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# Integration of an Adaptive Swing Control into a Neuromuscular Human Walking Model*

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***Abstract*** **Understanding the neuromuscular control underlying human locomotion has the potential to deliver practical controllers for humanoid and prosthetic robots. However, neurocontrollers developed in forward dynamic simulations are seldom applied as practical controllers due to their lack of robustness and adaptability. A key element for robust and adaptive locomotion is swing leg placement. Here we integrate a previously identi ed robust swing leg controller into a full neuromuscular human walking model and demonstrate that the integrated model has largely improved behaviors including walking on very rough terrain ($\pm$10cm) and stair climbing (15cm stairs). These initial results highlight the potential of the identi ed robust swing control. We plan to generalize it to a range of human locomotion behaviors critical in rehabilitation robotics.**

## I. INTRODUCTION

Understanding the neuromuscular control underlying human locomotion has the potential to advance the state of the art in different elds. It can lead to new rehabilitation methods [1], [2], provide simulation testbeds which realize virtual experiments dif cult or impossible to conduct with human subjects [3] [5], and deliver practical controllers for humanoid and prosthetic robots [6] [9]. Since the human neural control architecture is dif cult to identify directly, several research groups develop computational models of neuromuscular control to propose and test speci c control architectures. For instance, inspired by the observation of central pattern generators (CPGs) in neurophysiological studies [10], [11], neural control architectures based on CPGs and feedback pathways have been proposed in simulation studies and have been found to generate walking and running [12] [16]. Similar results have been obtained [17] testing neuromuscular control based on the equilibrium point hypothesis [18] and on interpreting principles of legged dynamics and control with muscle re exes [19], [20]. However, all these controllers have so far produced only limited robustness and adaptability required for real world applications and, thus, are seldom applied for control in robotic systems.

A key element for robust and adaptive locomotion is swing leg placement; it is, for instance, critical in maintaining gait stability in legged systems that encounter large disturbances. Simpli ed models of dynamic balance such as the linear inverted pendulum model and the spring-mass model [21] [23] can be used to predict these target placements. But these models do not reveal how humans control their segmented legs to reach the targets. In a recent study, we have idealized the problem of swing leg placement by studying a double pendulum system that is hinged at the hip and found a control that achieves robust leg placement into arbitrary target points on the ground under large disturbances [24].

Here we integrate this swing leg controller into our full neuromuscular human walking model, and explore its potential of generating different and robust locomotion behaviors. In section II, we brie y review the previous human model and the identi ed swing leg controller. We then present our work on integrating the two in section III and show in section IV that the resulting model not only generates steady locomotion at normal walking speeds, but also negotiates very rough terrain and climbs stairs. Finally, we discuss future directions of this work in section V.

## II. PREVIOUS NEUROMUSCULAR WALKING MODEL AND ROBUST SWING CONTROL

### A. Previous Neuromuscular Model

The musculoskeletal system of the previous model consists of 7 segments (trunk, thighs, shanks and feet) and 6 internal degrees of freedom (hip, knee and ankle joints) [19] (Fig. 1). The joints are actuated by seven Hill-type muscle models per leg, ve of which are monoarticular muscles (soleus, SOL; tibialis anterior, TA; vastus, VAS; gluteus maximus, GLU; and grouped hip exors, HFL) and two of which are biarticular ones (gastrocnemius, GAS and hamstring group, HAM). The contractile elements of the muscle models take stimulation signals $S_{\mathsf{m}}$ between 0 and 1, which generate muscle forces that translate into joint torques, $\tau_{\mathsf{m}} = F_{\mathsf{m}}\, r_{\mathsf{m}}(\varphi)$, where $r_{\mathsf{m}}(\varphi)$ estimate the variable moment arms observed in physiology. The ground contacts and joint limits are modeled as nonlinear spring-dampers.

The muscle stimulations $S_{\mathsf{m}}$ are the outputs of the neural control architecture of this model. The full control network can be categorized into four control groups based on their functionalities: trunk balance, stance, swing initiation, and swing control. The trunk balance control is active proportionally to the load the leg is bearing during the stance phase; the stance control is active throughout the stance phase; swing initiation is active during the double support phase of late stance phase; and the swing control is active during the swing phase (Fig. 1-a). Most of the sensory re ex pathways are local positive force or length feedbacks, $F+$ or $L+$ (Fig. 1-b). These sensory and stimulation signals are time-delayed, to model neural transport delays. (See [19] for more details on this model.)

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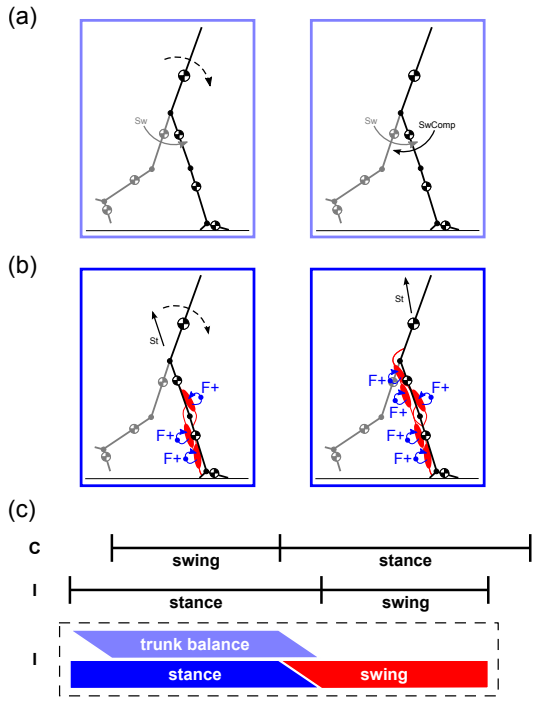


Fig. 3. Modi cations made in the neurocontroller. Feedforward controllers are added to (a) the trunk balance control to compensate moments induced by the swing leg and (b) the stance control to compensate ground reaction forces transferred by the stance leg. In addition, the explicit swing initiation control required in the previous neuromuscular control is now removed (c).

control activity reduces in proportion to the load that the other (front) leg is bearing while the activity of the swing control increases by the same amount (Fig. 3-c).

## IV. LOCOMOTION BEHAVIORS

We explore the swing controller's potential for generating different locomotion behaviors including steady walking on at ground, walking across very rough terrain, and climbing up stairs. For all three behaviors, a single set of internal swing control parameters is used based on the hand-tuned constant values identi ed in [24] (section II-B). In contrast, different sets of the swing control parameters $\alpha_{\mathsf{tgt}}$ and $l_{\mathsf{clr}}$ as well as the stance control parameters are identi ed for the three individual behaviors using optimization with the covariance matrix adaptation evolution strategy (CMA-ES, [27]). The optimization uses three different cost functions for the three behaviors, samples 64 sets of the parameters based on a covariance matrix of one generation, runs individual simulations to calculate corresponding values of the cost function, and uses the best 32 sets to update the covariance matrix for the next generation. The procedure repeats for a total of 400 generations.

The cost function we use for steady walking is

$$J_{\mathsf{steady}} = c_1 \ |\dot{x}_{\mathsf{avg}} - \dot{x}_{\mathsf{tgt}}| + c_2 \ C_{\mathsf{E}}, \qquad (2)$$

where $\dot{x}_{\mathsf{avg}}$ and $\dot{x}_{\mathsf{tgt}}$ are average and target walking speeds, $C_{\mathsf{E}}$ is the energetic cost, and the coef cients $c_1$ and $c_2$ are empirically determined constants (10 and 1, respectively). $\dot{x}_{\mathsf{avg}}$ and $C_{\mathsf{E}}$ are computed during multiple consecutive steps of steady walking. $C_{\mathsf{E}}$ is computed from $C_{\mathsf{E}} = E_{\mathsf{M}}/(m x_{\mathsf{d}})$, where $E_{\mathsf{M}}$ is the total metabolic energy consumed by all muscles (using the energy model in [28]), $m$ is the body mass, $x_{\mathsf{d}}$ is the walking distance traveled. For crossing very rough terrain and climbing stairs, the cost functions are de ned as

$$J_{\mathsf{rough}} = -x_{\mathsf{end}} \qquad (3)$$

and

$$J_{\mathsf{stair}} = \begin{cases} -c_{\mathsf{L}}, & \text{if steady} \\ -x_{\mathsf{end}}, & \text{o.w.} \end{cases} \qquad (4)$$

where $x_{\mathsf{end}}$ is the distance travelled by the human model in the forward direction and $c_{\mathsf{L}} = 100$ is a large constant rewarding steady stair climbing.

Without changing the internal swing control parameters, the model achieves all three locomotion behaviors (Fig. 4). The model generates steady walking at normal human walking speeds ($1.4ms^{-1}$, Fig. 4-a), robustly travels over terrain with randomly generated large and frequent changes in ground height (changes every $1m$, observed maximum height changes of $+12cm$ and $-9cm$, Fig. 4-b), and steadily climbs stairs (arbitrarily chosen to be $50cm$ apart and $15cm$ high, Fig. 4-c).

## V. FUTURE DIRECTION

The initial results show the potential of the identi ed swing control in generating robust and adaptive locomotion behaviors. Our long-term goal is to generalize this work and to identify a neuromuscular control architecture that combines a large range of robust and adaptive locomotion behaviors critical to humanoid and rehabilitation robotics. Toward this goal, our next step is to translate the torque controller of the swing leg into a neuromuscular one. For this, we plan to adapt the muscle-re ex control derived in [29] for the idealized double-pendulum swing leg system.

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