RESPONSE ANALYSIS OF THE KNEE JOINT IN FLEXION UNDER QUADRCEPS ACTIVATION

W. Mesfar and A. Shirazi-Adl
Génie mécanique, École Polytechnique, Montréal, Québec, Canada

The human knee joint is a complex structure with interactions between muscle forces, ligaments, menisci and articulations at different regions. Proper management of rehabilitation and treatment programs requires a solid understanding of such interactions in intact and injured conditions. Towards this goal, a realistic nonlinear 3-D finite element model of the entire knee joint is developed. In this work, the ligament forces and contact stresses/areas are computed as the unconstrained joint is flexed from 0° to 90° ± a constant 137 N quadriceps force. Predictions support the coupling between various components as a function of quadriceps exertion and flexion angle. The model is promising in augmenting our understanding of the joint function leading to improved design for rehabilitation programs and replacement procedures in active patients.

KEY WORDS: knee joint, biomechanics, quadriceps force, finite elements, ligaments.

INTRODUCTION: In various sport activities, both patello-femoral and tibio-femoral joints are subjected to very large articular contact loads far exceeding the body weight while at the same time accommodating the mobility demand placed on the knee joint. The primary function of the patella is to transmit and increase the effective lever arm of the quadriceps muscles forces. In doing so, the patello-femoral joint is frequently the source of the knee anterior pain related to disturbances in normal tracking, instability, and excessive pressure syndrome. Patello-femoral pain is a principal health problem accounting for 33.2% of all knee disorders in women and 18.1% of all knee disorders in men (Dehaven and Lintner, 1986). The study of biomechanics of the entire knee joint including the patello-femoral joint is of primary importance in understanding the joint function and interactions between various components in both intact and perturbed conditions. An example is the compensatory role of quadriceps and hamstrings in patients with ACL or PCL rupture among copers and non-copers and at different activity levels. Such knowledge would benefit preventive and treatment procedures, addressing improved joint and ligament recnstruction designs to more effective rehabilitation exercises.
Determination of the contact forces and laxities of the patellofemoral and tibiofemoral joints as well as forces in ligaments has been the subject of numerous experimental investigations. No realistic finite element model of the entire joint has, however, yet been reported in the literature. In this study, based on our validated model of the tibiofemoral joint (Moglo and Shirazi-Adl, 2003), we aim to develop a 3D finite element model incorporating both tibio-femoral and patello-femoral joints. The detailed response of the knee joint will subsequently be analysed in flexion from 0° to 90° in presence of quadriceps forces. This report concentrates primarily on forces in joint ligaments and patellofemoral contact forces as a function of joint felxion and quedriceps forces.

METHOD: The finite element model of the knee joint consists of three bony structures (tibia, femur and patella) and their articular cartilage layers, menisci, six principal ligaments (collaterals, cruciates, and medial/lateral patello-femoral ligaments), patellar tendon and quadriceps muscle vectors (divided into three groups) (Fig. 1). The non-linear elastostatic analysis is performed by employing ABAQUS finite element package program. The bony structures (i.e., femur, tibia and patella) are represented by rigid bodies due to their much greater stiffness compared with joint soft tissues. Menisci are modeled as a nonhomogeneous composite of a bulk material reinforced by radial and circumferential collagen fibres. Ligaments are each modeled by a number of uniaxial elements with different prestrain (or pretension) values. Nonlinear material properties are based on those used in our earlier model studies and those reported in the literature (e.g., Atkinson et al., 2000; Moglo and Shirazi-Adl, 2003; Staulbi et al.,1999). The medial patello-femoral ligament (MPFL) is represented by four elements, the lateral patello-femoral ligament (LPFL) by three elements and the patellar tendon (PT) by nine elements (Fig. 1). In order to couple the femoral displacements to the quadriceps forces, the quadriceps muscles are simulated by springs with constant forces inserted into the patella and the femur. The 3D motion of the joint is characterized by a proper joint coordinate system. To simulate an unconstrained flexion response, the femur is fixed while the tibia and the patella are left free. The joint reference configuration at full extension is initially established under ligament prestrains and quadriceps forces. The quadriceps load of 137 N accounts for 40N in vastus medialis obliqus (VMO), 60 N in rectus femoris/vastus intermidus medialis (RF/VIM), and 50 N in vastus lateralis (VL), selected according to the ratios of their physiological cross-sectional areas VMO:RF/VIM:VL=2:3:2.5 (Sakai et al.; 1996). Direction of force in each component of quadriceps muscle is derived from the Q-angle model (Sakai et al.; 1996) with the Q angle taken as 14°. The tibia is subsequently flexed to 90° in presence of prestrains (pretensions) in ligaments and 137 N quadriceps preload.

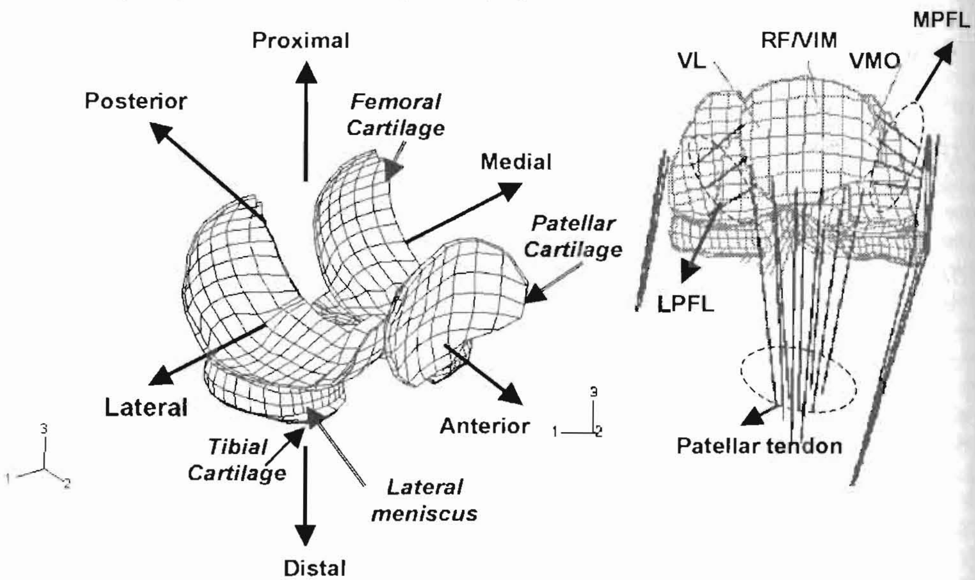

Figure 1: The knee joint finite element model showing cartilage layers, menisci, ligaments, patellar tendon and quadriceps muscles. Bony structures are not shown.

RESULTS: In reference configuration at full extension under ligament prestrains and quadriceps force, the ACL carries the largest load among ligaments while the PT resists almost the entire applied quadriceps muscle force (Fig. 2). In joint flexion from 0° to 90°, both these forces substantially diminish to 42 N in the PT and 14 N in the ACL at 90° flexion (Fig. 2). The PCL force remains zero at angles <65° reaching 13N at 90°. The patello-femoral ligaments (MPFL and LPFL) play a mechanical role only at smaller flexion angles.

As for contact forces at the patello-femoral joint, the resultant contact force increases from 65 N at full extension to 138 N at 90° (Fig. 3). The contact force is primarily in anterior-posterior direction at smaller flexion angles whereas it changes orientation to become axial at the larger flexion angles (~130 N at 90°) (Fig. 3). These changes are consistent with the fact that the joint flexion is applied by the rotation of the tibia while the femur is restrained.

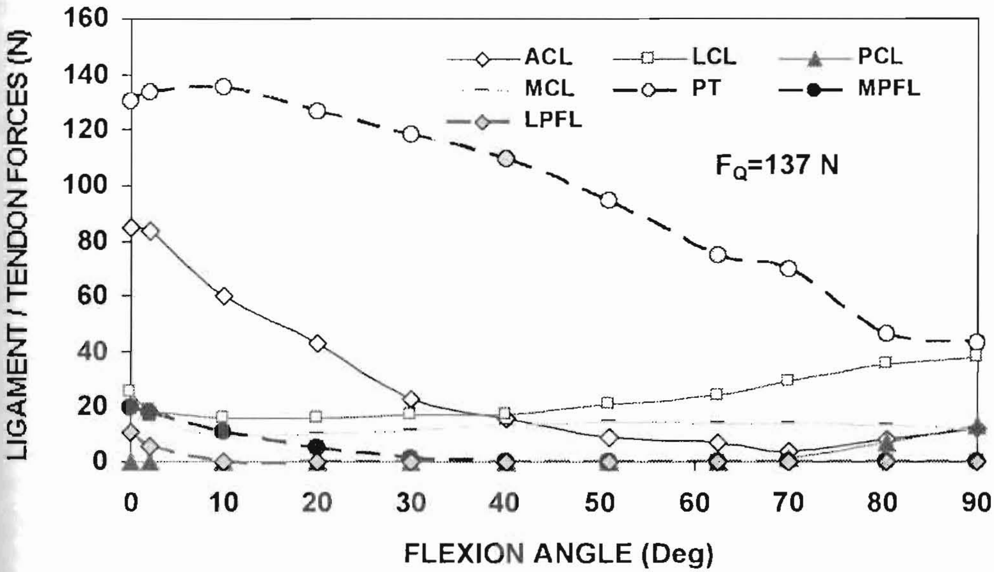

Figure 2: Computed forces in joint ligaments and patellar tendon at different flexion angles under quadriceps force of 137 N.

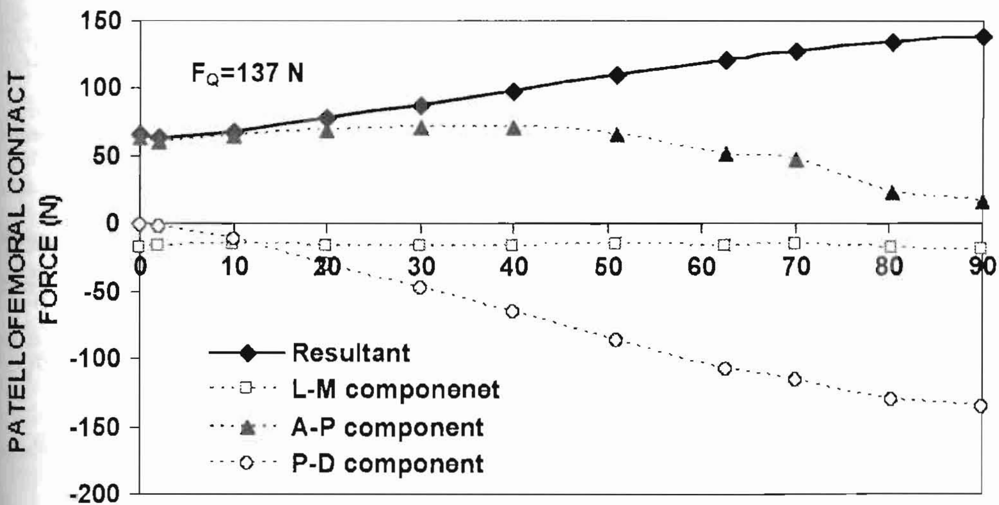

Figure 3: Resultant patellofemoral contact force and its components in different anatomical directions as a function of joint flexion under 137 N constant quadriceps muscle force.

DISCUSSION: This work was set to analyse the detailed response of the entire knee joint, in particular ligament/tendon forces and patello-femoral contact loads, under quadriceps forces at different flexion angles. Under constant quadriceps force, the PT and ACL forces are greatest at full extension substantially diminishing as the flexion increases to 90°. The predicted changes in ACL and PCL forces for the cases with and without quadriceps activation and their variations with flexion angle are found in good agreement with measured results (Beynnon and Fleming, 1998; Li et al., 1999 and 2004; Pandy and Shelburne, 1997). The increase in the ACL force due to quadriceps activations could indicate a higher risk for the ACL or its graft at smaller flexion angles in exercises that demand quadriceps exertions alone without hamstrings.

The resultant contact force at the patello-femoral articulation significantly increased with joint flexion, a prediction in agreement with reported measurements (Ahmed et al., 1983; Singerman et al., 1995). The dominant component was in the A-P direction at smaller flexion angles and in the axial direction at larger flexion angles. The relative values of these components would alter had we applied the flexion on the femur while fixing the tibia. Moreover, an injury to the ACL, though not studied, could alter the articular contact stresses.
The current model studies should delineate the likely interactions between quadriceps/hamstrings muscle exertions and cruciate ligament forces on one hand and their combined effects on articular cartilage contact stresses/areas on the other. The understanding gained would be of great help in the design of optimal rehabilitation programs and joint replacement procedures for active patients with the goal to restore near-normal joint activities while preventing joint degeneration, instability and pain.

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