Computational Biodynamics of Human Knee Joint in Gait Cycle: From Muscle Forces to Cartilage Stresses

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INTRODUCTION:
Due to large loads/motions in activities of daily living, human knee joint is prone to an elevated risk of injuries and degeneration. Effective preventive measures and conservative/surgical management of joint disorders depend on a sound knowledge of tissue-level stresses/strains and hence of muscle forces and joint motions. The vital role of computational modeling of the knee joint is recognized, as direct in vivo measurements (though valuable) remain limited, invasive, and costly. Due to technical difficulties in consideration of multi-articulations and physiological loads/motions, in vitro testing is also restricted especially when looking for cartilage/contact stresses and ligament forces. In the current work, we aim to use a validated detailed finite element (FE) model of the knee (with both tibiofemoral (TF) and patellofemoral (PF) joints) that includes novel depth-dependent nonlinear fibre-reinforced menisci and cartilage layers as well as major ligaments and muscles [1,2]. Experimental data on hip/knee/ankle joint moments/rotations [3] and ground reaction forces [4] recorded in asymptomatic subjects, we simultaneously compute unknown muscle forces and detailed joint response during the stance phase of gait. In this manner, biodynamics of gait, at both global musculature and local tissue-level, are investigated with a complex iterative FE model of the knee joint and lower extremity that is driven by measured in vivo kinematics and loads.

METHODS:
The FE model considers bony structures (tibia, patella, femur), TF and PF joints, major TF (ACL, PCL, LCL, and PF (MPFL, LPFL) ligaments, patellar tendon (PT), as well as quadriceps (3 distinct muscles: RF/VI, VM, VL), hamstrings (3 muscles: BF, SM, SR/GR/ST) and Gastrocnemius (2 muscles: GM, GL) muscle groups [1,2]. Depth-dependent isotropic hyperelastic material properties are considered for non-fibrillar solid matrix of PF and TF cartilage layers with the elastic modulus varying linearly from 10 MPa at the surface to 18 MPa at the deep zone. In the short-term gait loading of the current work, this model with a compressible material is verified to be equivalent both to an incompressible elastic model with much smaller moduli and a poroelastic model [1]. Cartilage matrices are reinforced by nonlinear collagen fibril networks parallel to the articular surface in the superficial zone, randomly oriented in the middle zone and vertical in the deep zone [1]. Menisci are also nonlinear anisotropic with depth-dependent collagen fibril networks. Ligaments are each simulated by a number of nonlinear axial elements with initial prestrains [2].

The hip/knee/ankle joint rotations/moments and ground reaction forces at foot during the entire stance phase (Fig. 1a) are based on reported in vivo measurements [3,4] taking a body weight of 61.9 kg for our knee model of a female subject. Analyses are performed at 5 time instances corresponding to beginning (heel strike, HS), 12.5%, 25%, 50%, 75%, and end (toe off, TO) points of the stance phase (Fig. 1a). The femur is fixed in its position while the tibia and patella are completely free except for the TF rotations (Fig. 1a). Under prescribed TF rotations, ground reaction forces, and leg/foot weights, unknown forces in all muscles are iteratively estimated by minimization of the sum of cubed muscle stresses as the objective function and reported joint moments (Fig. 1a) [3] as constraint equations. The MATLAB and ABAQUIS commercial programs are used iteratively at each step till convergence is reached.

RESULTS:
Forces in the GL and GM muscles initiated at 25% and reached their maximum (respectively, 92 N and 256 N) at 75% of the stance phase (Fig. 1b). Large activation levels were estimated in the RF/VI and VL at 25% and in the BF at 25%-75% periods (Fig. 1b). Force in PT increased from 202 N at HS to its peak of 946 N at 25%. ACL Force (PL bundle only) reached its maximum of 343 N at 25% and decreased thereafter. Force in the LCL reached its maximum of 206 N at TO while forces in other ligaments remained < 20 N at all times. Total contact forces were much greater in the medial plateau reaching 1856 N at 25% and 1770 N at 75% (Fig. 1c). Contact force in the lateral plateau also peaked at 25% but at a much smaller value of 595 N (Fig. 1c). Contact pressure peaked at the medial tibial/femoral cartilages at 25% reaching 8.1/10.1 MPa (Fig. 2a,b). TF load transfer occurred primarily via cartilage layers at uncovered areas. Much smaller contact forces and stresses were computed at the PF joint (Figs. 1c and 2a).

DISCUSSION:
A detailed kinematics-driven FE model of the entire knee joint combined with a biomechanical model of the lower extremity are employed simultaneously and iteratively to compute the joint nonlinear response throughout the stance phase subject to in vivo hip/knee/ankle rotations/moments [3] and ground reaction forces [4] recorded in gait. This is a novel complex model that accurately accounts for the joint passive properties while computing not only muscle forces but the tissue-level stresses/strains and load transmission. Muscle forces (Fig. 1b) are in overall agreement with reported estimations and follow the same trends as in measured EMG [3]. Similarly, load transmission and contact forces (Fig. 1c) corroborate data available in the literature [5].

In accordance with the joint adduction moment/rotation (Fig. 1a), much greater contact forces/pressures occur on the medial compartment (Figs. 1c and 2a) as opposed to the lateral one. The medial side has indeed been reported to have higher incidence of OA. Under relatively small knee flexion angles in the early stance phase (Fig. 1a) and subject to large quadriceps (and hence PT) forces, ACL posterolateral bundle supported the entire load of 231 N at HS that increased to its maximum of 343 N at 25% and thereafter decreased to 138 N at 50%, 107 N at 75% and finally 13 N at TO. These results highlight the crucial mechanical role of the ACL in early stance phase and call for caution in the post ACL reconstruction rehabilitation period assuming that the gait characteristics remain unchanged after the ACL replacement.

SIGNIFICANCE:
The current novel iterative kinematics-driven FE model that accounts for the synergy between passive structures and active musculature of the knee joint could provide vital results for ACL-deficient/reconstructed and OA joints and hence for the OA pathology and tissue engineering.

ACKNOWLEDGEMENTS: MUTAN-Tunisia and NSERC-Canada.