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Computational biodynamics of human knee joint in gait: From muscle forces to cartilage stresses

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ABSTRACT

Using a validated finite element model of the intact knee joint we aim to compute muscle forces and joint response in the stance phase of gait. The model is driven by reported *in vivo* kinematics–kinetics data and ground reaction forces in asymptomatic subjects. Cartilage layers and menisci are simulated as depth-dependent tissues with collagen fibril networks. A simplified model with less refined mesh and isotropic depth-independent cartilage is also considered to investigate the effect of model accuracy on results. Muscle forces and joint detailed response are computed following an iterative procedure yielding results that satisfy kinematics/kinetics constraints while accounting at deformed configurations for muscle forces and passive properties. Predictions confirm that muscle forces and joint response alter substantially during the stance phase and that a simplified joint model may accurately be used to estimate muscle forces but not necessarily contact forces/areas, tissue stresses/strains, and ligament forces. Predictions are in general agreement with results of earlier studies. Performing the analyses at 6 periods from beginning to the end (0%, 5%, 25%, 50%, 75% and 100%), hamstrings forces peaked at 5%, quadriceps forces at 25% whereas gastrocnemius forces at 75%. ACL Force reached its maximum of 343 N at 25% and decreased thereafter. Contact forces reached maximum at 5%, 25% and 75% periods with the medial compartment carrying a major portion of load and experiencing larger relative movements and cartilage strains. Much smaller contact stresses were computed at the patellofemoral joint. This novel iterative kinematics-driven model is promising for the joint analysis in altered conditions.

1. Introduction

Activities of daily living such as walking and stair climbing impose relatively large loads and movements on the human knee joint. This mechanical burden increases in many occupational and recreational tasks causing injuries and degenerations in joint ligaments, menisci, cartilage and bone. Any failure, degeneration or alteration in one of these components influences the response of the entire joint and likely increases the risk of further perturbations (Moglo and Shirazi-Adl, 2005). Effective preventive and conservative/surgical managements of joint disorders depend hence on a sound knowledge of stress and strain distributions in various components under both intact and altered conditions. These values, in turn, are heavily dependent not only on external loads and inertial forces but on muscle activities across the joint. As such, accuracy in estimation of muscle forces has a direct bearing on the reliability of stresses and strains. Since direct *in vivo* measurements of tissue stresses and muscle forces remain invasive and likely impossible, computational modeling is recognized as a vital complementary tool to estimate multiple variables of interest. Due to technical difficulties in measurements and consideration of physiological loads and motions, *in vitro* testing is also limited especially when looking for cartilage/meniscus stresses/strains and ligament forces.

Using instrumented knee implants, efforts have been directed to *in vivo* measurement of loads on the knee joint of subjects in daily activities (Kutzner et al., 2010). Forces up to ~3 times body weight have been recorded in walking. These measured data, though very valuable, are however collected on implanted joints and not intact ones. Moreover, they are restricted to the load components transmitted via the implant itself. Mathematical modeling along with joint kinematics and ground reaction forces are combined with such measurements to estimate muscle, compartmental, and ligament forces (Anderson and Pandy, 2001; Delp et al., 2007; Kim et al., 2009; Lin et al., 2010; Shelburne et al., 2004; Zhao et al., 2007). To gain further insight into biomechanics of the joint during gait, some recent works have employed MRI, video motion systems and fluoroscopy to

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estimate changes in ACL length, contact areas and cartilage thickness (Andriacchi et al., 2009; Coleman et al., 2011; Koo et al., 2011; Taylor et al., 2011) or to identify the effect of ACL-deficiency on joint kinematics (Chen et al., 2011). The medial plateau is found to experience larger tibiofemoral movements and cartilage strains.

In parallel, other studies have used 3D link-segment models (inverse dynamics and optimization) to predict load distribution between medial and lateral compartments as well as ligament forces during simulated gait (Shelburne et al., 2004, 2005, 2006). These models, however, explicitly neglect not only the compliant cartilage layers and menisci but the knee joint passive resistance. Similarly and using measured *in vivo* load in one subject with instrumented knee, joint contact forces have been estimated during the gait (Lin et al., 2010; Zhao et al., 2007). Simulating cartilage and menisci with linear elastic properties, a recent finite element (FE) model study in gait has on the other hand neglected joint rotations, redundancy in equations, patellofemoral joint, and out-of-sagittal plane moments (Yang et al., 2010).

Despite foregoing *in vivo* measurement and model studies, a detailed musculoskeletal FE model of the entire intact knee joint with depth-dependent nonlinear cartilage/meniscus properties remains yet to be developed. This model study should provide important data during gait on not only the muscle forces but the ligament forces, contact stresses as well as stresses/strains within the cartilage and menisci. Such is the aim of the current study that is based directly on earlier developments, validations and applications of a detailed FE model of the knee joint (Adouni and Shirazi-Adl, 2009; Bendjaballah et al., 1995; Mesfar and Shirazi-Adl, 2006; Moglo and Shirazi-Adl, 2005; Shirazi et al., 2008). In the current study, cartilage of both tibiofemoral (TF) and patellofemoral (PF) joints along with menisci are simulated as depth-dependent non-homogeneous composites enclosing nonlinear collagen fibril networks (Shirazi and Shirazi-Adl, 2009a,b; Shirazi et al., 2008). For the sake of comparison, a less-refined model with isotropic (no fibril networks) depth-independent cartilage is also considered. Both models incorporate the mean reported *in vivo* gait data on hip/knee/ankle joint moments/rotations (Astephen et al., 2008a) and ground reaction forces (Hunt et al., 2001) of asymptomatic subjects. In this manner, biodynamics of gait during the stance phase, at both global musculature and local tissue-level, are investigated with iterative FE models of the knee joint and lower extremity that are driven by measured *in vivo* kinetics/kinematics. We hypothesize that (1) muscle activation levels and contact stresses/areas substantially alter during the stance phase and (2) in contrast to contact stresses/areas, ligament forces and tissue stresses/strains, the muscle forces can as accurately be estimated by the less complex FE model.

2. Methods

2.1. Finite elements model

The FE model includes bony structures (tibia, patella and femur), TF and PF joints, major TF (ACL, PCL, LCL and MCL) and PF (MPFL and LPFL) ligaments, patellar tendon (PT), as well as quadriceps (3 distinct muscles), hamstrings (3 muscles), gastrocnemius (2 muscles), tibialis posterior and soleus (Adouni and Shirazi-Adl, 2009; Shirazi et al., 2008; see Fig. 1 for details). The bony structures are represented by rigid bodies due to their much higher stiffness (Donahue and Hull, 2002).

Muscle components are modeled by uniaxial elements with orientations at full extension taken from the literature. The Q angle model ($Q=14^\circ$; Sakai et al., 1996) is used for quadriceps muscles; orientations relative to the femoral axis in frontal/sagittal planes are: RF–VIM 0°/4° anteriorly, VL 22° laterally/0° and VMO 41° medially/0°. Orientations for hamstrings muscles relative to the tibial axis, respectively for BF, SM, and TRIPOD are taken (Aalbersberg et al., 2005) as 11.8° medially, 7° laterally, and 7.1° medially in the frontal plane whereas 0°, 16.1°, and 18.7° posteriorly in the sagittal plane. Gastrocnemius fascicles are parallel to the tibial axis in the sagittal plane while oriented (GM) 5.3° medially or (GL) 4.8° laterally in the frontal plane (Delp et al., 2007; Hillman, 2003). Tibialis posterior/Soleus are oriented 5.3°/4.1° laterally and 1.0°/4° anteriorly relative to the tibial

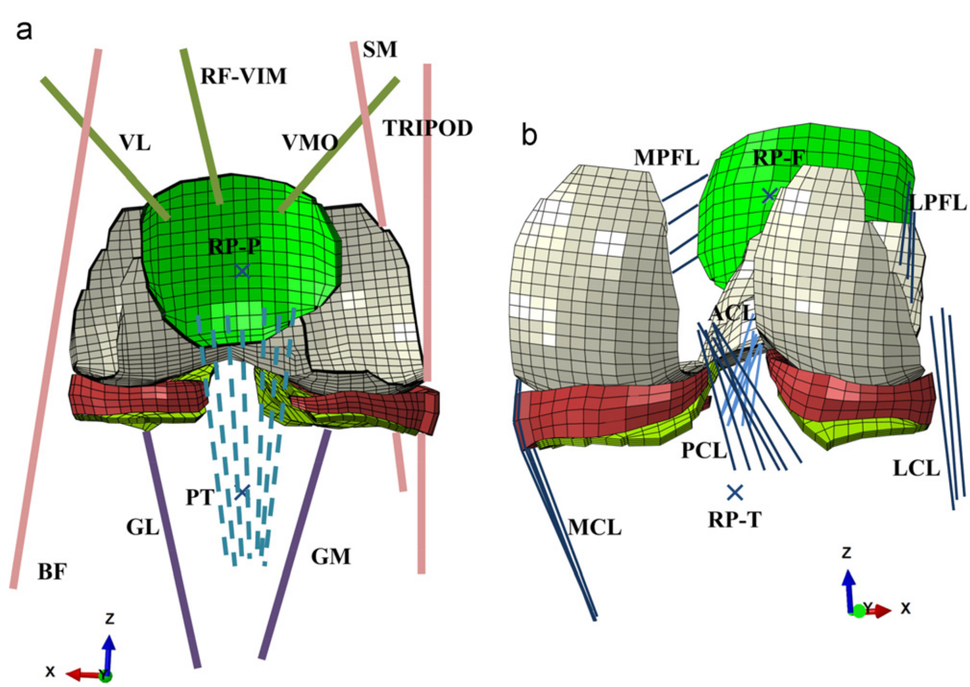

Fig. 1. (a) Knee joint FE model; tibiofemoral (TF) and patellofemoral (PF) cartilage layers, menisci, patellar tendon (PT), quadriceps, hamstrings and gastrocnemius muscles, (b) TF and PF cartilage layers, menisci and joint ligaments. Crosses indicate the reference points (RP) representing rigid bony structures that are not shown. Quadriceps components are vastus medialis obliquus (VMO), rectus femoris (RF), vastus intermidus medialis (VIM), and vastus lateralis (VL). Hamstring components include biceps femoris (BF), semimembranous (SM), and the TRIPOD made of sartorius (SR), gracilis (GR), and semitendinosus (ST). Gastrocnemius components are gastrocnemius medial (GM) and gastrocnemius lateral (GL). Tibialis posterior and soleus muscles of the ankle joint are not shown. Joint ligaments include lateral patellofemoral (LPFL), medial patellofemoral (MPFL), anterior cruciate (ACL), posterior cruciate (PCL), lateral collateral (LCL), and medial collateral (MCL).