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Partitioning of knee joint internal forces in gait is dictated by the knee adduction angle and not by the knee adduction moment

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ABSTRACT

Medial knee osteoarthritis is a debilitating disease. Surgical and conservative interventions are performed to manage its progression via reduction of load on the medial compartment or equivalently its surrogate measure, the external adduction moment. However, some studies have questioned a correlation between the medial load and adduction moment. Using a musculoskeletal model of the lower extremity driven by kinematics–kinetics of asymptomatic subjects at gait midstance, we aim here to quantify the relative effects of changes in the knee adduction angle versus changes in the adduction moment on the joint response and medial/lateral load partitioning. The reference adduction rotation of 1.6° is altered by ±1.5° to 3.1° and 0.1° or the knee reference adduction moment of 17 N m is varied by ±50% to 25.5 N m and 8.5 N m. Quadriceps, hamstrings and tibiofemoral contact forces substantially increased as adduction angle dropped and diminished as it increased. The medial/lateral ratio of contact forces slightly altered by changes in the adduction moment but a larger adduction rotation hugely increased this ratio from 8.8 to a 90 while in contrast a smaller adduction rotation yielded a more uniform distribution. If the aim in an intervention is to diminish the medial contact force and medial/lateral load ratio, a drop of 1.5° in adduction angle is much more effective (causing respectively 12% and 80% decreases) than a reduction of 50% in the adduction moment (causing respectively 4% and 13% decreases). Substantial role of changes in adduction angle is due to the associated alterations in joint nonlinear passive resistance. These findings explain the poor correlation between knee adduction moment and tibiofemoral compartment loading during gait suggesting that the internal load partitioning is dictated by the joint adduction angle.

1. Introduction

Medial knee osteoarthritis (OA) is a common disease afflicting a large portion of population with a gloomy prognosis in our ageing and obese populations. The higher incidence in the medial compartment is likely associated with the greater compartmental load in gait as suggested by biomechanical model studies (Adouni and Shirazi-Adl, 2013; Adouni et al., 2012; Shelburne et al., 2005; Winby et al., 2009) and in vivo investigations with instrumented implants (Kim et al., 2009; Kutzner et al., 2010; Zhao et al., 2007). At various stages of OA pathology, surgical (e.g., osteotomy) and conservative (e.g., knee braces, shoe soles, and gait modifications) interventions are routinely carried out to manage its progression via reduction of loading on the medial compartment. The external knee adduction moment has often been considered as the surrogate measure of this medial load and a consistent marker for OA disease and severity (Andriacchi, 2013; Butler et al., 2007; Kinney et al., 2013; Sharma et al., 2000; Shelburne et al., 2008; Zhao et al., 2007). As such, it is considered as the primary parameter when evaluating the efficacy of different treatment modalities performed with the objective to diminish compartmental contact force. It is worth however to raise two important concerns here; one is that any alterations towards a valgus alignment may trigger OA initiation or progression process at the lateral compartment (Andriacchi, 2013; Felson et al., 2013; Sharma et al., 2001). The second point is that the internal load distribution is influenced not only by the external moments but also muscle forces crossing the joint and joint kinematics. It is indeed conceivable to have conditions in which larger adduction moments do not actually yield greater medial contact forces.

Some recent in vivo studies using instrumented implants have indeed questioned a direct association between the knee adduction moment and the medial load and qualified such correlation as poor to average (Meyer et al., 2013; Walter et al., 2010; Winby et al., 2013). Walter et al. (2010) show that gait modifications (medial thrust and walking pole) can significantly reduce knee

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adduction moment without producing equally important reductions in medial contact forces. To reduce medial loading, this study suggests minimizing alterations in the knee flexion moment caused by gait modifications. Moreover, Meyer et al. (2013) indicate that external knee loads and EMG measures are not strong indicators of internal loads (medial and lateral contact forces) during gait activities. In addition, this investigation postulates that the external adduction moment correlates more with the medial to total contact force ratio than with the medial contact force. Winby et al. (2013) also call for caution when inferring joint contact loads from external measures (i.e., loads and EMG). The mechanisms governing any association between the external loads on the knee joint and resulting load distribution within the joint remain hence unclear.

Adequate understanding of the role of various external parameters, being displacement or load dependent, is crucial in proper prevention and management of knee OA. The efficacy of prophylactic wedge insole interventions as a remedy for medial OA (Russell and Hamill, 2011) has been questioned as it influenced neither the adduction moment (Abdallah and Radwan, 2011; Nester et al., 2003; Schmalz et al., 2006) nor the medial contact forces estimated by an instrumented implant (Kutzner et al., 2011). Varus alignment has however been identified as a significant risk factor in medial OA (Sharma et al., 2001). A parameter that likely plays a crucial role in joint internal loading is the passive moment resistance. Ligaments, menisci, articular cartilage and contact forces are known to markedly contribute to the adduction passive moment-carrying capacity of the joint (Bendjaballah et al., 1997; Markolf et al., 1981; Marouane et al., in press). Recently, Marouane et al. (in press) reported substantial passive resistance of the tibiofemoral joint in adduction moment that significantly increased with greater adduction rotation and compression force. Passive structures contribute both to the equilibrium of external moments thereby reducing muscle activation and to the stability by stiffening the joint (Markolf et al., 1981). Despite the strong sensitivity of the joint passive adduction resistance on the adduction angle, joint rotations are either not measured or involve relatively large errors when estimated using motion analysis systems (Gorton et al., 2009; Groen et al., 2012; Szczerbik and Kalinowska, 2011). Mean errors of up to 4.4° in adduction rotation were reported in gait when comparing intra-cortical pins versus skin markers (Benoit et al., 2006). Even with highly experienced testers, inter-tester differences of few degrees (2–6°) in peak rotations were documented (Benoit et al., 2006; Leigh et al., 2013; Pohl et al., 2010). Such variations in adduction rotation could significantly alter joint passive adduction moment and as a consequence the muscle activity and internal load redistribution (Marouane et al., in press).

Using a validated musculoskeletal model of the lower extremity including a detailed finite element (FE) model of the entire knee joint (Adouni and Shirazi-Adl, 2013) driven by reported kinematics–kinetics of asymptomatic subjects at midstance of gait (Astephen, 2007; Hunt et al., 2001), we aim here to quantify the sensitivity of knee joint response when altering either the knee adduction angle (by $\pm 1.5°$) or the knee adduction moment (by $\pm 50\%$). Apart from the effects on muscle activation and ligament forces, attention is focused on the alterations in contact forces on the medial and lateral compartments. It is hypothesized that the internal load partitioning is influenced primarily by changes in the adduction angle as compared to changes in the adduction moment.

2. Methods

2.1. Finite element model

An existing validated iterative kinematics-driven model that accounts for the active musculature of the lower extremity and detailed FE model of the knee joint is employed (Adouni and Shirazi-Adl, 2013, 2014; Adouni et al., 2012). This model incorporates the hip and ankle respectively as 3D and 1D spherical joints crossed by a total of 31 distinct muscles (Fig. 1). The knee joint is represented by a complex nonlinear model consisting of bony structures (tibia, patella, and femur), tibiofemoral (TF) and patellofemoral (PF) joints, major TF (ACL, PCL, LCL, and MCL) and PF (MPFL and LPFL) ligaments, patellar tendon (PT), as well as quadriceps (4 distinct muscles), hamstrings (6 muscles), gastrocnemius (2 muscles) (see Fig. 1 caption). The bony structures are represented by rigid bodies due to their much higher stiffness (Donahue and Hull, 2002). Details on the knee musculature and ligaments (Fig. 1) are available elsewhere (Adouni and Shirazi-Adl, 2013, 2014; Mesfar and Shirazi-Adl, 2005).

The depth-dependent fibrils networks at different regions of articular cartilage and menisci are considered. In superficial zones of femoral and tibial cartilage (15% of total thickness) as well as bounding surfaces of menisci, the collagen fibrils are simulated by membrane elements with uniform fibril distribution. In cartilage transitional zones (22.5% of thickness) with random fibrils (i.e., no dominant orientations), continuum brick elements that take the principal strain directions as the material principal axes represent collagen fibrils. In the deep zones (62.5% of thickness), fibrils are modeled with vertical membranes offering resistance only in their local fibril direction oriented initially normal to the subchondral junction. In the bulk region of each meniscus in between peripheral surfaces, collagen fibrils that are dominant in the circumferential direction are represented by membrane elements with local material principal axes defined in circumferential and radial directions. Thickness of membrane elements in cartilage and menisci is computed based on fibrils volume fraction in each zone. For cartilage, fibrils volume fractions of 15, 18 and 21% are considered in superficial, transitional and deep zones, respectively. In menisci, the collagen content is 14% in the circumferential direction and 2.5% in the radial direction of the bulk region along with 12% in the outer surfaces at both directions (Shirazi and Shirazi-Adl, 2009; Shirazi et al., 2008). The cartilage and menisci non-fibrillar matrices are simulated by continuum elements.

To study the short-term response of the joint, an elastic response (equivalent to a biphasic response) is taken with depth-dependent isotropic hyperelastic (Ogden-Compressible) material properties for the non-fibrillar solid matrix of cartilage with an elastic modulus varying linearly from 10 MPa at the surface to 18 MPa at the deep zone and a Poisson's ratio of 0.49. This model (considered here due to convergence difficulties) was initially verified to yield global displacements and stresses/strains almost identical to an earlier one having incompressible matrix with much lower moduli ($\sim$1 MPa) (Shirazi et al., 2008). The nearly incompressible hyperelastic model was initially also employed for the non-fibrillar menisci but due to convergence difficulties at contact areas, the matrix of menisci was represented, similar to our earlier studies (Mesfar and Shirazi-Adl, 2005), by a compressible elastic material with a Young's modulus of 10 MPa and a Poisson's ratio of 0.45.

2.2. Muscle force estimation

In each iteration, equilibrium equations are in the form of $\sum r \times f = M$ where r, f and M are respectively lever arms of muscles, unknown total muscle forces at the joint under consideration and associated moments. To resolve the redundancy, optimization algorithm with the cost function of sum of cubed muscle stresses (Arjmand and Shirazi-Adl, 2006) is employed along with inequality equations of muscle forces remaining positive but smaller than the maximum active forces (i.e. 0.6 MPa $\times$ physiological cross-sectional areas, PCSA). Muscle force passive components are neglected here due to negligible changes expected in muscle lengths.

2.3. Loading, kinematics and boundary conditions

Analyses are carried out at the mid-stance period of gait. The femur is initially fixed in its instantaneous position reported in gait while the tibia and patella are completely free except for the prescribed TF rotations. The hip/knee/ankle joint rotations/moments and ground reaction forces (GRF) at foot are taken from the mean data of in vivo measurements on asymptomatic subjects (Astephen et al., 2008; Hunt et al., 2001). The location of GRF at each instant is determined so as to generate reported joint moments (Astephen et al., 2008) accounting for the leg/foot weight (29.78 N/7.98 N). Since our model was constructed based on a female knee joint, a body weight of BW=606.6 N (61.9 kg) is considered (De Leva, 1996).

At mid-stance and subject to GRF and leg/foot weight, muscle forces at the hip, knee and ankle joints are predicted iteratively by counterbalancing moments in deformed configurations at each step. These muscle forces are subsequently applied as additional external loads and the procedure is repeated (8–10 iterations) till convergence (unbalanced moments < 0.1 N m). To investigate the effect of changes in the knee adduction rotation or moment on results, analyses are repeated under identical kinematics/kinetics except that the knee adduction angle or moment is altered one at a time; the reference adduction rotation of 1.6° is altered by $\pm 1.5°$ to 3.1° and 0.1° ($R \pm 1.5$) or the knee reference adduction moment of 17 N m is varied by $\pm 50\%$ to 25.5 N m and 8.5 N m ($M \pm 50\%$). These changes are chosen according to the reported variations in these quantities (Fig. 2) (Benoit et al., 2006; Leigh et al., 2013; Pohl et al., 2010). Matlab (Optimization Toolbox, genetic algorithms) and ABAQUS 6.11.2 (SIMULIA, Providence, RI) commercial programs are used.