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Integrating dynamic stereo-radiography and surface-based motion data for subject-specific musculoskeletal dynamic modeling

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Abstract

This paper presents a new subject-specific musculoskeletal dynamic modeling approach that integrates high-accuracy dynamic stereo-radiography (DSX) joint kinematics and surface-based full-body motion data. We illustrate this approach by building a model in OpenSim for a patient who participated in a meniscus transplantation efficacy study, incorporating DSX data of the tibiofemoral joint kinematics. We compared this DSX-incorporated (DSXI) model to a default OpenSim model built using surface-measured data alone. The architectures and parameters of the two models were identical, while the differences in (time-averaged) tibiofemoral kinematics were of the order of magnitude of 10° in rotation and 10 mm in translation. Model-predicted tibiofemoral compressive forces and knee muscle activations were compared against literature data acquired from instrumented total knee replacement components (Fregly et al., 2012) and the patient's EMG recording. The comparison demonstrated that the incorporation of DSX data improves the veracity of musculoskeletal dynamic modeling.

Keywords

Musculoskeletal Dynamic Model; Dynamic Stereo-Radiography; Integration; Tibiofemoral Kinematics; OpenSim

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Conflicts of interest statement

The authors have no conflict of interest related to the manuscript or the work it describes.

1. Introduction

Musculoskeletal modeling frameworks such as OpenSim (Delp et al., 2007) and AnyBody (Damsgaard et al., 2006) enable in-depth mechanistic understanding of normal or pathological movements. These musculoskeletal modeling tools require specification of skeletal kinematics and external loads to create subject-specific computer simulations. Skeletal kinematics data have traditionally been obtained with surface-based motion capture systems. These systems have many advantages such as being non-invasive, easy to use, and relatively inexpensive. However, the accuracy of surface-measured kinematic data is compromised by soft tissue artifacts (STAs). The magnitudes of STAs, as assessed using external fixation devices or radiographic imaging, can reach more than 15 mm at bony landmarks (Maslen and Ackland, 1994; Sati et al., 1996) and up to 40 mm on the thigh (Barre et al., 2013; Cappozzo et al., 1996; Tsai et al., 2009; Tsai et al., 2011).

The past decade has witnessed the increasing use of dynamic stereo-radiography (DSX) and similar (e.g., bi-planar fluoroscopy) systems for measuring *in vivo* three-dimensional (3D) joint kinematics (Berthonnaud et al., 2005; Brainerd et al., 2010; Hanson et al., 2006; Tashman and Anderst, 2003). A DSX system, for example, is capable of measuring joint kinematics with static accuracy of ± 0.2 mm in translation and ± 0.2 degree in rotation and dynamic accuracy of ± 0.4 mm in translation and ± 0.6 degree in rotation (Anderst et al., 2009). The differences between surface-based model-derived and DSX-measured kinematics have been found to be substantial—the overall mean ($\pm$ SD) RMS differences were up to 9.1 ± 3.2 degrees in rotation and 8.8 ± 3.7 mm in translation for tibiofemoral kinematics during running (Li et al., 2012). However, DSX and similar systems currently can image only one small body region or single joint at a time due to their limited fields of view, and only for a short duration due to radiation exposure restrictions.

This work aimed to establish an approach that incorporates *in vivo* joint kinematics obtained from a DSX system and full-body kinematics from a surface-based motion capture system to create subject-specific musculoskeletal dynamic models, and to evaluate the resulting model veracity in terms of joint force and muscle activation predictions.

2. Methods

We modeled a patient with subtotal left lateral meniscectomy (30-year-old male, height 187 cm, mass 112 kg) who participated in an Institutional Review Board (IRB) approved experimental study of meniscus transplantation efficacy. The patient performed a static upright standing trial and gait trials (at 1.0 m/s) on a dual-belt instrumented treadmill. A customized DSX system imaged the tibiofemoral motion of the meniscectomized knee in one gait trial and the intact knee in another. An eight-camera motion capture system (Vicon-MX, Oxford, UK) measured the full-body motion with a set of retro-reflective spherical surface markers (1 cm diameter) placed according to the Plug-in-Gait marker set protocol (Davis et al., 1991). The sampling frequency for both systems was set at 100 Hz. The ground reaction forces (GRFs) were measured at 1000 Hz by two force plates (Bertec Corporation, Columbus, OH) embedded in the treadmill. Electromyography (EMG) data for seven lower limb muscles—vastus medialis, rectus femoris, vastus lateralis, biceps femoris,

semimembranosus, tibialis anterior, and medial gastrocnemius—were collected at 1000 Hz using a wireless EMG system (ZW180, Zero Wire, Milano, Italy). The recorded data across different systems were synchronized using a precision pulse generator (Model 565, Berkeley Nucleonics Corporation, San Rafael, CA). High-resolution CT scans (slice spacing: 0.625 mm) of both knees were also collected.

A volumetric model-based tracking process determined 3D tibiofemoral kinematics with sub-millimeter accuracy using recorded DSX images and CT-acquired bone models (Anderst et al., 2009). The recorded surface-marker data, GRFs, and EMG were processed and prepared for subsequent modeling steps using a “Gait Extract Toolbox” (Dorn, 2008): the GRFs were low-pass filtered at 20 Hz, and EMG data were high-pass filtered at 20 Hz, rectified and low-pass filtered at 5 Hz.

Two distinct models, a default model and a DSX-incorporated (DSXI) model, were developed in OpenSim (Figure 1). The default model was based on the latest generic OpenSim model (Arnold et al., 2010) and made subject-specific by employing the surface-based kinematics data alone. The tibiofemoral joint was modeled as a 1-DOF joint: two rotations (external-internal and abduction-adduction) and three translations (anterior-posterior, lateral-medial and proximal-distal) between femur and tibia were constrained by cubic spline functions of flexion-extension knee angles based on literature data (Walker et al., 1988). The DSXI model was based on the same model by Arnold et al. (2010) but integrated the DSX-measured tibiofemoral kinematics with the surface-based whole-body kinematics: the tibiofemoral joint was defined as a 5-DOF joint (external-internal rotation, abduction-adduction, flexion-extension, and anterior-posterior and lateral-medial translations were independent); the proximal-distal translation was specified as a cubic spline function of the knee flexion-extension angle based on DSX-measured kinematics. We chose to model the tibiofemoral joint as a 5-DOF joint instead of a 6-DOF joint, which would have required inclusion of the knee ligaments as restraints. The chosen approach greatly reduced the computational cost and avoided introducing more modeling variables or unknowns (e.g., ligament force-length properties). The constraint of the proximal-distal translation as a function of flexion-extension accounted for the effect of ligamentous constraints without explicitly modeling the ligaments. The proximal-distal translation DOF was chosen, given that the residual force in that direction would otherwise be relatively large.

The local coordinate systems (CS) on the femur and tibia in the two models were unified into a common anatomical knee CS (see Fig. 1) defined based on consistently identifiable anatomical landmarks and embedded in the DSX data/model (Tashman et al., 2004). In doing so, the literature-based knee kinematic constraints in the default model remained unchanged in the common anatomical knee CS, while functions describing the constraints were transformed.

For both models, the standard OpenSim procedures, including scaling, inverse kinematics (IK), residual reduction algorithm (RRA), and computed muscle control (CMC) algorithm (Delp et al., 2007; Thelen et al., 2003), were applied to create dynamic simulation of the gait motion. The implementation of the DSXI model required modification of the scaling and IK

procedures that determined the joint kinematics input to the later dynamic simulation and prediction steps (i.e., RRA, CMC and Joint Reaction analysis). In the scaling procedure, both models were scaled according to the patient's anthropometric measurements and the surface marker data from the static standing trial; in the DSXI model, the DSX-measured knee position in the static trial was set as the neutral posture. In the IK procedure, the default model derived the joint angles of the entire body by matching the model virtual marker motions with the measured surface marker motions, whereas the DSXI model incorporated both surface-measured kinematics and DSX-measured knee kinematics—the knee joint angles were forced to equal DSX measurements while the remaining joints angles were determined such that the virtual marker motions best matched the measurements (Lu and O'Connor, 1999). The compressive tibiofemoral force was calculated using the Joint Reaction analysis in OpenSim (Steele et al., 2012).

The EMG data recorded simultaneously with the kinematics were used to compare the muscle activation predictions by the models. To estimate the muscle activations, EMG data were normalized according to the minimum and maximum values over multiple repeated gait cycles (the average $\pm$ one standard deviation). The tibiofemoral *in vivo* compressive load data measured from an instrumented total knee replacement (TKR, age: 83 years, weight: 64.6 kg, height: 166 cm) publicly available (<https://simtk.org/home/kneeloads>) (Fregly et al., 2012) were used for comparing the joint compressive force predictions by the two models (Figure 2).

3. Results

The tibiofemoral compressive forces predicted by the DSXI model were in closer agreement with *in vivo* TKR measurements than the default model (Figure 2). The maximum compressive forces during gait predicted by the DSXI model were 2.2 times body weight (BW) in both the meniscectomized and intact knees; the maximum compressive forces predicted by the default model were 3.8 times BW in the meniscectomized knee and 4.0 times BW in the intact knee. The default model not only over-estimated the tibiofemoral compressive forces, but also seemed to amplify the difference between the intact and injured knees (Figure 2).

The muscle activations predicted by both models were generally consistent with the muscle activation patterns estimated from the EMG data recorded during gait (Figure 3). The activations of quadriceps (e.g., rectus femoris) and hamstrings (e.g., biceps femoris) from the DSXI model had better matched onset timing and magnitudes at the heel strike.

There were substantial differences between the default and DSXI models in tibiofemoral kinematics resulting from the IK procedure (Figure 4). The time-averaged differences in flexion-extension, external-internal rotation, and abduction-adduction were $10.0 \pm 1.7^\circ$, $3.6 \pm 1.7^\circ$, and $2.2 \pm 0.3^\circ$ for the meniscectomized knee, and $12.1 \pm 2.8^\circ$, $9.2 \pm 1.3^\circ$, and $3.5 \pm 0.4^\circ$ for the intact knee; the differences in lateral-medial (LM), proximal-distal (PD), and anterior-posterior (AP) translations were 1.2 ± 0.3 mm, 13.5 ± 0.3 mm, and 1.5 ± 1.1 mm for the meniscectomized knee, and 0.3 ± 0.2 mm, 14.5 ± 0.5 mm, and 2.8 ± 1.3 mm for the intact knee.

4. Discussion

We demonstrated the feasibility of ‘seamless’ integration of DSX-measured tibiofemoral kinematics and surface-based whole-body kinematics for building subject-specific musculoskeletal dynamic models. The incorporation of more accurate tibiofemoral kinematics led to better predictions of tibiofemoral compressive forces and activation patterns for some of the muscles involved.

In vivo measurement of tibiofemoral contact forces via instrumented TKR components is the best available means to validate joint forces predicted by musculoskeletal models (Fregly et al., 2012), notwithstanding the limitation that the data were acquired from old TKR patients. In the present study, the model-predicted peak tibiofemoral compressive force was about 2.2 times BW for both knees in the DSXI model during level treadmill walking at 2.2 miles per hour. This was in close agreement with the reported peak force of instrumented TKR patients: 2.1 ± 0.2 times BW during level treadmill walking at 1-3 miles per hour (D’Lima et al., 2012). In contrast, the default model overestimated the tibiofemoral compressive force (3.8 or 4.0 times BW) as in many previous modeling attempts that predicted peak forces to be in a range from 1.8 to 8.1 times BW, mostly in a range from 3.0 to 3.5 times BW (Fregly et al., 2012). The comparison suggests that increased accuracy in skeletal kinematics, even limited to only the tibiofemoral joint, can improve the validity of rigid-body inverse dynamics prediction at least for that joint. It remains unknown and would be interesting to find out whether this ‘local effect’ percolates to other joints. Noteworthy also is that the predicted tibiofemoral compressive force was lower in the meniscectomized knee compared to the contralateral healthy knee during most of the gait, indicative of a compensatory mechanism to unload the affected limb (DeVita et al., 1998; Netravali et al., 2010). This was confirmed by the lower GRF and lower knee flexion moment in the meniscectomized limb — the reductions in peak GRF and peak moment were about 3% of BW and 2% of body weight times height (BW×H), respectively.

The somewhat underwhelming improvement in muscle activation prediction by the DSXI model over the default model may reflect several limitations of the optimization-based muscle force distribution algorithm employed. First, its inability to predict co-contraction by the antagonist muscles (Hughes et al., 1995) makes it more likely to equalize the two models. Alternative algorithms or ‘hybrid’ approaches (Amarantini et al., 2010; Cholewicki and McGill, 1994; Li et al., 1999) that can account for co-contraction and are computationally viable may be explored. Second, generic “average” muscle geometry (e.g., volume and attachment sites) and property data, which serve as the weighting factors in resolving the muscle force redundancy, are not sensitive enough to the skeletal kinematics difference to truly distinguish the musculoskeletal dynamics between the two models. Third, it remains unknown whether the optimality assumption or the same optimal criterion (minimal sum of squares of muscle activations) would still hold for joints with pathology (Fregly et al., 2012; Steele et al., 2012). This could inflate the uncertainty in model predictions involving an injured or diseased joint, thus increasing the potential to mask the true differences between models.

The discrepancies between the DSX-measured and surface-measured tibiofemoral kinematics in the present study are of a magnitude comparable to what we previously reported for cruciate ligament-deficient subjects during running and stair ascent activities (Li et al., 2012). This consistent order of magnitude (10° in rotation and 10 mm in translation) may be a typical level of inaccuracy to expect from surface-based (i.e., default) musculoskeletal dynamic models in general. This will be continually verified in our intended future endeavors to apply the integrative methodology described here to a wide variety of tasks and conditions.

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Figure 1. Flowchart comparing procedures for constructing the two subject-specific musculoskeletal dynamic models in OpenSim.

Figure 2.

Tibiofemoral compressive forces predicted by the two models, compared to instrumented TKR force measurements (Fregly et al., 2012).

Figure 3.

Muscle activation patterns for both knees predicted by the two models, compared to estimates based on EMG measurements.

Figure 4.

Tibiofemoral kinematics resulting from the IK procedures for the two models. Note that external-internal rotation and medial-lateral translation were constants for the default model.