Medical Engineering and Physics 000 (2017) 1–10

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Dynamic simulation of knee-joint loading during gait using force-feedback control and surrogate contact modelling

Jonathan P. Walter*, Marcus G. Pandy

Department of Mechanical Engineering, University of Melbourne, VIC 3010, Australia

A R T I C L E I N F O

Article history:
Received 24 January 2017
Revised 31 May 2017
Accepted 25 June 2017
Available online xxx

Keywords:
Knee model
Musculoskeletal simulation
Forward dynamics
CMC
Static optimization

A B S T R A C T

The aim of this study was to perform multi-body, muscle-driven, forward-dynamics simulations of human gait using a 6-degree-of-freedom (6-DOF) model of the knee in tandem with a surrogate model of articular contact and force control. A forward-dynamics simulation incorporating position, velocity and contact force-feedback control (FFC) was used to track full-body motion capture data recorded for multiple trials of level walking and stair descent performed by two individuals with instrumented knee implants. Tibiofemoral contact force errors for FFC were compared against those obtained from a standard computed muscle control algorithm (CMC) with a 6-DOF knee contact model (CMC6); CMC with a 1-DOF translating hinge-knee model (CMC1); and static optimization with a 1-DOF translating hinge-knee model (SO). Tibiofemoral joint loads predicted by FFC and CMC6 were comparable for level walking, however FFC produced more accurate results for stair descent. SO yielded reasonable predictions of joint contact loading for level walking but significant differences between model and experiment were observed for stair descent. CMC1 produced the least accurate predictions of tibiofemoral contact loads for both tasks. Our findings suggest that reliable estimates of knee-joint loading may be obtained by incorporating position, velocity and force-feedback control with a multi-DOF model of joint contact in a forward-dynamics simulation of gait.

1. Introduction

Accurate knowledge of muscle and joint loading during human locomotion is critical for improving the diagnosis of musculoskeletal conditions such as knee osteoarthritis and for evaluating the effects of implant design and surgical technique on the functional performance of total joint replacements. Direct measurement of muscle and articular joint contact forces is infeasible thus computational modelling is used in conjunction with gait analysis techniques to evaluate these quantities non-invasively. Tibiofemoral contact forces during gait are usually determined by applying a sequential modelling approach: a 1-degree-of-freedom (1-DOF) translating hinge-knee model is incorporated in a full-body model to calculate lower-limb muscle forces, and these results are then applied to a more complex multi-DOF knee model to determine contact forces at the tibiofemoral joint [1–4]. There are two main limitations associated with this approach. First, only the knee flexion–extension moment is considered in the calculation of muscle forces when the knee is modelled as a 1-DOF hinge joint. This simplification may lead to erroneous results as estimates of knee muscle forces are also influenced by the forces and moments (henceforth referred to as ‘loads’) acting in the frontal and transverse planes as well as the joint contact forces transmitted by the articulating surfaces of the femur and tibia [5,6]. Second, muscle and joint contact forces are evaluated independently in the sequential modelling approach in spite of the fact that these quantities are directly related to one another.

Muscle and joint contact forces can be calculated simultaneously using a forward-dynamics simulation approach [7–10]. Thelen et al. [10] used standard joint position and velocity control in a computed muscle control (CMC) algorithm to calculate the time histories of muscle excitations needed to track hip, knee and ankle joint angular displacements measured for five trials of level walking in one subject. Muscle forces were determined by minimizing an activation-based performance criterion while knee contact loads were simulated using an elastic foundation model of articular contact. They reported model-predicted tibiofemoral contact forces that were in close agreement with direct measurements obtained from an instrumented knee implant. Guess et al. [7] performed a whole-body forward-dynamics simulation of walking and calculated tibiofemoral contact forces by assuming rigid-body contact and applying Hertzian contact theory. Rather than using optimization theory to resolve the mechanically

* Corresponding author.
E-mail addresses: jwalter@cedtechnologies.com, walterj@unimelb.edu.au (J.P. Walter).

http://dx.doi.org/10.1016/j.medengphy.2017.06.043
1350-4533/

Table 1
Details of subjects participating in this study and of the tasks investigated.

	Walk speed (m/s)	Stair descent speed (m/s)	Age	Height (m)	Mass (kg)	Gender
Participant 1 (4th Competition)	1.3 (5 trials)	0.4 (2 trials)	80	1.7	68	Male
Participant 2 (3rd Competition)	1.2 (5 trials)	–	69	1.7	78	Female

redundant nature of the musculoskeletal system, these authors used a proportional-integral-derivative (PID) feedback control scheme to drive the model-computed joint angles and muscle-tendon lengths to a set of desired kinematics.

Multi-body contact problems can be solved efficiently using a method of interpolation called ‘Kriging’, which converts a computationally expensive elastic-foundation (bed-of-springs) model into a computationally cheap surrogate contact model [11,12]. Lin et al. [11] used a 3D surrogate contact model of the knee joint in conjunction with optimization theory to calculate muscle and joint contact forces simultaneously for one cycle of total knee arthroplasty (TKA) gait. Surrogate modelling enabled knee-joint contact mechanics to be simulated ~50 times faster than an elastic foundation model of equivalent complexity. No study to our knowledge has combined surrogate contact modelling with a forward-dynamics simulation approach to calculate muscle and joint contact forces during gait.

In a previous study we quantified the contributions of tibiofemoral joint contact loads to the net knee loads calculated from inverse dynamics for multiple subjects and multiple gait patterns [13]. We found that tibiofemoral contact loads contributed substantially to the net knee extension and knee adduction moments as well as to the total superior–inferior tibiofemoral force present during normal walking. We also developed generalized linear relationships between the measured joint contact and net loads at the knee for multiple subjects. The overall goal of the present study was to extend this work by using these experimentally-derived relationships at the knee to perform muscle-driven forward-dynamics simulations of gait, specifically, level walking and stair ambulation. Our specific aims were firstly, to calculate muscle and joint contact forces using a 6-DOF surrogate contact model of the knee in tandem with a forward-dynamics simulation method that incorporates position, velocity, and force-feedback control; and second, to compare these results against those obtained from a standard CMC algorithm with a 6-DOF knee contact model; CMC with a 1-DOF translating hinge-knee model; and static optimization with a 1-DOF translating hinge-knee model. We hypothesized that force-feedback control with generalized knee load relationships would yield more accurate estimates of knee loading during gait than those derived from either standard position and velocity control or static optimization.

2. Methods

2.1. Gait experiments

Experimental data obtained from the Third and Fourth “Grand Challenge Competitions to Predict In Vivo Knee Loads” [14,15] were used in this study. The data obtained were collected from two participants implanted with load-measuring knee replacements walking overground at their self-selected speeds. One of these participants also walked down a staircase at their self-selected speed (see Table 1). Body motion, ground reaction forces and tibiofemoral contact loads were measured simultaneously for each task. Institutional review board approval and informed consent were obtained prior to data collection. The tibial implants were custom tibial prostheses equipped with force transducers that measured dynamic loads applied to the tibial tray [16].

2.2. Musculoskeletal model

A generic musculoskeletal model was modified and used to simulate level walking and stair descent [17,18]. The lower-limb was represented as a 5-segment, 17-degree-of-freedom skeletal linkage actuated by 43 muscles. Segments representing the foot, shank and thigh segments, as well as a single segment representing the pelvis, torso and head combined, were scaled using participant-specific scale factors calculated from anatomical marker locations and CT images. The femur and tibia in the generic model were personalized to each participant by scaling and aligning the generic bone geometries to participant-specific geometries created from CT scans of the participant’s lower limbs [13,19]. The long axis and proximal end of the CT-based bone geometry were aligned to the long axis and proximal end of the generic bone geometry. A scale factor was then applied to the generic bone to ensure the proximal ends of the generic and CT-based bones were well aligned. The models of the hip, ankle and subtalar joints were identical with those represented in the generic model. The knee was modelled as a 6-DOF tibio-femoral joint by adding internal–external rotation, abduction–adduction rotation, superior–inferior translation, medial–lateral translation, and anterior–posterior translation to the 1-DOF translating hinge knee represented in the generic model [19,20]. Motion of the patella was represented as a function of the knee flexion–extension angle. The geometry and mechanical properties of the medial collateral ligament, lateral collateral ligament, and posterior cruciate ligament were included in the model. Attachment sites, reference strains, stiffness parameters, and nonlinear force-strain relationships were based on data reported by Shelburne et al. [2]. Ligament reference strains were then iteratively adjusted using Newton’s method until the peak values of ligament forces and strains calculated in the model for passive knee flexion matched corresponding data reported by Shelburne et al. [2].

2.3. Tibiofemoral joint contact model

A two-step procedure was used to model tibiofemoral joint contact during dynamic activity. First, an elastic-foundation contact model was created to calculate the tibiofemoral contact loads (3 forces and 3 moments) and the medial and lateral compartment forces as a function of tibiofemoral kinematics [12,21,22]. The elastic-foundation contact model was used to determine the interpenetration between the articular surfaces of the proximal tibia and distal femur and to calculate the corresponding contact forces. The geometry of the articulating implant surfaces were obtained from the Grand Challenge datasets [15]. Values for the stiffness and Poisson’s ratio of each articular contact surface were based on data reported by Zhao et al. [23]. Second, a surrogate contact model was developed and used as a substitute for the elastic-foundation model when simulating articular contact in the medial and lateral compartments of the tibiofemoral joint [11,12]. The surrogate contact model was created by fitting a Kriging model to a large set of sampling inputs (tibiofemoral joint kinematics) and the corresponding outputs (tibiofemoral contact loads) of the elastic-foundation contact model (see Lin et al. [11,12] for details). The surrogate contact model was used in conjunction with the 6-DOF tibiofemoral joint model to simulate articular contact loading during level walking and stair descent.