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Computation of the role of kinetics, kinematics, posterior tibial slope and muscle cocontraction on the stability of ACL-deficient knee joint at heel strike – Towards identification of copers from non-copers

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A B S T R A C T

Rupture of anterior cruciate ligament (ACL) undermines normal activity and function of the knee joint and places it at higher risk of re-injury and degeneration. ACL reconstruction surgery neither necessarily ensures return to pre-injury activities nor alleviates risk of long-term degeneration. Here in this computational investigation of a lower-extremity hybrid model at heel strike (HS) of gait, we search for factors that influence the stability of the joint and hence the distinct performances between post-ACL injury copers and non-copers. Due to the very unstable state of the joint under the mean gait input data, joint rotations-moments, posterior tibial slope (PTS), and cocontraction were altered within the reported data in the literature and the effects on the joint stability (anterior tibial translation (ATT) and critical muscle stiffness coefficient (q_{cr})) were investigated. Results indicate that, in presence of both a small extension moment (0.1 or 0.2 Nm/kg) and a flexion rotation (~5–8°), ACL-deficient (ACL-D) knee joint stability substantially improves to levels computed in the pre-injury intact joint. In addition, low cocontraction levels of 1–3% (in hamstrings and quads only and not in gastrocnemii) and reduced PTS (by 5°) further improve ACL-D joint stability. Therefore for a stable joint with ATT < 3 mm and q_{cr} < 25 similar to those in the intact knee at HS, higher flexion angles (>5°) and a small extension moment (~0.1–0.2 Nm/kg) (i.e., higher activity in hamstrings than quads) are required. A lower posterior tibial slope (by 5°) and a small minimum cocontraction level (1–3%) in hamstrings and quads (but not in gastrocnemii) are also beneficial. These results identify mechanisms likely in play at HS in gait of copers when compared to non-copers.

1. Introduction

Anterior cruciate ligament (ACL) makes primary contributions to the anterior translational (Butler et al., 1980; Fukubayashi et al., 1982) and rotational (Fleming et al., 2001; Kanamori et al., 2002; Markolf et al., 1990) stiffness and stability (capacity of a system to support physiological loads and small perturbations therein without exhibiting hypermobility (Preuss and Fung, 2005; Reeves et al., 2011; Shahvarpour et al., 2016)) of the knee joint. Occurring especially in younger athletes, ACL rupture is one of the most common (>250,000 cases reported yearly in the US alone) and serious knee injuries causing pain, reduced performance, instability and increased risks both of failure to remaining intact tissues and of early osteoarthritis (Griffin et al., 2006; Hewett et al., 2013; Stein et al., 2012). About half of ACL reconstructed knees (~150,000 each year (Griffin et al., 2006; Koh, 2005)) develop symptomatic osteoarthritis within 12–14 years of initial injury (Grossman et al., 2005; Lohmander et al., 2004; Micheo et al., 2010; Von Porat et al., 2004). A smaller proportion of injured individuals (<14% (Eastlack et al., 1999)) stabilize their joint, show near-normal gait kinematics and return to intensive pre-injury activities (i.e., copers). In contrast, however, the rest (i.e., non-copers) experience instability and episodes of giving way even in regular daily activities (Eastlack et al., 1999). Quantification of the stability and parameters affecting it in ACL deficient (ACL-D) knee joints is therefore crucial in not only the joint performance evaluation and treatment management but also the identification and exploitation of markers for foregoing post-injury distinct functional performances in copers and non-copers.

Both the stability and equilibrium of the knee, similar to other joints (e.g., spine), are maintained by a delicate interplay between

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active musculature and passive structures (Crisco Iii and Panjabi, 1991; El-Rich et al., 2004; Preuss and Fung, 2005; Reeves et al., 2011). Partial injury to ACL could perturb this balance affecting joint stability, kinematics-kinetics, and load partitioning in gait (Sharifi et al., 2017). The observation of generally higher antagonistic coactivity in ACL-D patients in gait (Ciccotti et al., 1994; Grabiner and Weiker, 1993; Serrancolí et al., 2016; Solomonow et al., 1987) is in line with the notion that cocontraction stabilizes human joints. In addition, higher activity in biceps femoris whereas lower in quads and gastrocnemii at early stance (Limbird et al., 1988) along with higher flexion rotation (Frank et al., 2016; Fuentes et al., 2011) have been indicated. In contrast, lower flexion rotations (Lewek et al., 2002) and no differences in hamstrings activity (Hurd and Snyder-Mackler, 2007) have also been noted in ACL-D subjects.

Post-injury passive anterior laxity tests are not able to differentiate copers from non-copers (Eastlack et al., 1999; Herrington and Fowler, 2006). Higher cocontraction of hamstrings (Alkjær et al., 2002; Courtney and Rine, 2006) and of both hamstrings and quads (Alkjaer et al., 2003) have been reported in copers versus non copers. Less knee flexion rotations and moments (Alkjaer et al., 2011; Di Stasi and Snyder-Mackler, 2012; Hurd and Snyder-Mackler, 2007; Kaplan, 2012) though higher muscle contraction (Hurd and Snyder-Mackler, 2007; Kaplan, 2012) in non-copers have also been recorded. Moreover, smaller flexion angles at all stance periods, especially an extension angle instead of a flexion one at HS, are found in non-copers (Rudolph et al., 1998). In another study on non-copers, smaller flexion excursion in between early and late stance periods are reported in injured knees (Hurd and Snyder-Mackler, 2007). Increased activation in hamstrings but decreased activation in quadriceps are reported in non copers (Shanbehzadeh et al., 2017). The underlying contributory mechanisms responsible for the distinct functional performances in copers and non-copers have, nevertheless despite foregoing studies, remained obscure in part due to the fact that the sample populations in many studies indiscriminately included both ACL-D subject groups (Herrington and Fowler, 2006; Hurd et al., 2008).

Due to the agonist and protective role of hamstrings to ACL versus antagonist role of quads (at smaller flexion angles) and gastrocnemii (Adouni et al., 2016; Fleming et al., 2005; Mesfar and Shirazi-Adl, 2006), quadriceps avoidance (Sandberg et al., 1987; Williams et al., 2004; Williams et al., 2005) and hamstrings cocontraction (Ciccotti et al., 1994; Knoll et al., 2004; Solomonow et al., 1987) are considered as compensatory adaptations in ACL-D joints. The effectiveness of hamstrings in resisting and of quads in generating anterior tibial translation (ATT) is however influenced by changes in both the posterior tibial slope (PTS) (Marouane et al., 2014, 2015) and the knee flexion-extension (F-E) angle (Adouni et al., 2016; Mesfar and Shirazi-Adl, 2006). On the other hand, larger PTS that has been observed in subjects with non-contact ACL injuries (especially in females, Brandon et al., 2006; Gwinner et al., 2017; Todd et al., 2010; Zeng et al., 2016) substantially increases ATT and ACL force in compression and in gait (Marouane et al., 2014; Shao et al., 2011). In addition, greater flexion angles (specifically at heel strike, HS) (Button et al., 2008; Chen et al., 2012; Frank et al., 2016; Fuentes et al., 2011; Georgoulis et al., 2003; Shabani et al., 2015) and tibial internal rotation (Andriacchi and Dyrby, 2005; Chaudhari et al., 2008; DeFrate et al., 2006; Gao and Zheng, 2010; Georgoulis et al., 2003) have been reported in ACL-D subjects as likely post-injury dynamic adaptations.

Bearing the complexity to identify individual factors affecting distinct post-injury performances in copers and non-copers and exploiting our recent novel stability analyses with a validated hybrid lower extremity musculoskeletal (MS) model (Sharifi et al., 2017), we aim (1) to quantify for the first time the mechanical stability and ATT of a ACL-D joint at HS under reported mean gait kinematics-kinetics data and (2) to investigate the effects of alterations, within limits reported in the literature, in knee joint kinematics-kinetics, PTS, and muscle cocontraction on the joint stability. In view of earlier studies presented above, we hypothesize that changes in the knee sagittal flexion-extension (F-E) angles and moments, muscle cocontraction and PTS could markedly influence ACL-D knee joint stability and ATT toward levels computed in the intact joints. This could then be exploited in distinguishing copers versus non-copers.

2. Method

2.1. Lower extremity MS model

A detailed iterative kinematics-driven finite element MS model of the lower extremity (Fig. 1) is employed. This model (refined model) consists of rigid bony structures and their articular cartilage layers (tibia, femur, and patella), menisci, patellar tendon (PT), primary ligaments (ACL, PCL, LCL, MCL, MPFL, LPFL), and 34 muscles. Knee joint is modeled as a complex 3-D nonlinear finite element model whereas hip and ankle joints are simulated as 3D and 1D frictionless spherical joints without passive resistance, respectively. Menisci and articular cartilage layers are modeled as non-homogeneous and nonlinear depth-dependent fibril-reinforced composites of collagen fibril networks and incompressible hyperelastic matrices (Shirazi et al., 2008). Ligaments are represented each by a number of nonlinear springs (tension only) with varying initial pre strains (Mesfar and Shirazi-Adl, 2006). More description of the model can be found elsewhere (Adouni and Shirazi-Adl, 2013; Adouni et al., 2012; Marouane et al., 2017).

To simulate ACL rupture (ACL-D), all 6 elements representing its anteromedial (am) and posterolateral (pl) bundles are removed. Moreover, to reduce the computational time and mitigate convergence difficulties in the unstable joint with ATT > 5 mm when ACL is removed, a coarser mesh of the model is employed (Marouane et al., 2015; Mesfar and Shirazi-Adl, 2008a,b) where articular cartilage layers are simplified as equivalent homogeneous compressible isotropic materials. To evaluate the relative accuracy of this coarse model, the refined model is used in a limited number of ACL-D cases with more stable knee joint at ATT < 5 mm.

2.2. Estimation of muscle forces at HS

The reference model is driven by mean gait kinetics (3 moments at the hip and at the knee and one at the ankle) and kinematics (3 rotations at the hip and at the knee and one at the ankle) data reported on 60 asymptomatic subjects at HS period of stance (Astephen, 2007). The ground reaction forces (Hunt et al., 2001) at the foot are applied at a pressure center as to generate these mean gait moments at the knee joint (Astephen, 2007; Astephen et al., 2008) while accounting for the leg/foot weights of 29.78/7.98 N in our female model with body weight of BW = 606.6 N and body height of 171 cm. In the iterative analyses of gait at HS, all joints are first rotated according to the reported rotations at HS, initial strains in ligaments are then applied followed by the application of ground reaction forces with the femur completely fixed in its rotated position and the knee joint left unconstrained in translations. The unknown muscle forces are iteratively calculated to counterbalance reaction moments at all joints and applied in the following step as additional external loads. This procedure continues till convergence is reached (i.e., all unbalanced moments < 0.6 Nm). The nonlinear elastostatic analyses are carried out using ABAQUS (version 6.12, Simulia, Inc., Providence, RI, USA) finite element package program. Matlab (R2013a Optimization

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