Alterations in Knee Contact Forces and Contact Centers During Gait in Normal, OA and Varus-Valgus Altered Subjects - A Detailed Lower Extremity Musculoskeletal Model Study

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Disclosures: M. Adouni: None. A. Shirazi-Adl: None.

Introduction: Alterations in knee joint kinetics-kinematics affect the initiation and progression of joint pathologies such as osteoarthritis (OA). Improved understanding and management of the risks involved requires a sound knowledge of the joint response in gait as the most common load-bearing activity. The knee contact mechanics is a crucial aspect dealing with contact pressure/area as well as contact forces (CFs) and contact centers (CCs) in various articulating compartments. Regions of contact have been correlated with cartilage thickness in normal subjects. Alterations in CF and CC on the articulating surfaces of cartilage have been indicated either in the initiation/acceleration of tissue degenerations or alternatively in the prevention of medial OA progression. Knowledge on CF and CC as well as factors affecting them could hence be used as a tool in evaluation, treatment and rehabilitation programs. Accurate computational modeling has the advantage to circumvent many difficulties and shortcomings in earlier investigations when attempting to quantify CFs and associated CCs [1-3]. Here, a lower extremity musculoskeletal model [4] including a validated complex finite element (FE) model of the entire knee joint driven by reported in vivo kinematics/kinetics [5,6] is employed to simulate the stance phase of gait. Attention is focused on the contact pressure distribution, CF and CC on each tibial plateau and on the entire tibiofemoral (TF) joint in normal, OA and varus-valgus altered subjects.

Methods: An existing validated FE model of the entire knee joint [4] made of bony structures (tibia, femur and patella) and their cartilage layers, menisci and ligaments is used (Fig. 1). Articular cartilage layers and menisci are simulated as non-homogeneous nonlinear depth dependent composites of collagen fibrils and incompressible matrices. Cartilage layers are reinforced by fibril networks parallel to the articular surface in the superficial zone, randomly oriented in the middle zone and vertical in the deep zone. Ligaments are each simulated by a number of nonlinear axial elements with initial pre-strains. This detailed knee model is introduced in a musculoskeletal model of the lower extremity including hip and ankle joints as well as uni- and bi-articular muscles to simulate the stance phase of gait under in vivo kinematics/kinetics reported for (1) normal subjects, N, (2) severe OA subjects, OA and (3) varus-valgus altered subjects, VV. Unknown muscle forces are evaluated iteratively and applied along with the ground reaction forces and kinematics/kinetics to investigate the knee joint biomechanics in gait. The input kinematics-kinetics data at 0%, 5%, 25%, 50%, 75% and 100% periods of stance are taken directly from in vivo measurements [5,6]. The VV cases are simulated by altering only the adduction-abduction rotation or moment at mid stance (50% period) in normal subjects (N model) by ±1.5 deg or ±50%, respectively. The calculated contact force/pressure distributions on medial and lateral plateaus are then used to determine total contact forces (CFs) and their locations (CCs) once on each plateau separately and once on the entire TF joint (moments of contact forces when evaluated at its corresponding CC would disappear).
**Results:** In both N and OA models, CFs were much greater on the medial plateau except at the early stance (0% and 5% periods). No contact occurred on the lateral plateau at the toe off (100% period) due to large adduction rotation (Fig. 2). With the exception of 50% period, CFs on both plateaus were greater in N model than OA model. The peak CFs were found on the medial plateau at 25% and 75% periods associated with peaks of ground reaction forces; the total CF on the TF joint reached its maximum of 2467 N (~4 body weight, BW, for a female specimen) at 25% period (Fig. 3). At the mid-stance period (50%), an increase in the adduction rotation almost unloaded the lateral plateau with little effect on CF at the medial plateau. In contrast, a decrease in the adduction rotation significantly increased CF on the lateral plateau with minor decreasing effect on the medial plateau. Changes in the adduction moment had smaller effects. On both medial and lateral plateaus (Fig. 2), largest contact forces (on the medial plateau at 25-100% periods whereas on the lateral plateau at 0% and 5% periods) were located more centrally on associated articular surfaces. In the entire TF joint (Fig. 3), CC was located on the lateral plateau (close to the tibial eminence) at 0% and 5% periods and shifted to the central medial surface thereafter at 25-100% periods. An increase in the adduction rotation at 50% period shifted CC medially on both plateaus. In contrast, CC moved laterally in opposite direction as adduction rotation decreased.

**Discussion:** Alterations in the magnitude as well as location (CC) of CFs on tibial plateaus during gait have been associated with the risk of OA. As such, quantification of contact mechanics is crucial in improved management of joint disorders, from evaluation to treatment and rehabilitation stages. Accurate determination of CFs and CCs involves however major difficulties and assumptions in both in vitro and in vivo studies. Although recent magnetic resonance and fluoroscopy imaging techniques have been very helpful, simplifying assumptions are often required that adversely affect estimations [1-3]. Our results, for example, demonstrate that CCs do not usually fall either on the centroids of contact areas or on the locations of maximum contact pressure/compressive strain (peak overlap of contacting surfaces). Moreover, CCs substantially alter in both ML and AP directions thus questioning the assumption of a path with fixed ML location during gait. Current investigation accurately quantified the CFs and CCs either separately on each plateau or on both together. The CF and CC in the entire TF joint (Fig. 3) highlight also the passive contribution of the knee articulation in the net joint moment-resistance. Higher passive resistance, for example, diminishes activation in muscles crossing the joint and hence the joint CF.

With the exception of mid-stance period, CFs were larger in N than OA group with the maximum of ~4 BW at 25% period. Larger CFs on both plateaus (at 0% and 5% periods on the lateral whereas at remaining 25-100% periods on the medial) occurred at more central articular regions that are reported to have thicker cartilage [3]. With OA, CCs shifted slightly by <3.5 mm. In both N and OA models at 25-100% periods associated with the largest TF and medial CFs, CCs when compared to early stance (0% and 5% periods) shifted medially (in entire TF and individual plateaus) and posteriorly (in plateaus only) in accordance with larger adduction and flexion rotations, respectively. Moreover in accordance with the joint internal rotation, in all cases and stance periods, CCs were more anterior on the medial plateau compared to the lateral plateau. Apart from rotations, muscle activities, joint moments and menisci also influence CCs.

**Significance:** Large contact forces (very early in stance on the lateral plateau and thereafter on the medial plateau) were located on the central articular regions with thicker cartilage. Changes in joint
rotations influenced the location of CC which in turn affected the joint passive resistance and activity level in muscles. Knowledge of CFs and CCs hence have important biomechanical and pathological consequences.
Figure 3

<table>
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<th>N / OA TF (N)</th>
<th>0%</th>
<th>5%</th>
<th>25%</th>
<th>50%</th>
<th>75%</th>
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ORS 2015 Annual Meeting
Poster No: 0767