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A VIRTUAL CRANKSHAFT THIGH MODEL TO ESTIMATE TIBIAL-FEMORAL TRANSVERSE PLANE KINEMATICS

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Sports injuries often require a thorough evaluation of the knee that includes transverse plane measurements, which are difficult to measure accurately using motion capture. We have developed a method to estimate thigh position modelling the lower limb as a modified slider-crank mechanism. Our model does not rely on cutaneous thigh markers; its motion is defined by a functionally determined hip joint center and constrained distally to the tibial plateau. Motion capture was used to acquire normal gait and counter-movement jump data from three unimpaired subjects. The transverse plane translations and rotation along with frontal plane rotation estimated by our model were shown to be reflective of those reported in literature. Our slider-crank model of the pelvis-femur-tibia complex has been demonstrated to perform well in both low and high impact motions.

KEY WORDS: marker set, skin movement artifact, tibial translation, knee, model.

INTRODUCTION: Knee injuries have been a focal point of sports-related biomechanics for several decades now. The treatment for such injuries, whether it be neuromuscular training or surgical intervention, often requires a biomechanical evaluation of the tibial-femoral joint. Such evaluation in a clinical setup is often used for the purpose of diagnosis to prescribe a treatment, surgery, or rehabilitation program. However, medical professionals rarely consider the accuracy and reliability of such data. The importance of transverse plane translations and rotations of the tibia with respect to the femur has been highlighted as an area of interest for both the prevention and post-treatment assessment of ACL traumas (Markolf et al., 1995). The challenge that arises is the accuracy of the measured data; specifically, the ability of marker-based motion capture to estimate underlying bone position. It has been shown that skin movement artifacts have a significant impact on experimental data and can introduce large errors of similar magnitude to tibial plateau translations and rotations on transverse plane (Cappozzo et al., 1996, Manal et al., 2003, Benoit et al., 2006).

Although solutions have been suggested, estimation of tibial-femoral translations and rotations are most often limited to the sagittal plane. The use of clusters to track segment position has been shown to reduce skin movement artifact; however, it does not necessarily solve the problem completely (Manal et al., 2003, Benoit et al., 2006). Currently, it has become commonplace to employ a method of global optimization with joint constraints, often referred to as inverse kinematics. This approach can be very beneficial for a joint possessing simple kinematics such as the hip. It is reasonable to model the hip as a ball-in-socket joint and therefore constrain it to rotate about each axis. Such a constraint would consider any translations measured by marker motion as an artifact (Lu & O'Connor, 1995). Unfortunately, since knee kinematics include rotations and translations in all three directions, applying inverse kinematic constraints to the knee would treat actual knee motion as an artifact. For this reason, inverse kinematics is not a viable option to reduce skin movement artifact when evaluating transverse plane knee kinematics.

We have developed the following model to address the issue of tibial-femoral translations and internal-external (IE) rotation estimation. Using classic mechanical kinematics, we treated the femur-tibia complex as a slider-crank mechanism. The advantage of this method is its ability to avoid tracking the motion of the thigh segment directly by estimating its position based on the motion of the pelvis and shank. The primary reason to employ this approach is that the large amount of soft tissue found in the thigh could result in significant marker artifact. Using primarily bony prominences, the focus of our model has been to reduce the impact of skin motion on transverse plane knee kinematics. The purpose of this study is to compare the slider-crank mechanism to *in vivo* kinematics data.

METHODS: Three unimpaired subjects (28±2.7 yo, 73±11.4 kg, 1.77±0.12 m) were tested at the Let People Move Biomechanical Laboratory, Perugia, Italy. Two types of trials were performed: normal gait and a counter movement jump (CMJ) recorded from propulsion to landing. The motion data was captured using a SMART-D 12 camera system (BTS Bioengineering, Padova, Italy). Cutaneous anatomical markers were placed on the anterior and posterior superior iliac spine (ASIS and PSIS), sacrum, greater trochanters (GT), medial and lateral epicondyles (MEP and LEP), and medial and lateral malleoli (MMAL, LMAL). Furthermore, four cutaneous tracking markers were placed on each segment: thigh, shank, and foot. Of these, it was ensured that one marker for each segment was placed in accordance with a modified Plug-in-Gait method (PiG) (Beaulieu et al., 2010). For the shank, one tracking marker was placed on the tibial tuberosity and one on the distal bony prominence of the tibia slightly medially to reduce the impact of the tibialis anterior tendon (Benoit et al., 2006). Joint centers for the knee and ankle were defined as half the distance between MEP and LEP, and half the distance between MMAL and LMAL respectively. Three different methods were used to define the hip joint center: using anthropoligical data to determine the location with respect to the ASIS and PSIS, measuring one-quarter the distance between the GTs, and using a functional method in Visual 3D (C-Motion, Germantown, MD).
We defined four marker sets in Visual 3D: a variant of PiG (Beaulieu et al., 2010) which used the anthropological hip joint center, a typical cluster approach (Cluster) (Cappozzo et al., 1995) which used the hip center from the GTs, and two in-house variants of the cluster method. For both of our slider-crank models, the hip joint center was defined using a functional dynamic trial (Schwartz, 2005): gait for the SCG model, and squat to 90° for the SCS model. The theory behind our approach was that defining the hip joint center (main bearing), rotation plateau (crank bearing), length of the femur (crankshaft), and position of the tibia (piston) would make it possible to determine the center of rotation. The crank bearing was modelled as a two-dimensional surface orthognal to the tibia, therefore representing the rotation plateau which is parallel to the tibial plateau but passing through the statically measured position of the epicondyles. The complexities of using a three-dimensional model required us to provide a constraint to ensure a unique solution; this was accomplished by requiring the axis of the virtual femur to pass through the tibial longitudinal axis. Finally, to model the effect of ligaments, the orientation of the femur was defined by a

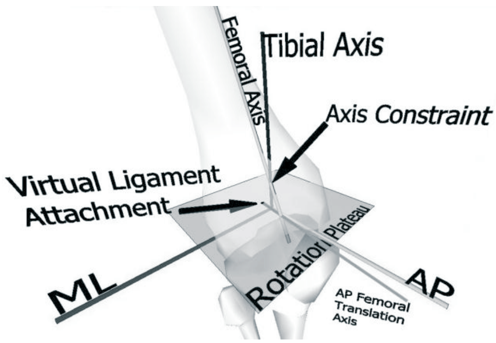

Figure 1: A graphic representation of the femur-tibia model constraints. The femoral axis was forced to cross the tibial axis, and the virtual ligament attachment was fixed to the tibial segment and used as a posterior orientation reference.

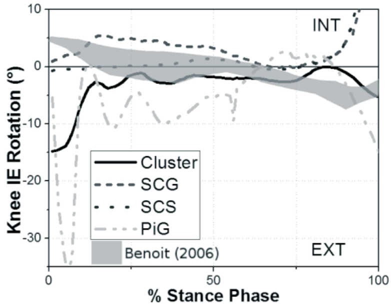

Figure 2: Comparison of average internal external (IE) knee joint angles during normal walking trials across all subjects. Shaded area representative of bone-pin data (Benoit, 2006)

landmark posterior to the knee joint center by one-quarter the depth of the shank on the tibial plateau (See Figure 1 for model constraints detail).

RESULTS: Transverse plane kinematics of the tibia with respect to the femur were calculated. In Figure 2, IE data are depicted for the modified PiG and Cluster models, as well as for the SCG and SCS models. Figure 3 presents tibial translation data of one subject during normal walking and countermovement jump trials; similar results were found for the other subjects as well. We also observed that during the last 10% of the stance phase, our models estimated significantly different rotations and translations than traditional models. Finally, it is important to note peak transverse plane data during the CMJ motion, in which the Cluster and and PiG marker sets show translations laterally of 10 mm and posteriorly of 20 mm, whereas our models both have peaks of 5 mm medially and 10-15 mm posteriorly.

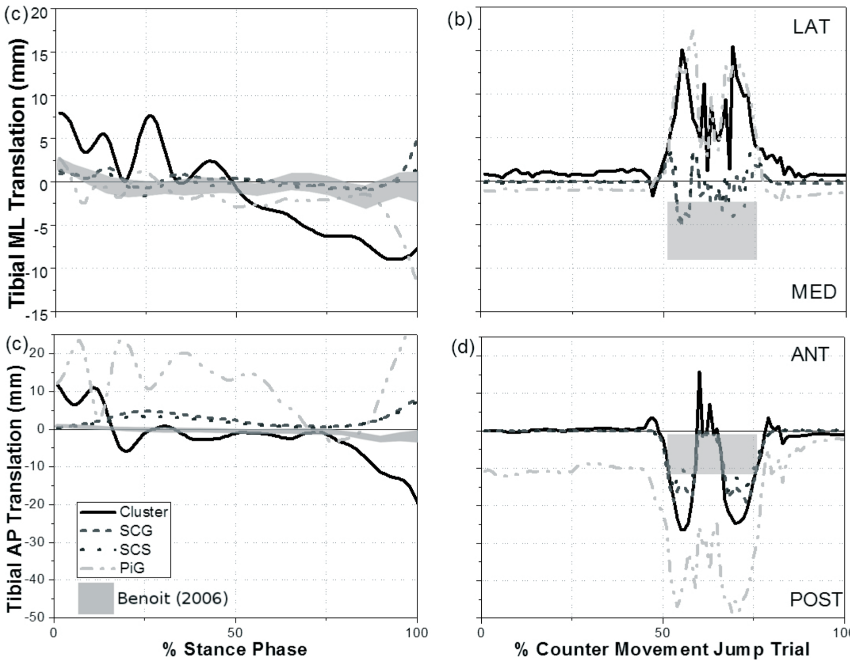

Figure 3: Model estimation comparison of medial-lateral (ML) (a, b) and anterior-posterior (AP) (c, d) translations during normal walking (a, c) and countermovement jump motions (b, d). Data shown for representative subject. Shaded area representative of bone-pin data (Benoit, 2006) adjusted by an offset to compensate for discrepancies in tibia center locations (for jump trials only range of cut data is shown).

DISCUSSION: The data estimated by our models were found to be consistent with previously reported bone-pin studies (Benoit et al., 2006). Of all the models discussed in this paper, the SCS model performed the most consistently in line with the literature in IE rotation, ML translation, and AP translation. The observed discrepancies between the literature and the four models tested for gait tibial translation are most likely due to knee center location offset between the studies by approximately 10 mm anteriorly and 2 mm laterally, which is supported by the discrepancies in Cluster data in the two studies. Differences found in the last 10% of the normal gait stance phase are attributed to a shortcoming of our models resulting from the combination of full hip extension and high knee flexion. The CMJ tibial translations are particularly interesting as they show clear differences between our models and traditional approaches. Benoit et al. reported translations during a cutting maneuver, which is high impact and kinematically similar to a CMJ. Figures 3b and 3d present the similarity between reported cutting maneuver peaks and peaks of our models during the CMJ. Of significant importance is the minimal variation of tibial translations throughout both the gait stance phase and the CMJ—data which were not replicated by PiG or Cluster models but are consistent with bone-pin data.

CONCLUSION: We have presented a novel approach for modeling femur motion that is based on a slider-crank mechanism intended to reduce the impact of skin movement artifact. Our results show good performance of our model on the transverse plane in both normal walking and high impact tasks. Future studies should improve model performance on the latter end of the gait stance phase and expand the model to include varus-valgus rotations. Regardless, this feasibility study demonstrates good performance in comparison to the literature and offers possibilities for more accurate knee kinematic evaluation.

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