

Dynamic Simulation of Isokinetic Knee Movements

Alan R. Morris^{□□□}, Stephen Naumann^{□□}, Tracy Lee[□], Gabriele D'Eleuterio[□], and John Wedge[□]

[□].Bloorview Macmillan Children's Centre

[□].Institute for Biomaterials and Biomedical Engineering, University of Toronto

[□].Institute for Aerospace Studies, University of Toronto

[□].Mechanical Engineering Department, University of Toronto

[□].Department of Surgery, University of Toronto

Introduction

The human body's musculoskeletal system is a complex series of passive and active soft-tissue, bones and articulating surfaces. The ability to predict a change in function following an orthopaedic procedure on any single muscle is complicated by muscular redundancy and the changing lever (moment) arms of each muscle. Cerebral palsy is a non-progressive motor disorder that manifests itself through atypical muscle recruitment profiles, hyper-reflexivity (spasticity), and reduced antagonistic inhibition. Individuals that are capable of ambulation exhibit reduced joint ranges-of-motion, greater variability of movement, and reduced range of gait speeds.

Our long-term research goal is to develop post-operative simulations of functional movements to assist in optimizing the choice of orthopaedic surgical interventions, when appropriate, on children with spastic-diplegic cerebral palsy. The framework for developing our predictive model is the development of a forward-dynamic model of the lower-leg, using electromyographic (EMG) data to drive 10 Hill-type dynamic muscle models geometrically mapped onto the various bones of the leg. As a first step, we have chosen to experimentally record and simulate isolated isokinetic knee movements on a functional testing dynamometer as our first step. In addition to the musculoskeletal model, control and inertial models of the Biodex testing dynamometer were also developed to implement the system's PID control of velocity during isokinetic testing.

Method

EMG muscle activities and dynamometer kinematics/kinetics were recorded during isokinetic knee flexion-extension exercises on a commercial (Biodex System-2) testing dynamometer. A unilateral musculoskeletal model was developed in Simulink v4.1 consisting of inertially modelled limb segments, anatomical and passive joint constraints, musculotendon lines-of-action, and Hill-type dynamic muscle models. The link-segment model was constrained such that only the knee and ankle joints were free to rotate.

Experimental Recording

An experimental protocol consisted of performing repeated isokinetic knee flexion-extension exercises on a Biodex System-2 dynamometer, that were a combination of concentric (CON) and eccentric (ECC) contractions, ranging in speed from 30-240 degrees/second. Four conditions were tested (CON-CON, CON-ECC, ECC-CON, ECC-ECC), with five extension-flexion repetitions per trial, and each were carried out at four different velocities (30, 90, 150, and 240 degrees/second). The movements were restricted to the same maximum angular velocity in both directions. The tests were carried out in an upright seated position with knee motion ranging between 0 and 90 degrees flexion. EMG signals were recorded from ten surface-lying muscles acting about the knee joint: tensor fascia latae, vastus lateralis, vastus medialis, rectus femoris, biceps femoris longus, semitendinosus, semimembranosus, lateral gastrocnemius, and medial gastrocnemius. The EMG signals were differentially amplified using a Bortec AMT-8 system (frequency response: 10-2000Hz, CMRR: 115dB, Gain: 500, Impedance: 10Gohm) and recorded at 2000Hz using a National Instruments DAQPad-6020E data acquisition board and using MATLAB software (Mathworks Inc.). Joint moment, angle, and angular velocity were simultaneously recorded from the Biodex system. To measure actual knee angle, a single-axis fibre-optic goniometer (accuracy ± 0.5 degrees) (S710, Measurand Inc.) was placed about the knee joint. EMG were digitally rectified and averaged using a 4th order zero-lag Butterworth filter with a 6Hz frequency cutoff. Biodex angle, angular velocity, joint moment, and goniometer signals were filtered using a 4th order generalized cross-validation (spline) filter (Woltring, 1986) with a cutoff frequency of 6Hz.

Limb Segments

The 4-segment model consisted of the torso, femur, tibia, and foot segments. Inertial estimates of torso centre-of-mass (COM) and moment of inertia (MI) were derived from regression equations (Yeadon and Morlock, 1989). The mass of the thigh and shank segments were estimated from regression equations (Jensen, 1989), and the COM and

MI were calculated through modelling the segments as frusta (Apkarian et al., 1989). Foot segment mass was estimated from regression equations (Jensen, 1989) while the COM and MI were calculated through modelling the segment as a cylinder (Meriam and Kraige, 1987).

The equations of motion for the multibody chain were implemented using SimMechanics Blockset (Mathworks Inc.) using Featherstone's Articulated-Body Method, a recursive Newtonian formulation (Featherstone, 1983).

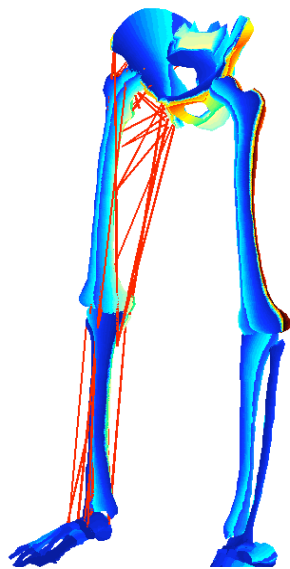
Anatomical Joints and Passive (Ligamentous) Constraints

The hip was constrained in place with the thigh flexed 90 degrees from the vertical. The knee joint was modelled as a single degree-of-freedom (DOF) polycentric joint where the shank displacement was a defined function of shank angle (Yamaguchi and Zajac, 1989; Delp, 1990). The patella was kinematically modelled, where patellar orientation and displacement relative to the tibia were defined as a function of shank angle (Delp, 1990). The ankle joint was modelled as two non-coincident, non-orthogonal axes, based on studies by Inman (1976).

Passive joint structures acting across joints contribute resistive forces near the limits of a joint's range-of-motion. The relationship has most often been described and modelled as an exponential function (Yoon and Mansour, 1982; Audu and Davy, 1985). The equation parameters for the hip, knee, and ankle were obtained from Yoon and Mansour (1982), Mansour and Audu (1986), and Weiss et al. (1986) respectively.

Musculotendon Lines-of-Action

In order to define anatomically accurate musculotendon lines-of-action, a scaleable geometric model of skeletal landmarks was developed using the Terry Database from the U.S. National Institute of Health (Kepple, 1998). This is a database of skeletal landmarks based on 52 cadavers. Origins of the skeletal segments are defined at the proximal end. To scale the geometric data to individual subjects, the three-dimensional coordinates were linearly scaled based on ratios derived from anthropometric data (skeletal lengths, widths, and depths) taken from measurements of the experimental subject's thigh, shank and foot. Musculotendon unit attachments were simplified as point contacts, constrained by as few as two contact points (origin and insertion) with additional wrapping points that act as pulleys. The musculotendon units may cross one or two joints. The lines-of-action of 41 musculotendon units were defined using anatomical landmarks and descriptions from the University of Washington's Lower Extremity Muscle Atlas (Richardson et al., 2000) and Basmajian (1982).



Mathematically, the coordinates of each muscles anchoring positions were converted from their local (bone-based) coordinate system (LCS) to a common global coordinate system (GCS) for each step in time in order to calculate both musculotendon lengths and velocities. Multiplication of LCS orientation and position, as expressed using the Denavit-Hartenberg notation, was used to carry out these rotations. To calculate the force on a bone-segment, the force vector components (in LCS) applied on a bone segment were summed. The moment of applied muscle forces about the proximal end of a bone-segment were calculated as the cross product of the force vector (in LCS) and the distance (in LCS) from the point-of-application to the proximal end of the bone-segment. The application of muscle actuators to skeletal segments is well described by Pandy (1999) and Zatsiorsky (2002).

Figure 1: Geometric lower-limb musculoskeletal model

Hill-type Muscle Model

Numerous numerical simulation models of empirical muscle dynamics have been developed. They are often based upon the experimental work of A.V. Hill (1938), vary in complexity, and are often implemented within customized software. We used the muscle modelling software developed Cheung, Brown and Loeb (Cheung et al., 2000) at USC. This group have made their Simulink-based muscle models freely available to encourage their use in musculoskeletal simulation efforts. The software developed allows the user to specify the properties of each fibre

type and the morphometry of each musculotendon element. The models have a fibre-type based recruitment scheme and can be driven by recorded EMG envelopes and calculated musculotendon lengths.

Biodex Dynamometer

To accurately model the experiment using forward-dynamics, it was necessary to model the control system and dynamics of the Biodex System-2 dynamometer. The details of the control system were derived from the inventor's patent (Boyd et. al., 1987) which detailed both the control strategy and hardware implementation. In isokinetic mode, the dynamometer's control system limits angular velocity via a clamping circuit that implements a PID controller to maintain angular velocity once the maximum velocity has been exceeded. The PID coefficients were chosen through trial-and-error to recreate the accuracy of control exhibited in experimental recordings. The dynamic characteristics of the dynamometer's DC motor (QMC MTE2309, QMC Technologies) were implemented using a two-pole Laplace transform incorporating both electrical and mechanical time constants. Other elements of the dynamometer model include the inertia of both the motor and the brace/interface between the subject's leg and the motor. The inertial parameters of the brace/interface was calculated experimentally using a custom made device that implemented an oscillation technique (Lephart, 1984).

Simulation

To drive the muscle using EMG data, a uniform gain of 3 was applied, through trial-and-error, to the filtered EMG data in order to achieve reasonably high recruitment levels. The system was simulated with a maximum step size of 0.005 seconds (i.e. 2000Hz) using MATLAB's ode45 integration solver with variable step size. The system was simulated for 3.5 seconds of data.

Results

Our preliminary results are based on the simulation of a set of isokinetic concentric-concentric extension-flexion repetitions performed at 150 degrees/second. Each simulation took approximately 30 minutes to run on 2.4Ghz Pentium-4 computer running Simulink 4.1. In order to compare simulation results, we have chosen to compare the angle and angular velocity profiles over 3.5 seconds of simulation. The angle-time profiles (Figure 2) demonstrate the desired extension-flexion motion of the knee, but with an increasing phase shift between the simulated and the experimental data. The simulated angular velocity-time profile (Figure 3) demonstrates a similar shape to the experimental data, but with also an increasing phase shift and overshoot of the velocity controller.

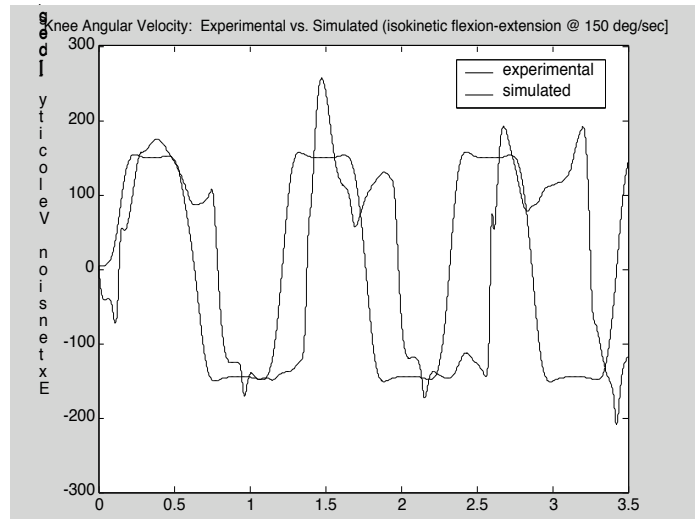
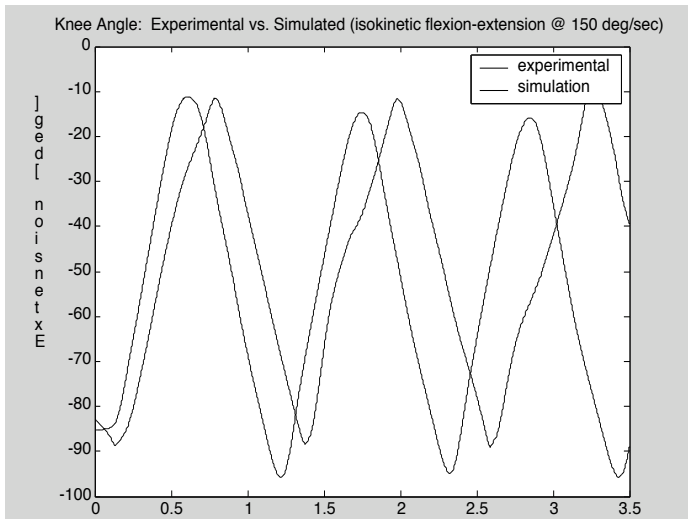


Figure 2: *Experimental vs. Simulated Knee Angle* **Figure 3:** *Experimental vs. Simulated Knee Angular Velocity*

Discussion

We have presented a forward-dynamic model to simulate knee extension-flexion experiments on a commercial dynamometer. We have tried to accurately model body-segment mass and inertia, joint stiffness, active-muscle recruitment, and the dynamometer control system. In order to achieve results that reasonably matched the experimental data, a factor of 3 had been introduced to the incoming EMG signals to the Hill model. As the

estimation of various parameters were necessary, the errors found in simulating the knee movements may be attributed to errors in estimating body-segment mass and inertia, Loeb-Brown muscle model parameters, and Biodex control parameters. Our future work will focus on improving the estimates of the various model parameters in question and investigating the simulation model across all of the experimental tests.

References

- Apkarian, J., Naumann, S., and Cairns, B. (1989)** A three-dimensional kinematic and dynamic model of the lower limb. *J. Biomech.* Vol. 22, No. 2, pp. 143-155.
- Audu, M.L. and Davy, D.T. (1985)** The influence of muscle model complexity in musculo-skeletal motion modelling. *J. Biomech. Eng.* 107: 147-157.
- Basmajian, J.V. (1982)** *Primary Anatomy*. 8th Ed. Baltimore: Williams & Wilkins.
- Boyd, R.L., Krukowski, R., and Gezari, D.Y. (1987)** U.S. Patent 4,691,694 *Muscle Exercise and Rehabilitation Apparatus*.
- Cheung, E.J., Brown, I.E., and Loeb, G.E. (2000)** Virtual muscle: a computational approach to understanding the effects of muscle properties on motor control. *J. Neurosci Methods* (101): 117-130.
http://ami.usc.edu/Projects/Muscular_Modeling/index.asp
- Delp, S.L. (1990)** *Surgery Simulation: A computer graphics system to analyze and design musculoskeletal reconstructions of the lower limb*. PhD Thesis: Stanford University.
- Featherstone, R. (1983)** The calculation of robot dynamics using articulated-body inertias. *Int. J. Robotics Res.* 2(1): 13-30.
- Inman, V.T. (1976)** *Joints of the Ankle*. Baltimore: Williams & Wilkins.
- Jensen, R.K. (1989)** Changes in segment inertia proportions between 4 and 20 years. *J. Biomech.* 22(6/7): 529-536.
- Kepple, T.M., et al. (1998)** A three-dimensional musculoskeletal database for the lower extremities. *J. Biomech.* 31(1):77-80.
- Lephart, S.A. (1984)** Measuring the inertial properties of cadaver segments. *J. Biomech.* 17(7): 537-543.
- Mansour, J.M., Audu, M.L. (1986)** The passive elastic moment at the knee and its influence on gait. *J. Biomech.* 19: 369-373.
- Meriam, J.L. and Kraige, L.G. (1987)** *Engineering Mechanics Vol.2 Dynamics* (2nd Ed.) Toronto: John Wiley & Sons.
- Pandy, M.G., (1999)** Moment arm of a muscle force. *Exer. Sports Sci. Rev.* 27: 79-118.
- Richardson, M.L., Teitz, C.C., Graney, D.O. (2000)** Lower Exterimity Muscle Atlas. University of Washington:
<http://www.rad.washington.edu/atlas2/>.
- Weiss, P.L., Kearney, R.E., and Hunter, I.W. (1986)** Position dependency of ankle joint dynamics. *J. Biomech.* 19: 727-751.
- Woltring, H.J. (1986)** A fortran package for generalized, cross-validatory spline smoothing and differentiation. *Adv. Eng. Software* 8(2): 104-113.
- Yamaguchi, G.T., and Zajac, F.E. (1989)** A planar model of the knee joint to characterize the knee extensor mechanism. *J. Biomech.* 22(1): 1-10.
- Yeadon, M.R. and Morlock, M. (1989)** The appropriate use of regression equations for the estimation of segmental inertia parameters. *J. Biomech.* 22(6/7): 683-689.
- Yoon, Y.S., and Mansour, J.M. (1982)** The passive elastic moment in the hip. *J. Biomech.* 15: 905-910.
- Zatsiorky, (2002)** *Kinetics of Human Motion*. Champaign: Human Kinetics.

Acknowledgements

The authors would like to thank the following organizations for their support:

Bloorview MacMillan Children's Foundation

Kids Action Research

University of Toronto