DEVELOPMENT OF A SUBJECT-SPECIFIC LOWER LIMB WITH A NATURAL KNEE MODEL AND EVALUATION DURING A LUNGE

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INTRODUCTION

Joint movements and loads are difficult to measure *in vivo* in complex musculoskeletal (MS) systems, and, therefore, MS models are applied to estimate internal kinematics and kinetics. Most models use idealized joints (such as a revolute joint knee) [1]. Although useful, such kinematic constraint-based models cannot estimate the effect of loads on the internal kinematics of non-conforming joints. More advanced MS models that include detailed joint representations allow for estimation of joint contact forces and secondary kinematics, which is necessary to study pathologies like osteoarthritis [2]. The geometries that these models are constructed upon has also been identified as a critical point for the accurate estimation of internal forces [3].

Recently, Marra et al. [4] proposed a methodology that included a morphing technique to use subject-specific geometries extracted from medical images, together with the application of the Force-dependent Kinematics (FDK) method [5]. This method proved accurate and allows to simultaneously estimate muscle, ligament and joint contact forces, and internal kinematics.

Although necessary validation efforts have been made to validate advanced MS models, most studies rely on instrumented prostheses [2]. However, results from such studies might not necessarily be transferable to healthy subjects, and therefore, further validation efforts in subject-specific natural detailed joint models with non-invasive validation methods that can be adopted in healthy subjects [6,7] should be conducted.

The aims of this study were (1) to apply a MS modeling workflow based on subject-specific magnetic resonance images (MRI), motion capture, and force plate data capable of estimating ligament, muscle, patellofemoral and tibiofemoral joint contact forces, and internal kinematics in a natural knee; and (2) to evaluate the estimated secondary joint kinematics based on experimental measurements using EOS^R biplanar X-rays during a quasi-static lunge.

METHODS

Four subjects' (age: 38 ± 10 years, body mass: 74 ± 7 kg, height: 1.86 ± 0.06 m) right leg lunges from roughly 0° to 90° of right knee flexion angle were recorded with a marker-based motion capture system (Qualisys, Sweden) and a force platform (AMTI Corp., USA).

To construct MRI-based models in the AnyBody Modeling System (AMS, Anybody Technology, Denmark), subjects underwent various MRI acquisitions that were manually segmented with Mimics 18.0 (Materialise, Belgium) to acquire subject-specific right leg bones (full pelvis, femur, patella, tibia, talus and foot). The bone geometries were used to morph the Twente Lower Extremity Model (TLEM) 2.0 in Mimics and AMS following an advanced morphing technique [4] developed by Materialise NV (Materialise, Belgium).

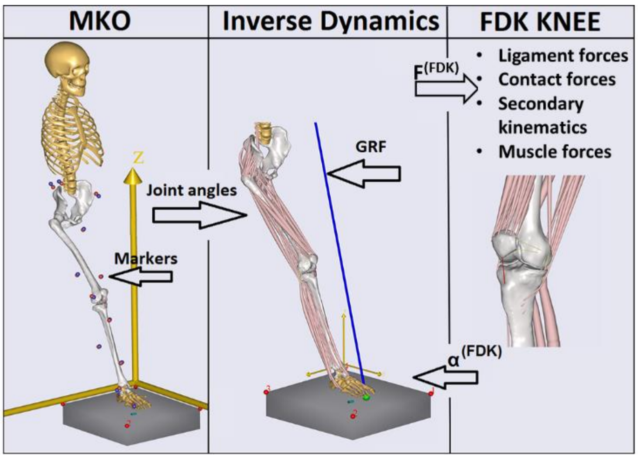

Fig 1. Simulation workflow in AnyBody Modeling system. Left: Multibody kinematics Optimization, Middle: FDK-based Inverse dynamic model. Right: Closer look to the subject-specific natural knee. $F^{(FDK)}$: FDK residual force, $\alpha^{(FDK)}$: FDK degree of freedom.

A Multibody Kinematics Optimization (MKO) is used in AMS to compute the positions and orientations of all segments from the motion capture data as first part of the simulation process. During this phase of the simulation (Fig. 1), all joints were assumed idealized (three degrees of freedom in the hip, one in the knee and two in the ankle).

The resulting optimized model kinematics from the MKO and experimentally recorded ground reaction forces and moments were used as input to the FDK-based inverse dynamic analyses (Fig. 1). For this phase of the simulation, the knee kinematic-based constraints were released and substituted by an 11 degree of freedom FDK knee joint (6 in the tibiofemoral joint and 5 in the patellofemoral joint, as the patellar ligament was assumed rigid). To restrict and stabilize the FDK knee joint, ligaments were included (anterior cruciate, posterior cruciate, medial collateral, lateral collateral, lateral epicondylo-patellar, lateral transverse and medial patellofemoral), characterized by three nonlinear force-displacement regions, and segmented from subject-specific MRI. Tibiofemoral and patellofemoral articular cartilages were also segmented from the MRI to compute contact forces based on an elastic foundation contact model. Three-element Hill-type muscle models defined muscle dynamics and muscle forces computed with a third order polynomial cost function. The FDK degrees of freedom were free to equilibrate between all acting forces (gravity, inertia, ligament, contact, and muscle forces).

To evaluate model performance, bi-plane X-ray images captured with EOS[R] (EOS Imaging, France) were used to reconstruct the position and orientation of femur, tibia and patella at certain knee angles during the right knee lunges [7] using Matlab (Mathworks, USA) and AMS.

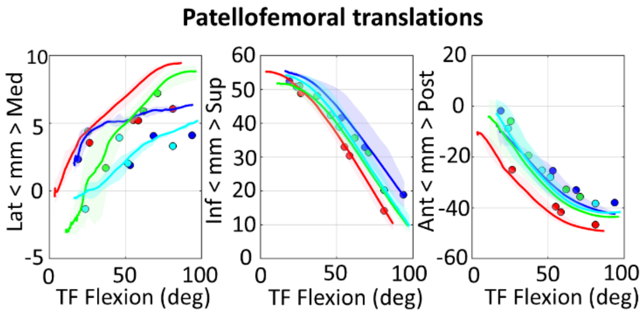

Fig. 2. Patellofemoral translations. Experimental measures in points and predicted results in lines. Left: medio-lateral. Middle: inferior-superior. Right: anterior-posterior. Each color represents a subject.

RESULTS AND DISCUSSION

Accurate estimates of patellofemoral translations (Fig. 2) and tibiofemoral secondary joint kinematics (Fig. 3) were found: translations were predicted with a mean difference (MD) and standard error (SE) of 2.13±0.22 mm between all trials and measures while rotations had a MD±SE of 8.57±0.63°.

The additional degrees of freedom in the tibiofemoral joint allowed for a more accurate estimation of the secondary joint kinematics during a lunge compared to simpler models [7], despite a much larger computational cost.

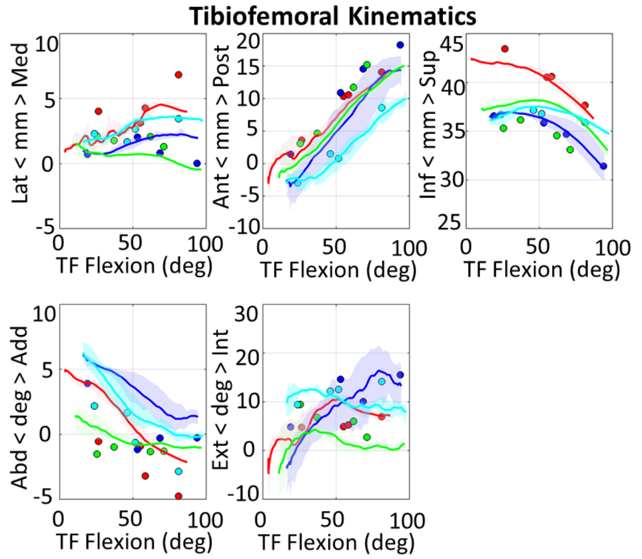

Fig 3. Tibiofemoral kinematics. Points represent the experimental measures and lines the predicted results from the simulation. Above: left, medio-lateral, middle, antero-posterior, and right, superior-inferior translations. Below: left, abduction-adduction, and middle, internal-external rotations. Flexion-extension is driven by the MKO.

CONCLUSIONS

In conclusion, we have developed subject-specific multibody MS lower limb and natural knee models, capable of simultaneously simulating internal tibiofemoral and patellofemoral secondary joint kinematics and forces. We have evaluated our model estimations against EOS[R] experimental data from the same subjects and found good agreement in tibiofemoral secondary joint kinematics and patellofemoral translations. The proposed modeling framework provides a powerful tool to simulate individualized knee mechanics and optimize clinical treatments.

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