Simulation of Bipedal Walking

Margaret M. Skelly[1] and Howard Jay Chizeck[2]
chizeck@ee.washington.edu
mxs70@po.cwru.edu

[1]Dept. of EECS, CWRU
Cleveland, Ohio 44106
[2]Dept of EE, University of Washington
Seattle WA, 98195-2500

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Department of Electrical Engineering
University of Washington
Box 352500
Seattle, Washington 98195-2500
PHN: (206) 543-2150
FAX: (206) 543-3842
URL: http://www.ee.washington.edu

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Margaret M. Skelly[1] and Howard J. Chizeck[2]
[1]Dept of Electrical Engineering and Computer Science, CWRU, Cleveland, OH
[2]Dept. of Electrical Engineering, University of Washington, Seattle, WA

Abstract

A 3-dimensional computer model of sustained bipedal walking is presented. It is intended be used as a development tool for walking controllers. The direct dynamic simulation has 8 segments, 19 degrees of freedom and is driven by prescribed joint moment and stiffness trajectories. Limited feedback in the form of a proportional-derivative controller provides upper body stability and allows walking to be sustained indefinitely. The joint moment and stiffness trajectories are specified in coarse block segments. By changing the intensity of hip extensor activity during terminal stance the walking stride length is modulated.

1 Introduction

The model developed here is designed to simulate human walking by modulating joint moment and stiffness while using limited feedback control. It is intended that this model be a development tool for walking controllers. Therefore, an important feature of the model is that no *a priori* limitations are placed on joint angle trajectories. This enables the model to respond freely to varying inputs and external disturbances. Joint moment and stiffness inputs are specified as open-loop trajectories, but in the future will be determined by an external walking controller. Some instantaneous feedback is used to maintain stability and an upright posture permitting the model to sustain locomotion, since only by analyzing the quality of walking over several consecutive steps can a controller's performance be assessed.

The philosophy behind the model's design is that it should be as simple as possible, but complex enough to exhibit the behavior that a walking controller must handle. To this end, the key features of the model are 1) the model is 3-dimensional; 2) it has 8 segments and 19 degrees of freedom; 3) each of the 13 degrees of freedom a the 7 joints has passive joint properties restricting movement to normal physiological ranges; 4) the foot is modeled as an ellipsoid and foot-floor contact is modeled as an external force; 5) muscle activation is supplied to the joints as moments and torsional stiffness and damping; and 6) it is a direct dynamic simulation with no *a priori* knowledge of walking kinematics.

The first three features ensure that the model reflects the movement of a human walking. Although most motion during walking is restricted to the sagittal plane, movement in the frontal and horizontal planes, especially of the pelvis, is important for good forward progression while minimizing upper body motion. The passive joint properties effectively limit their ranges of motion, an essential characteristic present in humans and a potentially exploitable feature for a controller.

The foot-to-floor contact computation is facilitated by the use of an ellipsoid shaped foot and thus a single point of contact. This non-physiological foot shape requires different kinematics from those measured in gait studies to walk without the swing leg contacting the floor. More complex foot models have been used [3] and would certainly lead to more physiologically reasonable joint angle trajectories, but the ellipsoid foot is considered an acceptable tradeoff since the objective of controllers will be to adapt and improve walking performance, not necessarily replicate human gait kinematics.

Previous research has demonstrated the importance of the inclusion of joint stiffness modulation for bipedal walking [1], and earlier versions of the model presented here used proportional-derivative (PD) controllers to simulate joint stiffness [8]. But because they require joint angle trajectories to follow, the PD controllers have been

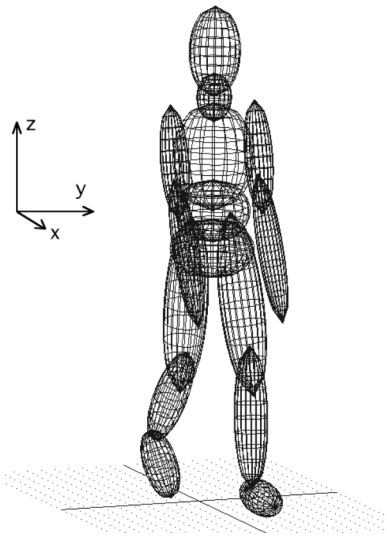

Figure 1: Biomechanical model. Middle and upper trunk sections along with the neck, head and arms are lumped into a single rigid body. Planes: sagittal = x-z, frontal = y-z, horizontal = x-y.

replaced by actuators that supply the joints with moment inputs along with torsional stiffness and damping. The reference angle for joint stiffness is no longer derived from measured walking kinematics, but is determined by the joint's passive properties. These actuators give the model the essential capability to modulate joint stiffness, but avoid the additional complexity and computation of more physiologically based muscle models.

Gilchrist and Winter's [3] model has similarities to the one presented here. It is a direct dynamic stimulation without prescribed joint angle trajectories, but it is driven only by joint moments. It uses some stabilizing feedback control at the hips, but no externally applied control. The joints use constant damping, but no stiffness to complete one step before being becoming too unstable to continue.

Dynamic programming and optimization methods have been shown to generate walking for multiple steps [2, 7, 14], but use measured joint angle trajectories as constraints or objectives. Thus, these models do not provide insight into the sensitivity of walking to muscle activation. However, the successful use of coarse patterns of muscle activation, *i.e.* block patterns, in an optimization formulation [2] implies that the ability to follow joint angle trajectories is not extremely sensitive to muscle activation.

The optimization problem can also be formulated without using any joint angle trajectories [5], although the search for optimal parameters becomes a demanding task, a candidate for the use of a genetic algorithm. But, the use of a genetic algorithm does not guarantee reproducibility of the solution or provide insight into how to change those parameters to adapt to varying model parameters, walking objectives or external disturbances.

The direct dynamic simulation presented here imposes no limitations on the joint angle trajectories to produce walking. The muscle activation patterns are defined in coarse blocks and are determined by trial and error. To demonstrate the type of influence a walking controller can display, a single parameter within the muscle activation pattern is altered resulting in measurable changes in stride length.

2 Biomechanical Model

The model consists of 8 rigid segments connected by joints allowing rotation in either one or three planes. The upper trunk segments, neck, head and arms shown in Figure 1 are lumped into a single rigid body (HAT). The lower trunk segment (pelvis) is connected to the HAT and both legs through joints allowing rotation in all 3 directions, while the ankles and knees are only allowed sagittal plane rotation.

The foot is approximated by an ellipsoid and foot-to-floor contact is modeled by a spring and damper activated by the penetration of the foot into the floor.

Passive joint moments are the total resistance to joint movement due to contact and deformation of tissues that cross the joint when there is no muscle activation. It is position dependent, displays a hysteresis dependent upon the direction of movement, and is not appreciably velocity dependent. Passive joint moments are modeled with the following two equations. One is used when the joint is moving from extension to flexion $M_{ext\to flex}$, the other for flexion to extension $M_{flex\to ext}$. When the joint velocity is near zero, the average of the two moments is used.

$$\begin{aligned} M_{ext\to flex}(\theta) &= A_1 e^{-k_1(\theta-\theta_1)} - A_2 e^{-k_2(\theta-\theta_2)} \\ M_{flex\to ext}(\theta) &= A_3 e^{-k_3(\theta-\theta_3)} - A_4 e^{-k_4(\theta-\theta_4)} \end{aligned} \tag{1}$$

where the subscripted A, k and θ are constants determined by fitting published experimental data [4, 9-11, 15]. Note that there is no linear damping term, but there is a loss of energy and thus a small damping effect produced by the switching between the two expressions.

The upper body controller is an external force and moment that provides stability to the HAT. Proportional-derivative (PD) controllers are active at the 3 rotational degrees of freedom at the model's center of mass. A PD controller along the global y-axis provides frontal plane stability. In addition there is a force proportional to the model's *velocity* in the x-direction; the position in the forward walking direction is not controlled. Since there is no control force in the vertical direction, the lower limbs must support the model's full weight.

2.1 Joint Actuators

When skeletal muscle is activated, not only does it contract generating force to move a joint, but its intrinsic properties change increasing its stiffness. In fact, two activated antagonist muscles may produce no joint movement, but substantially increase a joint's stiffness. The proper modulation of joint stiffness due to muscle activity could be such an important property that without it stable walking cannot be achieved.

One approach to include active muscle stiffness is to explicitly incorporate a muscle model with force-length and force-velocity characteristics. This type of model, however, incurs an additional computation burden for each muscle. In addition, there is the matter of determining the moment arm of each muscle. These complexities were avoided by the direct application of moment and stiffness to the joints.

Each actuated degree of freedom has two actuators, a flexor and an extensor. They model equivalent muscles, meaning they incorporate all the physiological muscles contributing to the joint dynamics. An individual actuator supplies the joint with a moment and torsional stiffness and damping.

Stiffness is defined as a function of the applied moment. Second order model fitting of isometric muscle contractions at the knee [16] and at the ankle [12, 13] reveals an approximately linear relationship between stiffness and moment and a nearly constant damping ratio.

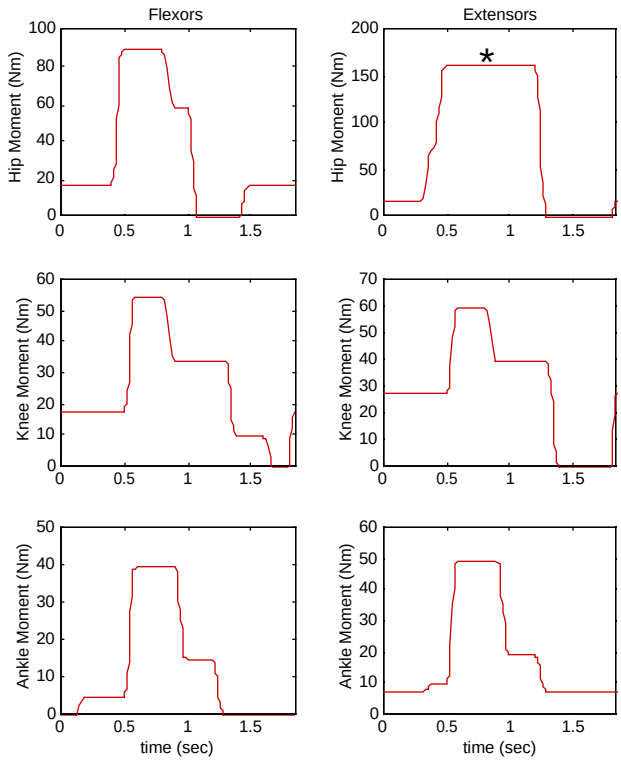

Figure 2: Actuator Inputs for the Right Leg. The asterisk (*) in the hip extensor plot indicates the segment of activation that is modified during the sensitivity analysis.

Hence, the joint's torsional stiffness is defined to be directly proportional to the actuator's applied moment.

$$K_f = c_f \cdot M_f$$

$$K_e = c_e \cdot M_e$$

K is the stiffness due to an actuator, M is the actuator's moment, c is a constant defined for each actuator and the subscript corresponds to the flexor or extensor actuator.

The constant damping ratio of $\zeta = 0.2$ is used.

$$B_f = 2\zeta\sqrt{K_f J_0}$$

$$B_e = 2\zeta\sqrt{K_e J_0}$$

B is the joint damping with the subscript again referring to flexor and extensor actuators. J_0 is the moment of inertia at the joint of all segments distal to the joint.

The final parameter that must be defined is the stiffness reference angle θ_{ref}. It is defined as a function of the net applied moment ($M_{net} = M_f - M_e$) using the passive joint properties. The inverse mapping of the passive moment (the mean of equations 1) is used to compute the angle the joint would be at when subjected to a net moment of M_{net} and no external disturbances. This angle is used as the stiffness reference angle.

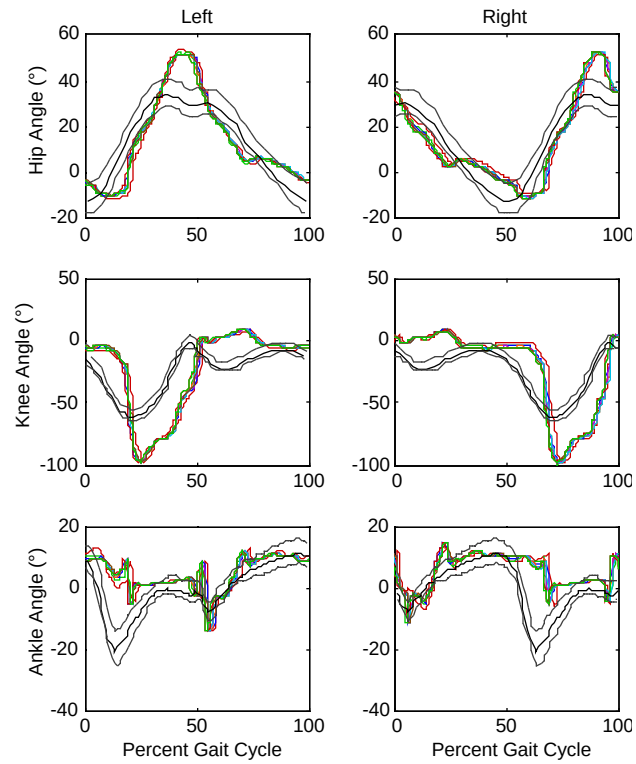

Figure 3: Simulated and Normal Joint Kinematics. The sagittal plane joint angles for each gait cycle (beginning and ending with right heel strike) are superimposed and plotted relative to percent. The solid line bounded by the dashed lines depicts normal data, mean ± 1 standard deviation. Axis orientation: hip flexion, knee extension and ankle dorsiflexion are positive.

The assumption is made that the flexor and extensor actuators' moments, stiffness and damping sum across the joint. Thus, the complete actuator input to a joint as a function of joint angle θ and angular velocity ω is

$$\begin{aligned} M(\theta,\omega) &= M_f - M_e \\ &+ (K_f + K_e)(\theta_{ref} - \theta) \\ &+ (B_f + B_e)\omega \end{aligned}$$

To generate locomotion, the actuator inputs are prescribed for a complete gait cycle beginning and ending close to right heel strike, Figure 2. The actuator input is repeated each 1.85 s, the chosen gait cycle duration. The inputs for the two legs are the same except for being offset by approximately 50% of the cycle. The activation pattern was determined by trial and error when the peak hip extensor level was 165 Nm. Once acceptable walking was achieved, additional simulations were conducted with different peak hip extensor activation to examine the changes in gait.

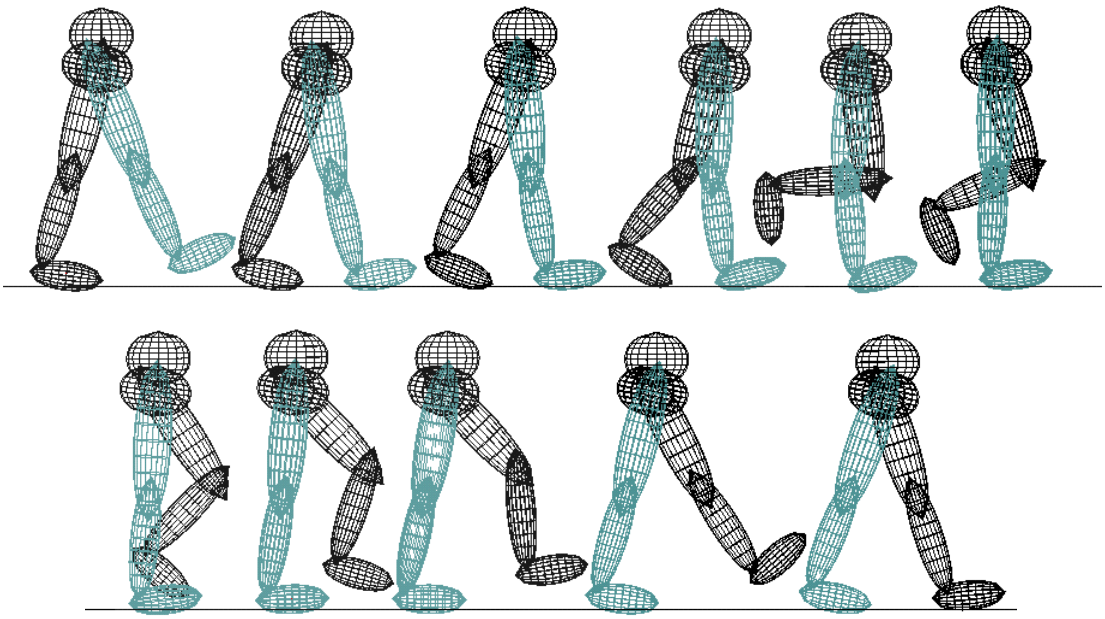

Figure 4: Snapshots of bipedal gait. The snapshots are spaced by 0.1 seconds. The right leg is cyan/gray. The HAT segment is a single ellipsoid.

3 Results

The model has attained continuous sustained locomotion. The longest simulation, 100 seconds and 54 gait cycles, exhibits cyclic kinematics that can be sustained indefinitely. Figure 3 shows the joint angle trajectories generated for 9 consecutive steps. The model exhibits some kinematic variability between steps, but not as much as normal gait [6]. Compared to normal gait the model has excessive hip and knee flexion during swing. There is additional ankle movement when the body's weight is transferred to the foot. In addition, the ankle does not plantarflex as much around toe off. At and shortly after heel strike, the knee hyperextends instead of flexing appropriately to absorb the energy from the impact of foot contact.

These deviations from normal gait are clearly depicted in the animation of one step in Figure 4. What appears to be excessive anterior pelvic tilt is an artifact of the ellipsoids used for visualization, the mean anterior pelvic tilt is near the normal 10°, while the magnitude of the angle excursion is 7°, slightly larger than the normal 4°.

The amount of upper body control used is depicted in Figure 5. The control is repeatable from cycle to cycle. The forces are cyclic, synchronized with each step, while the moments are not. The most control is used to correct the rotation in the model's sagittal plane.

Initial tuning of the actuator activation patterns was performed without the forward velocity controller resulting in between three and six steps. Walking was

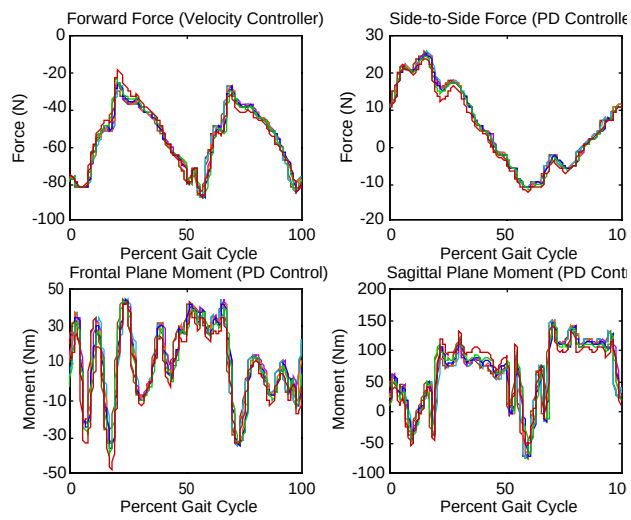

Figure 5: Upper Body Controller Input. Data for 8 consecutive cycles are superimposed and plotted vs. percent gait cycle. Horizontal plane moment is not shown.

much more sensitive to activation without the forward velocity controller. Once the velocity controller was implemented, sustained walking was readily achieved.

Figure 6 shows the effect of changing the hip extensor's activation. The activation was changed at the time within the gait cycle after the center of mass has passed over the supporting leg and before the foot lifts off to take a step, *i.e.* around terminal stance. The increase in

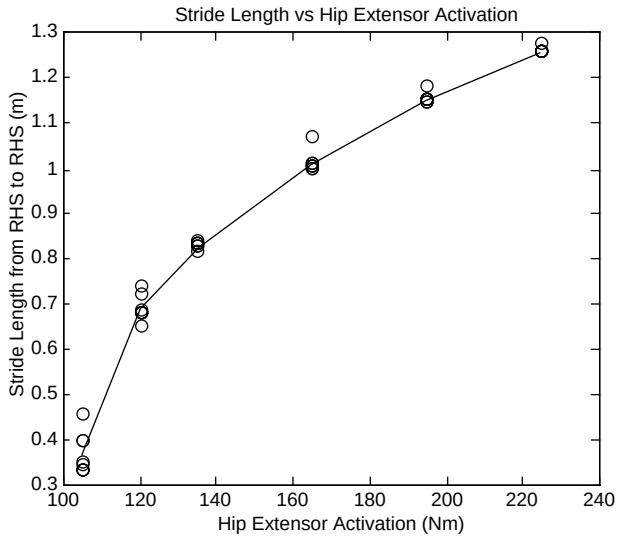

Figure 6: Stride Length vs. Hip Extensor Activation. There is a circle for each stride. The line connects the mean of the data points at each activation level.

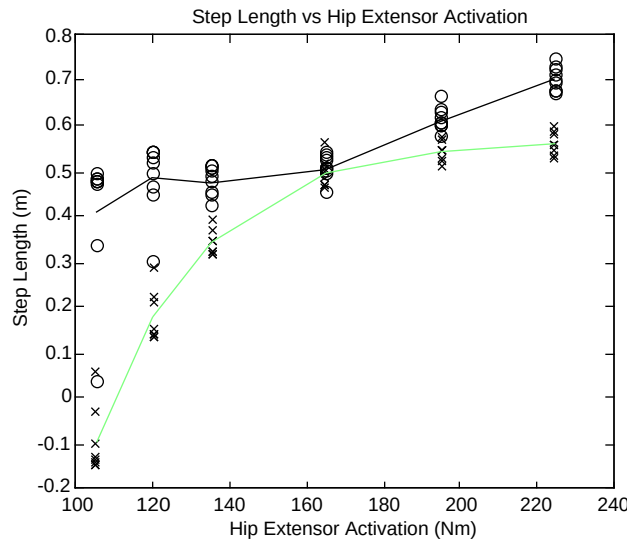

Figure 7: Left and Right Step Length vs. Hip Extensor Activation. Right steps are circles. A solid line connects their means. Left steps are marked with Xs and are connected by a (green) dash-dot line.

stride length as a function of the extensor activity is evident.

Figure 7 shows a similar relationship for the right and left step lengths. Despite attempts to design a symmetric activation pattern, the walking is not consistently symmetric. This asymmetry becomes more apparent at activation levels other than the one used to adjust the initial activation pattern.

4 Conclusion

During model development tradeoffs are made regarding simplifications to make the model tractable and complexity to accurately reflect the behavior being simulated. Since the model here will be used to develop walking controllers, the exact kinematics generated by the model are not as important as its ability to sustain walking and produce qualitative changes in response to varying input. In addition, it is essential that the walking be generated without *a priori* knowledge of the walking kinematics.

A mechanism to modulate joint stiffness as a model input is necessary to accomplish these goals. The approach chosen here is to design joint actuators that directly apply moments, stiffness and damping to the actuated joints. This simpler alternative is chosen over a more accurate and complete model that includes neural activation and force generating muscles. By this direct method of including joint stiffness and damping instead of being an emergent property of the muscles, it is possible to directly examine the role of joint stiffness and damping due to muscle activation during walking. The actuator model, however, does not include any activation dynamics such as the time delay between activation and force generation. The inclusion of these effects may be added as the capability of a walking controller to address them becomes more important.

The inclusion of passive joint properties is necessary for the model to emulate human gait. They most obviously come into play at the knee where the passive moment prevents the knee from further hyperextension in stance. Omission of the passive properties would require a very different pattern of activation to generate comparable output.

The ellipsoid foot shape is a modeling a tradeoff. In exchange for simplifying foot contact computation, joint kinematics, especially those at the ankle, are forced to deviate from normal trajectories. This tradeoff is acceptable since walking controllers will not be designed to replicate normal joint kinematics, but to achieve higher-level goals while adapting to model and environmental variability.

Ideally, the model would include no stabilizing feedback control. But to maintain an upright posture, feedback moments are required in all 3 directions at the body's center of mass while feedback force in the frontal plane must provide side-to-side stability. The inclusion of feedback control to check the forward velocity is essential for the model to sustain its locomotion for more than a few steps. It is unclear if this upper body feedback control is needed only to compensate for the coarse, manually tuned activation patterns attempted here, or if it is required for all but a highly refined, high resolution set of

inputs. An initial analysis of the feedback control indicates that the magnitude of the forces and moments varies with the quality of walking. The forward velocity controller uses more force at longer stride lengths. The stabilizing moment in the sagittal plane also exhibited variation at the observed stride lengths, but more investigation is needed to understand how the amount of upper body control relates to the quality of observed locomotion and how it might be used as a walking performance indicator.

The asymmetric gait is an unexpected finding. The step lengths are unequal especially as the hip extension deviates from its originally tuned value. Although the activation pattern is grossly symmetric between the two legs, limitations in the way it is represented means that at its finest resolution, it is not perfectly symmetric. The implication of this finding is that gait may be more sensitive to the timing than to the magnitude of activation. Changes of actuator timing on the order of 50-100 ms have measurable effects on walking quality.

The model does exhibit the anticipated behavior of longer stride lengths as a function of increasing hip extension. Since locomotion is sustained over a large range of hip extensor activity levels, a walking controller should be able to evaluate the gait and adjust the activation to make the model walk at a desired stride length.

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