Control of Leg Movements Driven by Electrically Stimulated Muscles

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Abstract - A planar biomechanical model of a human leg has been developed to investigate automatic control strategies for artificially stimulated muscle using electrical stimulation (ES). The model comprises the nonlinear multiplicative model of the muscles, tendons and two body segments (thigh and shank-foot complex), that is, the hip and the knee. Flexor and extensor muscles are included for each joint. The parameters determining the model used for the analysis can all be assessed; hence, the model can be customized to a particular subject. Simulation software is organized within the SIMULINK environment, and the user can easily change parameters of the model by using Dialogue Boxes. The simulation allows determining strategies for artificially stimulated muscle using electrical stimulation (ES). The model comprises the nonlinear multiplicative model of the muscles, tendons and two body segments (thigh and shank-foot complex), that is, the hip and the knee. Flexor and extensor muscles are included for each joint. The parameters determining the model used for the analysis can all be assessed; hence, the model can be customized to a particular subject. Simulation software is organized within the SIMULINK environment, and the user can easily change parameters of the model by using Dialogue Boxes. The simulation allows determining strategies for artificially stimulated muscle using electrical stimulation (ES).

Index Terms - biomimetic control, biomechanics, human standing, legged locomotion.

I. INTRODUCTION

The study of the biomechanics of walking provides extensive material for designing different aids for humans with disabilities, as well as for investigating the physiological processes and neural mechanisms controlling the system. The biomechanical models available today are generally far too complicated, and almost impossible to customize to fit a particular subject. Their simplification would help to derive better controllers for gait restoration. A customized model could be used to tune the controller off-line before trying it out on a spinal cord injured (SCI) person.

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A distinction can be made between attempts to deal with an essentially complete system (human body walking on the ground), and reduced models. In these models the problem is simplified by reducing the number of degrees of freedom in the model by, for example, treating the motion in only one plane at a time, or reducing the number of links in the model. Consequently, attempts to analyze multi-joint structures are often characterized by a search for ways of reducing the number of degrees of freedom to a manageable level. The dynamics of motion confined to a plane is much simpler than that in three dimensions. Locomotion can often be decomposed into a dominant component in the sagittal plane with much smaller components in the frontal and horizontal planes. Confining the model to a plane and using a small number of links usually achieve the reduction in complexity; yet, in many multi-joint structures (e.g., the vertebrate spine) drastic simplification is required [29].

The ways of managing complexity may be summarized as follows:

1. reducing the number of degrees of freedom analytically by finding approximations and constraints, and by designing systems with the minimum number of joints; e.g., 8 degrees of freedom are usually used to describe a seven segment body model [19,29];
2. decomposing a complex problem into several simpler ones by, for example, separating the control of quantities that do not interact significantly.

To model human limbs Stepanenko and Vukobratović [24] investigated a Newton-Euler approach to dynamics, instead of the somewhat more traditional Lagrangian approach. This work was revised for efficiency by Orin and colleagues [20] for the legs of walking robots. They improved the efficiency by writing the forces and moments in the local link reference frames instead of the inertial frame. They also noticed the sequential nature of calculations from one link to the next, and speculated that an efficient recursive formulation might exist. Armstrong [1] and Luh et al. [15] improved the computational efficiency and published an algorithm that reduces the complexity. This was accomplished by setting up the calculations in an iterative (or recursive) manner and by expressing the velocities and accelerations of the links in the local link frames. Koozekanani et al. [14] applied a recursive free body approach to estimate joint torque associated with observed human postural motion.

Three-dimensional simulation models have also received considerable attention. Huston et al. [10,11] developed a
general approach for studying a human body model using equations based on d’Alambert’s principle. A seven link planar model was developed by Onyshko and Winter [19]. Equations of motion formulated using Lagrangian mechanics consist of a 7 x 7 matrix of anthropometric constants and segment angles, a vector of the angular accelerations and a vector containing the torques acting on the segments. Hatze [9] used the traditional Lagrangian approach to define a mathematical model of the total human musculoskeletal system. The model comprised a linked mechanical and musculo-mechanical set of ordinary first-order differential equations, which describe the dynamics of the segment model and muscle model respectively. Hatze [8] also reported a simulation of a planar long-jump take-off with seventeen segments, driven by 46 myoactuators and controlled by a controller subsystem. Zheng et al. [33] studied the impact problem; they showed that an impact, acting on a biped subjects each link of the system to an instantaneous velocity change. Thus, an impact may cause large internal impulsive forces in the body.

Marshall et al. [18] used a general Newtonian approach to simulate an N-segment open chain model of the human body. The model simulated planar movement using data for joint torques and initial absolute angular displacements and velocities for each body segment. These values are used to solve the direct dynamics problem, expressed in the form of n simultaneous linear equations, to yield angular accelerations. Zajac with collaborators [12,13,29] developed a planar computer model to investigate paraplegic standing induced by Functional Electrical Stimulation (FES). The objective of the study by Yamaguchi and Zajac [29] was to determine a minimal set of muscles that could approximate able-bodied gait trajectories without requiring either higher levels of force or precise control of muscle activation. They suggested that gait was more sensitive to changes in the on/off timing of the muscle stimulus than to its amplitude. The process of adjusting the muscle set and the admissible activation levels was critically dependent upon accurately understanding the effect of each muscle on the dynamic response of the system. The existing biomechanical models [8,29] are very difficult to customize and very hard to work with, due to their complexity and poor user interface. The main idea here was to develop interactive, user friendly, interface software for biomechanical simulations.

II. MODEL

Muscle has been modeled at a variety of levels of detail [32]. Many models of whole muscle in the literature are based on the work of Hill [2,28,31]. These models represent muscle dynamics by a contractile component and nonlinear series and parallel viscoelastic elements. The force generated by the nonlinear contractile component depends on neural activation, and is also length and velocity dependent.

A. Three Factor Muscle Model

The nonlinear model of muscle dynamics used for joint angle control in this simulation software is a modified version of the Hill model. Active muscle force depends on three factors: neural activation, muscle length and velocity of shortening or lengthening [25]. The model is formulated as a function of joint angle and angular velocity, rather than muscle length and velocity. The relations between velocity and angular velocity are determined by the joint angle-dependent moment arm. The three-factor model is given by:

\[ \tau_a = A(u)f(\phi)g(\phi) \]  

where \( \tau_a \) is the active torque generated by the muscle contraction; \( A(u) \) is the dependence of torque on the level of evoked muscle activity \( u \) (which in turn depends on the stimulus amplitude, pulse width and stimulus frequency); \( f(\phi) \) is the dependence on the angle \( \phi \) and \( g(\phi) \) is the dependence on the angular velocity \( \dot{\phi} \). The parameter \( u \) is normalized to the range \( 0 \leq u \leq 1 \) and the function \( g \) has the value of 1 under isometric conditions \( \phi = 0 \). According to the literature [25], the muscle model described by (1) can predict the muscle torque with 85-90% accuracy during simultaneous, independent, pseudo-random variations of recruitment, angle and angular velocity.

Fig. 1 gives a schematic of the total model of muscle and load. The active torque \( \tau_a \) of the muscles and the passive torque \( \tau_p \) contributed by passive elastic properties of muscles and other passive tissues of joint, sum to produce the total torque \( \tau \) that acts on the load to produce movement which is then fed back to the muscle. The passive torque produced by all the muscles acting on a joint and the properties of the joint cannot be distinguished in the intact system and have been lumped together as will be described later.

B. Activation Dynamics \( A(u) \)

In accordance with the literature [3,4,6,25] the muscle response to electrical stimulation was approximated by a second order, critically damped [3,25], low pass filter with a delay. Thus, the activation dynamics was assumed to be expressed by Equation 2:

\[ \frac{A(j\omega)}{U(j\omega)} = \frac{\omega_p^2}{\omega_p^2 + 2j\omega_p\omega_0 + \omega_0^2}e^{-j\omega_0t} \]  

where \( A(j\omega) \) is the Fourier transform of the muscle's contractile activity, \( U(j\omega) \) is the Fourier transform of the muscle's electrical activity, \( \omega_p \) is muscle's natural (pole)
frequency (typically in the range of 1-3 Hz), \( \tau_4 \) is the excitation-contraction and other delays in the muscle (typically 20-50 ms). The nonlinear model of muscle dynamics used is a modified, discrete time version of Hill’s model.

C. Muscle Torque vs. Joint Angle and Velocity

The nonlinear function relating torque and joint angle is simulated by a quadratic curve:

\[
F = a_0 + a_1 \phi + a_2 \phi^2
\]

The torque generated can not be negative for any angle, so:

\[
f(\phi) = \begin{cases} 
F & \text{if } F \geq 0 \\
0 & \text{if } F < 0
\end{cases}
\]  

(3)

Coefficients \( a_0, a_1 \) and \( a_2 \) define the shape of the torque-angle curve. The angle and angular velocity are in rad and rad/s respectively.

The nonlinear curve relating torque and angular velocity of the joint is simulated using a piece-wise linear approximation:

\[
g(\phi) = \begin{cases} 
 b_1 \phi & \text{if } G \geq b_2 \\
G \phi & \text{if } 0 < G < b_2 \\
0 & \text{if } G \leq 0
\end{cases}
\]

(4)

where the function \( G = 1 + b_1 \phi \).

D. Two Link Model of the Skeleton

Fig. 2 shows the model of the skeleton and the usual definitions for the hip flexion angle \( \phi_H \) and knee flexion angle \( \phi_K \). As either of these angles increases, the flexor muscles will shorten and the torque will be reduced. Conversely, as the joint flexes the extensor muscles will be stretched and torque will be increased. Thus, the coefficient \( b_1 \) is positive for extensor muscles and negative for flexor muscles, as seen in Fig. 3. Typical values for the maximum torque \( b_2 \) are from 1.2 to 1.8 [2]. We adopted the mechanical structure as a double pendulum moving in the X-Y plane. The upper segment represents the thigh, while the lower is for the shank and foot. Ankle dynamics could also be added, but the ankle is often fixed by an ankle-foot orthosis or limited by spasticity in the population of interest. The dynamic equations of a planar two-link pendulum are found in many textbooks, and we have mainly used the notation of Yoshikawa [30]. The following notations are used in the Fig.: \( \phi_2 \), the flexion angle of the knee joint; \( \phi_H \), the flexion angle of the hip; \( \phi_0 \), the angle of the shank measured with respect to the X axis direction; \( \phi_t \), the angle of the thigh measured with respect to the X axis direction; \( m_t \), the mass of link i; \( J_t \), the moment of inertia of link i about the X axis; \( L_i \), the length of link i; \( d_i \), the distance between joint i and the center of mass of link i. Two torques \( M_t \) and \( M_t \) are net joint torques at the hip and knee parallel to the Z-axis. It was also assumed that gravitational force acts in the negative Y direction, and that the hip joint is immobile. The equations of motion for the two-link skeleton are then:

\[
M_t = M_{t1} \phi_H - M_{t2} \phi_K + h_{122} \phi_K^2 + g_1
\]

\[
M_s = M_{s1} \phi_H - M_{s2} \phi_K + h_{211} \phi_K^2 + g_2
\]

(5)

where the coefficients are:

\[
M_{t1} = m_t d_i^2 + J_t + m_s ( L_i^2 + d_i^2 + 2 L_t d_s \cos \phi_K ) + J_{CS}
\]

\[
M_{t2} = M_{s1} = m_s d_i^2 + J_{CS}
\]

\[
h_{122} = h_{112} = h_{211} = m_s L_t d_s \sin \phi_K
\]

\[
g_1 = m_t g d_t \sin \phi_K + m_s g ( L_t \sin \phi_H + d_s \sin ( \phi_H - \phi_K ) )
\]

\[
g_2 = m_s g d_s \sin ( \phi_H - \phi_K )
\]

The quantities in Equation 5 can be identified as the effective inertia \( M_{t0} \), the coupling inertia \( M_{t0} \), the centrifugal and Coriolis acceleration coefficients \( b_{ij} \) and the gravitational term \( g \). The definition of the angles in Fig. 2 differs from Yoshikawa [30] in order to match the conventional anatomical definitions of flexion and extension angles at the hip and knee. The net torque at the i-th joint \( M_{t0} \) is obtained by taking the difference of the active torques of the flexor and extensors muscles that acts at the segment. In this study we reduced the dimensionality of the system by introducing the equivalent flexor and extensor muscles, being equal to the net sum of all muscles contributing to flexion or extension of the i-th joint.

The active hip flexor contributes to flexion (positive direction), while the active hip extensor acts opposite at the hip joint. The other components acting within about the hip joint are the passive stiffness and viscosity. These

Fig. 2: Model of the two-link skeleton. See text for details.
joint torques result from elastic and viscous properties of muscles and tendons exposed to external stretching, and viscoelastic behavior of the surrounding tissue. They act in the opposite direction of the active torques. The active knee flexor and passive knee extensor acts in positive direction, while the active knee extensor and passive knee flexor act oppositely. In other words

\[ M_\tau = \tau_{f,e} - \tau_{e,f} - \tau_{p,e} \]
\[ M_\beta = -\tau_{e,f} + \tau_{e,p} + \tau_{p,e} \]

where the active flexor and extensor torques are given by Equation 1. The passive stiffness is modeled as [2]:

\[ \tau_{p,e} = c_i \exp(c_j \theta) \phi \]

where the \( c_i \) and \( K_i \) are constants. Equation 7 has six constants and an alternate form with six constants would be a fifth order polynomial. However, our extensive analysis showed that the exponential fit is more effective, particularly for the knee. The two last coefficients provide an offset and a linear term, while the nonlinearities are fitted with the sum of two exponential curves. The constants \( K_i \) give the passive viscosity.

E. User Interface

The actual implementation of the model was developed using a SIMULINK interface (MathWorks, Inc.). Flexor and extensor muscles are defined as muscle*.m functions and they can be modified using the dialogue box to change any of the active or passive characteristics to include individual characteristic of a specific object analyzed. The model of the system (Equation 5) is defined in a block described with a skeleton.m file. The block can be approached using a dialogue window where all the individual parameters can be put: masses, lengths, moments of inertia, coefficients of passive stiffness and viscosity, second order damping parameters and initial conditions (joint angles and angular velocities at time \( t=0 \)). The skeleton.m file contains the system of differential equations, which is automatically solved using the Runge-Kutta or other integration methods (part of the SIMULINK program). The skeleton.m is represented with an icon allowing users to generate any combination of muscles and skeleton by manipulating with described blocks and other blocks from the SIMULINK library. Every dialogue box is opened by double clicking at the corresponding box, following the methodology of SIMULINK. The program is available for any additions and can be used for extensive studies on effects of different control strategies, limitations etc. The inputs, \( u_i \) can be set as required and the outputs from single muscles can be viewed (Scope) as well as an animated, stick Fig. of the movement. Various parameters are also sent to a workspace for further analysis or plotting.

Fig. 4 shows a configuration with four muscles around the two links of the skeleton. The muscle model has three inputs: the level of activation, the angle and the angular speed at the joint. The output from the muscle block is the active torque. The skeleton block has two inputs for each joint. The first input is reserved for the flexor torque, while the second is for the extensor torque generated by the equivalent muscles. Each joint has two outputs: the first output is joint angle \( \phi_i \) (rad), while the second is the joint angular speed \( \dot{\phi}_i \) (rad/s).

The final block is called Animation, and it contains a program that graphically presents the actual movements of the segment. It allows the user to visually inspect the performance while the system is simulating. This block also allows replay, so that some specific phases during the movement can be analyzed. A feature of the program is that it is possible to monitor any of the variables of interest by connecting as many Scope blocks as needed.

III. EXPERIMENTAL METHODS

The methods for determining the passive properties of a joint have been described previously [23]. Briefly, the joint is slowly pulled over its full range of motion (5-10 s) to measure the passive elasticity in what is referred to as a "Pull Test". The force is measured with an in-line strain gauge and the joint angle with a goniometer (Penny & Giles, U.K. or equivalent). This is repeated several times to check for the stability of the curves. From the data the parameters can be calculated to fit the passive stiffness (Equation 7) using a simple MatLab program that is available from the authors.

Segment lengths and body mass can be measured and from these the distances to the center of mass of the segments, as well as the mass and moment of inertia of the segments can be calculated using the standard formulae given in Winter [28]. These values are general approximations and we also use a "drop" test to verify the values. For the knee the subject lies supine and the shank is either flexed or extended from equilibrium and then dropped. Again an in-line strain gauge can measure the force of the full and a goniometer the resultant angular change. The damped oscillation that results gives a measure of the inertia of the shank and foot, as well as the stiffness and passive viscosity of the knee joint. Small drops (5 - 10°) and drops starting from a flexed position are less likely to elicit reflexes. This is important since reflexes will affect the damping and other properties of the oscillation and thus lead to erroneous
values of the biomechanical parameters. The methods are described in Stein et al. [23] and again programs are available from the authors.

The whole leg can also be "dropped" after bracing the knee. The subject can stand with one leg on a stair, so that the leg of interest moves freely after being dropped. This can also lead to reflexes, particularly in subjects who suffer from spasticity, but the drop can also be done with the subject lying on one side. The leg is suspended and moved laterally in an arc. When dropped it will oscillate and the biomechanical parameters of the whole leg can then be determined. With the parameters known for the whole leg and the shank, the parameters of the thigh can be calculated, as described in [23], and compared with the values predicted anthropometrically.

Finally, the parameters of the active torque-angle and the torque-velocity curves can be determined by using a CYBEX, KINCOM or similar machine. The subject contracts maximally or is stimulated maximally (in the case of paralyzed subjects) at a variety of different angles. The torque generated can then be fitted as a function of angle using Equation 3. Similarly, at an angle where close to maximal force is generated, the effect of contractions at different speed are determined and fitted by Equation 4. In entering the parameters it is important to remember that these values should be normalized to the isometric value (see Equation 4). Also, some subjects with severe osteoporosis run a risk of bone breakage, so stimulation may have to be applied with less than maximal strength.

IV. RESULTS

Fig. 5 shows the data from a "pull" test and a "drop" test at the knee, as described under Experimental Methods. The pull test measures the passive stiffness of a joint and muscles. The stiffness can be fitted with Equation 7 (see also [16]) and the fitted curve is also shown. There is some hysteresis, since the force is greater when the muscle is being stretched than when it passes through the same angle during a release. Extra terms can be used to fit the hysteresis [21], but we have used a static curve collected during stretch and release.

The drop test measures the passive dynamics and can be fitted by an under damped second order system, as described previously by several authors [e.g., 5,7,23,26]. The three parameters of the fitted curve in Fig. 5B, the magnitude, natural frequency and damping ratio can be used to measure the stiffness, inertia and viscous damping of the joint. These values of inertia and stiffness can be compared with the values calculated from measurements of body segments and mass and from the pull test. In general, the values agree well in normal individuals, although the calculated inertia may be lower in disabled individuals, because of atrophy in limb muscles [23].

Having measured the properties of individual joints, the model can then be used to predict the movement of the whole leg when dropped. The leg will then show movements typical of a complex pendulum with interactions between the joints arising from the coupling inertia, centrifugal force, etc. (see terms in Equation 5). Three examples are shown in Fig. 6, in which A) the whole leg was moved into flexion, B) the whole leg was moved into extension and C) only the knee was flexed. The starting position is indicated by a stick Fig., which is the output of an animation routine, which is included in the simulation package. There is generally good agreement between the predicted and observed movements, although deviations are seen both in magnitude and timing. The visco-elasticity of the knee joint was measured at the middle of its range with the drop test with the subject lying supine. However, when standing, the knee operates near the end of its range (full

Fig. 5: Experimental data (dots) and fitted curves (line) of a pull test (A) and a drop test (B) for the knee joint of a normal subject. Before fitting the data in (A), the torque necessary to overcome the gravitational component has been subtracted to yield the applied torque needed to overcome the passive stiffness.

Fig. 6: Experimental data (dots) and simulations (lines) when the whole leg was moved into flexion (A), when the whole leg was moved into extension (B) and only the knee was moved into flexion (C). The initial angles are shown by the stick Figures on the right. Note the difference in scale for the knee angle in (C).
the hamstring and quadriceps muscles were stimulated alternately (B). Muscles were stimulated intermittently using surface electrodes (A) and when

![Fig. 7](image)

Fig. 7: Experimental data (dots) and simulations (lines) when the hamstring muscles were stimulated intermittently using surface electrodes (A) and when the hamstring and quadriceps muscles were stimulated alternately (B).

extension). Hence, the visco-elasticity can be somewhat different which will affect the magnitude and damping of the oscillations. The parameters were not varied to optimize the fit to the data, but were those measured from single joint movements, as in Fig. 6.

As a final example, Fig. 7 shows the response of the model to A) intermittent stimulation of the hamstring muscles (knee flexors) and alternation of stimulation to the quadriceps (knee extensors) and hamstrings. Again there is reasonable agreement between the observed and predicted results. Any input can be tested as long as its waveform is fed to the input terminals of the respective muscles, including the pattern of EMG recorded from the muscles during normal walking, but the examples shown here should suffice to illustrate the potential of the model for a variety of tasks.

V. DISCUSSION

The model described here is intentionally quite simple with only two links and four muscles operating in a single plane. The muscles and joints are also relatively simple with a limited number of parameters. The whole model is imbedded as a block diagram (Fig. 4) with pop-up dialogue boxes in a flexible package under SIMULINK and an animation routine to visualize the movements as they are computed. There are several advantages to this approach. First, all of the parameters of the model are easily measurable, as described in the Experimental Methods and Results sections. Thus, it can be customized for each normal or disabled individual, rather than relying on "standard" values. Second, the simple structure means that it is straightforward to test the effects of varying the inputs or particular patterns on the resultant movements. Thirdly, the form as a block diagram under SIMULINK means that new blocks can be easily added, as required by the interests of the user. For example, the model as it stands is invariant over time, but a time constant could be added for the effects of fatigue as a result of stimulating muscles. This may be important in applications to electrical stimulation of muscles that have atrophied and become more fatigable after spinal cord injury [22]. If the model is going to be used extensively to study the effects of stimulation with implanted wires using pulse width modulation, then a block can be added that converts pulse width to an activation level according to a suitable sigmoid function [17].

This approach also has some disadvantages. For example, by using the standard integration routines under SIMULINK, the solutions are not optimized for speed. Thus, with a step size of 0.01s, using a Runge-Kutta 5th order integration routine on a Pentium 1 GHz machine with the Animation routine, the model still runs slower than real time (a 5 seconds integration takes about 2 seconds). However, there are a variety of integration routines available which may be helpful to maintain stability under some conditions.

We have also not tried to optimize the fit of the data to the model in Figs. 6 and 7 for example, but it would be feasible to do so with the limited number of parameters. Multi-parameter optimization works better, if many of the variables are fixed. It would, of course, be an individual decision to decide which parameters are well known and can be fixed and which are somewhat uncertain and can be varied to obtain an optimal solution.

Optimization techniques could also be added to the input block to determine the pattern of inputs that would give the best swing phase for walking, in terms of distance covered in a short period of time. At present, ground reaction forces and accelerations at the hip have not been included, so that the model is not suitable for the stance phase of walking. Ground reaction forces can be measured and added to the model and the accelerations at the hip are not large in disabled individuals, who walk slowly with a quasi-static gait. However, the movements at the hip, both in the forward and the lateral plane, will affect whether the leg clears the ground satisfactorily. These can be added, as required, but will increase the complexity of the model. Double joint muscles are not included but could be added, if required. Overall, we feel that the model offers a combination of simplicity and flexibility that will be useful for a number of purposes and that it can be modified to suit the requirements of various applications.

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