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Alterations in knee contact forces and centers in stance phase of gait: A detailed lower extremity musculoskeletal model

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ABSTRACT

Evaluation of contact forces-centers of the tibiofemoral joint in gait has crucial biomechanical and pathological consequences. It involves however difficulties and limitations in *in vitro* cadaver and *in vivo* imaging studies. The goal is to estimate total contact forces (CF) and location of contact centers (CC) on the medial and lateral plateaus using results computed by a validated finite element model simulating the stance phase of gait for normal as well as osteoarthritis, varus-valgus and posterior tibial slope altered subjects. Using foregoing contact results, six methods commonly used in the literature are also applied to estimate and compare locations of CC at 6 periods of stance phase (0%, 5%, 25%, 50%, 75% and 100%).

TF joint contact forces are greater on the lateral plateau very early in stance and on the medial plateau thereafter during 25–100% stance periods. Large excursions in the location of CC (> 17 mm), especially on the medial plateau in the mediolateral direction, are computed. Various reported models estimate quite different CCs with much greater variations (∼15 mm) in the mediolateral direction on both plateaus. Compared to our accurately computed CCs taken as the gold standard, the centroid of contact area algorithm yielded least differences (except in the mediolateral direction on the medial plateau at ∼5 mm) whereas the contact point and weighted center of proximity algorithms resulted overall in greatest differences. Large movements in the location of CC should be considered when attempting to estimate TF compartmental contact forces in gait.

1. Introduction

Alterations in knee joint kinetics-kinematics influence the initiation and progression of joint pathologies such as osteoarthritis (OA) that afflicts a considerable portion of adult population (Dillon et al., 2006). Effective preventive measures and treatment managements of such disorders require a sound knowledge of the joint behavior in both healthy and pathologic conditions. Changes in contact forces (CF) and contact centers (CC) on the cartilage articulating surfaces have been indicated as important markers either in the prevention/initiation/progression or alternatively in the evaluation of treatment stages of joint disorders (Andriacchi et al., 2006; Engel et al., 2015; Harris et al., 1999). Accordingly, quantification of the joint contact mechanics in gait has been the focus of a number of investigations.

Cadaveric studies have measured contact pressure distribution across the knee joint using pressure-sensitive sensors (Gilbert et al., 2014; Wang et al., 2014) and films (Engel et al., 2015; Seitz et al., 2012). To identify the location of CC, the pressure weighted center of contact (pWCoC) is quantified by assigning different weights (proportional to recorded contact stresses) to each sensing element on the tibial plateau (Gilbert et al., 2014; Wang et al., 2014). These studies remain limited to *in vitro* investigations and hence subject to associated short comings especially on muscle forces and kinematics when simulating gait. Alternatively using bony landmarks and cartilage layers, imaging techniques allow *in vivo* the recording of joint articular surface interactions in static and dynamic conditions in order to quantify various measures such as joint overlap, minimum joint space and contact area/center (Coleman et al., 2011; Koo and Andriacchi, 2008; Wretenberg et al., 2002).

To determine dynamic joint CC location, several models have been developed based on imaged 3D geometry of contacting bones (Anderst et al., 2005; Anderst and Tashman, 2003; Asano et al., 2001; Farrokhi et al., 2014; Li et al., 2004; Wang et al., 2014) and

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cartilage layers (DeFrate et al., 2004; Han et al., 2005; Li et al., 2005a). Different techniques are employed in these studies to locate the joint CC during various activities. Shortest (minimum interface) distance gap was used to identify the location of CC at the TF joint (Asano et al., 2001; Li et al., 2004). DeFrate et al. (2004) developed a cartilage-overlap method in which the location of CC was estimated as the geometric centroid of the cartilage at overlapping (contact) areas. Anderst and Tashman (2003) proposed the distance weighted center of proximity (WCoP) in which higher weights were assigned to points on the tibial plateau located in closer proximity to the opposing femoral subchondral surface. During simulated walking, Wang et al. (2014) identified joint CC using either this image-based WCoP method or an alternative pressure weighted center of contact (pWCoC) accounting for measured contact pressures over contacting areas using cadaver specimens and pressure sensors. The anteroposterior (AP) location of WCoP was found to significantly correlate with that of pWCoC on both tibial plateaus but the correlation was very poor in the mediolateral (ML) direction.

In parallel, lower extremity musculoskeletal models often consider fixed paths and orientations for TF CFs independent of the external loads, kinematics and joint articular structures (Gerus et al., 2013; Winby et al., 2009). In fact, the contact point method employed in these studies to compute CFs assumes that, in the frontal plane, the compartmental CFs remain on fixed midpoints of each condyle (Winby et al., 2009). Using the OpenSim musculoskeletal model, Lerner et al. (2015) reported substantial differences in computed medial and lateral CFs when using radiographic images to identify CC locations instead of the foregoing approximate approach. Each 1 mm deviation in the medial-lateral location of CC altered peak medial CF by 41 N.

Accurate estimation of CCs markedly influence predictions of joint passive/active response affecting thus the subsequent evaluation of preventive, treatment and rehabilitation programs. Detailed computational modeling has the advantage to circumvent many shortcomings in earlier investigations when quantifying CFs and associated CCs. Here, results of a lower extremity musculoskeletal model (Adouni et al., 2015; Marouane et al., 2014) including a validated complex finite element model of the entire knee joint driven by mean reported values of gait kinematics/kinetics (Astephen et al., 2008; Hunt et al., 2001) are employed to initially quantify these contact quantities and then compare computed CCs with those using existing methods. Attention is focused on the CF and CC on each tibial plateau and on the entire TF joint in the normal as well as OA, varus-valgus and posterior tibial slope altered subjects. We hypothesize that the location of CC markedly alters in gait and by the algorithm used.

2. Methods

This study exploits the predicted results of our earlier model studies on biomechanics of the knee joint during the stance phase of gait in normal and OA subjects (Adouni and Shirazi-Adl, 2014a), varus-valgus altered subjects (Adouni and Shirazi-Adl, 2014b) and subjects with different posterior tibial slope (Marouane et al., 2014, 2015b). Here, a short description of the model is provided for completeness.

The FE model of the knee joint (Fig. 1) is made of bony structures (tibia, femur and patella) and their compliant cartilage layers as well as menisci, major tibiofemoral (TF: ACL, PCL, LCL, MCL) and patellofemoral (PF: MPFL, LPFL) ligaments, patellar tendon (PT), quadriceps (four components), hamstrings (six components) and gastrocnemius (two components). Articular cartilage layers and menisci are simulated as non-homogeneous depth-dependent composites of nonlinear collagen fibril networks and hyperelastic matrices while the bony structures are represented as rigid bodies. Ligaments are each simulated by a number of nonlinear

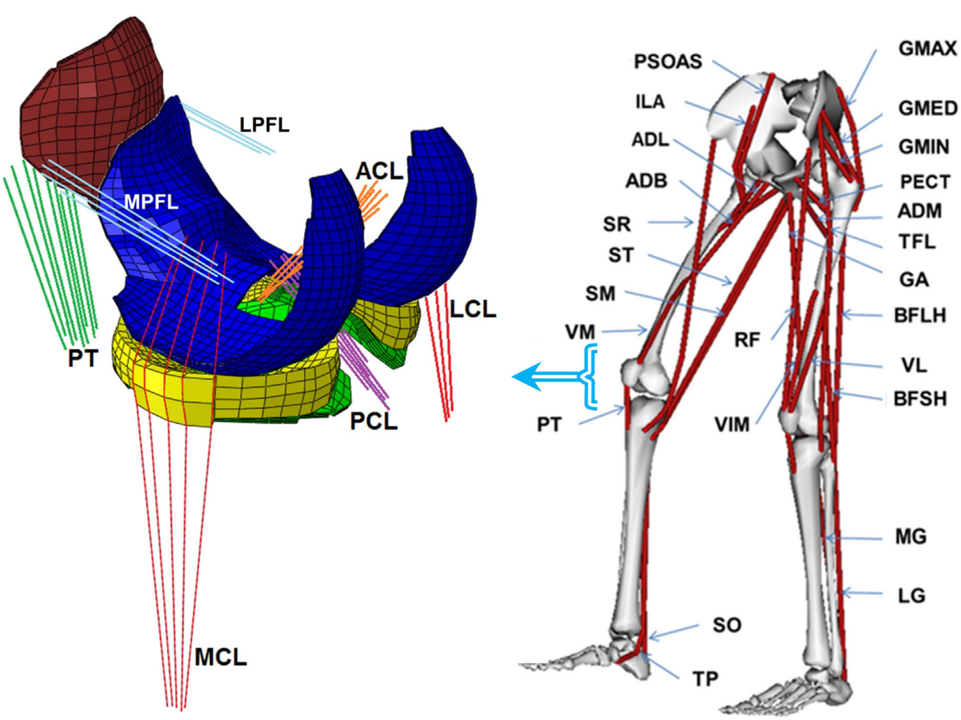

Fig. 1. (a) Knee FE model; tibiofemoral (TF) and patellofemoral (PF) cartilage layers, menisci, patellar Tendon (PT). Joint ligaments include lateral patellofemoral (LPFL), medial patellofemoral (MPFL), anterior cruciate (ACL), posterior cruciate (PCL), lateral collateral (LCL) and medial collateral (MCL). (b) Schematic diagram showing the 34 muscles incorporated into the lower extremity model (Open Sim, Delp et al., 2007). Quadriceps components are vastus medialis obliquus (VMO), rectus femoris (RF), vastus intermidus medialis (VIM) and vastus lateralis (VL). Hamstrings components include biceps femoris long head (BFLH), biceps femoris short head (BFSH), semi membranous (SM) and TRIPOD made of sartorius (SR), gracilis (GA) and semitendinosus (ST). Gastrocnemius components are gastrocnemius medial (GM) and gastrocnemius lateral (GL). Soleus (SO) muscle is uni-articular ankle muscle. Hip joint muscles (not all shown) include adductor, long (ADL), mag (3 components ADM) and brev (ADB); gluteus max (3 components GMAX), med (3 components GMED) and min (3 components GMIN), iliacus (ILA), iliopsoas (PSOAS), quadriceps femoris; pectineus (PECT), tensor facia lata (TFL), periformis.