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Entrainment of Overground Human Walking to Mechanical Perturbations at the Ankle Joint

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Abstract— **Unlike upper-extremity robotic therapy, robotic therapy of lower extremities has not matched the effectiveness of human-administered approaches. We hypothesize that this may stem from inadvertent interference with natural movement control and investigated the oscillatory dynamics of human locomotion. Specifically, we assessed gait entrainment to periodic mechanical perturbations. Because the treadmills used in most studies necessarily interact with the dynamics of natural walking, we compared our experimental intervention during gait entrainment in treadmill and overground walking. Fourteen healthy subjects walked overground and on a treadmill while wearing an exoskeletal ankle robot which exerted short plantarflexion torque pulses at periods 50 ms shorter or longer than the subjects' preferred stride period. Entrainment to the periodic perturbation occurred in all conditions, however more readily in overground walking. In all cases, the stride period phase-locked with the torque pulse at 'push-off' such that it assisted propulsion. This entrainment of the stride period and its sensitivity to context indicate the subtlety and adaptability of human walking. Our observations suggest new avenues for gait rehabilitation and implications for exoskeleton design and legged locomotion research.**

I. Introduction

Whereas upper-extremity robotic therapy has proven more effective than human-administered therapy, lower-extremity robotic therapy has not [1]-[10]. One possible explanation for the diminished effectiveness of robotic therapy for walking might be the use of human-interactive robots that may suppress the expression of the natural oscillatory dynamics of walking. Most current therapeutic robots for walking emphasize tracking of pre-planned trajectories, discouraging (often preventing) voluntary participation of subjects. Additionally, the majority of experiments in the robotic gait rehabilitation field involve the use of treadmills that interact with natural movement control. In fact, even without a robot involved, body-weight-supported treadmill training has not proven superior to therapy that did not involve locomotion [11]-[12].

Does treadmill training obstruct the functional outcomes of walking therapy? Evidence from various studies suggests the importance of further investigating the dynamic and mechanical differences between treadmill and overground walking. For instance, walking on a standard motorized treadmill imposes a constant speed that may interfere with the natural variability of human locomotion. Additionally, the dynamics of the foot-ground interactions in these two walking environments appear to be unalike—possibly due to differences in the stiffness properties of the foot and ground.

To provide an effective rehabilitation strategy for neurologically impaired subjects, it is essential to investigate the minimal "mechanical components" that contribute to robust stable human locomotion [13]. An effective strategy needs to allow the impaired subjects to re-learn how to take advantage of the natural oscillatory dynamics that result from their foot-ground interaction.

While there is evidence that neural control of the upper extremity predominantly dictates hand kinematics in world coordinates, the dominant control scheme of human locomotion remains unclear. In contrast to reaching, walking is a rhythmic process that combines continuous and discrete dynamics [14]-[16]. Competent mathematical models of rhythmic locomotion have been developed using nonlinear limit-cycle oscillators, such as the van de Pol oscillator or the half-center Matsuoka oscillator [17]-[22]. To design more effective technology for lower-limb therapy, we need to investigate the natural oscillatory dynamics in human locomotor control. Various modeling studies have demonstrated that a combination of the inertial and gravitational mechanics of the legs and intermittent foot-ground collisions with energy dissipation can generate a stable limit-cycle [23]-[26].

One characteristic of limit cycles is dynamic entrainment to external perturbation: they synchronize their period of oscillation to that of an imposed rhythmic perturbation. In contrast to linear systems[1], nonlinear systems only exhibit entrainment when the perturbation frequency is close to their unperturbed oscillation frequency. Ahn and colleagues proposed a novel approach to entrain human gait to periodic torque pulse perturbations at the ankle via a wearable robot. Ahn and Hogan [27] demonstrated entrainment to external periodic plantarflexion torque pulses during treadmill walking, evidence of an underlying nonlinear limit-cycle oscillator. More interestingly, it appeared that subjects' gaits synchronized with the plantarflexion perturbations at 'push-off', suggesting that the torque pulses assisted propulsion.

To further explore that neuro-mechanical oscillator, this study tested the effect of periodic perturbations to the ankle joint in a similar fashion, but this time not only during treadmill walking but also overground. The novelty of our work is to examine the differences and limitations of gait synchronization in treadmill versus overground walking.

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[1] Stable linear systems—not necessarily oscillatory—will *entrain* to inputs of all all frequencies.

II. METHODS

Fourteen healthy subjects participated in an experimental study. All participants gave informed consent in accordance with procedures approved by the Institutional Review Board (IRB) of the Massachusetts Institute of Technology (MIT). The purpose of the study was to compare the subjects' performance on a standard treadmill versus overground.

A. Equipment and Protocols

Each subject performed 2 trials on a Sole Fitness F80 treadmill (with a 0.84 m X 1.90 m deck), and another 2 trials walking overground in a large corridor at MIT. For both walking conditions, shorter and longer perturbation periods were delivered. In all trials, subjects performed a cognitive distractor task that consisted of listing countries, cities, animals, etc. in alphabetical order (one category at a time).

The robot used in these experiments was the Anklebot by Interactive Motion Technologies, Inc. (**Figure 1**). This wearable therapeutic robot attached to the leg via a knee brace and a shoe. A potentiometer embedded in the knee brace recorded the subjects' knee angle profile during walking. The Anklebot's highly back-drivable linear actuators were capable of actuating the ankle in dorsi-plantarflexion and inversion-eversion. In all trials subjects wore a harness to distribute the weight of the Anklebot over the upper body. The robot was preprogrammed to deliver periodic square torque pulses of magnitude 10 N-m and duration 100 ms in the same fashion as in Ahn and Hogan [27]. In addition to exerting the torque pulses, the robot behaved like a torsional spring-and-damper with 5 N-m/rad stiffness, 1 N-m-sec/rad damping, referenced to a constant equilibrium position measured from the subject's upright posture (see also Ahn and Hogan [27]).

B. Treadmill Trials

Subjects were asked to adjust the speed of the treadmill to a comfortable walking speed. The selected speed was recorded and maintained throughout the duration of any one trial. A treadmill trial (TM) began with subjects walking at their preferred speed. Subjects' preferred stride duration (τ_0) was measured as the average duration of 15 consecutive strides as in [27]. In previous work [27], entrainment during treadmill walking was observed only when the perturbation period (τ_p) was sufficiently close (~6.7%) to the pre-perturbation stride duration (τ_0); τ_p had been varied from shorter to longer in different trials with a resolution of 50 ms. For this particular study, we did not aim to determine the basin of entrainment in OG vs. TM walking. Hence, to maintain similarity with previous work, τ_p was discretized again with a 50 ms resolution, but this time to allow only two variations: one shorter (TM-shorter) and one longer (TM-longer). Each trial was divided into 3 sections: *before*, *during*, and *after*. As in [27] the *before* section consisted of 15 strides with no perturbation (these strides were used to determine τ_0). The *during* section comprised 50 consecutive perturbations, which was the greatest number of torque pulses possible within the length of the hallway used for the overground trials. In the *after* section the robot stopped exerting the torque pulses but maintained its spring-damper behavior while subjects walked another 15 strides, in the same fashion as in [27]. Subjects stopped walking and the trial terminated immediately afterwards.

Figure 1: An unimpaired human subject wearing the Anklebot.

C. Overground Trials

Overground trials (OG) differed from treadmill trials mainly in that there was no fixed-walking-speed constraint. The overground trials began by asking the subjects to walk at their preferred walking speed. Once comfortable walking speed was achieved, their preferred walking period (τ_0) was measured using the subsequent 15 strides. Overground trials were conducted in the same fashion as treadmill trials, for shorter (OG-shorter) and longer (OG-longer) perturbation periods. Throughout all overground trials subjects were followed by the experimenter from a very close distance who moved the computer equipment on a rolling cart.

III. DATA ANALYSIS

The gait cycle was defined based on knee angle measurements recorded by a potentiometer embedded in the Anklebot's knee brace. All data collected from onboard sensors were recorded at a sampling rate of 200 Hz. Subjects' stride durations before, during, and after the perturbations were compared to evaluate whether mechanical perturbations sped up or slowed down the subjects' walking cadence. Statistical significance was set at a 5% significant level.

A. Gait Cycle

The gait cycle was estimated from extrema in the knee angle profile. Four landmarks were used in subsequent analyses: maximum knee flexion during stance phase, maximum knee extension during terminal stance phase, maximum knee flexion during swing phase, and maximum knee extension during terminal swing phase before heel strike. The knee angle profile was normalized from 0 to 100% to define a gait phase for each stride, with 100% defined as the maximum knee extension adjacent to heel strike (**Figure 2**).

B. Assessment of Entrainment

The plantarflexion perturbations were delivered at a constant period throughout each trial; however, the onset of the torque pulses could vary with respect to landmarks in the gait cycle (e.g. the maximum knee flexion) given its 50 ms difference from the preferred stride period. Hence, the phase of the gait cycle at which perturbations occurred would not necessarily be constant. In order to entrain to the applied perturbations, subjects' gait period must be the same as the period of the imposed torque pulses; entrainment requires each pulse to occur at the same phase of the gait cycle.

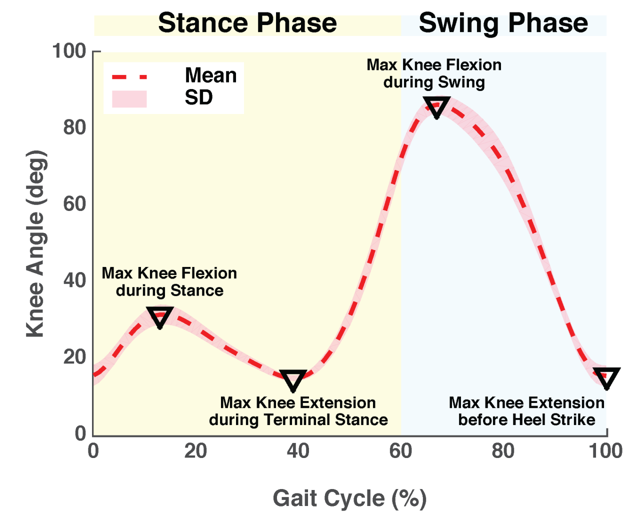

Figure 2: Typical knee angle across the gait cycle. The trajectory illustrates the four extrema (∇) that defined the gait cycle from 0 to 100%.

The gait phase of each perturbation was determined as the percentage of the gait cycle that coincided with the onset of the torque pulse. The gait phases related to the 50 consecutive perturbations were calculated in reverse order starting from the 50th perturbation. To avoid sudden jumps in the gait phases when the onset of a perturbation crossed the 0 or 100% boundaries, wrap-arounds in the gait cycle were allowed (i.e. gait phases greater than 100% or less than 0%).

A linear regression of gait phase (Y) onto perturbation number (x) should evidence entrainment as a zero-slope segment (**Figure 3**). This regression ($Y = mx + b$) was applied to the last 10 perturbations in each trial; entrainment was indicated if the 95% confidence interval included zero slope. If the null hypothesis was accepted (H_0: $m = 0$), then the gait was considered *entrained*. Trials for which H_0 was rejected were defined as *not entrained to shorter* τ_p ($m < 0$) or *not entrained to longer perturbations* ($m > 0$).

C. *Converged Gait Phase*

To evaluate gait phase convergence for each entrained gait, the phase and the onset of phase convergence were determined. To calculate these two measures the standard deviation (σ) of the gait phases at which the last 10 perturbations occurred was determined. The converged gait phase value (φ_{conv}) was determined as the mean gait phase corresponding to the greatest number of consecutive perturbations lying within an interval $\varphi_{conv} \pm 2\sigma$. When determining φ_{conv}, it was deemed acceptable for up to 3 consecutive perturbations to lie outside the interval, provided the subsequent perturbation re-entered the interval. The onset of the converged gait phase or *phase-locking* was determined as the first perturbation to lie within the defined interval. For each subject the torque pulse number of the onset of entrainment was recorded in all 4 trials (except those trials that did not entrain. The dependent measures, gait phase and onset of phase convergence, were submitted to a 2 (TM vs. OG) x 2 (shorter vs. longer) ANOVA using SAS JMP® statistical software package [28].

IV. Results

A. *Entrainment*

Entrainment was observed in 46 out 56 total trials. One subject (S6) did not entrain in any of the 4 different trials. The remaining 6 trials identified as not entrained were all TM-longer trials; i.e. entrainment was not observed in 50% of the TM-longer trials. **Figure 4** shows the perturbation torque phase—gait phase at which torque pulse occurred—as a function of the perturbation number for all entrained gaits.

B. *Phase-Locking in Entrained Gaits*

In the 46 entrained trials, subjects synchronized their gaits with the torque pulses at ~50% of the gait cycle. Histograms of gait phase in the last 10 perturbations of entrained gaits are shown in **Figure 5**. The mean φ_{conv} across all entrained gaits was 51.64% (± 2.36%), which was near the boundary between the terminal stance and pre-swing phases. This coincides with the interval of maximum ankle plantarflexion torque, known as 'push-off' [29].

Figure 6 shows the mean onset of phase convergence between subjects for the four conditions. The two-factor ANOVA evaluating the onset of phase convergence revealed significant main effects for both walking environment and perturbation period ($p < 0.001$, $F_{1,42} = 19.61$ and $p < 0.001$, $F_{1,42} = 19.01$ respectively). Onset of phase convergence was earlier in OG (Mean = 24.12, SD = 10.26) than in TM trials (Mean = 32.90, SD = 7.13). Similarly, a more rapid gait

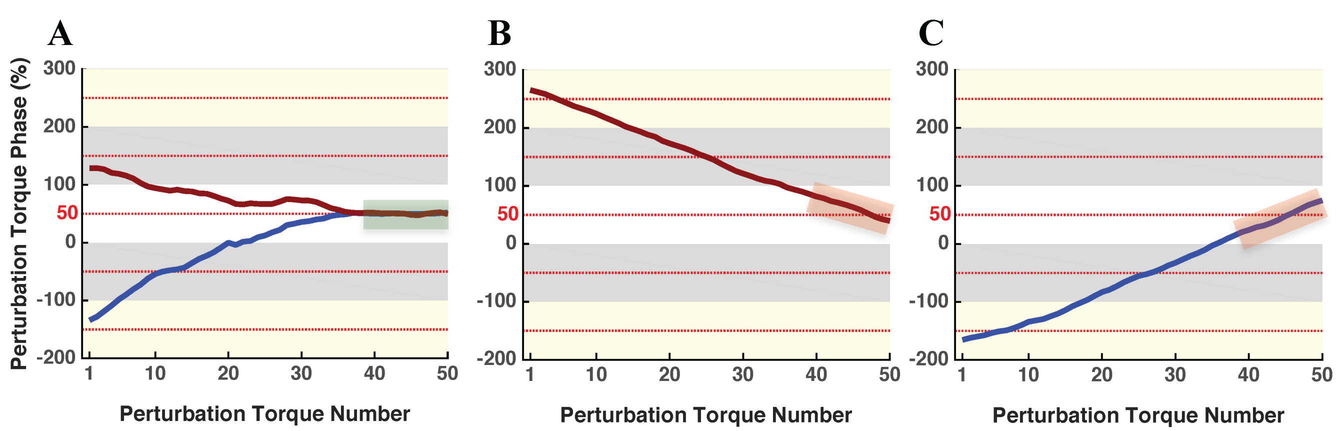

Figure 3: Regression of perturbation torque phase vs. perturbation number in the last 10 torque pulses for three representative cases. (A) Two regression slopes not significantly different from zero, in entrained gaits to shorter (burgundy) and longer (blue) perturbation periods; **(B)** Significantly negative regression slope in a gait that did not entrain to shorter perturbation periods; **(C)** Significantly positive regression slope in a gait that did not entrain to longer perturbation periods. The alternating regions shaded in light gray and yellow correspond to *wrap-arounds* in the gait cycle.

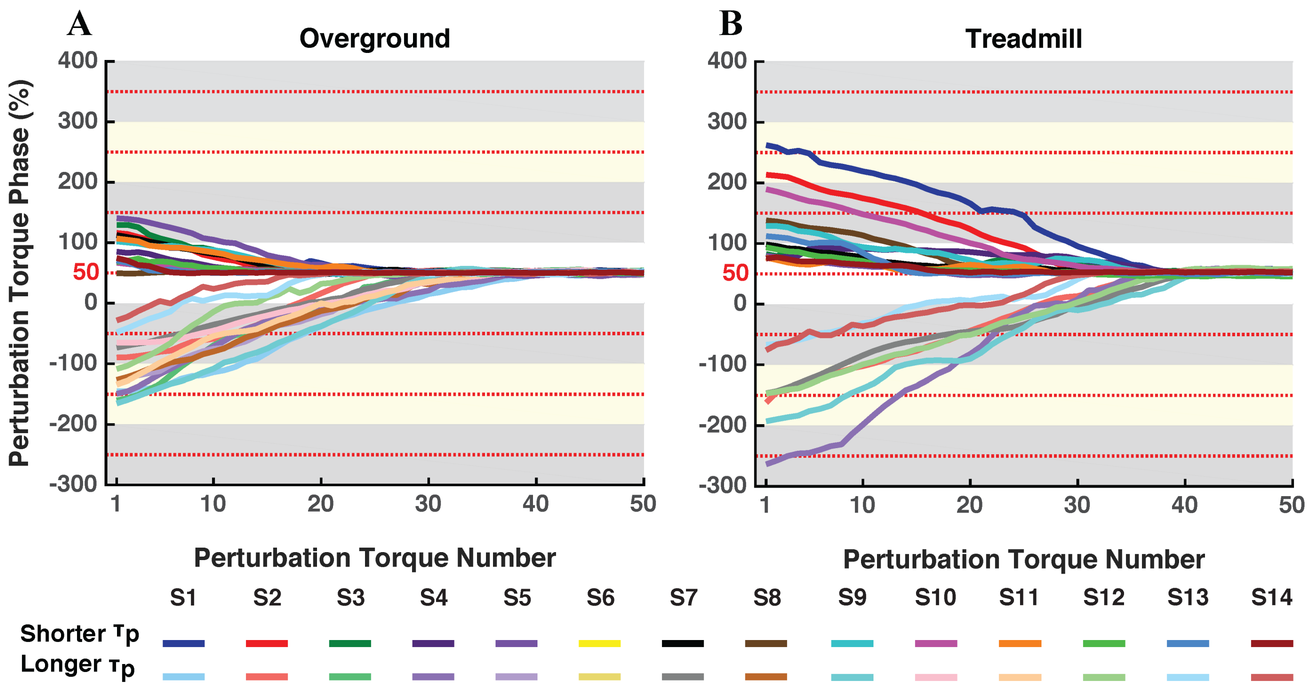

Figure 4: Perturbation torque phase as a function of perturbation torque number for all entrained gaits. (A) Entrained gaits during overground trials. **(B)** Entrained gaits during treadmill trials. Each color corresponds to a different subject, with a dark and a light shade corresponding to the trials with shorter and longer perturbation periods respectively. Subject 6 (S6) is not shown since she did not entrain in any of the 4 trials. Other missing lines correspond to trials in which subjects did not entrain.

phase convergence was detected in trials with shorter τ_p (Mean = 24.04, SD = 10.71) in comparison to those with longer τ_p (Mean = 33.00, SD = 6.05). No significant interaction was found between the two factors ($p = 0.098$).

V. Discussion

A. Nonlinear Neuro-Mechanical Oscillator Underlying Human Locomotion

Human bipedal locomotion displays some fundamental features indicative of an underlying nonlinear limit-cycle oscillator. In fact, nonlinear oscillators with a limit-cycle have served as competent models of rhythmic pattern generators (including CPGs) [17]-[20] and stable bipedal walkers (e.g. "passive walkers") [23], [24], [26]. A distinctive characteristic of nonlinear limit-cycle oscillators is entrainment to an external rhythmic perturbation. Our experiments demonstrated gait entrainment to periodic perturbations (i.e. plantarflexion torque pulses at the ankle joint) in both treadmill and overground walking, together accounting for 46 entrained trials out 56 total trials.

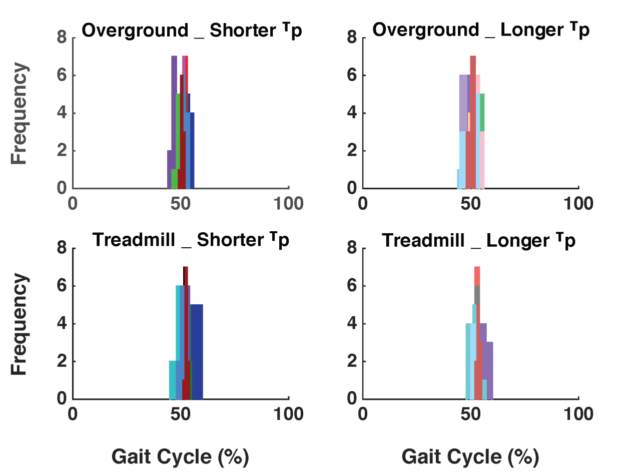

Figure 5: Histograms of gait phase in the last 10 torque pulses of entrained gaits. Distribution of the gait phase (φ_{conv}) for each of the four conditions for all 14 subjects. Colors in the histogram bars correspond to different subjects as in **Figure 4**.

To minimize voluntary gait synchronization to the imposed perturbations, we asked subjects to perform a distractor task. If gait entrainment was a result of voluntary synchronization, then the onset of phase convergence should have occurred within the first few perturbation cycles. Instead, a rather moderate-to-slow convergence was observed in overground and treadmill trials, occupying 24 and 32 perturbation cycles on average respectively.

To our knowledge, this is the first study demonstrating dynamic entrainment to external periodic plantarflexion perturbations at the ankle joint during overground walking. We submit that these results show clear, behavioral evidence that a nonlinear neuro-mechanical oscillator with a limit-cycle plays a significant role in human locomotion.

B. Gait Entrainment in Overground vs. Treadmill Walking

Previous studies reported no significant differences in the gait patterns of chronic stroke survivors with and without the Anklebot on the paretic leg during treadmill and overground walking [30]. In our experiments, TM and OG trials were conducted in the same fashion, but without constraint on fixed speed in the OG trials. However, our gait entrainment results differed significantly in TM and OG trials: gait phase converged faster in OG than in TM trials, taking an average of 24 and 32 perturbation cycles respectively (**Figure 6**).

Furthermore, a larger number of trials entrained in OG compared to TM trials. This difference appears to be due to

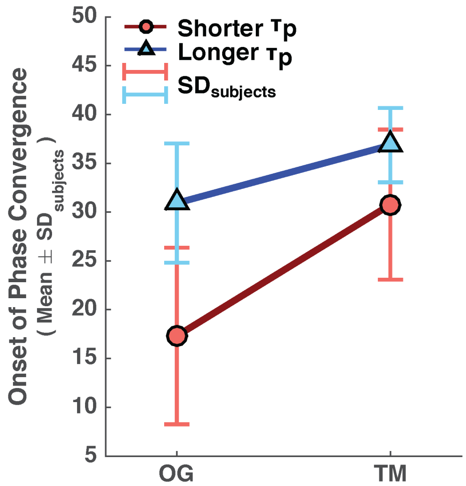

Figure 6: Mean perturbation torque number corresponding to the onset of phase convergence. Convergence in trials with longer τ_p was slower than in trials with shorter τ_p. Treadmill trials (TM) showed slower phase convergence than overground trials (OG). Error bars indicate the standard deviation of the onset of phase convergence across subjects by perturbation period and walking environment.

the fixed-speed constraint in TM trials. In order for subjects to entrain to periodic perturbations 50 ms different from their preferred stride period, they either had to change their speed, and/or their stride length. Given that the speed was kept constant in TM trials, subjects could only adjust their stride period to eventually match the perturbation period. Therefore, the constant speed of the treadmill belt, including the limited length of the treadmill deck, appears to have influenced the gait phase convergence in TM trials. Further experiments are required to address this substantial difference in gait entrainment during treadmill and overground walking.

C. *Faster Phase-Locking at Ankle 'Push-off' in OG Trials*

While entrainment requires phase convergence of the subject's stride duration to the perturbation period, it does not limit such phase convergence to any particular, constant phase. Ahn and Hogan [27] previously reported that gait synchronized with the perturbation at approximately 50% of the gait cycle when subjects walked on a treadmill. Our experiments not only replicated their findings on treadmill walking, but also extended their observation to walking overground. Analysis of gait phase convergence revealed that the average gait phase in the 46 entrained trials was 51.64% (± 2.36%) (**Figure 5**). It must be emphasized that the final gait phase value was independent of the gait phase at which perturbations were initiated. Perturbations were randomly initiated at various phases of the gait cycle across all trials, which can be seen in **Figure 4**. Hence, the end of double stance (~50% of the gait cycle) may be regarded as the "global" attractor for phase-locking in gait entrainment to periodic ankle plantarflexion perturbations.

In normal walking, the maximum ankle plantarflexion torque is exerted at 'push-off' (47-62%), which begins near the end of terminal stance phase (31-50%) and extends to the duration of the pre-swing phase (50-62%) [29]. Phase-locking occurring consistently at ankle 'push-off' in our experiments suggests that gait adapted so that the periodic perturbations mechanically assisted plantarflexion at the ankle, thus facilitating forward propulsion. This observation is of significance for lower-extremity robotic rehabilitation and exoskeleton design since the mechanical perturbations could supply the additional torque needed by patients who cannot produce sufficient propulsion to swing their paretic leg forward. Similarly, we reason that varying the magnitude and frequency of the torque pulses that provide assistance as needed may stimulate voluntary participation [31].

In several entrained trials, it was noted that a torque pulse occurring at 'push-off' was not always accompanied by immediate gait synchronization (phase-locking). This can be seen in **Figure 4** as several regression curves crossed the red horizontal lines corresponding to 50% of the gait cycle, yet there was no entrainment until further along in the trials. If subjects synchronized their gaits with the perturbation where it assisted propulsion –or did not oppose ankle actuation– then why did they not do so at the very first opportunity? Perturbation periods (τ_p) were strictly 50 ms shorter/longer than preferred stride duration (τ_0). However, τ_0 was determined as the averaged duration of 15 consecutive strides, measured by a stopwatch and visually estimating the moment of heel strike. Hence, not only could the preferred period be non-stationary, but it also had a variability. As a result, τ_p could be further apart from or closer to subjects' walking cadence when perturbations were initiated. Synchronization to perturbation periods further apart from subjects' stride period required greater changes in cadence. Given the nonlinear nature of the limit-cycle oscillator postulated to underlie human locomotion, entrainment could only occur when τ_p was sufficiently close to the subjects' stride period. Hence, the perturbation period could have been significantly different from a subject's stride period at the very first opportunity a torque pulse occurred at 'push-off', thus making phase-locking unattainable. In these cases, gradual changes in walking cadence eventually reduced the difference between τ_p and subjects' stride period, leading to entrainment.

D. *Entrainment to Shorter vs. Longer Perturbation Periods*

The plantarflexion torque pulses applied to the ankle during double-stance can only act as mechanically assistive pulses, adding positive work. Entrainment to fast perturbation periods ($\tau_p = \tau_0 - 50$ ms) required subjects to speed up cadence. Hence, gait entrainment to fast perturbation periods might be due to the positive work added by the mechanically assistive perturbations. A simple model presented by Ahn and Hogan reproduced this behavior [29].

In contrast, entrainment to longer perturbation periods ($\tau_p = \tau_0 + 50$ ms) cannot solely be attributed to a mechanical response to assistive perturbations. Ahn and Hogan's model was capable of reproducing entrainment and phase-locking only when the perturbation periods were faster than preferred stride period [29]. However, our experiments also demonstrated gait entrainment to longer perturbations, which therefore cannot be attributed to mechanics alone. Entrainment to longer perturbation periods required subjects to slow down their cadence, even though the mechanically assistive plantarflexion torque pulses caused them to speed up, at least locally. The fact that phase-locking occurred significantly later in trials with longer τ_p is consistent with this reasoning. In fact, in treadmill walking entrainment was only detected in 50% of the total trials. In all, our results suggest that gait entrainment may not simply be the result of peripheral mechanics in human walking. Instead, gait entrainment seems to require a more complex interaction between the neuro-muscular periphery and the gravito-inertial mechanics in human locomotion.

VI. Conclusions and Implications

Results of this study suggest that a nonlinear neuro-mechanical oscillator that is sensitive to subtle differences in context—treadmill vs. overground walking—plays a non-negligible role in human locomotion. Our experiments further suggest that intermittent foot-ground collisions with concomitant energy dissipation may be a key element of legged locomotion that needs to be further explored. In fact, it appears that these collisional interactions capable of generating a stable limit-cycle may determine the stability and robustness of locomotor control. These observations should be considered when designing therapeutic robots and exoskeletons to improve human locomotion and/or walking efficiency. Although further investigation is required, our study indicates gait entrainment to mechanical perturbations at the ankle may be a feasible approach for walking rehabilitation.

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