Muscle-reflex model of human locomotion entrains to mechanical perturbations

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***Abstract*— Prior experiments have shown that human gait synchronizes to periodic torque pulses applied about the hip and ankle joints by robotic exoskeletons. Importantly, entrainment occurred even when the pulse period differed slightly from the user's preferred stride period, making it a viable approach to increase gait speed. As gait speed is an important outcome of gait therapy, gait entrainment to mechanical perturbations may serve as a promising new method of robot-aided therapy. Still, an understanding of the underlying neuromechanical processes that give rise to gait entrainment is needed to fully evaluate its therapeutic potential. To gain such insight, the goal of this paper was to evaluate whether an existing neuromechanical model of human locomotion exhibited entrainment behavior similar to that observed in the prior human experiments. Simulation results showed that the model entrained to pulses applied at both the ankle and hip joints. The convergence of relative phase between model gait and hip perturbations was similar to that observed with the human gait, but differed slightly for ankle perturbations. Thus, models that can more accurately describe neuromechanical interactions between human gait and robotic exoskeletons are still needed. Nevertheless, the simulation results support the notion that the limit-cycle behavior observed during locomotion does not require supra-spinal control or a self-sustaining oscillatory neural network, which has important implications for improving gait therapy.**

I. INTRODUCTION

Locomotion is a fundamental behavior of biological systems. Impairments in locomotor function can present for a multitude of reasons, including injury, neurological disease, or even simply aging, and when they do, it can significantly impact one's quality of life. Thus, highly effective methods for gait rehabilitation are greatly needed.

Wearable robotic exoskeletons hold great promise for improving gait therapy. However, current approaches to robot-aided gait rehabilitation have shown limited efficacy [1], [2], especially when compared to the successful use of robots in upper-limb therapy [3]–[6]. One possible reason is that current approaches, which impose kinematic trajectories onto the limbs, might disrupt the natural oscillatory dynamics of walking. Unimpaired human walking is inherently rhythmic and exhibits features consistent with those of nonlinear limit cycle oscillators. One such feature is dynamic entrainment, where the period of an oscillator synchronizes to that of an imposed oscillation with a particular phase relation (i.e. phase-locking). The use of rhythmic cues has been shown promising for improving gait. For instance, rhythmic auditory stimulation has been shown to increase gait speed and reduce gait asymmetry in stroke patients [7]. Rhythmic, mechanical perturbations applied by an ankle exoskeleton have also been shown to increase stride frequency in those with neurological impairments [8]. Still, a better understanding of underlying neuromechanical dynamics and control of human locomotion is required to develop effective gait therapies based on entrainment.

Along with behavioral experiments, modelling can be a complementary source of insight. Not only can models aid our understanding of how the behavioral observations arise, but they can also produce new predictions to inform the design of subsequent experiments. In previous work, Ahn and Hogan [11] identified a minimal, one-degree-of-freedom model (point mass with two massless legs with ankle actuation via a linear torsional spring) that could reproduce the limit-cycle behavior of human locomotion, including entrainment to mechanical perturbations about the ankle joint. The advantage of highly simplified models such as this is that often they afford a more intuitive understanding of system behavior. Reduced complexity, however, often comes at the cost of accuracy. For instance, the aforementioned model ignores human physiology. To fully understand the emergence of entrainment and how it might be utilized for locomotor therapy, more-detailed, neuro-mechanical models are also needed.

The purpose of this study was to determine whether the muscle-reflex model of locomotion developed by Geyer and Herr [12] could reproduce the entrainment to periodic torque pulses delivered by robotic exoskeletons about the ankle and hip joints as observed in prior human experiments. The version of muscle-reflex model examined here did not have a supra-spinal controller or rhythmic central pattern generator, similar to the simplified model proposed by Ahn and Hogan [11]. This allowed us to examine how entrainment may emerge from peripheral neuromechanics during locomotion.

The remainder of the paper is organized as follows. Section II summarizes the key features of experimentally observed behavior used to evaluate the competency with which the model can describe entrainment. Section III briefly describes the muscle-reflex model and details simulation conditions evaluated. Section IV presents the simulation results, and Section V discusses their implications and the insights they provide for improving robot-aided gait therapy.

*B.A. was supported by a Dean's Fellowship from the College of Engineering at UMass Amherst. J.L. was supported by a Samsung scholarship. N.H. was supported by the Eric P. and Evelyn E. Newman fund, NIH-R01-HD087089, NSF-NRI 1637824, and NSF-CRCNS-1724135.

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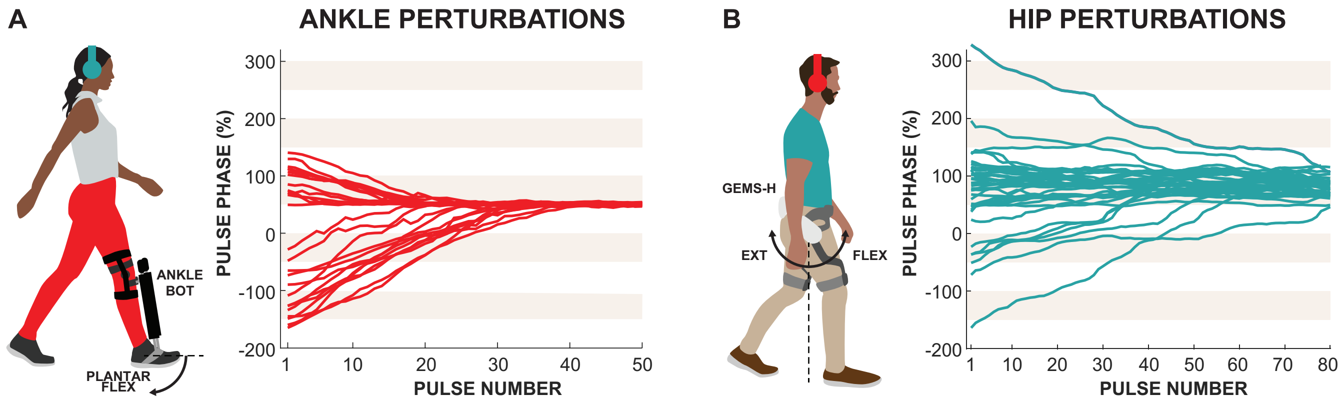

Fig. 1. Experimental setup and plot of pulse phase over pulse number from all entrained trials in a representative (A) ankle perturbation experiment [9] and (B) hip perturbation experiment [10]. Each line represents one entrained trial. Headphones were used in both experiments to block any sound from the robotic exoskeletons or the environment.

II. SUMMARY OF PRIOR EXPERIMENTAL RESULTS

Behavioral experiments have shown that human gait entrains to periodic torque pulses applied to either the hip or ankle by lower-extremity robotic exoskeletons. Here, we provide a summary of those experimental results and outline the key features used to evaluate a model's ability to describe human gait entrainment.

A. Human Gait Entrainment to Ankle Perturbations

Ahn and Hogan showed that unimpaired humans entrained their gait to a series of periodic mechanical perturbations at the ankle joint delivered by a robotic ankle exoskeleton (InMotion Ankle, Bionik Laboratories Inc.) while walking on a treadmill [13]. These perturbations consisted of torque pulses applied unilaterally to the right ankle joint in the plantarflexion direction (pulse magnitude: 10 Nm; pulse duration: 100 ms). Entrainment was observed for pulse periods both shorter and longer than a person's preferred stride period, but only when the pulse period was sufficiently close (within 6.7%) to that person's preferred stride period. In a follow-up study, Ochoa et al. confirmed the findings of [13] and showed that humans also entrained their gait to ankle perturbations during overground walking [9]. In fact, entrainment occurred more often and more rapidly during overground walking. Entrainment was observed for pulse periods 50 ms shorter and longer than the person's preferred stride period. Another robust observation across both studies in all walking environments was that human subjects entrained their gait such that the pulse phase occurred at 50% of the gait cycle (Fig. 1A). This phase coincided with right leg terminal stance, or ankle push-off, where the plantarflexion torque from the robot would assist forward propulsion.

The simplified model proposed by Ahn and Hogan [11] could reproduce human entrainment to ankle perturbations for pulse periods shorter than the preferred pulse period and phase-locking to ankle push-off. It could not, however, replicate human entrainment to pulse periods longer than the preferred stride period.

B. Human Gait Entrainment to Hip Perturbations

After observing entrainment to periodic ankle perturbations, Lee et al. [10] found that human gait also entrained to periodic torque pulses applied to the hip joints. In that study, periodic hip torque perturbations were applied bilaterally (flexion to the right hip and extension to the left hip) using the Samsung GEMS-H exoskeleton (pulse magnitude: 6 Nm, pulse duration: 250ms). The pulse period was set to 25 ms shorter than the person's preferred stride period; longer pulse periods were not tested. Similar to the ankle perturbation experiments, entrainment was observed and the relative phase between gait and the perturbations converged to $\sim$78% on average[1] (Fig. 1B). This phase coincides with the midswing phase of the right leg, where the flexion robotic torque applied to the right leg would be assistive.

C. Summary of Model Evaluation Criteria for Entrainment

Based on the aforementioned experimental results, a successful model of human locomotion should produce stable walking patterns and exhibit the following: (1) entrainment to periodic perturbations delivered about the ankle and the hip joints within a finite basin of entrainment, and (2) consistent phase-locking to a specific phase relation (ankle perturbations: right leg terminal stance; hip perturbations: right leg midswing).

[1] In Lee et al. [10], 0% of the gait cycle was defined as left leg toe-off, whereas in this paper, 0% is defined as right heel strike. Thus, phase values from [10] have been converted to follow the convention of this paper. It was also previously reported in [10] that phase-locking occurred at two distinct phases in the gait cycle. Estimating phase, however, is notoriously difficult and noisy. Gregg and colleagues [14]–[16] recently proposed reliable and robust methods to determine a phase variable for human walking, using hip angle. When gait phase was recalculated using their improved method, lower variability was observed since the noise introduced by numerical analysis was reduced [17]. As a result, the distribution of terminal pulse phases became more 'clustered' towards a unimodal distribution, indicating that phase-locking occurred at one distinct phase in the gait cycle, rather than two as previously reported. Interestingly, the mean terminal pulse phase was minimally affected.

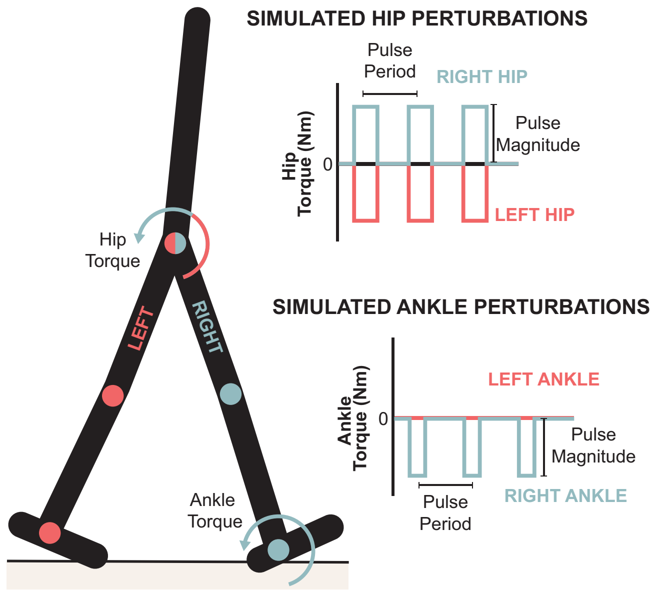

Fig. 2. Simulation setup. The muscle-reflex model developed by Geyer and Herr [12] was simulated with periodic torque pulses applied either (1) unilaterally to the right ankle joint in plantarflexion or (2) bilaterally to the right and left hip joints in flexion and extension, respectively.

III. METHODS

A. Muscle-Reflex Model of Locomotion

Geyer and Herr [12] developed the muscle-reflex model to translate a simplified locomotion model (bipedal spring-mass model [18]) into one that more accurately reflects human physiology. The model encodes human locomotor behavior using muscle reflexes based on known neurophysiology. The model depicts human locomotion with a trunk, two three-segments legs and seven Hill-type muscles on each leg. Overall, this model can generate stable walking behavior that is also robust to ground disturbances. Importantly, it also replicates the joint kinematics, joint moments, ground reaction forces, and muscle activity of human walking with reasonable accuracy. Here, we examined whether this model could competently describe entrainment of human gait to mechanical perturbations.

B. Simulation Methods

The muscle-reflex model was simulated in Simulink (MATLAB R2019a, Mathworks, Natick, MA, US). The model has several hyperparameters that can be tuned (e.g., geometric characterizations and mechanical properties of the segments, the kinematic initializations, joint motion limits, muscle controls and foot-ground interaction parameters). For this study, all hyperparameters were kept at their default values. As described in [12], these default values have been shown to produce human-like gait. Note the mass of the exoskeleton devices was not included in the model.

For each simulated trial, the simulation stop time was set to 160 s, which resulted in approximately 140 walking strides. The model was not perturbed during the first 40 strides until it reached steady-state behavior. Over the next 80 strides, perturbations in the form of torque pulses were applied either to the ankle or hip joints as described below. The perturbations were removed for the remaining 20 strides.

C. Ankle Perturbation Simulations

A total of 55 simulated trials were run to assess how the magnitude and period of the ankle torque pulses affected entrainment. To match the human experiments [9], [13], the torque pulses were delivered unilaterally about the right ankle joint in the plantarflexion direction (Fig. 2). Torque pulse magnitude ranged from 2 Nm to 10 Nm with a resolution of 2 Nm. Torque pulse period ranged from $\sim$88 ms shorter to $\sim$17 ms longer than the natural stride period of the model of 1.168 s (i.e., 1.08 s to 1.185 s) with a resolution of 10 ms. The range of pulse periods to investigate was based on an initial low-resolution parameter sweep in order to focus the analysis on a region of the parameter space where the model did not fail. The duty cycle of the pulses was set to 10% of the natural stride period to mimic the human experiments. In all simulated trials, the initial pulse occurred at 46.5% of the gait cycle.

To assess phase-locking, an additional eleven trials were simulated. In these trials, the initial phase of perturbation onset was varied by changing the start time of initial pulse (with a 100ms resolution). The pulse magnitude and period were held constant (10 Nm and 1.15 s, respectively). These values were chosen to mimic the conditions of the human experiments as best as possible without causing the model to fall.

D. Hip Perturbation Simulations

A total of 75 simulated trials were run to assess how the magnitude and period of the hip torque pulses affected entrainment. This time, however, the torque pulses were delivered bilaterally to the hip joints to mimic the human experiments [10]. The torque pulses were applied in the flexion direction for the right hip joint and in the extension direction for the left hip joint (Fig. 2). The magnitude values ranged from 2 Nm to 10 Nm with a resolution of 2Nm, and the period values ranged from 1.05 s to 1.415 s (i.e., $\sim$118 ms shorter to $\sim$247 ms longer than the natural stride period of the model) with a resolution of 25 ms. The duty cycle of the pulses was set to 22.7% to mimic the human experiments. In all simulated trials, the initial pulse occurred at 17.1% of the gait cycle.

To assess phase-locking, an additional nine trials were simulated. In these trials, the pulse magnitude and period were held constant (10 Nm and 1.15 s, respectively) to approximate the conditions of the human experiments, and the initial phase of perturbation onset was varied as in the ankle perturbation simulations.

E. Gait Phase Definition

Within every trial, each stride was time-normalized from 0 % to 100 % to define gait phase. 0 % of the gait cycle was defined as the right leg's heel strike. From 0 % to 60 % of the gait cycle, the right leg was in stance phase, while from 60 % to 100 % of the gait cycle, the right leg was in swing.

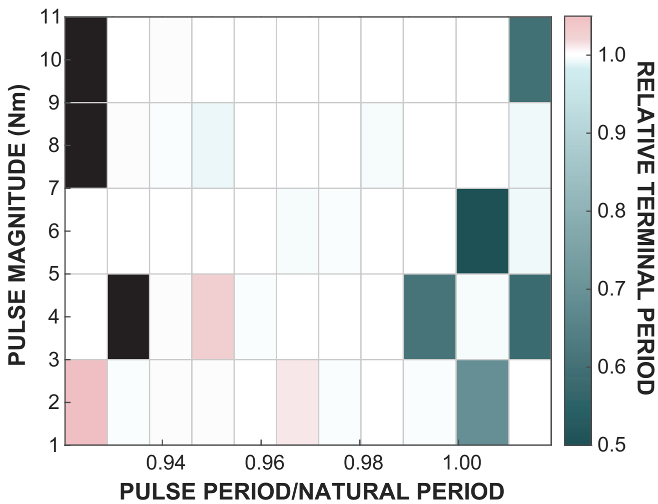

Fig. 3. The effect of pulse magnitude and period on model entrainment in ankle perturbation simulations. Each colored square depicts a simulated trial with a given torque pulse magnitude and period. The color axis depicts the relative terminal period. A ratio of approximately unity indicates entrainment with 1:1 frequency-locking (shown in white). Squares in black indicate trials where the simulation failed (i.e., when the model became unstable before the simulation stop time). In trials where relative terminal period significantly deviates from unity (i.e., squares in dark green), the model can exhibit behavior that diverges from typical walking (e.g., shuffling or a non-periodic pattern), especially if it is on the verge of falling over.

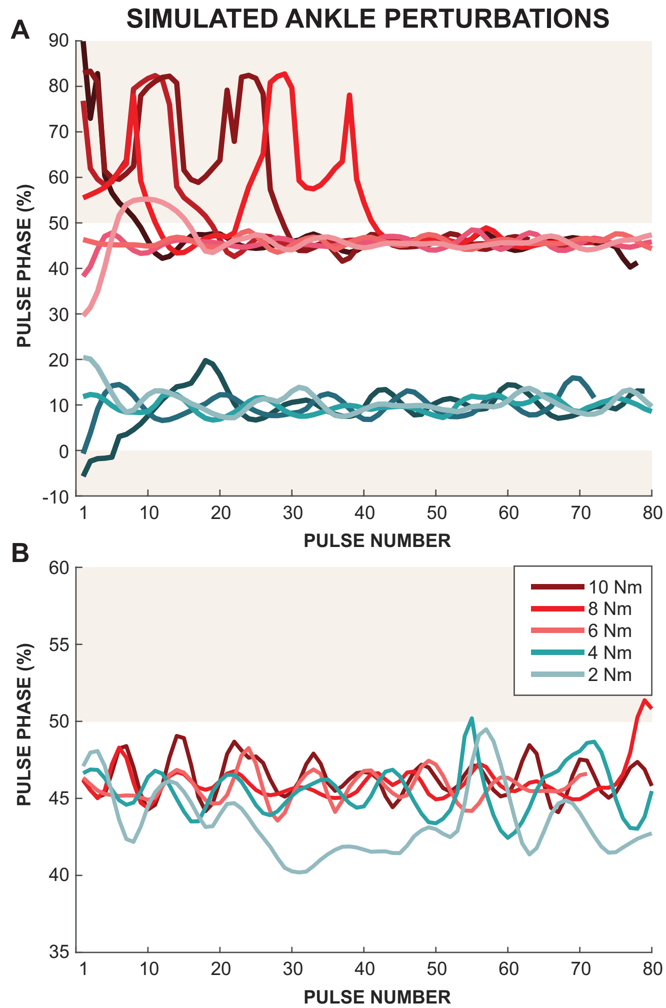

Fig. 4. (A) The effect of initial pulse phase on terminal pulse phase in the ankle perturbation simulations. Each line represents a trial simulated with a different initial pulse phase. Torque magnitude was set to 10 Nm. (B) The effect of pulse magnitude on entrainment in ankle perturbation simulations. Each line represents a trial simulated with a different pulse magnitude. For all trials in panels (A) and (B), the pulse period was set to 1.15 s (i.e., ∼18 ms shorter than the natural period of the model).

F. Assessment of Entrainment

1) Basin of Entrainment: For a given pulse magnitude, the basin of entrainment was defined by the range of pulse periods in which entrainment with 1:1 frequency-locking was observed. To assess entrainment, the relative terminal period, defined as the ratio of the mean terminal stride period to torque pulse period, was used. Mean terminal stride period was defined as the average stride period of the model from the last 10 strides before the offset of the torque pulses. A ratio of approximately unity indicated entrainment with 1:1 frequency-locking.

2) Phase-Locking: Entrainment with 1:1 frequency-locking also implies 1:1 phase-locking (i.e., convergence to a constant phase relation). It does not, however, imply that the model must converge to any one, particular phase. Because convergence to a specific phase was observed in human experiments, mean pulse phase (i.e., phase at the onset of each pulse) during the terminal strides was also assessed.

IV. SIMULATION RESULTS

A. Entrainment in Ankle Perturbation Simulations

1) Basin of Entrainment: The muscle-reflex model did entrain to the periodic torque pulses applied about the ankle. Consistent with nonlinear limit-cycle behavior, the range of pulse periods to which the model entrained was finite. This basin of entrainment in terms of pulse period was narrower than what has been previously observed experimentally. For a pulse magnitude of 10Nm as used in the human experiments, the model entrained to pulse periods between ∼78 ms shorter and ∼7 ms longer than its natural period (Fig. 3). In the human experiments, gait entrainment to pulse periods 50 ms shorter and longer was observed [9]. As seen in Fig. 3, lowering the pulse magnitude tended to reduce the basin of entrainment.

2) Phase-locking: In the human experiments, the pulse phase consistently locked to ∼50% of the gait cycle (right leg terminal stance). In the model simulations, however, the pulse phase converged to two distinct phases: 10.7% (right leg loading response) and 45.5% (right leg terminal stance) (Fig. 4A). The converged phase relation depended on the initial phase (Fig. 4A), but not pulse magnitude (Fig. 4B).

B. Entrainment in Hip Perturbation Simulations

1) Basin of Entrainment: Simulation results showed that for a pulse magnitude of 6 Nm (as used in the human experiments), the model entrained at pulse periods between ∼93 ms shorter and ∼197 ms longer than its natural period (Fig. 5). To date, human experiments have only tested entrainment

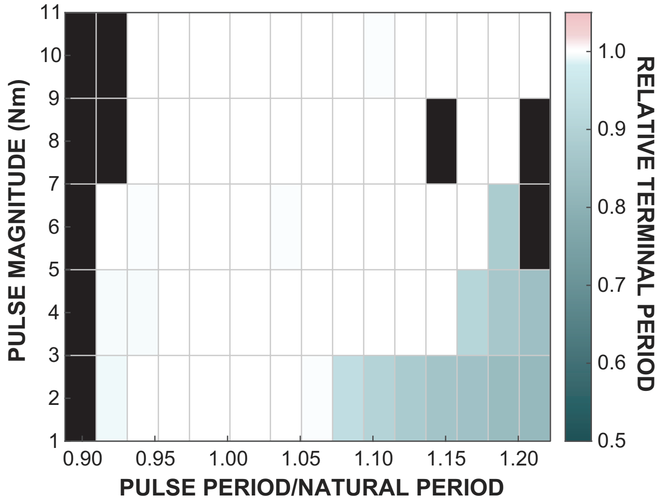

Fig. 5. The effect of pulse magnitude and period on model entrainment in hip perturbation simulations. Each colored square depicts a simulated trial with a given torque pulse magnitude and period. The color axis depicts the relative terminal period. A ratio of approximately unity indicates entrainment with 1:1 frequency-locking (shown in white). Squares in black indicate trials where the simulation failed (i.e., when the model became unstable and fell before the simulation stop time).

to hip pulses with a period of 25 ms shorter than preferred stride period, which is within the basin of entrainment of the model. For all pulse magnitudes tested, the boundary of the basin of entrainment to shorter pulse periods resulted from pulse periods that destabilized the model, causing it to fall (Fig. 5). The basin of entrainment was also reduced for lower pulse magnitudes (Fig. 5).

2) Phase-locking: In the human experiments, the pulse phase locked to ~78% of the gait cycle (right leg mid swing). The model exhibited similar phase-locking behavior. In the simulated trials, the pulse phase consistently converged to 94.9% of the gait cycle (right leg terminal swing) (Fig. 6A). Initial pulse phase and pulse magnitude did not affect the converged phase relation, but the degree of entrainment did weaken with lower pulse magnitudes (Fig. 6A,B).

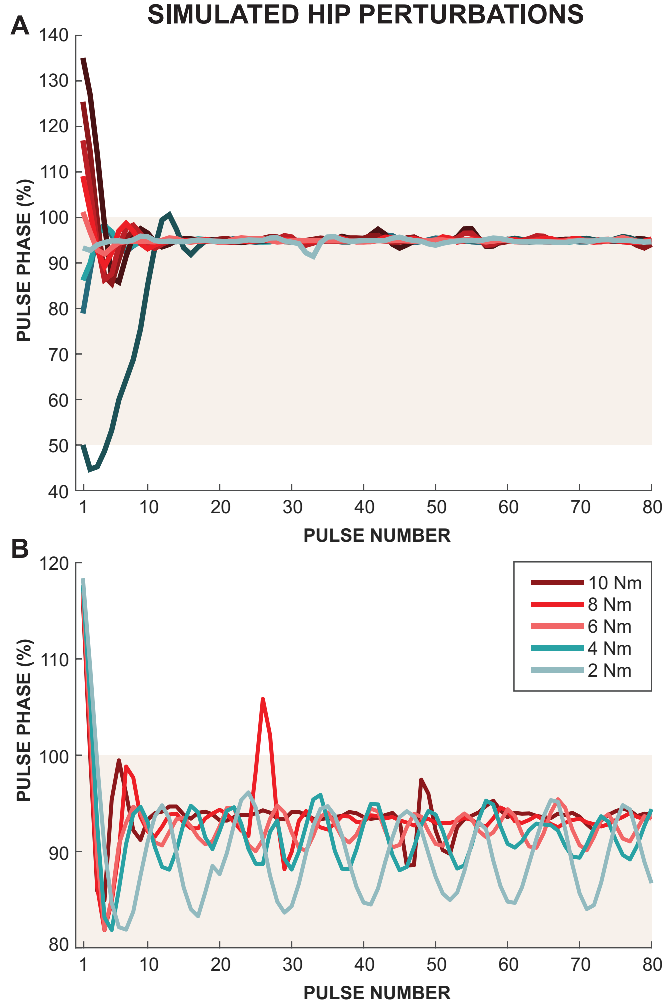

Fig. 6. (A) The effect of initial pulse phase on terminal pulse phase in the hip perturbation simulations. Each line represents a trial simulated with a different initial pulse phase. Pulse magnitude was set to 6 Nm. (B) The effect of pulse magnitude on entrainment in hip perturbation simulations. Each line represents a trial simulated with a different pulse magnitude. For all trials in panels (A) and (B), the pulse period was set to 1.15 s (i.e., ~18 ms shorter than the natural period of the model).

V. DISCUSSION

In this work, we examined whether an existing model of human locomotion rooted in neurophysiology could competently describe gait entrainment to periodic mechanical perturbation as seen in prior behavioral experiments. Exploiting rather than suppressing the natural oscillatory dynamics of walking to induce entrainment by coupling with an exoskeleton robot may provide a novel approach to gait therapy. A model of this phenomenon, in conjunction with behavioral experiments, would allow us to fully explore the extent to which gait entrains and inform how it arises from a neuromechanical standpoint.

As outlined in Section II-C, the first criterion used to evaluate the model was its ability to entrain to periodic perturbations within a finite range of pulse periods (i.e., basin of entrainment). The muscle-reflex model successfully entrained to both ankle and hip perturbations.

For ankle perturbations shorter than its natural period, the model's basin of entrainment resembled the experimental results. However, the model only entrained to pulse periods up to ~7 ms longer than its natural period. Interestingly, the same limitation was observed for the simplified model proposed by Ahn and Hogan [11]. To overcome it, Rigobon et al. [19] introduced additional regulation of stride-to-stride energetics into the simplified model, which resulted in entrainment to ankle perturbations with shorter and longer periods within a finite basin of entrainment as observed in human experiments. Song and Herr introduced an updated version of muscle-reflex model capable of 3D walking with an additional level of supraspinal control [20]. Future work will evaluate whether the inclusion of spinal feedback similarly improves the model's ability to competently describe human entrainment.

While the basin of entrainment to hip perturbations has yet to be established experimentally, these simulation results suggest that entrainment to a larger range of pulse periods is possible. The model's basin of entrainment to hip perturbations was larger than to ankle perturbations. The results also suggest that increasing the magnitude of the torque pulses beyond 6Nm to increase gait period may not substantially widen the basin of entrainment.

The second criterion used to evaluate the model was convergence to a consistent phase relation between its gait and the perturbations. In human experiments, ankle perturbations consistently phase-locked to $\sim$50% of the gait cycle, which is during right leg terminal stance or ankle push-off. The simplified model of Ahn and Hogan [11] similarly exhibited phase-locking to a single phase $\sim$50% of the gait cycle, matching the human experiments. While the muscle-reflex model similarly phase-locked $\sim$46% of the gait cycle, it also phase-locked to $\sim$11%, depending on the initial pulse phase. Recent work has shown that human gait also entrains to rhythmic electrical stimulation of the gastrocnemius muscle, which plays a crucial role in ankle plantarflexion [21]. In those experiments, phase-locking to two distinct pulse phases was also observed. The two phases ($\sim$56% and $\sim$100%) were similar, but slightly off in different directions than those observed in the simulation results. Although the exact cause for this discrepancy between the model and human behvaior is still unknown, one possibility is that it may due to assumptions made in the foot-ground contact model, such as how energy is dissipated during interaction. In the simplified model described in [11], dissipation of kinetic energy during foot-ground contact was critical for locally stable entrainment, for example. In contrast, the model's phase-locking behavior with hip perturbations was similar to that observed in human experiments. First, convergence to a single phase relation was observed. Terminal pulse phase was slightly later in the gait cycle for the model ($\sim$95%) compared to human gait ($\sim$78%). However, both occurred during right leg swing where the flexion torque pulse applied about the right hip was assistive.

While the simple model proposed by Ahn and Hogan [11] and the muscle-reflex model proposed by Geyer and Herr [12] greatly differ in terms of complexity, a feature common to both is that they produced limit cycle behavior during walking without a self-sustaining oscillatory neural network or supra-spinal control. Instead, this behavior emerged from peripheral neuromechanics and foot-ground interaction. Even though the the muscle-reflex model could not accurately describe human gait entrainment to ankle perturbations, it replicated human gait entrainment to hip perturbations reasonably well. Thus, the muscle-reflex model may serve as useful tool to determine how the potential therapeutic effects of gait entrainment can be maximized.

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