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TIBIO-FEMORAL CONTACT FORCE DURING GAIT: AN ITERATIVE METHOD USING EMG-CONSTRAINED MULTI-BODY SIMULATION AND FINITE ELEMENT ANALYSIS

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INTRODUCTION

In this study, we present an innovative methodology (Figure 1) to calculate patient specific tibio-femoral (TF) contact forces by integrating medical image data, 3D skin-mounted marker trajectories, ground reaction forces, electromyography (EMG) data and finite element analysis (FEA). The muscle redundancy problem is solved through an EMG-constrained optimization approach. Calculated muscle forces are input to a FEA to calculate TF contact forces. Kinematics of the degrees of freedom (DOFs) of the knee that cannot be accurately assessed from the trajectories of skin-mounted markers, are estimated using a novel iterative procedure which combines muscle force calculation with dynamic FEA. The presented methodology is applied to analyze TF contact forces of a walking trial performed on an instrumented treadmill of which the speed was sequentially ramped up and down. The results presented in this abstract will be validated against the in-vivo measured TF contact forces.

METHODS

Based on the marker trajectories, ground reaction forces, EMG signals, implant geometry and medical image data provided by the ASME Grand Challenge organizers [1], medial and lateral TF contact forces were calculated for subject JW (height 168cm, weight 66.7kg) during one gait cycle of the accelerating phase (gc1) and one gait cycle of the decelerating phase (gc2) of a walking trial on an instrumented treadmill.

Figure 1 outlines the innovative methodology that has been applied for calculating TF contact forces. All multi-body simulations were based on the provided OpenSim [2] model in which the TF joint has six DOFs. Musculotendon (MT) parameters were taken from the model of Delp [3] and scaled with respect to the MT lengths in the anatomic position.

During inverse kinematics (step1), the patella-femoral flexion was constrained to be two thirds of the knee flexion [4] while the other translations and rotations describing patella motion with respect to the femur were kept constant. Kinematics, of all remaining DOFs in the model, was estimated using a Kalman smoother algorithm [5] implemented as an OpenSim tool.

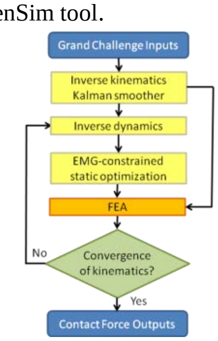

Figure 1: Schematic outline of the implemented methodology.

Joint reaction moments were calculated using an inverse dynamic analysis (Opensim, step 2) and were input to two consecutive static optimization procedures to calculate individual muscle activations (step3):

1. First, muscle activations were constrained to lie between 0 and 1, i.e. the physiological limits.
2. Second, the muscle activations were constrained below based on the experimental surface EMG. The EMG was rectified and scaled to the maximal muscle activations calculated by the first

static optimization (1). The lower bound on the muscle activation was set to half of the scaled EMG signal. Since EMG was only available for 14 superficial muscles, muscles were grouped with relevant synergists. For some muscles, no relevant synergist could be identified and muscle activations were unconstrained.

In step 4 of the procedure, a FEA model was built based on the available CT image data and the implant geometry. Bone models representing the pre and post-operative geometry were obtained from pre and post-operative CT images using Mimics (Materialize, Leuven, Belgium). The insertion point of the medial and lateral collateral ligament, patellar and posterior cruciate ligament were defined in the pre-operative bone models based on the centroid of the respective attachments. Relevant soft-tissue attachment sites were transferred to the post-operative model by rigidly registering the corresponding 3D bone models. The defined bone and soft-tissue geometries were assembled with the TKA components following the surgical cuts and positioned in accordance with the reference position of the provided OpenSim model. For further FEA simulations, bone models were removed from the assembly to reduce the computational cost. For each component of the assembly, the material models and properties were based on the literature [6,7]. The tibial component, underlying the tibial tray, was fixed. The ligaments were pre-strained [8]. Muscle force directions and attachment points were calculated based on the static optimization results from step 3 [9]. The contributions of the four quadriceps muscles were combined into one single equivalent force and point of application on the patella, whereas the knee flexor forces were separately applied to the femur.

During the consequent explicit dynamic FEA (Abaqus 6.12-1, Dassault Système, France), the varus/valgus and internal/external rotation of the femoral component was calculated while the flexion/extension was constrained to follow the knee flexion angle obtained from the inverse kinematics (step1). The patella was free to move.

FEA-based kinematics of the unconstrained DOFs of the patella and of the femoral component (step 4) was compared to the initial kinematics that served as input to the muscle force calculation (step 3). An iterative procedure between both steps was implemented until convergence of the unconstrained kinematics. At convergence, the associated TF contact forces are saved.

RESULTS AND DISCUSSION

Figure 2 shows TF contact forces for the first cycle of gc1 (a) and gc2 (b).

The first novelty of our methodology lies in the successful integration of and iteration between multi-body simulations and FEA. Secondly, the applied Kalman smoother algorithm results in more accurate estimates of the DOFs with a limited range of motion. Finally, the EMG-constrained static optimization procedure combines the advantages of optimization procedures and EMG-driven simulations. It calculates individual muscle forces for a large number of muscles while accounting for the experimentally measured muscle activation patterns. Further steps in our analysis include the assessment of the subject-specific ligament properties based on the laxity tests and the validation of the predicted kinematics based on the fluoroscopy images.

Computational methods can complement surgical decision making and patient rehabilitation; this potential is recognized by those developing such methods although a significant portion of the clinical community remains skeptical. The onus is on us to convince our critics about the validity of the model with respect to relevant output parameters. The Grand Challenge provides such an opportunity. We hope this effort will be continued and that extended data collection in patients with an instrumented knee implant will allow us to validate the tools we developed for MRI-based musculoskeletal modeling and MT parameter estimation based on extended dynamometer experiments.

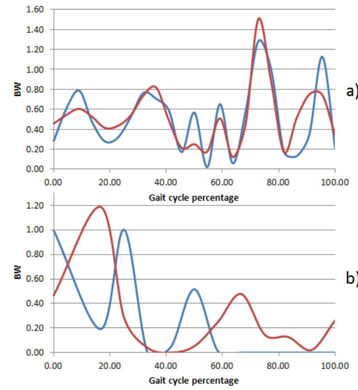

Figure 2: Lat (blue) and Med (red) TF contact forces: (a) gc1, (b) gc2.

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