COMPRESSION FORCE SUBSTANTIALLY INCREASES THE TIBIOFEMORAL JOINT PASSIVE STIFFNESS AND MOMENT IN SAGITTAL AND FRONTAL PLANES

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INTRODUCTION

In various activities of daily living, the human knee joint experiences large loads and motions. Recent in vivo measurements using instrumented knee implants have reported large compression forces on each tibiofemoral (TF) joint that could exceed 3 times total body-weight during level walking [1]. Experimental investigations of the knee joint, however, have commonly neglected the presence of such physiological compression forces due primarily to associated technical difficulties, joint instability and artifact loads. Similar to spinal intervertebral joints [2], such compression preloads likely substantially increase the passive stiffness and moment resistance of the knee joint in various directions. Nevertheless, musculoskeletal model studies of the knee joint routinely overlook the passive resistance of the joint when attempting to estimate muscle forces [3]. This assumption can markedly influence estimated muscle, contact and ligament forces. This study hence aimed to determine the effect of varying compression force (0-1800 N) on the passive response of the TF joint at different knee flexion angles (0-45°). For this purpose, two validated finite element (FE) models of the TF joint were employed [4]. To simulate the transient response, the articular cartilage layers and menisci are simulated as a non-homogeneous nonlinear depth-dependent composite of collagen fibrils and an incompressible matrix. For comparison an equivalent but less refined model with isotropic depth-independent cartilage properties is also used. The TF joint response is studied at four flexion angles; 0 (full extension), 15, 30 and 45° under compression preloads varying from nil to 1800 N (i.e., 3 times body weight of a female subject in our model constructed based on a female cadaver knee). Due to convergence problems, the response under 1800 N compression was obtained only with the less refined model.

METHODS

Two validated FE models of the TF joint consisting of two bony structures (tibia and femur) and their compliant articular cartilage layers as well as menisci and four principal ligaments (ACL, PCL, LCL, MCL) are employed [4]. To simulate the transient response, the articular cartilage layers and meniscus are simulated as a non-homogeneous nonlinear depth-dependent composite of collagen fibrils and an incompressible matrix. For comparison an equivalent but less refined model with isotropic depth-independent cartilage properties is also used. The TF joint response is studied at four flexion angles; 0 (full extension), 15, 30 and 45° under compression preloads varying from nil to 1800 N (i.e., 3 times body weight of a female subject in our model constructed based on a female cadaver knee). Due to convergence problems, the response under 1800 N compression was obtained only with the less refined model.

To circumvent the artifact moments under compression, initially the MBP position (on the tibia) is determined as a point where a compression force does not generate any frontal and sagittal rotations in the unconstrained joint. This position alters, as expected, with the magnitude of applied compression force and the joint flexion angle. In the current simulations, the femur is fixed while the tibia remains unconstrained. With compression preloads applied at their respective MBP, the joint response is subsequently determined in the sagittal (flexion-extension) and frontal (varus-valgus) planes. The TF instantaneous (tangent) angular stiffness in these planes under various compression preloads is also determined by the perturbation method at the loaded configurations.

RESULTS AND DISCUSSION

Both nonhomogeneous and less-refined homogeneous models yielded similar results. The MBP location shifted medially with the compression force and posteriorly as the joint flexed from the full extension position. Flexion moments
carried by the femur at its reference point (ie, mid-distance between epicondylar centers) increased significantly with the compression preload at all joint flexion angles (Fig. 1). The femoral valgus moments also increased with compression preload but were highest at the full extension. The instantaneous (tangent) angular rigidities in both sagittal (Fig. 2) and frontal planes also markedly increased with the compression preload and were highest at near full extension. The TF moment resistance at full extension substantially increased in both varus and valgus rotations as the compression preload increased (Fig. 3). Despite greater moments, forces in collateral ligaments decreased in varus/valgus rotations as the compression increased.

Results under compression forces and moments acting alone or combined were in good agreement with those available in the literature. Determination of the joint MBPs allowed for the first time in the current study the proper application of physiological compression forces without any artifact moments. Under all joint flexion angles, the joint MBP shifted medially with compression thus increasing load transmission via the medial compartment. The compression preload substantially increased TF moment carrying capacity and instantaneous angular stiffness in both sagittal and frontal planes. Despite greater varus-valgus moment resistance, forces in collateral ligaments dropped as the compression force increased suggesting alterations in moment resistance from ligaments to contact forces and the protective role of compression force.

In conclusion, a novel TF joint MBP is identified at which location a compression force does not cause rotations in sagittal/frontal planes. The compression force substantially increases TF joint moment resistance and angular rigidities in frontal and sagittal planes. While the passive angular stiffness enhances the joint stability, the augmented passive moment resistance under compression preloads plays a role in supporting external moments and should as such be considered in the musculoskeletal models of the joint aiming for an accurate estimation of muscle forces and joint response.

REFERENCES

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