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Users Maintain Task Accuracy and Gait Characteristics During Missed Exoskeleton Actuations Through Adaptations In Joint Kinematics

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***Abstract*— In operational settings, lower-limb active exoskeletons may experience errors, where an actuation that should be present is missed. These missed actuations may impact users' trust in the system and the adapted human-exoskeleton coordination strategies. In this study, we introduced pseudo-random catch trials, in which an assistive exoskeleton torque was not applied, to understand the immediate responses to missed actuations and how users' internal models to an exoskeleton adapt upon repeated exposure to missed actuations. Participants (N = 15) were instructed to complete a stepping task while wearing a bilateral powered ankle exoskeleton. Human-exoskeleton coordination and trust were inferred from task performance (step accuracy), step characteristics (step length and width), and joint kinematics at selected peak locations of the lower limb. Step characteristics and task accuracy were not impacted by the loss of exoskeleton torque as hip flexion was modulated to support completing the stepping task during catch trials, which supports an impacted human-exoskeleton coordination. Reductions in ankle plantarflexion during catch trials suggest user adaptation to the exoskeleton. Trust was not impacted by catch trials, as there were no significant differences in task performance or gait characteristics between earlier and later strides. Understanding the interactions between human-exoskeleton coordination, task accuracy, and step characteristics will support development of exoskeleton controllers for non-ideal operational settings.**

I. INTRODUCTION

One common use case for lower-limb exoskeletons is to assist or augment a user's motor performance. There is an underlying assumption that the user's physical effort will be reduced by the exoskeleton and the human will adapt to reduce their muscle activation to coordinate with the system. However, across one set of users, the initial response to the exoskeleton was to oppose the system by increasing muscle activation and energy expenditure, therefore counteracting the exoskeleton design goal [1]. These initial responses can be participant-specific, with individuals exhibiting different responses during early adaptation to exoskeletons [2]. In these studies, the systems performed as designed, which may not always occur in operational settings. For example, exoskeletons may miss an actuation during a stride and human-exoskeleton trust may be reduced [3]. When the exoskeleton does not apply an expected torque, users' trust may decrease. It is important to understand how users respond when an expected actuation is not provided for each step to support co-adaptive control design that enables improved human-exoskeleton coordination.

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Human motor strategies are constructed based on internal mental models formed during adaptation to a novel environment [4]. Internal models are state-dependent approximations of external forces that inform motor commands and predictions [4], [5]. In the presence of an external force, the central nervous system will aim to minimize movement errors by either stiffening the limb or learning internal models to respond to the force [6]. These changes are due to motor adaptation, which is the modification of movement in response to trial-to-trial error feedback [7], or the changes in strategy over time. To understand generalizations within an internal model, random trials with unexpectedly altered dynamics (catch trials) are introduced. If an internal model was learned, catch trials would result in endpoint errors without feedback compensation as the person has adapted their motor strategies to account for the external force.

Previous studies have introduced catch trials to understand gait adaptation when exposed to external forces. Bucklin et al. [8] showed that participants adapt motor strategies for center-of-mass trajectories as they walk towards a target in an uniform force field. During catch trials, participants' trajectories deviated in the direction opposite of the force, indicating a learned internal model corresponding to the force field. Similarly, Cajigas et al. [9] observed that participants modified their hip and knee kinematics when walking in a constant force field. Similar to these force-field studies, exoskeletons apply external torques that can be learned and anticipated. Catch trials involving the absence of external torques can be used to understand the internal models developed with exoskeletons and goal-directed tasks. However, catch trials may affect users' trust and thus system usage.

In addition to human-exoskeleton coordination and trust, there exist several other factors that can impact motor strategies. To perform a goal-oriented task, considerations include task performance time and accuracy. Gait strategies are also linked with energy expenditure, stability, and comfort. These considerations may interact. Different motivational prompts have supported the interaction between metabolic cost and stability [10]. These interactions are beginning to be examined with exoskeletons [11]. It is important to understand the interaction of human-exoskeleton coordination and task accuracy as the system behavior is manipulated, which can affect trust and the internal model developed.

We introduced catch trials to understand the internal model developed when adapting to a powered exoskeleton and how that model changes with missed actuations. We explored human-exoskeleton coordination and gait characteristics when completing a task that required foot placement accuracy. We hypothesized that there would be time-dependent changes in (1) step characteristics (step length and width), (2) task performance (step accuracy), and (3) joint kinematics (selected peak hip, knee, and ankle angles within the stride) in response to catch trials. These measures will be interpreted in the context of human-exoskeleton coordination, trust, and internal model development.

II. Methods

A. Participants

Participants (N = 15, age = 30.7 $\pm$ 9.9 years, height = 1.73 $\pm$ 0.10 m, mass = 74.9 $\pm$ 13 kg, 6 female (mean $\pm$ SD)) provided written informed consent. Participants were excluded if they had a lower extremity injury within the past 6 months or used an assistive walking device. Leg dominance was assessed by asking which foot participants use to perform a kick. The experimental protocol was approved by the Committee on the Use of Humans as Experimental Subjects at the Massachusetts Institute of Technology.

B. Experimental setup

Participants walked on a split-belt, instrumented treadmill in a Computer Assisted Rehabilitation Environment (CAREN) System (Motekforce Link, Amsterdam, the Netherlands), which included an 18-camera optical motion capture system (Vicon Motion Systems Ltd, Oxford, UK) and a 24-foot diameter virtual reality dome with a 360 degree projection screen. Reflective markers were placed on the participants according to the Vicon Plug-in Gait model, adjusted for the exoskeleton by placing the lower limb markers on the lateral side of the exoskeleton when necessary. Markers were also placed on the treadmill on the four corners of the stepping target. Motion capture data were collected at 100 Hz. The treadmill, motion capture, and exoskeleton were controlled and time-synchronized via D-Flow software. Study participants wore the Dephy ExoBoot on both legs (DpEb45, Dephy Inc, Maynard, MA, USA) [12]. Torque was applied at push-off during the stance phase of the gait cycle, learned from 25 strides (Fig. 1).

C. Protocol

Participants were given a targeted stepping task (a 320 mm long projected rectangle that spanned the treadmill width), while walking at 1.25 m/s through a virtual city scene. Each participant underwent a training and testing period, separated by a 5-minute break. During training, participants walked with the stepping target for 30 minutes with the exoskeleton powered on, with torque applied during each stride. During testing, catch trials were randomly dispersed among normal strides (1900 strides total). The number of strides between each catch trial were generated by selecting 32 values from a uniform distribution bounded by (20, 100) strides. A total of 32 catch trials (16 per limb) were induced by not actuating the exoskeleton for a single stride. The exoskeleton's algorithm included a 1-3 stride "recovery period" after missed actuations, in which the assistive torque was reduced and ramped back up to baseline levels of torque.

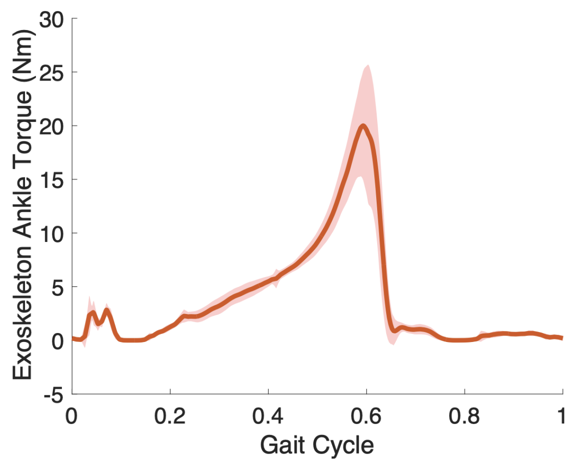

Fig. 1: (top) Powered ankle exoskeleton, which provides assistance by applying torque via the inelastic belt attached to the lever arm. (DpEb45, Dephy Inc) [2]. (bottom) Exoskeleton torque profile applied at all non-catch trial strides for one representative participant. The solid line is the average torque applied throughout the stride and the shaded region includes $\pm$1 standard deviation.

D. Data analysis and statistical methods

Gait cycles were segmented with a custom MATLAB script using the treadmill's force plates to identify heel strikes. Normalized step length (NSL), normalized step width (NSW), and task accuracy were calculated using heel marker positions and treadmill velocity. Step length was the sum of the difference in anterior footfall location and the distance that the treadmill moved between heel strikes. Step width was calculated as the difference between the lateral position of each heel marker. Step length and width were normalized by leg length, measured from the anterior superior iliac spine to the medial malleolus. Task accuracy was calculated as the difference between each heel strike and the center-line of the stepping target. Acceptable task error was within $\pm$160 mm of the center-line, which was derived from the size of the 320 mm-long stepping target, corresponding to the length of the largest shoe size. Joint kinematics were calculated per stride according to the Plug-in Gait model. Three metrics of

	NSL		NSW		Task Accuracy		Peak Hip Flexion		Min Knee Flexion		Peak Plantarflexion	
	F	p	F	p	F	p	F	p	F	p	F	p
Participant	285.44	<0.001	331.21	<0.001	335.14	<0.001	1.20	0.275	0.29	0.994	12.92	<0.001
Trial Type	<0.01	1.0	0.01	1.0	<0.01	1.0	88.30	<0.001	321.11	<0.001	233.95	<0.001
Timing	<0.01	0.963	<0.01	0.781	<0.001	0.984	0.36	0.554	11.96	<0.001	0.12	0.735
Leg	-	-	-	-	-	-	6.47	0.015	25.44	<0.001	23.83	<0.001
TT*Timing	<0.01	1.0	<0.01	1.0	<0.01	1.0	0.83	0.560	0.92	0.488	0.46	0.864
TT*Leg	-	-	-	-	-	-	2.27	0.027	3.81	<0.001	35.94	<0.001
Timing*Leg	-	-	-	-	-	-	7.41	0.007	7.20	0.007	0.08	0.786
TT*Timing*Leg	-	-	-	-	-	-	0.44	0.879	1.39	0.203	0.41	0.893

TABLE I: Summary of statistics for step characteristics, task accuracy, and joint kinematics for 15 participants, with a total of 57000 steps for NSL, NSW, and Task Accuracy and a total of 28500 strides for Peak Hip Flexion, Minimum Knee Flexion, and Peak Plantarflexion.

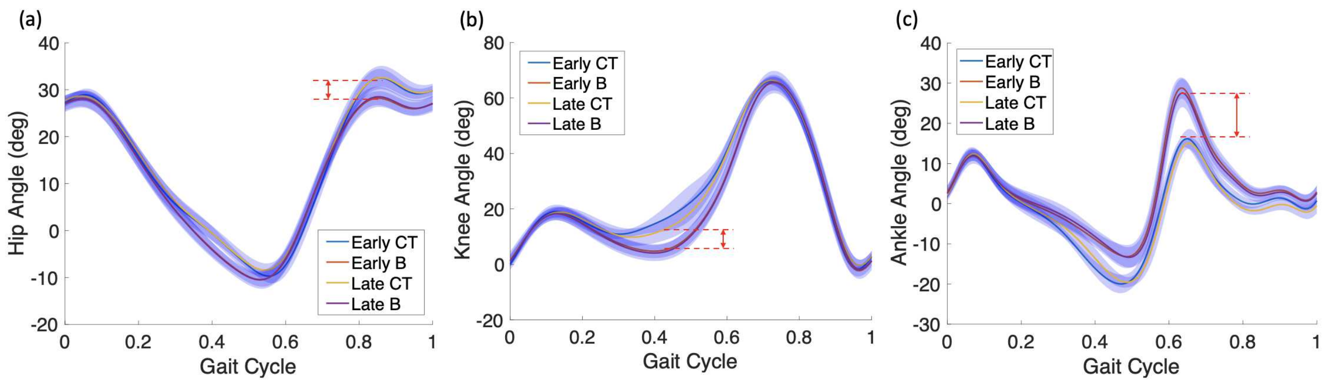

Fig. 2: (a) Hip, (b) knee, and (c) ankle kinematics for ipsilateral leg, where flexion/plantarflexion is positive and extension/dorsiflexion is negative. Regions of interest include peak hip flexion during swing, knee flexion after loading response, and peak ankle plantarflexion, indicated by the red arrows and dotted lines.

interest were identified for each stride (Fig. 2) – maximum hip flexion during swing, minimum knee flexion after loading response, and maximum ankle plantarflexion.

Steps and strides were segmented into the following Trial Type: 1 step/stride prior to the catch trial (CT-1), step/stride of the catch trial (CT), subsequent 5 steps/strides after the catch trial (CT+1 to CT+5). Baseline (B) was calculated using all steps/strides outside of the CT-1 to CT+5 conditions. NSL, NSW, and task accuracy were evaluated when parsed by step. Maximum hip flexion during swing, minimum knee flexion after loading response, and peak plantarflexion were evaluated when parsed by stride.

Initially, multi-factor ANOVAs were performed for each participant for all metrics to evaluate the effect of catch trials, timing, and leg for each individual, as previous studies have shown individual strategies for adaptation [2]. For these data, all participants showed similar trends, so independent repeated-measures ANOVAs were performed for each metric with 3 fixed factors of Trial Type (7 levels), Timing (2 levels), and Leg (for joint kinematics only, 2 levels), and a random factor of Participant. The Timing levels were early and late, where "early" data was defined as all steps/strides prior to the 17th catch and "late" data included all steps/strides on and after the 17th catch trial. Post-hoc pairwise comparisons between Trial Type, Timing, and Leg were performed with false discovery rate (FDR) correction. Cohen's d effect sizes were calculated for all post-hoc comparisons.

III. RESULTS

There was an effect of Participant on NSL, NSW, and Task Accuracy, but no effect due to Trial Type or Timing (Table I). Participants maintained their step characteristics and task performance with the absence of the actuation (Fig. 3). However, there were main effects of Trial Type and Leg on hip, knee, and ankle angle metrics, as well as interaction effects between Trial Type and Timing with Leg.

Catch trials resulted in a significant difference between CT and all other trial types (CT-1 to CT+5, B) on the ipsilateral leg (Fig. 3). During CT, hip flexion during swing increased ($p < 0.001$, $0.95 < d < 1.46$), minimum knee flexion after loading response increased ($p < 0.001$, $1.23 < d < 1.72$), and peak plantarflexion decreased ($p < 0.001$, $-0.69 < d < -1.60$).

At CT+1, hip flexion ($p < 0.01$, $0.64 < d < 1.46$) and minimum knee flexion ($p < 0.001$, $1.23 < d < 1.56$) were also significantly higher than all other trial types except CT. Maximum plantarflexion at CT-1 was greater than all other non-baseline trial types ($p < 0.05$, $0.95 < d < 1.60$). Additionally, maximum plantarflexion at CT+1 to CT+5 trials were each different from all other conditions ($p < 0.05$, $0.53 < |d| < 1.29$). Minimum knee flexion on the non-dominant leg was overall higher than the dominant leg ($p = 0.002$, $d = 0.74$), especially during later strides compared to earlier strides ($p = 0.007$, $d = 1.57$). The leg dominance had no effect on response to catch trials.

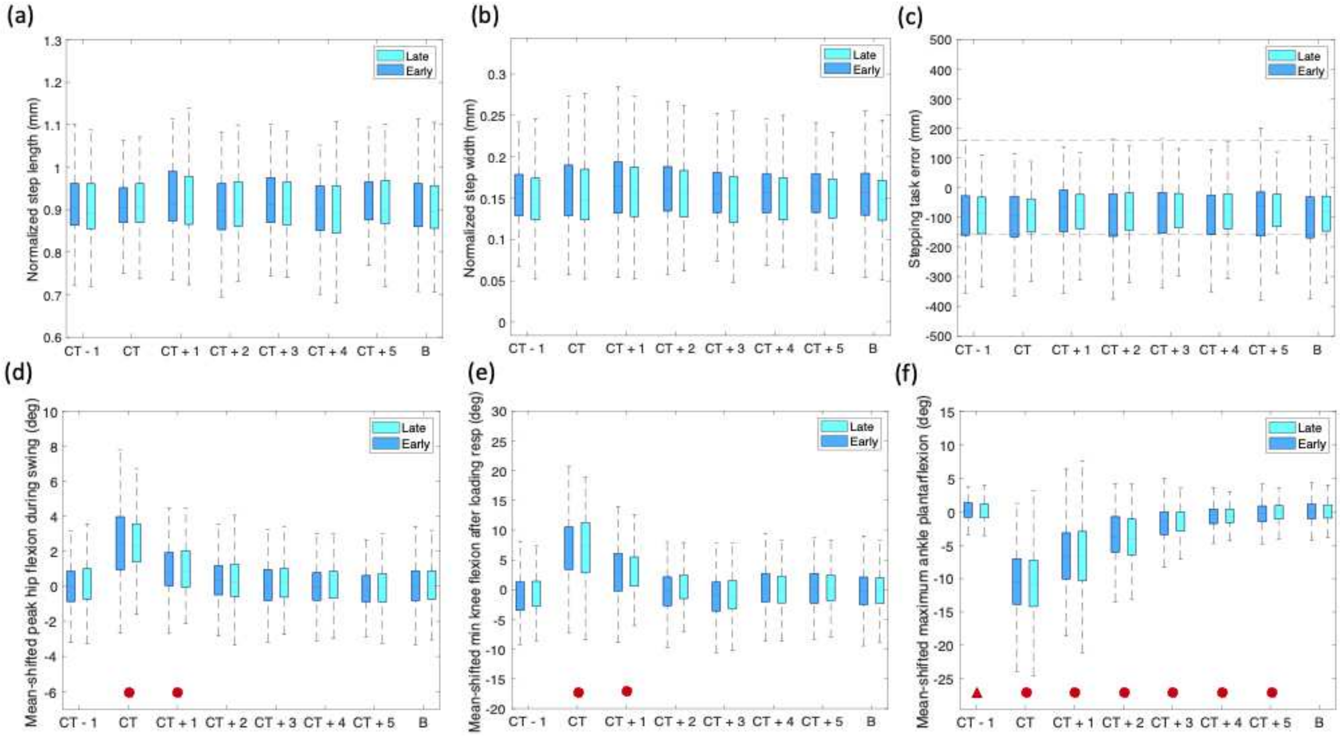

Fig. 3: (a) NSL, (b) NSW, (c) task accuracy, (d) peak hip flexion during swing, (e) minimum knee flexion after loading response, and (f) maximum ankle plantarflexion in response to catch trials (N=15). Joint angles were shifted by subtracting the mean of each participant's baseline. CT-1 to CT+5 conditions include 16 steps/strides each and B contains remaining steps/strides within experiment. Each box includes the 25-75th percentiles and whiskers are 1.5 IQR. Circles denote conditions significantly different than all other conditions, triangles denote conditions different than all catch conditions.

IV. Discussion

This study explored the interaction of human-exoskeleton coordination, task accuracy, and gait characteristics in response to missed exoskeleton actuations during walking. External torques from the exoskeleton can be learned and anticipated, with internal models developed representing their anticipated dynamics. However, human-exoskeleton coordination may shift across the continuum of trust with the exoskeleton. Changes in trust may be evaluated by differences in performance, which were assessed here using task accuracy, joint kinematics, and step characteristics.

Our first two hypotheses posited that there would be changes in step characteristics and task accuracy in response to catch trials. However, the data do not support effects of missed actuations on these metrics. While NSW can be considered an indicator of mediolateral gait stability [13], there was no effect of catch trials or timing on NSW, indicating that this stability characteristic was not impacted. While gait strategies can be influenced by a dual task [14], the dual task of walking with an exoskeleton and reaching a stepping target in the present study led to consistent step characteristics. Yet, with a loss of exoskeleton torque, a modulation in motor actions is needed to maintain task goals.

Our third hypothesis that joint kinematics would have time-dependent changes in response to catch trials was supported by the data. Peak ipsilateral ankle plantarflexion decreased during catch trials when the exoskeleton's assistive torque was not applied (Fig. 2c, 3f). A previous study observed that as participants adapted to walking with this same exoskeleton, they reduced their muscle-generated plantarflexion [2]. These reductions were interpreted as the participants coordinating with the exoskeleton, as defined through the relative levels of muscle activation accommodating to the provided exoskeleton torque. During catch trials in the present study, the exoskeleton did not generate assistive plantarflexion torque; thus, participants likely relied on their muscle-generated plantarflexion. Reductions in this muscle activity and the resulting plantarflexion moment align with the decreased peak ankle angles observed. Participants may have learned internal models that support human-exoskeleton coordination by lessening their contribution to total ankle plantarflexion. Ipsilateral knee flexion also increased after loading response during catch trials (Fig. 2b, 3e) which may arise due to a diminished extension moment about the knee from the decrease in ankle plantarflexion.

The changes in ankle and knee kinematics align with the feedforward mechanisms of an internal model. Schaefer et al. [15] reported that adaptation occurred during visuomotor reaching tasks when exposed to visual perturbations. When the perturbation was relevant to the task, participants experienced trajectory errors from feedforward internal models, which were corrected to reach the endpoint using visual feedback. In our study, visual feedback was also used to modulate kinematics such that participants corrected for potential endpoint errors due to the lower ankle torque. Participants increased their hip flexion during swing with the ipsilateral leg to place their heel within the stepping target during catch trials compared to baseline (Fig. 3d, 2a). The increase in hip flexion compensated for the decreased ankle torque during push-off, resulting in no net change in NSL and task accuracy. The modulation of hip flexion suggests an increased hip flexion moment, which opposes the intended design goals of the exoskeleton to reduce energy expenditure and impacts human-exoskeleton coordination.

All joint kinematics experienced a time effect where the peak values influenced by the catch trial returned to baseline after one or more steps (Fig. 3). Hip and knee flexion returned to baseline levels after CT+1, while ankle plantarflexion recovered later at CT+5. However, there were no significant differences in step characteristics, task performance, or joint kinematics between earlier and later strides (Fig. 3). A possible explanation for the unchanged step characteristics at CT+2 to 5 is that the decreases in ankle plantarflexion from baseline were relatively small and resulted in non-significant changes in NSL and task accuracy. We hypothesized that as the user's trust decreased, there would exist lasting changes in kinematics or gait characteristics. Acosta-Sojo et al. [2] showed that with consistent actuations using the same Dephy exoskeleton, users can exhibit different physiological responses. Possible indicators of decreased trust may include significant changes between early and late kinematics, such as greater hip flexion and ankle plantarflexion as the user anticipates the loss of torque and attempts to compensate. The unchanged gait characteristics and kinematics between earlier and later strides suggest that trust was maintained throughout the study. While exoskeletons ideally do not experience errors, missed actuations can occur during steady-state operation [16], but may not always be documented. This study had catch trials for 1.68% of the strides (32 of 1900 strides). Higher frequencies of catch trials may impact users' trust in the system and should be further studied.

These data suggest that exoskeleton users adjust their joint kinematics to compensate for the absence of exoskeleton torque to meet task performance goals, impacting human-exoskeleton coordination in that process. A limitation of this study is the assumption that all participants were fully adapted to the exoskeleton after the 30-minute training period, which could influence their motor strategies. Future work will evaluate muscle activation around the hip, knee, and ankle joints to further explore human-exoskeleton coordination with respect to the exoskeleton's design goals. While the exoskeleton in this study was intended to reduce energy consumption, different and more complex systems may result in other strategies. Contralateral leg data will also be analyzed in future work to understand its response to catch trials compared to the ipsilateral leg shown in this study.

V. Conclusion

This study explored the impact of missed exoskeleton actuations on users' human-exoskeleton coordination, task performance, and gait characteristics. When the exoskeleton did not provide an assistive torque, participants maintained acceptable task accuracy by increasing hip flexion to adjust for the missed actuation. Human-exoskeleton coordination was impacted by catch trials, but trust was maintained throughout the study. Understanding the interactions between human-exoskeleton coordination, task accuracy, and step characteristics will enable the design of exoskeleton controllers to support users in non-optimal operational settings.

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