the ankle load falls within the range of ± 13 N in 95% of system measurements from repeated squat trials. The hip position is accurate within 2 mm maximum error between targeted and measured hip position.

Passive motion was performed first, after attaching the femoral container to the free rotating "hip" of the rig. Specimens were cycled manually five times from full extension to maximum flexion. The specimen was then attached to the "ankle" of the rig. Testing was performed at constant speed with a constant vertical ankle load of 130 N from about 25 to 120° flexion. Higher extension positions could not be tested dynamically as the knee could be pulled in hyperextension by the quadriceps mechanism, damaging the specimen. Quadriceps loaded simulated squats were recorded without hamstrings loads, and with sequential loading of the medial and lateral hamstrings separately. These conditions were obtained by releasing or attaching the hamstrings sutures to the constant force springs.

Five calibrated infrared cameras (Vicon Motion Systems®, Los Angeles, CA, USA) recorded the motion of femur, tibia, and patella through the rigidly attached optical reflective markers. The markers were fixed to the anteromedial aspect of the femur and tibia, 10 cm proximal and distal of the joint line. The patella marker was fixed with a screw and spike plate to the patellar bone.

The motions of the bones in each trial then were digitally reconstructed. CT data and motion capture data were combined such that the coordinates of the optical markers in both systems were superimposed. This allowed the reconstruction of the anatomic landmarks during motion trials. Anatomic bone coordinate systems and joint rotations were calculated based on the Grood & Suntay[3] open-chain knee model, which was modified to use more accurate femoral landmarks. To define internal–external rotation of the femoral frontal plane, the line joining the medial and lateral condyle centers was used instead of the posterior condyles tangent line.[33] Additional measurements were taken, including the AP translations of the medial and lateral femoral condyle centers (MFT, LFT), projected onto the tibial horizontal plane. The translations were measured as the perpendicular distance of each condyle center to the line connecting the MFT and LFT, also projected onto the tibial horizontal plane.[33]

The kinematics data (mean $\pm$ SD) were plotted as functions of flexion angle in 10° intervals from 30–120°, except for the passive flexion trials, which were available from 0–140°. The kinematics from different muscle action conditions were compared using general linear model analsis of variance (ANOVA) for unequal sample sizes, across muscle action conditions and the flexion angles. Pairwise comparisons were made with the Tukey method. All statistical tests were performed with computer software (Minitab, State College, PA, USA) with significance set at $p = 0.05$.

RESULTS

Inter- and intraobserver variability of marking the geometrical reference points for defining planes and axes has been published previously.[33] Accuracy and precision of the motion analysis system, used for the kinematic recordings of the markers, was on the order of 0.2 to 0.3 mm.

Passive Motion (Fig. 1 and Table 1)

From 0–130° flexion, the tibia rotated internally with flexion to an average of $-16 \pm 3.0°$ (range: -12 to $-20°$). Between 0 and 40°, the medial condyle translated forward 4 ± 2.5 mm (0.8 to 5.5 mm), followed by a gradual posterior translation, totaling -9 ± 4.9 mm (-5.8 to -18.5 mm) between 40 and 140°. The lateral femoral condyle translated posteriorly with increasing flexion completing -25 ± 2.5 mm (-22.6 to -28.2 mm).

Loaded Quadriceps Only, No Hamstrings

From 30 to 70° flexion, the tibia was more anterior than during the passive motion (Fig. 1). When only the quadriceps was active, tibial axial rotation was inverted (Fig. 2). Between 30 and 60°, the tibia rotated slightly internally over 2°, but beyond 60°, the tibia rotated externally over 6.5° and ended in neutral axial orientation at 120° of flexion. This behavior resulted from anterior translation of the MFT between 30 and 60°, whereas the LFT remained relatively stable. Beyond 60° of flexion, the LFT started moving anteriorly, whereas the MFT remained stable (between 60 and 80°) or moved posteriorly (between 80 and 120°, where the movement was almost identical to the movement in passive conditions).

Overall, the position of the MFT was relatively stable with an average total excursion of -0.5 ± 2.4 mm (range:

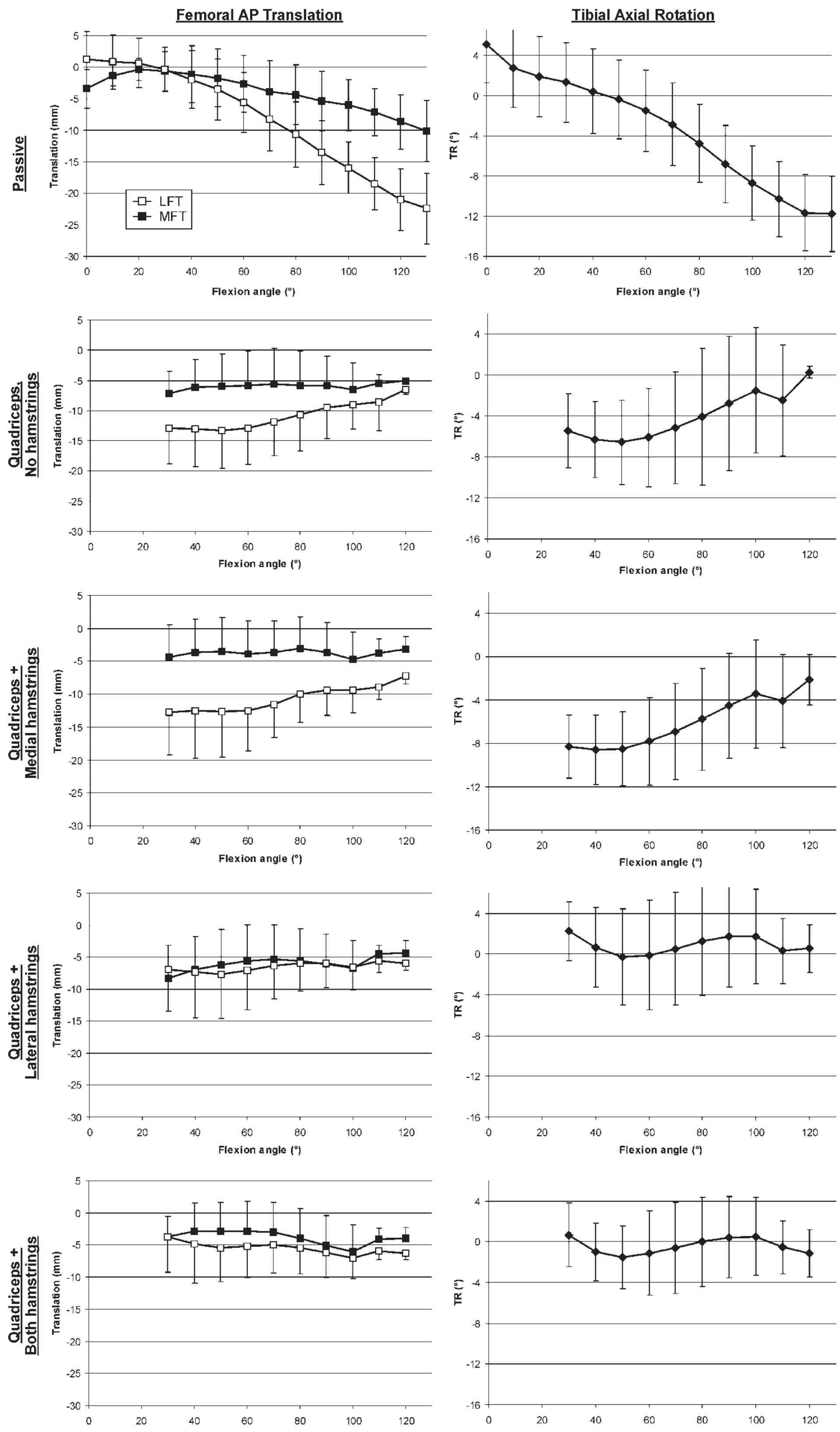

Figure 1. Translation (mm) and rotation (°) plots as a function of the flexion angle. MFT: medial femoral condyle center translation, LFT: lateral femoral condyle center translation, TR: tibial rotation, negative values represent internal rotation, positive values external rotation.

−4.3 to +2.4 mm). The LFT translated anteriorly with flexion with an average total excursion of 3.3 ± 5.1 mm (range: −3.6 to +10.6 mm). All parameters (TR, MFT, and LFT) differed significantly from their values during passive flexion (Table 2). Average LFT total excursion during passive motion in the same flexion range was significantly greater than in this loaded case without hamstrings ($p = 0.01$). Average excursion of MFT during

Table 1. Ranges of Movement for Tibial Axial Rotation and Femoral Condyle Centers

	TR (°) Average (SD) Min/Max	MFT (mm) Average (SD) Min/Max	LFT (mm) Average (SD) Min/Max
Passive (between 30° and 120° of flexion)	−12.3 (3.6)	−8.5 (3.0)	−21.0 (3.4)
	−18.5/−9.7	−12.9/−5.2	−24.3/−17.1
Quad. only	4.7 (5.4)	−0.5 (2.4)	3.3 (5.1)
	−3.3/11.8	−4.3/2.4	−3.6/10.6
Quad. and Med. Hamstrings	5.1 (5.1)	−2.2 (2.4)	2.2 (4.7)
	0/13.1	−5.0/0.6	−1.2/10.4
Quad. and Lat. Hamstrings	−1.2 (5.5)	0.9 (4.0)	−0.3(6.5)
	−7.8/−1.6	−3.5/6.1	−6.1/8.1
Quad. and both Hamstrings	−0.5 (5.8)	−3.5 (2.3)	−4.3(5.0)
	−6.2/8.8	−6.1/0	−7.9/4.6

rotation: − = int, + = ext.
translation: − = post, + = ant.
TR = tibial axial rotation; MFT = medial condyle center translation; LFT = lateral condyle center translation.

passive motion was also significantly greater than in this case ($p = 0.01$).

Effect of the Hamstrings

Active flexion with both hamstrings attached showed little rotation (Figs. 2 and 3). However, when no hamstrings were attached or when only the medial hamstrings were attached, the tibia was more internally rotated (Fig. 2). Passive flexion was significantly different from active flexion in all muscle load conditions, with greater differences at deep flexion (Table 2).

For the translations, significant differences were found among the conditions for both MFT and LFT. However, less pairwise differences among conditions could be detected for MFT than for LFT. In active flexion with the lateral hamstring attached, LFT was closest to zero. Compared to this case, LFT was 2.2 mm more anterior when only the medial hamstrings were attached and 3.3 mm when no hamstrings were attached ($p = 0.0004$ and $p = 0.0119$, respectively.). For LFT, passive flexion was significantly different from active flexion under all load combinations, with greater differences at deep flexion.

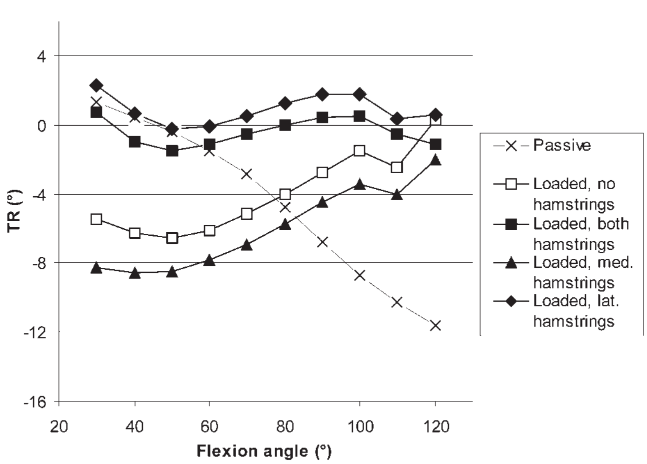

Figure 2. Average tibial rotations for the five conditions superimposed.

Patellar Tendon Angle

The angle between the patellar tendon and the tibial mechanical axis in the sagittal plane was positive between 0 and 65° flexion and negative from 65° to maximum flexion (Fig. 5b).

DISCUSSION

We investigated the effect of muscle action on knee kinematics. Based on the results, the null hypothesis is refuted. Muscle action significantly changed kinematics and affected the medial and lateral side of the joint differently. Significant differences existed between the passive condition and all loaded conditions and between the loaded condition with quadriceps only and the loaded condition with quadriceps + both hamstrings or quadriceps + lateral hamstrings. Adding or removing medial hamstrings did not induce significant kinematic changes.

Several limitations must be considered when interpreting the results. The specimens could be weakened by age or chronic illness. Six specimens are relatively few, given the reported variability in knee kinematics. Tissue quality can degrade over time during the experiment. The loaded squat is a reproducible motion, but does not reflect the full spectrum of knee motions of daily life. The interaction between hamstrings and quadriceps cannot be fully modeled, and the forces exerted in vivo by the hamstring muscles are unknown. The hamstrings load of 50 N medially and laterally was based upon previous cadaver studies.[14,21,24,25,34]

Churchill et al.[21] studied knee kinematics based in a loaded rig with 100 N ankle load and 30 N combined hamstrings load. Kwak et al.[25] used the relative physiological cross-sectional areas to simulate muscular

Table 2. *p*-Values of Pairwise Comparisons Across Different Load Conditions for Tibial Rotation (TR), Medial Condyle Center Translation (MFT) and Lateral Condyle Center Translation (LFT)

		Quad. only	Quad. and Med. Hamstr.	Quad. and Lat. Hamstr.	Quad. and both Hamstr.
TR	Passive	**0.0147**	**0.0024**	**0.0113**	**0.0148**
	Quad. only	—	0.9712	**0.0171**	**0.0092**
	Quad. & Med. Hamstr.		—	0.0037	**0.0048**
	Quad. & Lat. Hamstr.			—	0.3014
MFT	Passive	**0.0129**	0.0572	0.0718	**0.0317**
	Quad. only	—	0.1124	**0.0263**	0.4321
	Quad. & Med. Hamstr.		—	0.2484	**0.0347**
	Quad. & Lat. Hamstr.			—	**0.0205**
LFT	Passive	**0.0116**	**0.0001**	**0.0005**	**0.0020**
	Quad. only	—	0.4308	**0.0119**	0.1396
	Quad. & Med. Hamstr.		—	**0.0004**	0.2811
	Quad. & Lat. Hamstr.			—	0.1094

Bold values are significant.

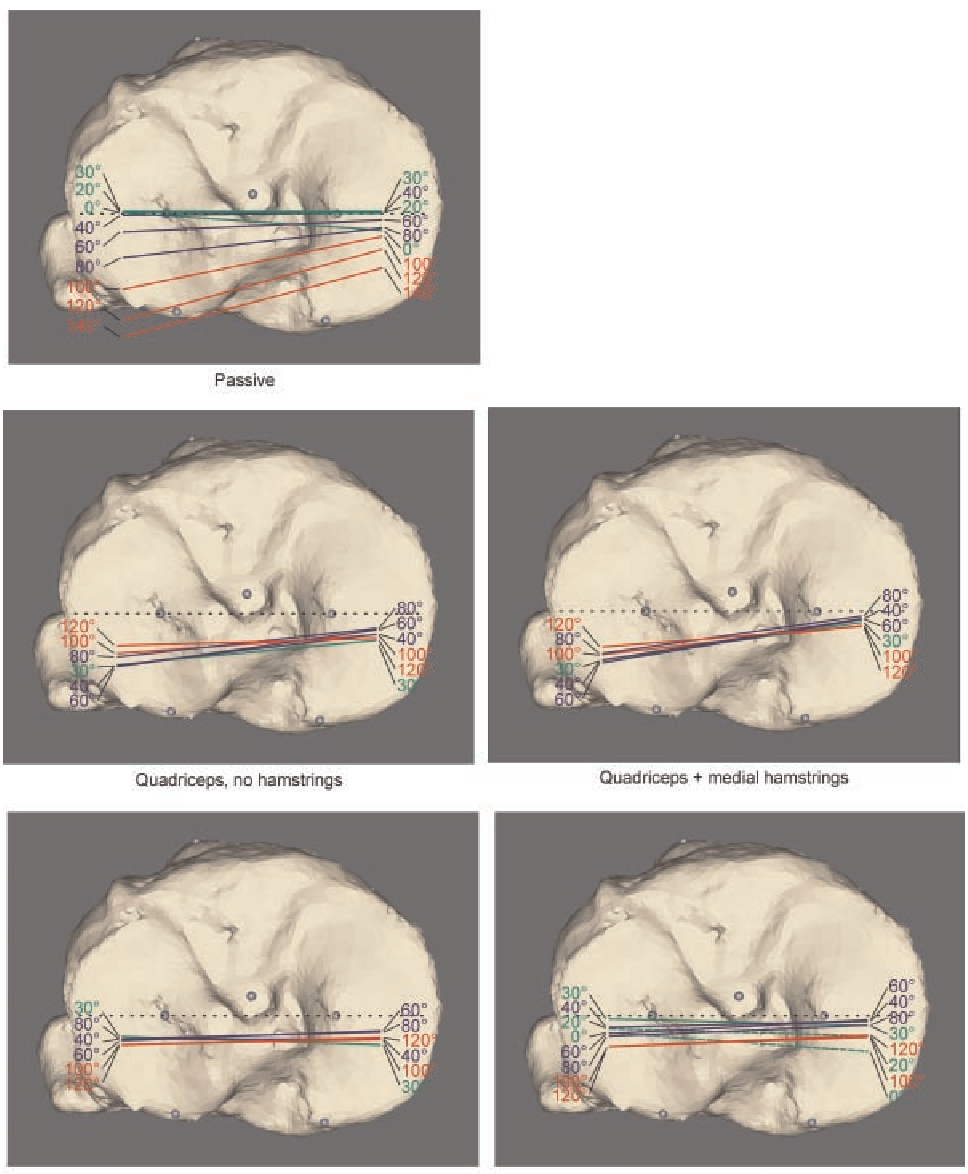

Figure 3. Projection of medial and lateral femoral condyle centers on the horizontal tibial plane for the different load cases.

cocontraction. For a quadriceps force of 534 N, they applied 111 N to the biceps femoris, 111 N to the semimembranosus, and 45 N to the pes anserinus muscles. Other authors used forces ranging from 45 N to 90 N for the hamstrings.[14,34] As we assumed that antagonist cocontraction should equal agonist action in physiologic motion, we considered the Kwak hamstrings values too high and chose the range proposed by the other authors.

Our data from the loaded setting (quadriceps + ham-hamstrings) seem at odds with several in vivo experiments. A partial explanation can be found in the weaknesses of the in vitro setup, not making use of the triceps surae action and ankle forces to replicate a deep knee bend. Also, the timing and magnitude of hamstrings cocontraction is chosen arbitrarily and might not resemble the in vivo setting. As our targeted ankle load was limited to 130 N, the mean quadriceps load observed at 30° was 263 N, rising to 1149 N at 110° flexion, creating a relatively low quadriceps contraction/hamstrings cocontraction ratio at low flexion angles. In addition, in in vivo studies the activities are often arbitrary because of methodological and technical constraints. In the study by Nakagawa et al.,[26] measurements were made with subjects "lying on the side" with the investigated leg supported by the table. External forces are likely to be small and muscle forces probably low. Johal et al.[27] reported kinematics during a loaded squat in an open coil MRI. They described the subject's body position during the experiment; volunteers performed a "wall sit" with the back supported by a board that was inclined backwards at 10°. Depending on the friction between the back of the subject and the wall, part of the subject's weight is removed from the knee joint. Moreover, the contact force with the wall has a horizontal component affecting the forces on the knee. Moro-Oka et al.[32] described three activities. The greatest tibial internal rotation was seen in the knee and lunge activities. According to their description, either the contralateral limb or the upper limbs contributed significantly in load sharing during these locomotor tasks, reducing the load on the studied knee. The stair activity resembled best the in vitro simulated knee bend as performed in our study. For this in vivo activity, the least tibiofemoral rotation and translation was reported.

Several authors described passive knee kinematics (Table 3).[6–12] Figure 4a shows a plot of our tibial rotations superimposed on the previously published data. The trend is the same for all studies, but the values show variation depending on methodology. The stable position of the medial condyle over the flexion arc from full extension to 110°, as shown by the group of Freeman,[7,10,27] is not confirmed in our study. Our parameters are close to those reported by Most,[6] with initial forward sliding of the medial femoral condyle to a maximum anterior position at 40°, followed by posterior translation, completing −12 mm. On the lateral side, posterior translation starts immediately, totalling −25 mm. The tibia undergoes internal rotation of 16° relative to the femur over the flexion arc. Most internal rotation occurs between 0 and 20° (screw-home) and between 70 and110°. From 110° to full flexion, no further rotation occurs because of equal posterior translation on the medial and lateral side.

Considering our results, the influence of muscle action on knee kinematics is undeniable. Recognition of this phenomenon is not new, and the clinical consequences can be extensive. Our findings on the effect of quadriceps loading confirm previous experiments. Beynnon et al.[35] reported that quadriceps contraction strained the anterior cruciate ligament (ACL) at 30° flexion but not at 90°, using in vivo techniques. Arms et al.[36] reported ACL strain to increase to 45° flexion and to decrease beyond 60°. This corresponds with our findings of anterior translation of the tibia up to 60° flexion, followed by posterior translation, if the quadriceps is loaded in isolation. As isolated quadriceps contraction can produce forces beyond those required for ACL tensile failure,[37] the role of hamstrings cocontraction is likely of significant clinical importance. Hamstring cocontraction has indeed been recognized as a stabilizing factor[38] and a protagonist of the ACL as it limits tibial anterior translation and internal rotation.[24,25,39] Renström showed in a cadaveric study, an increased ACL strain upon quadriceps loading between 0 to 45° in comparison to the passive condition. Simultaneous hamstrings action did not counteract this phenomenon unless flexion exceeded 30°.[39] MacWilliams et al.[24] tested the effect of hamstrings force in a loaded dynamic rig in two conditions: one loaded condition (hip vertical load and

Table 3. Overview of Recently Published Studies Reporting Passive Knee Kinematics

Authors	Materials	Motion	Measurement Technique
Blankevoort	4 cadavers	series of static positions	RSA
Hill	6 cadavers & 13 subjects	series of static positions	MRI
Iwaki	6 cadavers	series of static positions	MRI
Johal	10 subjects	series of static positions	MRI
Li	13 cadavers	least resistance path, robot	MRI
Lu	8 subjects	dynamic tests	Fluoroscopy CT models
Most	6 cadavers	least resistance path, robot	Digitised models
Wilson	15 cadavers	dynamic tests with rig	MRI

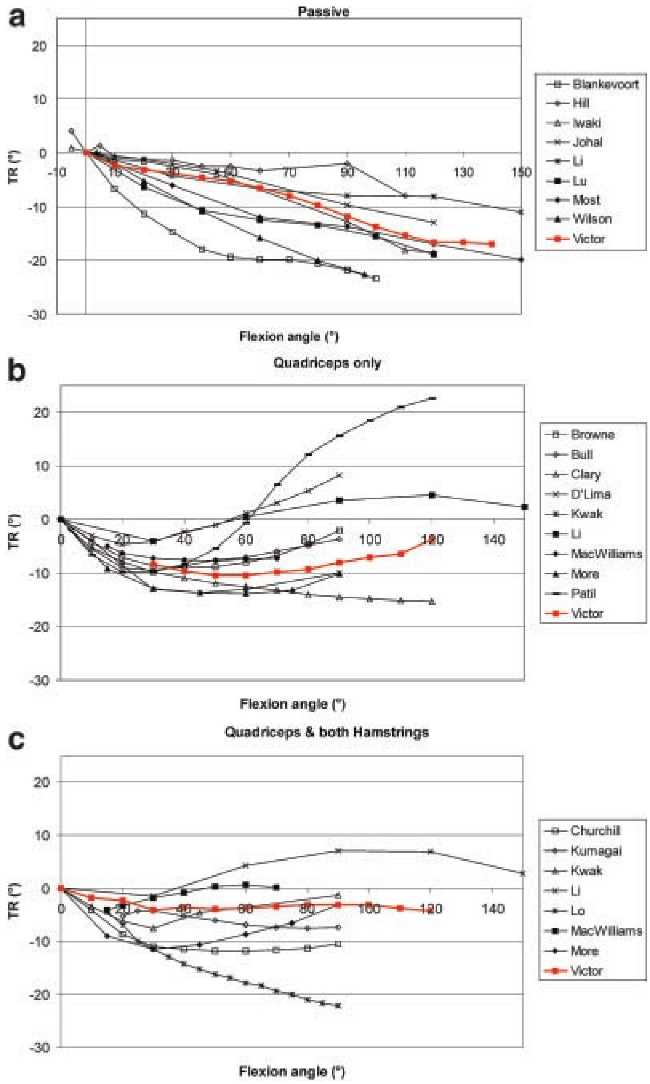

Figure 4. Tibial rotation as a function of the flexion angle based on previous publications. All curves were offset to pass through 0 at 0° of flexion. If no data were available at 0° flexion, the curve was offset to pass through the average TR value of all other papers at the lowest flexion angle where data were available. (a) Passive; (b) quadriceps only; (c) quadriceps + hamstrings.

variable quadriceps load) without hamstrings load and with hamstrings load, equivalent to the vertical load. They found the hamstrings to significantly reduce internal tibial rotation, anterior tibial translation and caused a reversed direction of shear force on the tibia. Similar results were reported by More et al.,[14] using a closed chain experimental setup combining quadriceps load with a constant hamstrings load of 0, 45, or 90 N. Kwak et al.[25] tested five cadaveric specimens statically in open chain between 0 and 90° of knee flexion under different muscle loading modes: quadriceps loading, in combination with or without hamstrings and iliotibial band loading. Despite a greater force being applied on the medial hamstrings (156 N) than on the biceps tendon (111 N), the tibia rotated externally with hamstrings force. The authors explained this by "a larger effective moment arm about the tibial shaft" because of the eccentric attachment of the biceps on the fibular head. The authors considered the hamstrings the most effective knee stabilizers (with respect to ACL function) and warned against the prelevation of hamstrings grafts for ACL reconstruction. They did not, however, distinguish between the separate role of the medial hamstrings versus the biceps. To our knowledge, no study examined sequentially passive motion versus loaded conditions separating quadriceps and medial and lateral hamstrings force. The available data on tibial rotation obtained in kinematic knee rigs in loaded conditions are summarized in Figure 4b and c. The strength of our study is the controlled measurement of the kinematic effects of sequential modes of loading in a continuous closed chain model.

We were able to calculate the patellar tendon angle, which is important in analyzing the effect of quadriceps loading, because anterior or posterior tibial translation is governed by force components in the sagittal plane. The patellar tendon pulls the tibia forward between 0 and 65° flexion (Fig. 5). In deeper flexion, the patellar tendon load causes the tibia to translate in a posterior direction. These results closely match the data previously published by Herzog et al.[41]

To help understand the effects of muscle loading on tibiofemoral kinematics, the results were reformatted (Fig. 6). The projections of the MFT and LFT on the tibial horizontal plane for the different load cases are superimposed for 30, 60, 100, and 120°. At 30°, the effect of the different muscle loads is explained as follows:

- The quadriceps (Q) induces anterior tibial translation. Consequently, the projections of the MFT and LFT shift posteriorly on the tibia. Because the medial compartment is inherently more stable, translation is less on the medial side, inducing internal tibial rotation.
- Adding 50 N medial hamstrings (Q + mH) load only affects the medial compartment. The hamstrings action counteracts the translation induced by the quadriceps, but not fully; medial hamstrings cocontraction fails to fully compensate the anterior shear and *increases* the internal rotation initiated by the quadriceps load.
- Adding 50 N lateral hamstring load (Q + lH) only affects the lateral compartment. The biceps action counteracts the translation induced by the quadriceps, but not fully: biceps cocontraction fails to compensate the anterior shear but *decreases* internal tibial rotation.
- Adding medial and lateral hamstrings cocontraction (Q + H) translates the tibia posteriorly, bringing it close to its relative position in the passive condition; both hamstrings effectively counteract shear and rotation induced by the quadriceps.

At 60° of flexion, the effects are similar, but with smaller absolute values. In deeper flexion (100 and 120° flexion), all combinations of muscle loads cause posterior

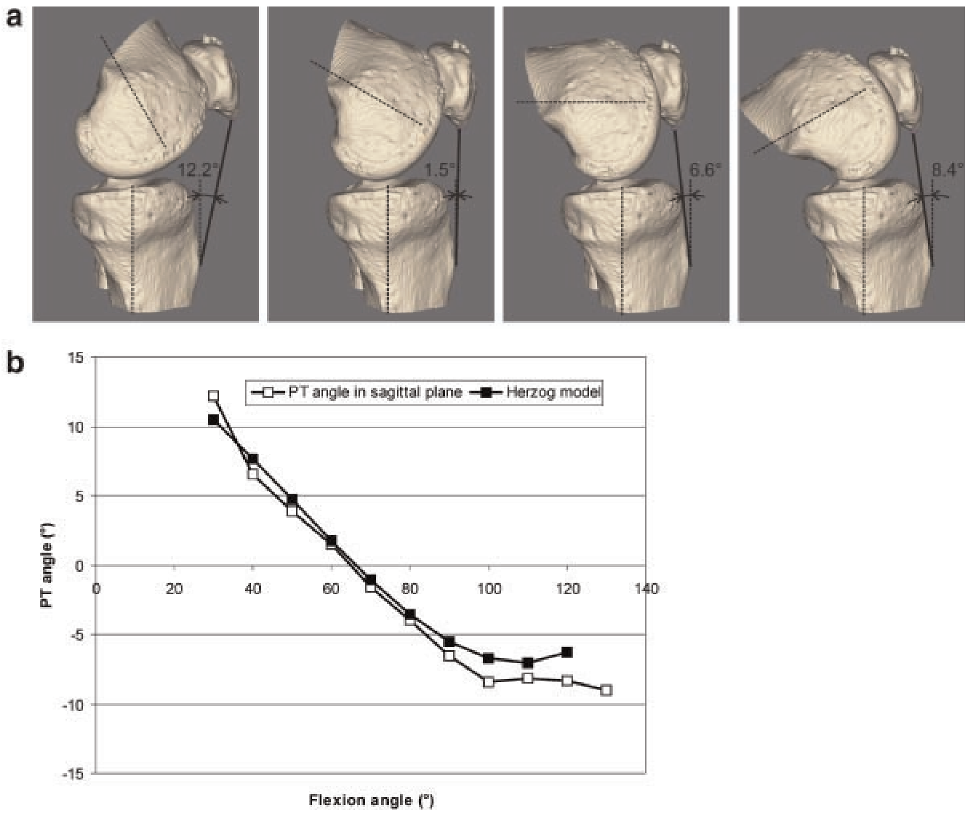

Figure 5. (a) Surface reconstructions showing the measurement of patellar tendon angle to the tibial mechanical axis in the sagittal plane. (b) Angle between patellar tendon and tibial mechanical axis in the sagittal plane of the tibia as a function of flexion angle.

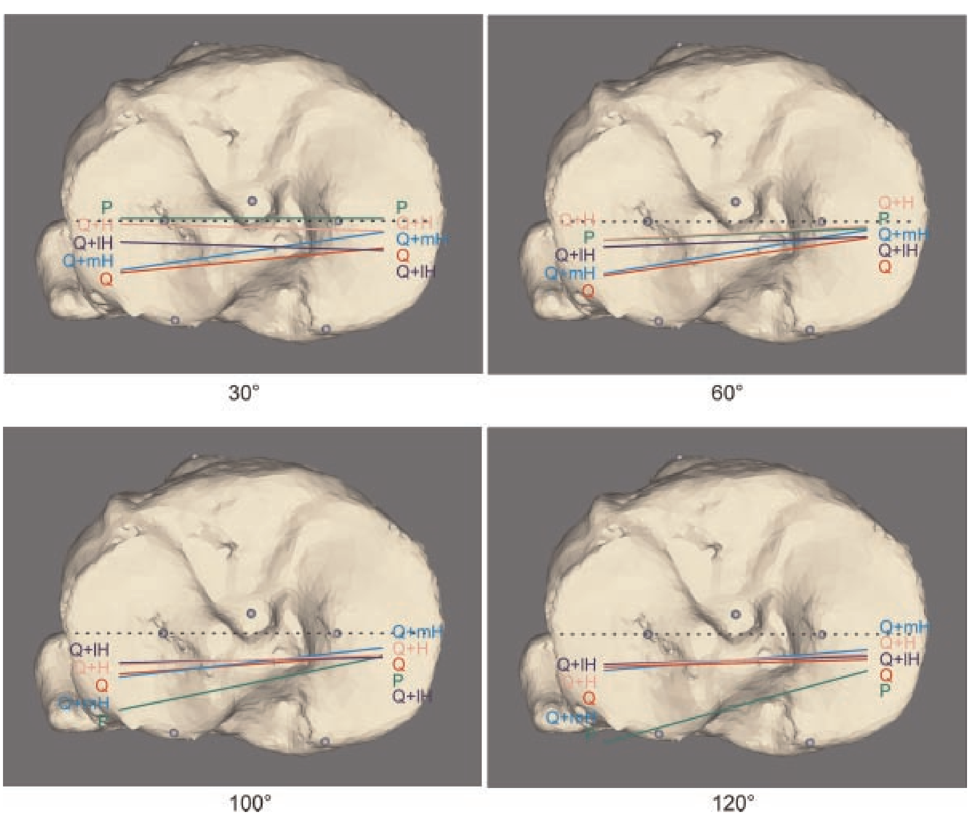

Figure 6. Projection of medial and lateral femoral condyle centers on the horizontal tibial plane for the different load cases grouped for four selected flexion angles.

tibial translation compared to passive conditions. Quadriceps loading reduces tibial internal rotation and anterior translation ("femoral roll-back"). Biceps loading enhances this reduction in tibial internal rotation.

These results confirm earlier work by MacWilliams et al.[24] and Kwak et al.,[25] among the active flexion trials, the action of both hamstrings together limits femoral AP translation and rotation. As such, the hamstrings action is protective of ACL loading. The greater effect of the biceps compared to the medial hamstrings can be explained by the larger effective moment arm[25] and by the greater natural laxity of the lateral compartment.[42] We disagree with Kwak et al. when it comes to the clinical consequences of ACL graft choice. The greatest muscle induced shear force on the ACL occurs in early flexion when the quadriceps translates the tibia forward and induces internal rotation. Our results demonstrate the greater effect of the biceps compared to the medial hamstrings in reducing tibial anterior translation. In addition, at 30° flexion, the medial hamstrings cocontraction increase tibial internal rotation in contrast to the biceps, which decreases tibial internal rotation. As for graft choice, no apparent biomechanical reason exists against selecting the semitendinosus or gracilis tendon for ACL reconstruction.

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