Introduction:

Contact forces (CF) and contact centers (CC) on the articulating surfaces of the human knee joint and alterations therein are important markers in both prevention/initiation/progression and treatment evaluation of joint disorders [1, 2]. The position of CC also plays a crucial role when estimating tibiofemoral CFs. To identify CC, cadaver studies estimate the pressure weighted center of contact by assigning weights according to contact stresses measured on the tibial plateau [3]. Alternatively using bony landmarks and cartilage thickness, imaging techniques have employed various measures to quantify CC using joint overlap, minimum joint space, contact area/center and distance weighted center of proximity [3-4]. In addition, lower extremity musculoskeletal models often consider fixed paths and orientations for TF contact forces independent of loads, kinematics and articular structures [5]. In the current study, we use a validated lower extremity model including a detailed knee joint finite element (FE) model [6] to initially compute CFs and CCs during the entire stance period of gait in normal and OA subjects. These predictions are then employed to determine and compare with CCs based on 6 other commonly reported routines.

Methods:

An existing validated iterative kinematics-driven neuro-musculoskeletal FE model of the lower extremity is employed [6]. This model simulates the hip and knee as 3D and 1D spherical joints, respectively, and the knee joint in details as a complex nonlinear FE model consisting of bony structures (tibia, patella, and femur), their articular cartilage layers, menisci, major ligaments and patellar tendon. Cartilage layers and menisci are represented by nonlinear depth-dependent fibril-reinforced tissues while ligaments are modeled with distinct nonlinear properties in tension and initial strains. The mean hip/knee/ankle joint rotations/moments and ground reaction forces (GRF) at the stance phase of gait collected on normal and severe OA subjects drive the model [7]. Subject to gait kinematics-kinetics, the model is iteratively analyzed and the muscle forces are estimated using optimization. At each stance phase (0, 5, 25, 50, and 100%), the calculated contact force/pressure distributions on the medial and lateral plateaus are then used to determine total CFs and their locations (CC-Ref) once on each plateau separately and once on the entire tibiofemoral joint. For the sake of comparison and based on our predictions, various methods commonly reported in the literature (Centroid of Contact Area, CCA; its modified version accounting also for contact stresses under menisci, CCA-m; Contact Point, CP; Maximum Strain, MS; Minimum Joint Space, MJS; Weighted Center of Proximity, WCoP) are also used to calculate the location of CC at all stance periods. Correlation coefficients are used to compare CC in various approaches relative to our CC-Ref (normal subjects) taken as the gold standard. Our CC-Ref is calculated in a manner such that the vector summation of moments of all nodal contact loads about this point disappears.

Results:

Tibiofemoral joint CFs are greater on the lateral plateau very early in stance and on the medial plateau thereafter during 25-100% stance periods (Fig. 1). Overall, gait of subjects with severe OA showed much smaller CFs except at mid-stance period. Large excursions in CC exceeding 10 mm, especially on the medial plateau in the ML direction, are computed. Various reported models estimate quite different contact centers with much greater variations in the ML direction than in the AP direction on both plateaus (Fig. 2). Using our computed results and compared to our CC-Ref as the gold standard, coefficients of determination are found lowest on the lateral plateau in the ML direction.

Discussion:

Contact forces and centers (CFs and CCs) are evaluated using a lower-extremity model driven by mean kinematics-kinetics of gait recorded in normal and severe OA subjects. Using the same FE predictions in normal subjects, six methods commonly used in the literature (CCA, CCA-m, CP, MS, MJS, and WCoP) are also applied to estimate and compare CC locations (at all 6 periods of stance). In both subject groups at 25-100% stance periods with much larger forces on the medial plateau, location of CC when compared to the early stance (0% and 5%) periods shifts medially and posteriorly in accordance with larger adduction and flexion rotations. Moreover due to the joint internal rotation, in all cases and periods, CCs are more anterior on the medial plateau when compared to the lateral plateau. Larger AP excursions on the medial plateau corroborate earlier reports that the knee center of rotation is on the lateral side for most of the time during stance [3]. Relatively large differences in CC position, particularly in the ML direction, are found when comparing various algorithms. On the tibial medial plateau, the location of CC in all approaches is strongly correlated with CC-Ref except for the CP and MJS models. The correlations are overall poorer on the lateral plateau especially in the ML direction. The MS approach nearly coincides with the location of maximum contact pressure; a physical interpretation that the MJS approach lacks. Its reliability however deteriorates when contact stresses are not symmetrically distributed about their peak value. The CCA approach, being based on the overlapped cartilage areas, neglects the pressure transmission on covered areas and hence the crucial load bearing role of menisci. Additional consideration of the covered areas in the CCA-m approach, as expected, markedly shifts the CCA estimation medially on the medial plateau and laterally on the lateral plateau towards their respective covered cartilage-meniscus areas. MS and WCoP approaches depend only on the relative locations of opposing bony surfaces and account neither for menisci nor for changes in cartilage thickness. The substantial ML excursion of CC-Ref on the medial plateau in stance phase raises serious concern on the assumption of a fixed ML location made in the CP model.

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References: