Journal of Biomechanics 63 (2017) 110–116

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Computational stability of human knee joint at early stance in Gait: Effects of muscle coactivity and anterior cruciate ligament deficiency

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ARTICLE INFO

Article history:
Accepted 7 August 2017

Keywords:
Knee Joint
Musculoskeletal model
Stability
Gait
Finite element
Anterior cruciate ligament
Biomechanics
Muscle coactivation

ABSTRACT

As one of the most complex and vulnerable structures of body, the human knee joint should maintain dynamic equilibrium and stability in occupational and recreational activities. The evaluation of its stability and factors affecting it is vital in performance evaluation/enhancement, injury prevention and treatment managements. Knee stability often manifests itself by pain, hypermobility and giving-way sensations and is usually assessed by the passive joint laxity tests. Mechanical stability of both the human knee joint and the lower extremity at early stance periods of gait (0% and 5%) were quantified here for the first time using a hybrid musculoskeletal model of the lower extremity. The roles of muscle coactivity, simulated by setting minimum muscle activation at 0–10% levels and ACL deficiency, simulated by reducing ACL resistance by up to 85%, on the stability margin as well as joint biomechanics (contact/muscle/ligament forces) were investigated. Dynamic stability was analyzed using both linear buckling and perturbation approaches at the final deformed configurations in gait. The knee joint was much more stable at 0% stance than at 5% due to smaller ground reaction and contact forces. Muscle coactivity, when at lower intensities (<3% of its maximum active force), increased dynamic stability margin. Greater minimum activation levels, however, acted as an ineffective strategy to enhance stability. Coactivation also substantially increased muscle forces, joint loads and ACL force and hence the risk of further injury and degeneration. A deficiency in ACL decreases total ACL force (by 31% at 85% reduced stiffness) and the stability margin of the knee joint at the heel strike. It also markedly diminishes forces in lateral hamstrings (by up to 39%) and contact forces on the lateral plateau (by up to 17%). Current work emphasizes the need for quantification of the lower extremity stability margin in gait.

1. Introduction

Knee joint is one of the most complex structures of human body that undergoes large motions and loads in various activities of daily living. As in every similar biomechanical system under loads, its normal and efficient performance within physiological range of displacements require simultaneous maintenance of equilibrium and dynamic stability in different directions. The latter requirement is essential for a safe performance both in voluntary movements as stair ascent/descent and gait and in perturbations under sudden loading and disturbances. Dynamic stability of the knee joint in daily activities is maintained by a delicate interplay between the passive tissues (i.e., articular surfaces, menisci, ligaments, passive muscles) and active musculature (voluntary and reflex activation). Knee joint instability accompanies excessive laxity, manifests itself in "giving way" episodes and has been associated with pain and osteoarthritis (OA) (Mulvey et al., 2013; Rudolph et al., 2001; Schmitt and Rudolph, 2008). Therefore, quantification of the knee joint stability reserve or margin (i.e., additional force that can be supported without instability) and factors affecting it is crucial in performance evaluation/enhancement, injury prevention and treatment managements.

Stability of a musculoskeletal (MS) system similar to any other mechanical system is defined as its ability to withstand small perturbations without exhibiting unbounded or growing response (displacements). In clinical context, the mechanical stability of a biological system is usually examined by its hypermobility when excessive laxities are detected under external physiological loads and disturbances. The Lachman and pivot shift tests are for example commonly performed to access passive knee laxity particularly following anterior cruciate ligament (ACL) injuries (Torg et al., 1976; Galway and MacIntosh, 1980). They remain however limited when dealing with the dynamic knee stability and the distinction between copers versus non-copers (Rudolph et al., 2001).

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http://dx.doi.org/10.1016/j.jbiomech.2017.08.004
0021-9290/

To assist passive structures to maintain and enhance control, stiffness and stability while performing various tasks, musculature plays a significant role through coactivation (Baratta et al., 1988; Grabiner and Weiker, 1993; Alkjær et al., 2012). Role of antagonistic coactivity in knee joint stability has been demonstrated in isometric knee extension (Hirokawa et al., 1991) and gait (Schmitz et al., 2009; Peterson and Martin 2010). Its role in the trunk biomechanics has also been recognized during trunk voluntary movements (Gardner-Morse and Stokes, 1998; Granata and Orishimo, 2001; El Ouaaid et al., 2009, 2013) as well as a preparatory strategy ahead of likely disturbances (Shahvarpour et al., 2015). Greater trunk muscle coactivity has been observed in exercises performed on unstable surfaces (Vera-Garcia et al., 2000; Lehman et al., 2005; Imai et al., 2010). Higher antagonistic coactivity recorded in patients with various knee injuries is usually interpreted as a stabilizing neuromuscular strategy; Mari et al. (2014) reported higher antagonistic coactivity in ataxic patients. OA has been associated with lower knee joint stability because of weakened quadriceps strength, pain, and altered joint structure (McAlindon et al., 1993; O'reilly et al., 1997; Sharma, 2001; Hurley, 2003) requiring thus, as an adaptive strategy, higher muscle coactivity (Childs et al., 2004; Hortobágyi et al., 2005; Schmitt and Rudolph, 2007; Heiden et al., 2009). Greater muscle coactivity has also been observed in ACL-deficient non-copers (Limbird et al., 1988; Rudolph et al., 2001). On the other hand and as an undesired consequence, higher muscle coactivation increases joint loads and hence the risk of further injury, degeneration and even instability (Griffin and Guilak, 2005; Hodges et al., 2016; Schmitt and Rudolph, 2008).

Despite numerous computational studies on the dynamic stability of the human trunk (Bazrgari and Shirazi-Adl, 2007; Crisco and Panjabi, 1991; El Ouaaid et al., 2009; Granata and Orishimo, 2001), there has hardly been any computational study on the mechanical stability of the knee joint or lower extremity using MS models. Goals of this study are hence set as follows: (1) to quantify the mechanical stability of the human knee joint as well as the lower extremity at early stance of gait (0% and 5% stance periods) using a hybrid MS model of the lower extremity (Adouni et al., 2012; Adouni and Shirazi-Adl, 2013) and (2) to investigate the role of muscle coactivity set as a minimum activity level and ACL partial ruptures on the joint stability margin as well as biomechanics (muscle/contact/ligament forces) at early stance periods of gait.

As a first study on the knee joint stability in gait, we hypothesize that (1) higher antagonistic coactivity increases the system stability margin but at the cost of greater muscle forces, joint internal loads, ACL force and tissue stresses and hence the risk of (further) injury and (2) partial ACL rupture destabilizes the joint while increasing tibial anterior translation (ATT).

2. Methods

2.1. Lower extremity MS model

A detailed iterative kinematics-driven finite element (FE) model of the lower extremity (Fig. 1) including 3 rigid bony structures (tibia, femur, and patella) and their articular cartilage layers, major TF and PF ligaments (ACL, PCL, LCL, MCL, MPFL, LPFL), menisci, and patellar tendon (PT) is utilized (Adouni et al., 2012; Adouni and Shirazi-Adl, 2013). In total, there are 34 muscles in the hybrid MS model (Fig. 1). Hip and ankle are respectively modeled as 3 D and 1 D spherical joints without any passive resistance and knee

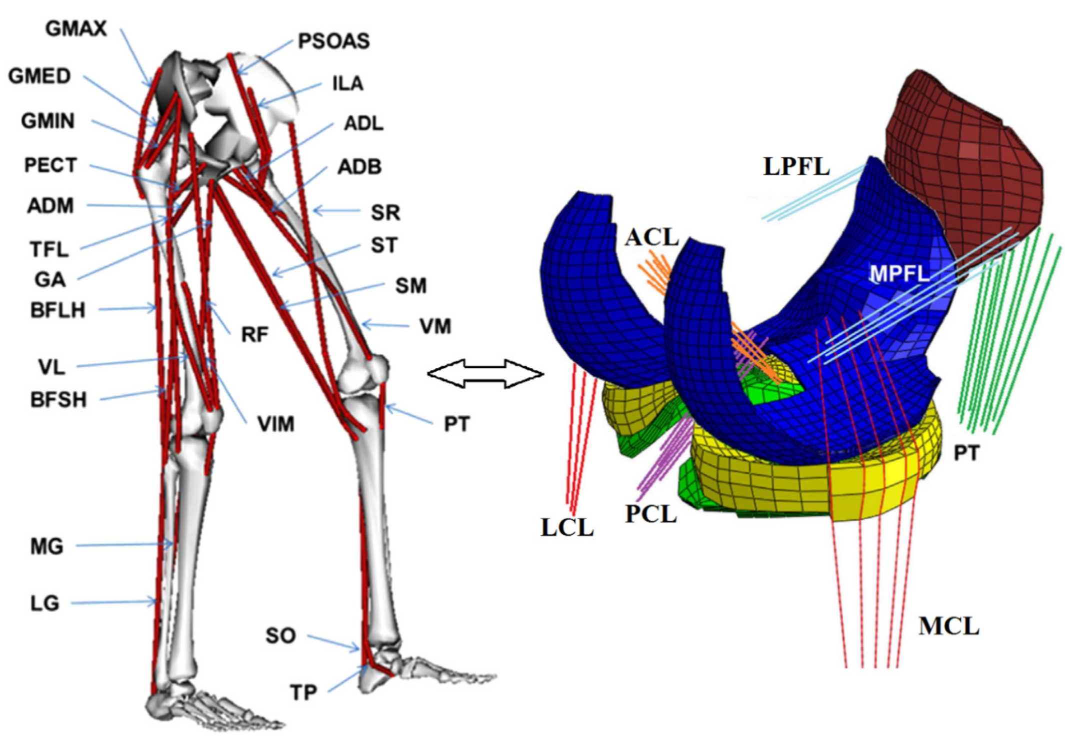

Fig. 1. Knee FE model including tibiofemoral (TF) and patellofemoral (PF) cartilage layers, menisci, patellar Tendon (PT) and ligaments: lateral patellofemoral (LPFL), medial patellofemoral (MPFL), anterior cruciate (ACL), posterior cruciate (PCL), lateral collateral (LCL) and medial collateral (MCL). A total of 34 muscles were incorporated into the lower extremity model including (Delp et al., 2007) quadriceps components: vastus medialis obliquus (VMO), rectus femoris (RF), vastus intermidus medialis (VIM) and vastus lateralis (VL), hamstrings components: biceps femoris long head (BFLH), biceps femoris short head (BFSH), semimembranous (SM) and TRIPOD madeofsartorius (SR), gracilis (GA) and semitendinosus (ST), gastrocnemius components: gastrocnemius medial (GM) and gastrocnemius lateral (GL) and soleus (SO) muscle as a uni-articular ankle muscle. Hip joint muscles (not all shown): adductor, long (ADL), mag (3 components ADM) and brev (ADB); gluteus max (3 components GMAX), med (3components GMED) and min (3 components GMIN), iliacus (ILA), iliopsoas (PSOAS), quadriceps femoris; pectineus (PECT), tensor facialata (TFL), periformis.