Computer Simulation of Human Lower Limb Motions Using a Three-Dimensional Musculoskeletal Model

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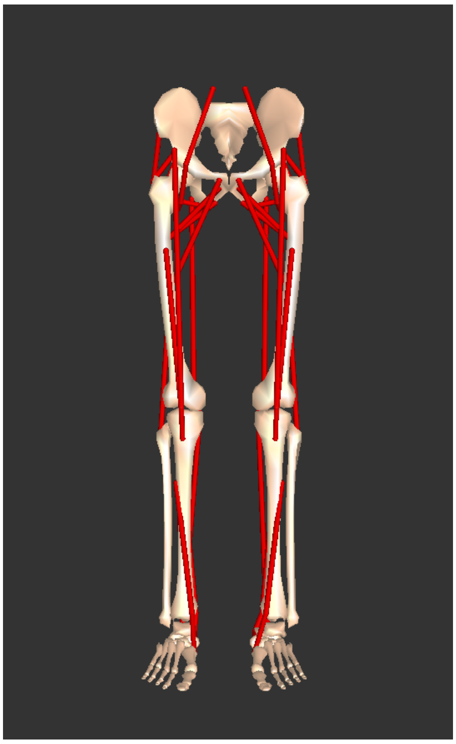
Abstract: For the purpose of studying human biomechanics, it is very advantageous to obtain real-time data of kinetic variables of the musculoskeletal system. More specifically, such variables as forces developed by muscles, forces applied on bones, length change of muscle fascicles and tendons can provide precious information for the evaluation of human whole body biomechanics. However, in reality, these are the variables that can hardly be obtained from human subjects without causing serious technical and ethical problems. For example, direct force measurement from human muscles is so invasive that it is very difficult to perform this methodology, especially outside of research laboratories. As an alternative, the methodology of computer simulation can provide useful data regarding the behavior of the musculoskeletal system. Those variables (i.e., forces and length changes in the musculoskeletal system) can be obtained easily by examining the state variables of the simulation model. In order to perform this type of study, it is necessary to develop a sophisticated model that has enough degrees of freedom and complexity to represent the human musculoskeletal system. To date, this requirement has discouraged many researchers from choosing this direction, except for a few research groups that have successfully developed a procedure to accomplish this goal. Our group is one of them. We have developed a musculoskeletal model of the human body (trunk and legs) that has 9 rigid body segments, 20 degrees of freedom and up to 86 muscles. The first purpose of this report was to provide a detailed description regarding the structure of the model. For each element of the musculoskeletal model, a detailed description was given or relevant articles were introduced. We have utilized the musculoskeletal model to study various kinds of human leg motions. Research results regarding three of those motions, i.e., ankle bending, jumping and walking have been published in international scientific journals. The second purpose of this report was to introduce the essential points of those three studies. Finally, parts of currently ongoing projects were discussed.

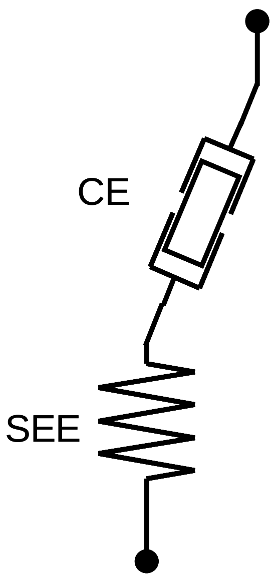
1. Construction of the Lower Limb Neuromusculoskeletal Model

1.1. Skeletal Model



The three-dimensional (3D) musculoskeletal model of the human body was developed using DADS-3D (LMS CADSI, Coralville, Iowa, USA) with the FORTRAN-based USER.FORCE option. The skeletal model consisted of nine rigid body segments (head-arms-trunk (HAT), right and left upper legs, right and left lower legs, right and left feet and right and left toes) connected with frictionless hinge joints. Body segmental parameter values such as mass and moment of inertia were derived based on the anthropometric data published by De Leva (1996). The whole body mass was set at 73.1 kg. Hip joints were modeled as universal joints that have three degrees of freedom. Knee joints were modeled as simple hinge joints. Ankle joints were modeled as biaxial joints as reported by Inman (1976). Metatarsophalangeal joints were modeled as hinge joints with a tilted axis (Delp, 1990). Therefore the total number of degrees of freedom of the skeletal model was 20.

1.2. Musculotendon Model



Up to 86 lower limb muscles (43 muscles in each leg) were implemented into the skeletal model. These include most of major muscles found in the human leg (Delp, 1990). Musculotendon parameter values, i.e., the optimal contractile element length (L_{CEopt}), the maximal isometric force of the contractile element (F_{max}), the pennation angle (α_{pen}) and the unloaded length of the series elastic element (L_{slack}), were derived from Delp (1990) and Friederich and Brand (1990). The specific tension value of 31.5 N/cm^2 (Brown et al., 1998) was utilized. Coordinate data of the origin, insertion and via-points of these muscles were derived from Delp (1990). A bilateral symmetry was assumed between the right and left legs. A musculotendon complex was composed of a contractile element (CE) and a series elastic element (SEE) serially connected with a pennation angle (α_{pen}). The mathematical model of the contractile element represented the force-length-velocity relations discussed by Hill (1938). The nonlinear stress-strain property of the series elastic element was modeled with a quadratic function. A detailed mathematical representation of these components can be found in Nagano and Gerritsen (2001).

1.3. Muscle Excitation-Contraction Model

The excitation dynamics of the contractile element was modeled with a first order ordinary differential equation as discussed by He et al. (1991). The time constant that specify the delay between an increase in the neural input signal and the following increase in the excitation level of the contractile element was set at 55 ms. The time constant that specify the delay between a decrease in the neural input signal and the following decrease in the excitation level of the contractile element was set at 65 ms. A detailed mathematical representation of these components can be found in Nagano and Gerritsen (2001).

1.4. Ground Reaction Force Model

The basic equation to describe ground reaction forces was adopted from Anderson and Pandy (1999):

$$GRF_y = \alpha_1 \cdot \exp(-\alpha_2 \cdot (y - \beta_1)) - \frac{\alpha_3 \cdot \dot{y}}{1 + \alpha_4 \cdot \exp(\alpha_5 \cdot (y - \beta_2))}$$

where GRF_y is the vertical component of ground reaction force and y is the vertical position of the foot / ground contact point (floor = 0.0 m). The model parameter values were determined through numerical optimization such that a computer simulation of a drop-test on this particular ground reaction force model gave results similar to those of pendulum testing on human heel pad tissue (Aerts and DeClercq, 1993). The optimization resulted in the following parameter values: α_1 = 1.039 N; α_2 = 491.804 m^{-1}; α_3 = 963.321 $N \cdot m^{-1}$; α_4 = 44.715 (no units); α_5 = 706.924 m^{-1}; $\beta_1 = 0.857 \times 10^{-4}$ m; and $\beta_2 = -2.325 \times 10^{-3}$ m. Five contact points in each foot (2 on the heel, 2 on the metatarsophalangeal joint and 1 on the toe) were modeled similar to Anderson and Pandy (1999).

1.5. Passive Joint Properties Model

The passive joint properties that function to limit the joint range of motion were adopted from Anderson and Pandy (1999). More specifically, the joint angle – passive joint moment profiles were represented with exponential functions whose coefficients have been adjusted to represent the normal range of motion for individual joints. A full description of this passive joint properties model can be found in Anderson (1999).

These are the construction of the musculoskeletal model. Research results obtained using this model are introduced in the following sections.

2. Ankle Bending

In this study, contribution of series elasticity of human mm. triceps surae was investigated in cyclic ankle bending (heel-raise) exercise. Major modifications were made on the musculoskeletal model in order to set the focus on the behavior of mm. triceps surae during the motion. Body segmental motions were constrained in the sagittal plane. Hip joint motions, knee joint motions and ankle joint inversion / eversion were prohibited. Therefore the ankle joint (dorsiflexion / plantarflexion) was the only movable joint in the model. Muscles other than mm. triceps surae were removed. An upright posture was maintained throughout the simulation to prevent the model from falling. The mm. triceps surae was activated by the neural activation input signal with a time resolution of 0.050 s. Cyclic ankle bending exercise (similar to hopping except for that the feet did not leave the floor) of a various range of cycle durations (0.300 sec (200 cycles/min) – 0.900 (66.7 cycles/min)) were generated using numerical optimization. The goal of the numerical optimization was to generate cyclic motions with as large range of motion as possible. As a result, realistic heel-raise motions were generated with the range of motion between 0.0023 m and 0.0414 m. It was found that the contribution of the series elasticity in positive mechanical work output of the muscle-tendon complex during the push-off phase increased as the motion frequency increased (3% at 66.7 cycles/min to 47% at 200 cycles/min). When the motion frequency was higher, relatively higher muscle activation was found during the downward moving phase. These tendencies are consistent with the findings reported in preceding studies performed using experimental animals as well as human subjects. (Nagano et al., 2003a.)

3. Squat and Countermovement Jumping

In this study, jumping motions generated with the musculoskeletal model were utilized in order to address a technical question regarding the methodology of digital filtering. Specifically, the applicability of the optimal cutoff frequency of the Butterworth digital filter determined through residual analysis was evaluated. A squat jumping motion (SQJ) and a countermovement jumping motion (CMJ) were chosen as the subject. Realistic SQJ and CMJ motions were generated using forward dynamic computer simulation through numerical optimization. Residual analysis was performed on each coordinate of anatomical landmarks in order to obtain the optimal cutoff frequency for the digital filter. Effects of artificially introduced experimental noise were considered. It was found that the optimal cutoff frequency was underestimated when the experimental noise was imposed on the kinematic data. The possibility of an information loss was suggested as a result of digital filtering with the cutoff frequency determined through residual analysis. It was implied that the optimal cutoff frequency obtained with residual analysis should be compared with the residual-cutoff frequency curves obtained through analyzing noise-free jumping motion data. (Nagano et al., 2003b.)

4. Walking of a Human Ancestor

The skeleton of *Australopithecus afarensis* (A.L. 288-1, known as "Lucy") is by far the most complete record of locomotor morphology of early hominids currently available. Even though researchers agree that the postcranial skeleton of Lucy shows morphological features indicative of bipedality, only a few studies have investigated Lucy's bipedal locomotion itself. To gain further insights into how Lucy may have walked, we generated a full 3D reconstruction of upright bipedal locomotion of this human ancestor. Laser-scanned 3D bone geometries were combined with the modeling and simulation techniques. A detailed full 3D musculoskeletal model was developed that encompassed all major bones, 10 joints and 52 muscles of the leg. A model of muscle heat production was used to estimate total energy expenditure during walking. Optimal neural activation inputs to each of the muscles that produced a single step of locomotion, at the same time as minimizing the energy expenditure per meter traveled, were searched through numerical optimization. The procedure resulted in smooth kinematics. As well, the predicted energy expenditure was appropriate for upright bipedal walking of an individual of Lucy's body size. (Nagano et al., in press.)

5. Ongoing Projects

We are currently making progresses in several areas of human biomechanics. Our short-term goal is to expand our research in those areas, firstly through completing the ongoing projects. (1) We are continuing research efforts to investigate the optimal coordination of the whole body during explosive activities such as jumping. For example, the importance of the activities of the non-extensor muscles of the leg (e.g., m. gluteus medius) is being examined. As well, difference in the optimal coordination patterns between a standing vertical jump and a standing broad jump is being studied. (2) Following the study introduced in the section **4.** of this report, we are expanding our walking research using the musculoskeletal model of the human body anthropometry. Many of preceding studies that investigated walking mechanism have simulated one step only assuming that the motion can be continued in a cyclic manner. However, considering the sensitivity of simulation kinematics to slight modifications of state and input variables, the appropriateness of this assumption is not obvious. Therefore we are currently generating multiple-steps simulation of human walking. So far, four “awkward” steps have been generated through numerical optimization. We are currently continuing the numerical optimization process to improve the kinematics and kinetics of this walking motion. (3) Another computer simulation model is being developed, that has not only legs but also arms. Technically speaking, it is not difficult to implement arm segments and muscles into the musculoskeletal model. However, as this modification will more than double the complexity of the model, it will necessitate much greater computation time. Therefore we are currently exploring the possibility to implement the model in a more powerful environment both in terms of hardware and software. When this problem is cleared, more interesting and valuable findings will be obtained in the area of upper extremity / whole body biomechanics.

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