

THE INFLUENCE OF INDIVIDUAL ANTHROPOMETRIC AND MECHANICAL VARIATIONS ON FUNCTIONAL INSTABILITY IN THE ACL-DEFICIENT KNEE

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Individual variations in anatomy, structure and strength of the human knee and their effect on the functional instability of the knee was systematically examined in this 2D modeling study. The model included the tibiofemoral and the patellofemoral joint articulations, four ligaments, the medial capsule, and four muscle units surrounding the knee. Simulations were conducted to determine tibial anterior translation and internal joint loading at a single selected position during early stance phase of gait when the knee was under a peak external flexion moment. Incremental hamstring muscle forces were applied to the modeled ACL-deficient knee in order to examine the level of the hamstring muscle forces required to prevent abnormal tibial anterior translation. It was found that bony geometry of the knee joint, especially the slope of tibial plateau, strongly affected tibial translation and hamstring compensation in the ACL-deficient knee.

KEYWORDS: functional instability, knee, ligament, ACL, injury

INTRODUCTION: Functional stability of the ACL-deficient knee, defined here as abnormal anterior translation of the tibia relative to the femur during functional activities, may be achieved by compensation through the neuromuscular system, specifically the co-contraction of the hamstring muscles (18). However, there are still little-understood factors that have caused some patients to cope poorly with their ACL deficiency, while others do well without the ACL. The factors that account for different outcomes have been suggested to include anthropometric variations, differences in mechanical restrains of surrounding soft tissues, and varied capabilities of muscle compensation among different individuals. Anthropometric variations of individual knees have been reported in the past. (10). Mechanical properties of soft tissues in the knee joint have been shown to vary between test specimens from *in vitro* studies. (8) (6). The variations in anthropometric and mechanical properties of the knee joint have been shown to affect joint function (48, 10). Beynnon et al. (4) found that anthropometric input parameters of the model had the most pronounced effect on the output of the model. Their model, however, did not include the patellofemoral joint and was not used to simulate knee motion during functional activity. The objectives of the current study using a two-dimensional anatomical knee model were twofold: to examine changes in joint motion during gait as a result of individual variations in joint anthropometric and mechanical properties of the ACL-deficient knees; and to determine the influences of these variations on hamstring compensative function.

METHODS: The influences of individual variations in joint anthropometric and mechanical parameters were examined by comparing knee motion at a single selected position of early stance phase during gait. At this position external flexion moment on the knee joint reached a peak value. The quadriceps muscle force was assumed at its peak value too in order to balance the peak external flexion moment. A set of normal gait data (46) was used in the simulations. The subject's body weight was 56.7 kg with a calculated shank weight of 3.5 kg. The peak flexion moment on the knee occurred at 13% of gait cycle starting from heel strike. At this position, the tibia was aligned almost vertically relative to the ground (0.3° tilting backward) while the knee was flexed at 15.6°. External loading applied on the tibia included also the resultant force and moment at the ankle joint. The resultant forces on the ankle joint were 595.7 N in vertical direction and 97.1 N in horizontal direction pointing posteriorly (46). The resultant ankle dorsiflexion moment was 0.2 Nm.

A two-dimensional knee model in the sagittal plane was used in this investigation with details provided elsewhere (30). The model included both the tibiofemoral and patellofemoral joints

and was comprised of three rigid bodies, the femur, the tibia and the patella. The contour of the femoral condyle in the tibiofemoral joint and the femoral contour of the patellofemoral were represented by ellipses (48). The contour of tibial plateau was represented by a straight line sloping posteriorly (33). The patella was approximated as a rectangular (48). The model included seven ligament and five muscle units: anterior and posterior bundles of the ACL, anterior and posterior bundles of the posterior cruciate ligament (PCL), medial collateral ligament (MCL), lateral collateral ligament (LCL) and the medial capsule, the rectus femoris, vasti (a combination of vastus lateralis and vastus medialis), long head of hamstrings, and short head of hamstrings. The initial strain, stiffness, and positions of insertion sites of ligaments and medial capsule were adapted from literature (4, 5, 6, 47). The origin and insertion sites, and the maximal isometric forces of muscles were adapted from Friederich & Brand (14) and Delp (11). The force of gastrocnemius muscle does not directly affect the motion of the shank, therefore, not included in the model. The forces from two hamstring muscle units were assumed to be approximately proportional to each other with a force ratio ($r_{si} = 0.194$) defined as the ratio of maximal isometric forces of hamstring short head unit relative to hamstring long head unit. Similarly, forces from two quadriceps muscle units (rectus femoris and vasti) were assumed to be proportional to each other with a force ratio ($r_v = 0.172$) defined as the ratio of maximal isometric forces of rectus femoris relative to vasti muscles.

Simulations were conducted on the modeled normal and ACL-deficient knees. Incremental hamstring muscle forces ranged from 0 to 100% of the maximal isometric force, at 2% increments, were applied to the knee in the simulations. Anthropometric parameters included bony geometric parameters, coordinates of insertion sides of ligamentous elements and medial capsule, and origin and insertion sides of muscles. A plus and a minus variations relative to a typical set of system parameters were simulated. The variations of bony geometric parameters of the knee were estimated from the data in the literature (10, 14, 18). The variations of ligament insertion sides were estimated from Blankevoort et al. (5). The variations of the coordinates of muscle origin and insertion centers were estimated to be 5 mm. The variations of mechanical characteristics of the MCL and medial capsule included their initial strain and stiffness were estimated from Blankevoort et al. (6). Comparisons were made on the levels of the hamstring muscle force that was required to compensate for abnormal tibial anterior translation due to the ACL deficiency.

RESULTS AND DISCUSSION: The results of simulations showed that changes to the slope of tibial plateau dramatically altered the tibial anterior displacement for the ACL-deficient knee, but not the normal knee. For the normal knee, the tibial anterior displacement was -2.4 mm, -1.5 mm, and -0.2 mm corresponding to the tibial slope angles of 4°, 8°, and 12° tilted posteriorly, respectively. However, for the ACL-deficient knee, the corresponding tibial anterior displacements were 6.8 mm, 10.5 mm and 15.1 mm, respectively, without hamstring muscle force (Figure 1). The effectiveness of hamstring muscle force to compensate for ACL deficiency depended strongly on the slope of tibial plateau. Only 22% of the maximal hamstring muscle force was required for the knee with 4° of the tibial slope to completely restore the tibia to a normal contact position for the ACL-deficient knee. The required hamstring muscle force increased to 55% of its maximal force when the tibial slope was 8°. However, for the knee with 12° of tibial slope, the tibial anterior displacement (2.6 mm) under the maximal hamstring muscle force was still greater than that for the normal knee, indicating that the tibial anterior translation due to ACL deficiency could not be completely compensated by hamstring muscle forces.

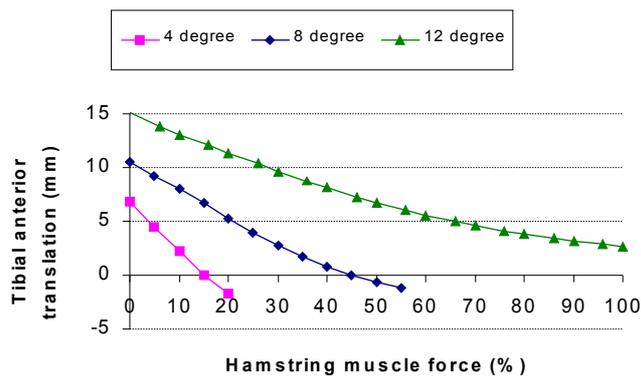


Figure 1 - The corresponding tibial anterior displacements without hamstring muscle force

In addition to the slope of tibial plateau, the most sensitive parameters were the distal attachment site of the hamstring long head unit and the coordinates of tibial tubercle. Simulations found that a plus variation in the coordinates of tibial tubercle produced an increase in the required hamstring muscle force by 15%, and a plus variation in the distal attachment site of the hamstring long head unit produced an increase of 16% of the required hamstring muscle force.

The variations in the slope of the tibial plateau among 281 knees have been reported to tilt posteriorly from 0° to 19° (10). This angle was shown, in the same study, to correlate significantly with anterior translation of the tibia in lateral x-ray films with the subjects standing. The results of the present study demonstrated the importance of this angle in relation to functional stability of the ACL-deficient knee during level walking. The hamstring muscle force could translate the tibia posteriorly to its normal position only if the angle of tibial slope was within a certain range. The simulations showed also that other bony geometric parameters and muscle attaching sites might have important influence on hamstring compensative function as well. A systematic examination may need to be developed for future clinical application of the result of this study.

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