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Relative contribution of the ACL, MCL, and bony contact to the anterior stability of the knee

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Abstract Ligaments and other soft tissues, as well as bony contact, all contribute to anterior stability of the knee joint. This study was designed to measure the in situ force in the medial collateral ligament (MCL), anterior cruciate ligament (ACL), posterolateral structures (PLS), and posterior cruciate ligament (PCL) in response to 110 N anterior tibial loading. The changes in knee kinematics associated with ACL deficiency and combined MCL+ACL deficiency were also evaluated. Utilizing a robotic/universal force-moment sensor system, ten human cadaveric knee joints were tested between 0° and 90° of knee flexion. This unique testing system is designed to determine the in situ forces in structures of interest without making mechanical contact with the tissue. More importantly, data for individual structures can be obtained from the same knee specimen since the robotic manipulator can reproduce the motion of the intact knee. The in situ forces

in the ACL under anterior tibial loading to 110 N were highest at 15° flexion, 103 ± 14 N (mean \pm SD), decreasing to 59.2 ± 30 N at 90° flexion. For the MCL, these forces were 8.0 ± 3.5 N and 38.1 ± 25 N, respectively. Forces due to bony contact were as high as 34.1 ± 23 N at 30° flexion, while those in the PLS were relatively small at all flexion angles. Combined MCL+ACL deficiency was found to significantly increase anterior tibial translation relative to the ACL-deficient knee only above 60° of knee flexion. These findings confirm the hypothesis that there is significant load sharing between various ligaments and bony contact during anterior tibial loading of the knee. For this reason, the MCL and osteochondral surfaces may also be at significant risk during ACL injury.

Key words Anterior knee stability · Anterior cruciate ligament · Medial collateral ligament

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Introduction

Complete ligament ruptures, especially those of the anterior cruciate ligament (ACL), are often associated with other ligamentous injuries, leading to the clinical complaint of knee instability. It has been reported that the incidence of the ACL rupture combined with medial collateral ligament (MCL) tear is up to 20% of all knee ligament injuries [15]. Combined ACL and MCL disruption

has been found to seriously compromise patients' joint stability and also lead to gross deterioration of both the menisci [5, 11, 17, 23] and articular cartilage [9, 14, 19, 24]. For these reasons, the status of other structures in the knee joint may largely influence the outcome of cruciate ligament reconstruction [18].

While the ACL is well recognized to provide the majority of the resistance to anterior tibial translation [4], the biomechanical roles of the MCL and posterolateral structures (PLS) have also been explored to determine their

roles in resisting this knee motion. The MCL and PLS have also been further examined during anterior-posterior (A-P) tibial loading of the cruciate-deficient knee to assess their interaction [10, 12, 22]. One approach to assess the contributions of these structures has been determination of their in situ forces in response to external loads applied to the intact joint [1-3, 6, 7, 16].

We hypothesize that there is load sharing between the ACL, MCL, PLS, and bony contact in an effort to stabilize the knee under anterior tibial loading, and this load sharing is dependent on the angle of knee flexion. If this is the case, simultaneous injury to these structures may occur with ACL tears. This study was therefore designed to assess the in situ load sharing between these structures, in response to 110 N A-P tibial loading. To do this, we utilized a robotic/universal force-moment sensor (UFS) system, which can determine the in situ forces in all knee structures within the same knee specimen, without making mechanical contact with the structure of interest. This system further enabled us to assess the associated changes in multi-degrees of freedom (DOF) knee kinematics with ligament deficiency under the same external loading within the same knee specimen.

Materials and methods

Ten fresh-frozen human cadaveric knee joints (age range, 71-80 years) were used in this study. Following thawing at room temperature, specimens were dissected free of all musculature (except the popliteus) leaving the joint capsule intact. The tibia and femur were cut 20 cm from the joint line and secured within 6-mm-thick-walled aluminum cylinders using epoxy potting compound. The femoral cylinder was then secured relative to the base of the robotic manipulator (Puma model 762, Unimate, Inc., Pittsburgh, Penn.) through an adjustable clamp. The tibial cylinder was attached to the universal force-moment sensor (UFS) (model 4015, JR3, Inc., Woodland, Calif.), which was in turn fixed to the end-effector of the six-joint robotic manipulator (Fig. 1). The repeatability

of the robotic manipulator for position and orientation is less than 0.5 mm and 0.5°, respectively [20]. The UFS can measure three forces and three moments along and about the axes of an embedded Cartesian system with a repeatability within 0.2 N for forces and 0.1 Nm for moments [20].

Following specimen preparation and mounting, the robotic/UFS system used force-moment control to find the 6-DOF path of passive flexion-extension of the intact knee joint from 0° to 90° [21, 25]. This was done by seeking a joint position of "no force, no moment" at a specified flexion angle, and then incrementing the flexion angle by 1° and seeking a new joint position. Once this path of passive flexion-extension was determined, the joint could be moved to the various flexion angles for testing. The 0° flexion position was defined using the long axes of the femur and tibia, and provided a reference for the other flexion angles.

The robotic manipulator was then used to place the joint at 0° flexion and apply external A-P tibial loads to ± 110 N at a rate of 1 mm/s guided by the force/moment measurements from the UFS. Using the UFS to control the application of the external loading to the joint allows only the desired anterior loading to be applied while permitting knee motion in the remaining 5-DOF. During this test, the position and orientation of the tibia were recorded by the robotic manipulator, and the forces and moments applied to the joint were measured by the UFS. A-P tibial loading to ± 110 N was also applied to the intact joint at flexion angles of 15°, 30°, 60°, and 90° [21, 25].

The ACL was then transected through a medial parapatellar incision and the robotic manipulator reproduced the previously recorded motion of the intact knee for 0° flexion. By measuring the forces required to move the ACL-deficient joint through the same path of motion as recorded for the intact knee with the UFS, the in situ force in the ACL was determined as shown in Fig. 2. This was done by comparing the forces measured by the UFS before and after cutting the ACL - for the same joint position. Calculating this difference in forces represents the in situ force in the ACL [8, 13, 20, 21]. To determine the in situ force in the ACL at the other flexion angles, the robotic manipulator then reproduced the recorded motions of the intact joint at 15°, 30°, 60°, and 90° flexion while forces were recorded by the UFS. For each flexion angle, the forces measured by the UFS before and after cutting the ACL were compared to determine the in situ forces in the ACL.

To assess the changes in knee kinematics associated with ACL deficiency, the robotic/UFS system was used to apply 110 N A-P tibial loading to the ACL-deficient joint [20]. The kinematics of the ACL-deficient joint were recorded by the robotic manipulator

Fig. 1 The robotic/universal force-moment sensor (UFS) testing system showing a right knee mounted for testing. The arrows detail the anatomical axes of the knee joint

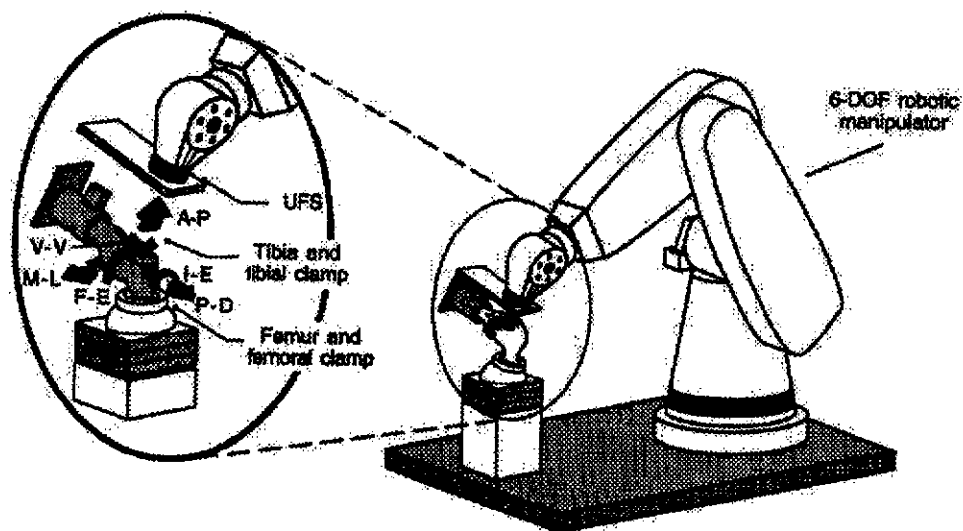


Table 1 In situ forces (N) in the various structures at the knee in response to 110 N anterior tibial load (mean \pm SD, $n = 10$) (ACL anterior cruciate ligament, MCL medial collateral ligament, PLS posterolateral structures, PCL posterior cruciate ligament)

Structures	Knee flexion angle				
	0°	15°	30°	60°	90°
ACL	92.8 \pm 15	103.1 \pm 14	90.9 \pm 17	70.0 \pm 24	59.3 \pm 30
MCL	11.1 \pm 3.8	8.0 \pm 4.0	14.3 \pm 12	28.7 \pm 15	38.1 \pm 25
PLS	6.8 \pm 6.6	8.8 \pm 8.0	10.4 \pm 12	9.4 \pm 12	9.6 \pm 17
PCL	7.7 \pm 2.5	4.6 \pm 1.9	5.7 \pm 3.4	6.5 \pm 2.8	6.1 \pm 2.8
Bony contact	16.7 \pm 11	21.5 \pm 9.3	34.9 \pm 23	29.2 \pm 21	17.0 \pm 15

and could then be compared with the kinematics of the intact joint under the same external loading (see Fig. 4). To further consider the effect of MCL deficiency in addition to ACL deficiency, the MCL was transected and the same external loading was applied to the knee. The changes in kinematics due to ACL+MCL deficiency were then examined for comparison with the intact and ACL-deficient knee.

The in situ forces in the MCL, PLS (posterolateral structures), and PCL were determined under anterior tibial loading to 110 N as described above for the ACL. To evaluate the forces due to bony contact under anterior tibial loading to 110 N, all soft tissues, including the menisci, were removed from the joint. The robotic manipulator was used to reproduce the motions of the intact joint, and the forces recorded by the UFS therefore directly represented bony contact.

Since the above tests were conducted within the same knee specimen, a repeated one-factor ANOVA with multiple contrasts was utilized to analyze the variations of the in situ forces at different knee flexion angles for the ACL, MCL, PLS, and bony contact. Statistical significance was set at $P < 0.05$.

Results

The magnitudes of in situ forces for all knee structures studied at all flexion angles are detailed in Table 1. The in situ forces in the ACL in response to an anterior tibial load of 110 N reached a peak of 103 ± 14 N at 15° flexion (mean \pm SD), and were reduced to 59.2 ± 30 N at 90° flexion. The in situ forces in the MCL displayed an opposite trend with knee flexion: only 8.0 ± 3.5 N at 15° of knee

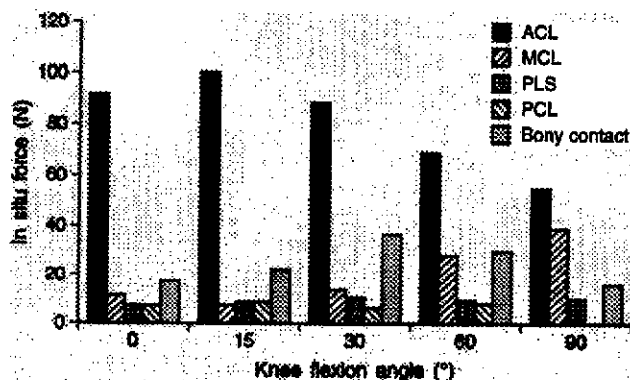


Fig. 2 Variation of in situ forces with knee flexion in response to 110 N anterior tibial load ($n = 10$). Note: With this methodology, all ligament forces are obtained within the same knee during testing

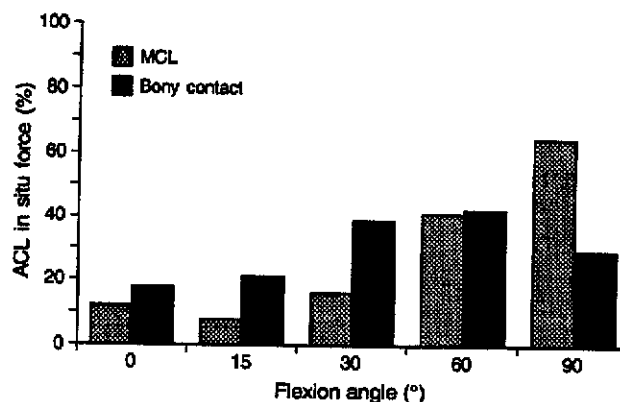


Fig. 3 Ratio of MCL and bony contact forces relative to the force in the intact ACL under ± 110 N anterior tibial loading as a function of flexion angle ($n = 10$)

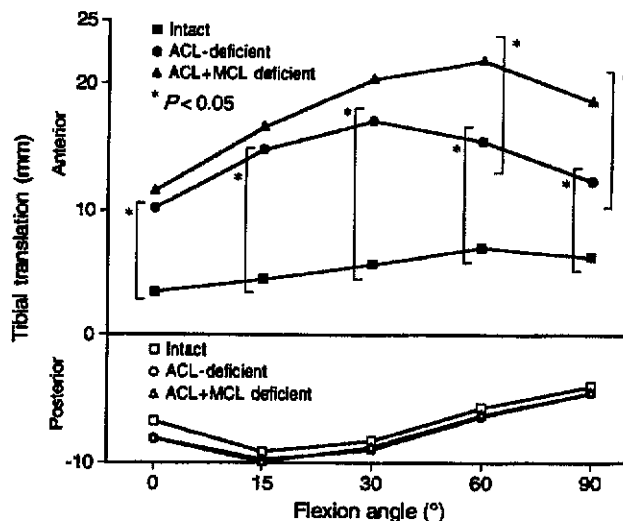


Fig. 4 Anterior-posterior (A-P) tibial translations under ± 110 N A-P tibial loadings as a function of flexion angle ($n = 10$)

flexion and increasing to 38.1 ± 25 N at 90° flexion (Fig. 2). The in situ forces for the PCL and PLS were found to be minimal under anterior tibial loading, indicating a negligible role in resisting anterior tibial translation. In contrast, bony contact generated in situ force as high as 34.1 ± 23 N under anterior tibial loading at 30° flexion. These

Table 2 Anterior tibial translations (mm) in response to 110 N anterior tibial load (mean \pm SD, $n = 10$)

Knee condition	Knee flexion angle				
	0°	15°	30°	60°	90°
Intact	3.2 \pm 1.0	4.4 \pm 2.2	5.8 \pm 3.5	7.0 \pm 2.8	6.1 \pm 1.8
ACL-deficient	10.1 \pm 3.0	14.6 \pm 4.0	16.8 \pm 5.2	15.3 \pm 5.8	12.2 \pm 3.7
ACL+MCL-deficient	11.4 \pm 3.7	16.4 \pm 4.8	20.0 \pm 5.4	21.6 \pm 5.6	18.3 \pm 4.5

Table 3 Posterior tibial translations (mm) in response to 110 N posterior tibial load (mean \pm SD, $n = 10$)

Knee condition	Knee flexion angle				
	0°	15°	30°	60°	90°
Intact	6.8 \pm 2.6	9.4 \pm 3.0	8.5 \pm 3.2	5.9 \pm 3.4	4.2 \pm 2.0
ACL-deficient	8.2 \pm 2.8	10.0 \pm 3.0	9.0 \pm 3.3	6.4 \pm 3.5	4.6 \pm 2.4
ACL+MCL-deficient	8.0 \pm 2.7	10.0 \pm 2.9	9.0 \pm 3.0	6.4 \pm 3.6	4.7 \pm 2.7

forces were reduced with both increases and decreases in flexion angle (Fig. 2). When the forces in the MCL and those due to bony contact are considered relative to the ACL, the MCL is found to carry forces more than 60% of those of the ACL at 90° flexion (Fig. 3). Bony contact generated forces as much as 42% of the ACL at 60° of flexion.

Changes in the kinematics of the joint following removal of various structures provide an alternative view of the contribution of these structures to the joint (Fig. 4). Anterior and posterior tibial translations in response to A-P loading to \pm 110 N, for the intact, postarthrotomy, ACL-deficient, and ACL+MCL-deficient knees are shown in Tables 2 and 3. The introduction of an anterior arthrotomy did not alter either anterior or posterior tibial translations under A-P tibial loading to \pm 110 N. Transection of the ACL significantly increased anterior tibial translation at all flexion angles, although no changes were observed for posterior tibial translation (Table 3). Anterior tibial translation of the ACL-deficient knee was largest at 30° of flexion, reaching 16.8 \pm 5.2 mm. Interestingly, transection of the MCL in combination with ACL (ACL+MCL-deficient) significantly increased anterior tibial translation relative to the ACL-deficient knee only at 60° and 90° of knee flexion ($P < 0.05$). At 60° of flexion, the introduction of MCL deficiency produced more than a 6-mm increase over the ACL-deficient knee, with anterior tibial translation of 21.6 \pm 5.6 mm (Table 2). Posterior tibial translation was not affected by the additional transection of the MCL, indicating a minimal role of the MCL in restraining posterior tibial loading.

Discussion

During anterior tibial loading, the ACL plays an important role in limiting anterior tibial translation, especially when the knee is near extension. The role of the MCL during this loading is minimal towards extension, but be-

comes more significant when the knee is flexed above 60°. The present findings agree with those of Shapiro et al. [22], who reported that the detrimental effect of MCL transection on anterior knee stability was larger at higher flexion angles. These results both underscore the importance of the Lachman test over the anterior drawer test for the evaluation of isolated ACL ruptures. However, if the absolute displacement found during the anterior drawer test is larger than that in the Lachman test, this may indicate a combined ACL+MCL injury. The importance of the meniscus under the same loading has been emphasized by Levy et al. [12], who found the largest contributions of this structure at 90° of flexion. This also matches with the present work, as the magnitude of the in situ force in the ACL is just over 50% of the applied anterior tibial load at 90° of flexion, indicating that other structures (such as the MCL or the medial meniscus) are contributing to resist anterior tibial loads in this range of knee flexion.

The results of the present work suggest that the MCL, menisci, and joint capsule are at risk of concomitant injuries if the ACL is injured when the knee is flexed beyond 60°. In the lower range of flexion between 0° and 30°, the ACL plays a dominant role, with the next largest in situ forces occurring due to bony contact. Therefore, the osteochondral surfaces may be at risk in this range if the ACL is injured. Studies using magnetic resonance imaging have demonstrated that most of the occult osteochondral injuries associated with acute ACL rupture are seen at the terminal sulcus of the lateral condyle, indicating anterior subluxation of the tibia, which may occur throughout the range of flexion [9, 24].

We have also found that the forces in the PLS are low during anterior tibial loading at all flexion angles. O'Brien et al. have previously reported that failures of ACL reconstructions may be the result of unrecognized and/or untreated posterolateral instability [18]. Although the A-P loading examined in this study is certainly not representative of all loadings of the knee, the fact that the

PLS does not appear to play a significant role may suggest that PLS deficiency in the ACL-reconstructed knee may not be responsible for the described deterioration of ACL graft function in the reconstructed knee. However, to completely characterize the interaction between the ACL and PLS, additional data on the in situ forces under other loading conditions (e.g., combined anterior tibial loading/tibial rotation) for the intact, ACL-deficient, and reconstructed knee will be needed.

Our findings confirm the hypothesis that there is significant load sharing between the ACL, MCL, and bony contact in the intact joint, and that this relationship is dependent on both knee position and external loading condi-

tion. For this reason, the MCL and osteochondral surfaces may also be at significant risk during ACL injury at certain knee positions. Further, bony contact provided a sizeable contribution to the anterior-posterior knee stability, even in non-weight-bearing conditions. The results of this study suggest future investigation using younger cadaver material that includes more complex loading conditions and the introduction of joint compressive forces (i.e., muscle contraction and/or weight bearing).

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