Robot-aided Training of Propulsion During Walking: Effects of Torque Pulses Applied to the Hip and Knee Joints During Stance

Robert McGrath, *Student Member, IEEE,* Barry Bodt, and Fabrizio Sergi, *Member, IEEE*

***Abstract*—We sought to evaluate the effects of the application of torque pulses to the hip and knee joint via a robotic exoskeleton in the context of training propulsion during walking. Based on our previous study, we formulated a set of conditions of torque pulses applied to the hip and knee joint associated with changes in push-off posture, a component of propulsion. In this work, we quantified the effects of hip/knee torque pulses on metrics of propulsion, specifically hip extension (HE) and normalized propulsive impulse (NPI), in two experiments. In the first experiment, we exposed 16 participants to sixteen conditions of torque pulses during single strides to observe the immediate effects of pulse application. In the second experiment, we exposed 16 participants to a subset of those conditions for 200 strides to quantify short-term adaptation effects.**

During pulse application, NPI aligned with the expected modulation of push-off posture, while HE was modulated in the opposite direction. The timing of the applied pulses, early or late stance, was crucial, as the effects were often in the opposite direction when changing timing condition. Extension torque applied at late stance increased HE in both experiments — range of change in HE: $(2.9 \pm 0.4$ **deg,** 7.7 ± 1.0 **deg),** $p <$ **0.001. The same conditions resulted in a negative change in NPI only in the single pulse experiment — change in NPI for knee torque:** -3.0 ± 0.4 **ms,** $p <$ **0.001 — and no significant change for hip torque. Also, knee extension and flexion torque during early and late stance, respectively, increased NPI during single pulse application — range of change in NPI:** $(3.8, 4.6)$ ± 0.8 **ms,** $p <$ **0.001. During repeated pulse application, NPI increased for late stance flexion torque — range of change in NPI:** $(4.5 \pm 0.7$ **ms,** 4.8 ± 0.8 **ms),** $p <$ **0.001, but not late stance extension torque. After exposure, we observed positive after-effects in HE in three conditions — range of change in HE:** $(2.0 \pm 0.3$ **deg,** 3.7 ± 0.7 **deg)** $p <$ **0.001 — and significant positive after-effects in NPI for early stance flexion torques — change in NPI:** $(2.7 \pm 0.6$ **ms,** $p <$ **0.001). These results indicate that positive propulsive after-effects can be achieved through repeated exposure to torque pulses.**

***Index Terms*—Exoskeleton, adaptation, propulsion, hip extension, propulsive impulse**

I. Introduction

In recent years, robot-assisted gait training (RAGT) has been devised and implemented in both research and clinical settings. Currently, the majority of RAGT devices, designed specifically to rehabilitate gait, utilize one of the various controller forms (e.g., force control, position control, or impedance/admittance control), and controller update methods (e.g., assist-as-needed control, inter-limb coordination, or finite state machine), to ultimately promote specific features of gait kinematics [1]. Despite the convergence of research on these methods, the efficacy of RAGT based on kinematic control has not yet exceeded that of conventional gait therapies [2], [3]. The limited efficacy of these methods could be due to their lack of targeting specific functional mechanisms of gait, which are only partially described by joint kinematics.

Gait speed (GS) is a primary indicator of walking function in rehabilitation, as it is associated with a better quality of life [4]. To increase GS during walking, it is necessary to increase the anterior component of the ground reaction force, and specifically its time integral, referred to as propulsive impulse [5]. Propulsive impulse is generated during late stance, right before push-off. Propulsive impulse is determined by push-off kinetics and kinematics; specifically by the moment applied by ankle plantarflexor muscles, and by the posture of the limb during push-off [6]. An established metric for quantifying push-off posture is the trailing limb angle (TLA), defined as the angle between the line formed by the foot center of pressure and hip joint center, at peak propulsive force, and the vertical laboratory axis [7]. In healthy control participants and post-stroke patients, it was observed that the increase in TLA contributes more than the increase in ankle moment to the resulting increase in propulsive force [7], [8]. As such, both push-off posture and propulsive impulse are possible targets for modulation of propulsion during robot-aided training. While it is known that push-off posture and propulsion are associated, it is unclear how to apply a robotic intervention to modulate propulsion, and whether a robotic intervention modulating propulsion during training will ultimately influence walking after training.

As such, we sought to develop a controller capable of training propulsion during walking. We began by investigating the changes in joint moments associated with the experimentally imposed factorial modulation of GS and TLA [9], [10]. We then approximated the effects of push-off posture on joint moments with brief pulses of joint torque. We compiled the amplitude and time of the resulting torque pulses to analyze patterns associated with TLA modulation for each joint. We observed that at the knee, an increase in TLA was associated with extension torque in early stance and flexion torque in late stance. At the hip, early and late stance extension torques were associated with an increase in TLA.

In this work, we seek to determine whether the modulation in joint moments associated with a change in propulsion dynamics, can be applied by a robotic exoskeleton in the form of torque pulses to the hip and knee joints to modify propulsion dynamics in healthy participants.

First, we wished to investigate the instantaneous effects of single-pulse torque intervention on the kinematics and kinetics of gait. As such, for the first protocol of this study, we formulated a set of sixteen conditions of torque pulses, based on the torque patterns associated with TLA modulation in our previous study [9], [10]. The protocol consisted of applying torque pulses to the hip and knee joints of healthy individuals during single strides, using a lower extremity exoskeleton, the ALEX II, while they walked on an instrumented treadmill.

We observed the effects of torque pulses on the outcome measures of hip extension angle (HE) and normalized propulsive impulse (NPI) and determined the predominant factors influencing these measures at the stride of pulse application and the following strides. We hypothesized that each pulsed torque condition would modulate HE and NPI in the same direction as the expected modulation of TLA, for the stride of application and the following three strides.

Second, we wished to measure how participants would adapt to pulsed torque training, and how they would respond right after exposure to training. As such, for the second protocol of this study, we selected a subset of eight torque pulse conditions with the largest effects on HE and NPI observed in the first protocol. The protocol consisted of applying the selected eight pulse conditions to the hip and knee joints for consecutive sets of 200 strides with the ALEX II before and after participants were walking with the exoskeleton controlled to display minimal interaction forces. We hypothesized that each torque pulse condition would exhibit short-term adaptation to pulse application and de-adaptation following pulse application removal. Furthermore, we hypothesized that the outcome measures would exhibit the same direction of modulation during intervention as the single-pulse experiment.

II. Methods

A. Study Participants

The single-pulse experiment included sixteen healthy adult participants (13 male, 3 female) of age (mean $\pm$ std) 25 $\pm$ 2 yrs, height 178 $\pm$ 5 cm and mass 75.6 $\pm$ 8.5 kg. The repeated-pulse experiment included sixteen healthy adult participants (7 male, 9 female) of age 24 $\pm$ 3 yrs, height 171 $\pm$ 10 cm and mass 70.3 $\pm$ 16.7 kg. Three participants (1 male, 2 females) were common to both experiment. Participants were only included if naive to the purpose of the experiment and free of neurological and orthopedic disorders that would affect normal walking function. All participants gave informed consent according to the IRB protocol number 929630 at the University of Delaware and wore their own comfortable lightweight athletic clothing.

B. Equipment

Data collections were conducted on an instrumented split-belt treadmill (Bertec Corp., Columbus OH, USA) that measured analog force/torque data. The ALEX II robot [11], a powered unilateral lower extremity exoskeleton, as seen in Fig. 1, was utilized to apply torque pulses about the right knee and hip joints of participants. The exoskeleton is suspended by a mobile carriage over the instrumented treadmill and secured from moving during experimentation by locking casters. A custom real-time controller written in MATLAB & Simulink (MathWorks Inc., Natick MA, USA) acquired signals from the instrumented treadmill and ALEX II and sent command signals to the two motors at a frequency of 500 Hz. The ALEX II contains two Kollmorgen ACM22C rotary motors with integrated Smart Feedback Devices (Danaher Corporation, Washington D.C., USA). These provide an emulated encoder resolution of 4096 pulses per revolution providing an effective hip and knee angle resolution of 4.4×10^{-4} deg.

Fig. 1: Experimental setup consisting of a participant in the Active Leg EXoskeleton II (ALEX II) while on the instrumented split-belt treadmill.

C. Controller

A high level controller utilizes the vertical ground reaction force, as measured by the instrumented treadmill, to determine the time of right heel strike events and send desired torque values to the low-level controller. Heel strike events define the onset of the gait cycle and the time between events define the duration of the gait cycles. The average of the previous six gait cycles yields an estimated gait cycle time which is used with pulse time values, as percentages of the gait cycle, to determine controller event timing. The high-level controller's selection of pulse condition, as per experimental protocol, specifies the joint torque amplitudes, onset as percentage of gait cycle, and duration as 10% of gait cycle.

A low-level torque controller featuring gravity and friction compensation was developed utilizing direct torque feedback, from two 6-axis force/torque sensors located between the exoskeleton structure and the shank/thigh cuffs. The implementation of force/torque sensor feedback allows for partial compensation of the high friction present in the geared motors and the inertia of the exoskeleton structure [12]. Section D of the Supplementary Materials describes the controller compensation for the delay in torque pulse application and a comparison between prescribed torque pulses and resulting joint torque.

D. Experimental Procedures

1) Single-pulse Experiment: In this experiment, torque pulses were applied only at one stride, followed by 5-7 strides of no pulse application to measure participants' response. Sixteen pulsed torque conditions were tested in this experiment, as shown in Fig. 2. Each condition consisted of square waves with a duration of 10% gait cycle, and a timing of early or late stance, initiating at 10% or 45% gait cycle, respectively. The amplitude of these square wave pulses are 10 N·m or -10 N·m for knee extension or flexion, respectively, and 15 N·m or -15 N·m for hip extension or flexion, respectively.

All participants performed two separate sessions, all while walking at their predetermined self-selected gait speed. Each

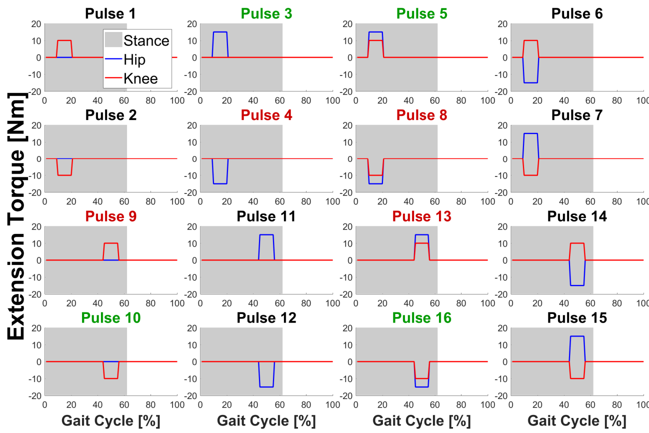

Fig. 2: 16 pulse conditions implemented in the single-pulse experiment and the 8 pulse conditions, with colored titles, implemented in the repeated-pulse experiment.

session consisted of 5 repetitions of each of the 16 pulse conditions in a pseudo-randomized order. Both sessions featured the same sequence of pulses, lasted for a duration of 8-10 minutes, and were separated by a minimum of 5 minutes of rest outside of the exoskeleton. The participants walked at their self-selected speed, determined before the experiment while wearing the exoskeleton in zero-torque mode. Self-selected speed was determined and exoskeleton fitting was performed as in our previous work [13].

2) Repeated-pulse Experiment: In this experiment, torque pulses were applied for 200 consecutive strides, preceded and followed by 100 strides of no pulse application, to measure participants' response. A subset of 8 pulse conditions, out of the full set of 16 pulse conditions, were selected for testing in the repeated-pulse experiment (Fig. 2). The four pulse conditions with green and red titles increased and decreased, respectively, NPI in the single-pulse experiment. The selection process of these pulses is explained further in Sections II-E2c & III-A3

Each session consisted of a total of 700 strides in which strides 101-300 and 401-600 each consisted of a different pulse applied in consecutive strides. The first, middle, and last 100 strides were performed in the absence of pulse application, to allow for measurement of baseline and effects for each pulse condition. All participants performed a total of four separate sessions, two on each of the two days of experimentation while walking at their self-selected gait speed. Each session was separated by a minimum of 5 minutes of rest outside of the exoskeleton and lasted for 12-14 minutes. Each day of experimentation contained a grouping of 4 pulses with the same directional effect on NPI. Both the sequence of day (exposure to pulses increasing or decreasing NPI), and the sequence of the pulses within each day were pseudorandomized across participants.

E. Data Analysis

1) Outcome Measures: Two outcome measures were selected to describe the effects of the intervention on walking kinematics and kinetics. For kinematics, we selected hip extension angle (HE), as the angle of the hip of the right (robot-assisted) leg at the instant of peak anterior ground reaction force (GRF). For kinetics, we selected normalized propulsive impulse (NPI), defined as the integral of the anterior-posterior component of GRF over the time interval that the component is positive. Propulsive impulse was then normalized by the participant's body weight (in N). Changes in outcome measures reported as percentages are normalized by the average baseline value of that outcome measure of the respective experiment.

2) Single-pulse Experiment: In the single-pulse experimentation, outcome measures were observed at a total of five strides per applied individual pulse. This includes the prior baseline stride (-1), the stride of (0), and three strides following (1, 2, 3) application of torque pulses.

a) Linear Mixed Effect Models: We used linear mixed effect models to determine which factors of torque pulse application influenced the outcome measures of HE and NPI. As such, we utilized JMP Pro Version 14 (SAS Institute Inc., Cary, NC, USA) to fit a linear mixed model to each of the single-pulse outcome measure data sets consisting of 1280 data points (16 participants x 16 pulse conditions x 5 strides x 1 mean value). In the linear mixed effects models, each mean data point was assigned a unique combination of values from the following effects: stride number (-1, 0, 1, 2, or 3), pulse timing as a phase of gait cycle (Early Stance or Late Stance), knee torque amplitude in N·m (-10, 0, or 10), hip torque amplitude in N·m (-15, 0, or 15), and participant (1 through 16). We added a null data set for hypothetical pulses 17 and 18 in order to implement the linear mixed effect model which requires a full factorial dataset. The hypothetical pulses 17 and 18 consist of zero torque amplitudes in early and late stance, respectively, by averaging measures from pulses 1 through 8 and 9 through 16 from stride -1, respectively. The averaged data from stride -1 was copied to the remaining four strides (0, 1, 2, 3) within each of the 16 participants.

The fixed effects include stride number (Stride), pulse time as phase in gait cycle (Phase), knee torque (Knee), and hip torque (Hip) which therefore provide 4 main effects, 6 two-way effects, 4 three-way effects, and 1 four-way effect. The random effects include the main effect of participant and 4 two-way effects of participant by stride number, pulse time, knee torque, and hip torque. Main and interaction effect terms that are significant at a false positive rate of $\alpha < 0.05$ will be reported.

b) Pairwise Tests: To determine if any pulse condition significantly modulated the outcome measure, we performed pairwise tests at the group level between the baseline stride (-1) and each of the following strides (0, 1, 2, 3), at a false positive rate of $\alpha < 0.05/16$, given Bonferroni correction for 16 comparisons (one per pulse condition, within a stride condition). The performed pairwise tests were selected on a test by test basis depending upon the normality of the baseline and compared stride data sets. The Shapiro-Wilk parametric hypothesis test of composite normality was utilized to determine the normality of each data set. If both compared data sets were normal, a paired t-tests was performed, otherwise a Wilcoxon signed-rank test was performed.

c) Pulse Selection for Repeated-pulse Experiment: Due to concerns on possible participant fatigue and time constraints, only a subset of the original 16 pulse conditions could be tested in the repeated-pulse experimental protocol. As such, we selected a subset of 8 pulse conditions which had modulated the outcome measure in an amplitude-dependent way (positive effect for pulse A, negative effect for pulse A with negative magnitude).

3) Repeated-pulse Experiment: In the repeated-pulse data analysis, the 200 strides of pulse torque intervention is referred to as pulse application, the preceding 100 stride sub-section is baseline, and the 100 stride sub-section following pulse application is after-effects. To perform statistical analysis, we defined five time points of measurement (TP) within each section: baseline (BL) - last 20 strides before intervention, early pulse application (P-E) - strides 2-6 after start of intervention, late pulse application (P-L) - last 5 strides of intervention, early after-effects (AE-E) - strides 2-6 after the end of intervention, and late after-effects (AE-L) - last 5 strides of no pulse condition after intervention. At each of these time points, we obtained the outcome measure as the mean for the designated strides. Stride-averaged outcome measures are used for analysis in Sections II-E3a & II-E3c.

a) Linear Mixed Effect Models: The pulses considered in the repeated-pulse experiment do not span factorially all combinations of pulse factors like in the single-pulse experiment. As such, to determine which factors of torque pulse application influenced the primary outcome measures of HE and NPI in the repeated-pulse experiment, we developed two separate linear mixed models, where each model examines the data of four of the eight pulse conditions. Therefore, each model examines a data set of 320 points (16 participants x 4 pulse conditions x 5 time points). The linear mixed effect Model A includes the four pulses where torque is applied to both the hip and knee, at the same time (Table I). Model A effects are pulse time as the phase of gait cycle (Early or Late Stance), knee & hip torque direction (Flex or Ext), time point of measurement (BL, P-E, P-L, AE-E, or AE-L), and participant (1 through 16). The fixed effects include the main, two-way, and three-way effects of pulse time, direction, and time point. The random effects include the main effect of participant and two-way interaction of participant and the main effects. The linear mixed effect Model B includes the four single joint pulse conditions (Table II). Model B effects include knee & hip torque direction, and joint & phase combination (Hip & Early Stance or Knee & Late Stance), time point of measurement, and participant. The fixed effects include the main, two-way, and three-way effects of direction, joint & pulse time combination, and time point of measurement. The random effects include the main effect of participant and two-way interaction between participant and the main effects. Both models for both measures have fixed effects tests conducted with a false positive rate of $\alpha < 0.05$.

b) Pairwise Tests: Pairwise tests were performed on the measured data, not the modeled values, to establish whether any pulse condition significantly modulated the outcome measures during pulse application (2 paired tests per pulse pairing P-E with BL and P-L with BL), and after pulse application (2 paired tests per pulse pairing AE-E with BL and AE-L with BL). The Shapiro-Wilk test was used to detect normality of the paired samples. If the samples were normal, a t-test was performed, otherwise a Wilcoxon signed-rank test was performed. For either test, a false-positive rate of $\alpha = 0.05/32$ was selected, using a Bonferroni correction to account for 32 comparisons (4 comparisons per pulse and 8 pulses).

Model A		Phase	
		Early Stance	Late Stance
Direction	Ext	Pulse #5	Pulse #13
	Flex	Pulse #8	Pulse #16

TABLE I: Model A consists of the four double-joint pulses with the shown break-down of factors.

Model B		Joint & Phase	
		Hip & Early	Knee & Late
Direction	Ext	Pulse #3	Pulse #9
	Flex	Pulse #4	Pulse #10

TABLE II: Model B consists of the four single-joint pulses with the shown break-down of factors.

c) Responder Analysis: We sought to establish whether the response of individual participants to the intervention followed a pattern of adaptation or learning (i.e., whether after-effects were in the opposite or in the same direction of effects measured during training). To establish patterns at the individual participant level, we defined the Z-score for phase P-E and AE-E (early pulse application and early after-effects) of each pulse condition for all 16 participants as:

$$Z_{Phase} = \frac{\mu_{Phase} - \mu_{BL}}{\sigma_{BL}} \quad (1)$$

Based on the relative signs of change in outcome measures in the two phases, each participant's response will follow one of four patterns: positive learning, negative learning, positive adaptation, or negative adaptation (Table III). Each pulse will then have n responders for each pattern, with n defined as the number of participants whose response to a condition follows a specific pattern. For each pulse, we report which pattern was the one with the largest number of responders, and the mean Z-scores of responders.

Pattern	**P-E Z-Score**	**AE-E Z-Score**
Positive Learning	Positive	Positive
Negative Learning	Negative	Negative
Positive Adaptation	Negative	Positive
Negative Adaptation	Positive	Negative

TABLE III: Definition of response patterns based on Z-scores measured in the early pulse application (P-E), and early after-effects (AE-E).

III. Results

A. Single-pulse Experiment

1) HE:

a) Linear Mixed Effect Models: The linear mixed effect model for HE had an R^2 adjusted of 0.91. All of the statistically significant fixed effect terms are shown in Table S1 for

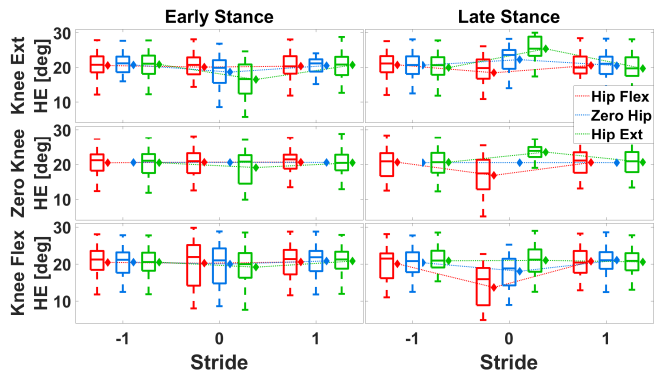

Fig. 3: Breakdown of HE measurements (boxplots) and model estimates (diamonds) for different levels of the four factors considered in the single-pulse experiment.

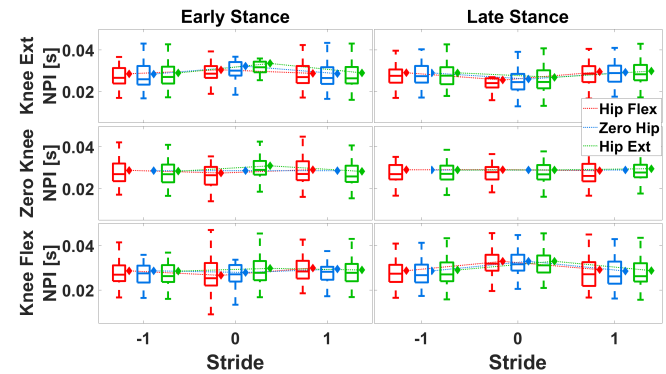

Fig. 4: Breakdown of NPI measurements (boxplots) and model estimates (diamonds) for different levels of the four factors considered in the single-pulse experiment.

HE. The linear mixed model means for HE and measurement distributions are shown in Fig. 3. For the significant effects involving stride, the least squares means estimates of levels or post-hoc Tukey HSD pairwise comparisons are presented.

The significant main effect of stride was driven by a reduction in HE during the stride of pulse application (Stride -1: 20.5 ±1.0 deg, Stride 0: 19.7±1.0 deg, $p < 0.001$). The only significant paired comparisons were measured between HE at stride 0 and all other strides. The significant two-way interaction of stride and knee torque was driven by a significant decrease in HE from stride -1 to 0 for knee flexion torque (change in HE, Knee Flex: -1.8±0.2 deg, $p < 0.001$) which is significantly greater than the decrease in HE for knee extension torque (change in HE, Knee Ext: -0.4±0.2 deg, $p < 0.001$). The significant two-way interaction of stride and hip torque was driven by a significant decrease in HE from stride -1 to 0 for hip flexion torque (change in HE, Hip Flex: -2.2±0.2 deg, $p < 0.001$) which was significantly more negative than the change in HE from stride -1 to 0 for hip extension torque (change in HE, Hip Ext: 0.2±0.2 deg, $p < 0.001$). The significant two-way interaction of stride and phase was driven by a significant decrease in HE from stride -1 to 0 in early stance (change in HE: Early Stance: -1.1±0.2 deg, $p < 0.001$). The significant three-way interaction of stride, phase, and knee torque was driven by a significant decrease in HE from stride -1 to 0 for knee extension torque at early stance (change in HE: Early Stance, Knee Ext: -2.3±0.3 deg, $p < 0.001$), by a significant increase in HE for knee extension torque at late stance (change in HE, Late Stance, Knee Ext: 1.6±0.3 deg, $p < 0.001$), and by a significant decrease in HE for knee flexion torque at late stance (change in HE, Late Stance, Knee Flex: -2.9±0.3 deg, $p < 0.001$). The significant three-way interaction of stride, phase, and hip torque was driven by a significant decrease in HE from stride -1 to 0 for hip extension torque at early stance (change in HE: Early stance, Hip Ext: -2.4 ± 0.3 deg, $p < 0.001$), by a significant increase in HE for hip extension at late stance (change in HE: Late stance, Hip Ext: 2.8 ± 0.3 deg, $p < 0.001$), and by a significant decrease in HE for hip flexion at late stance (change in HE: Late stance, Hip Flex: -4.1 ± 0.3 deg, $p < 0.001$).

b) Pairwise Tests: The outcome measure of HE had statistically significant pairwise comparisons present in seven of the sixteen total pulse conditions (Fig. 5). The only significant pairwise differences were between baseline stride (-1) and the stride of pulse application (0), as seen in Table IV. Pulse conditions 1 and 5, both containing knee extension torque pulses in early stance, decreased HE. Flexion pulses at late stance, conditions 10 and 12, both decreased HE, and combine to form pulse condition 16 which had an even greater decrease in HE. Pulse conditions 11 and 13 both feature late stance hip extension torque pulses and increased HE.

2) *NPI:*

a) Linear Mixed Effect Models: The linear mixed effect model for NPI had an R^2 adjusted of 0.98. All of the statistically significant fixed effect terms are shown in Table S2 for NPI. The linear mixed model means for NPI with measurement distributions are shown in Fig. 4. For the significant effects involving stride, the least squares means estimates of levels or post-hoc Tukey HSD pairwise comparisons are presented.

The significant main effect of stride was driven by an increase in NPI during the stride of pulse application (Stride -1: 28.7±2.1 ms, Stride 0: 29.5±2.1 ms, $p < 0.001$). The significant paired comparisons were measured between NPI at stride 0 and all other strides. The significant two-way effect of stride and knee torque was driven by a significant increase in NPI from stride -1 to 0 for knee flexion torque (change in NPI: Knee Flex: 1.8±0.2 ms, $p < 0.001$) which was significantly greater than the change measured for knee extension torque from stride -1 to 0 (change in NPI: Knee Ext: 0.28±0.20 ms, $p < 0.001$). The significant two-way interaction of stride and hip torque was driven by a significant increase in NPI from stride -1 to 0 in hip extension torque (change in NPI: Hip Ext: 1.7±0.2 ms, $p < 0.001$) which is significantly greater than the change measured for hip flexion torque (change in NPI: Hip Flex 0.1±0.2 ms, $p < 0.001$). The significant two-way interaction of stride and phase was driven by a significant increase in NPI from stride -1 to 0 in early stance (change in NPI: Early Stance: 1.1±0.2 ms, $p < 0.001$) which was a significantly greater than the increase measured in late stance (change in NPI: Late Stance: 0.5±0.2 ms, $p = 0.008$).

The significant three-way interaction of stride, phase, and

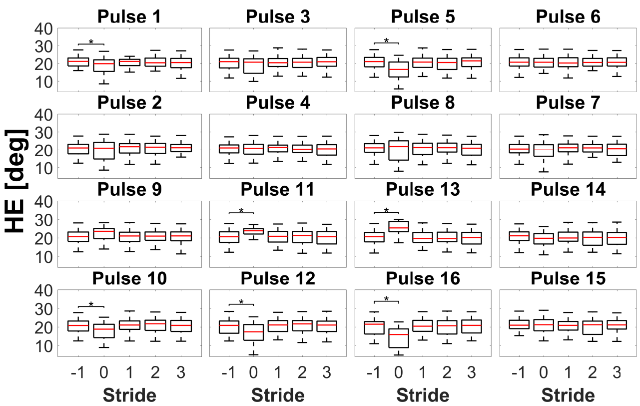

Fig. 5: Group-wise depiction of HE by stride for all pulse conditions in the single-pulse experiment. Asterisks indicate a statistically significant comparison between the baseline stride and specified following stride.

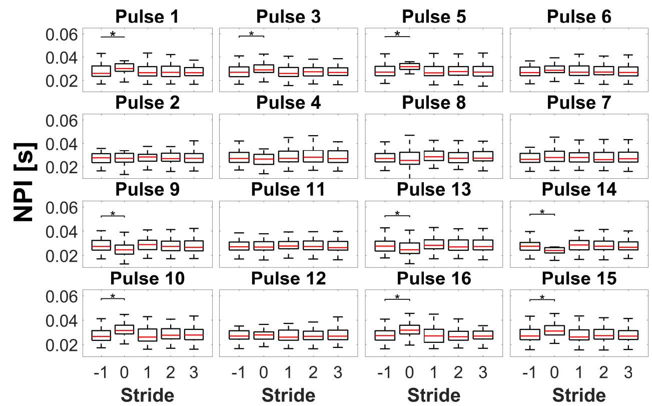

Fig. 6: Group-wise depiction of NPI by stride for all pulse conditions in the single-pulse experiment. Asterisks indicate a statistically significant comparison between the baseline stride and specified following stride.

Pulse #	Expected effect on TLA	HE Effect Size	HE p Value	NPI Effect Size	NPI p Value
1	Increase	-0.999	**0.001**	1.198	**<0.001**
2	Decrease	-0.184	0.472	-0.297	0.379
3	Increase	-0.540	0.047	1.852	**<0.001**
4	Decrease	0.095	0.709	-0.692	0.014
5	Increase	-1.599	**<0.001**	1.413	**<0.001**
6	Unknown	-0.436	0.101	0.786	0.010
7	Unknown	-0.487	0.070	0.927	0.003
8	Decrease	-0.057	0.823	-0.629	0.023
9	Decrease	0.583	0.034	-1.836	**<0.001**
10	Increase	-1.317	**<0.001**	3.530	**<0.001**
11	Increase	1.781	**<0.001**	0.065	0.836
12	Decrease	-1.487	**<0.001**	0.037	0.679
13	Unknown	1.597	**<0.001**	-1.190	**<0.001**
14	Decrease	-0.508	0.020	-2.193	**<0.001**
15	Increase	0.081	0.749	3.392	**<0.001**
16	Unknown	-1.622	**<0.001**	2.081	**<0.001**

TABLE IV: Expected effects on TLA based on our previous work [10], effect size, and p value for the pulse of application relative to baseline for measured HE and NPI, (threshold $p = 0.05/16 = 0.003$) for each of the 16 pulse conditions in single-pulse application

knee torque was driven by a significant increase in NPI from stride -1 to 0 for early stance knee extension torque (change in NPI: Early Stance, Knee Ext: 3.4±0.3 ms, $p < 0.001$), by a significant decrease in NPI for late stance knee extension torque (change in NPI: Late Stance, Knee Ext: -2.9±0.3 ms, $p < 0.001$) and by a significant increase in NPI for late stance knee flexion torque (change in NPI: Late Stance, Knee Flex: 4.2±0.3 ms, $p < 0.001$). The significant three-way interaction of stride, phase, and hip torque was driven by a significant increase in NPI from stride -1 to 0 for early stance hip extension torque (change in NPI: Early Stance, Hip Ext: 2.8±0.3 ms, $p < 0.001$).

b) Pairwise Tests: The outcome measure of NPI had statistically significant pairwise comparisons in seven pulse conditions, only between baseline stride (-1) and the stride of pulse application (0) (Fig. 6). Hip extension at early stance lead to a significant increase in NPI. Pulse conditions 9, 13, and 14 all contained late stance knee extension torque and lead to a decrease in NPI. Conversely, pulse conditions 10, 15, and 16 all contain late stance knee flexion torque and lead to an increase in NPI.

3) Pulse Selection for Repeated-pulse Experiment: The selected subset of 8 pulse conditions include four pulse conditions which increased NPI and decreased HE: 3, 5, 10, and 16, during the stride of pulse application, relative to the baseline stride. The other four pulse conditions decreased NPI and increased HE: 4, 8, 9, and 13, during the stride of application, relative to the baseline stride. These selected pulse conditions are depicted in Fig. 2 where the pulse title colors are red or green corresponding to a decrease or increase, respectively, of NPI in single pulse application.

B. Repeated-pulse Experiment

The group means by stride for HE and NPI are shown for all eight repeated-pulse conditions in Figs. S1-S2.

1) HE:

a) Linear Mixed Effect Models: The linear mixed effect Models A & B for HE had an R^2 adjusted of 0.81 and 0.68, respectively. The statistically significant fixed effects tests for both HE models are given in Table S3. The linear mixed model estimates of means with standard errors is given in Fig. 7; in which the first and second subplot columns correspond to Models A & B, respectively.

The significant two-way interaction in Model A for HE of time point of measurement and direction was driven by multiple pairwise comparisons within direction and between BL and post-BL time points. For extension pulses, HE significantly increased at all time points, compared to baseline: P-E (2.0±0.6 deg, p = 0.030), P-L (3.8±0.6 deg, $p < 0.001$), AE-E (2.0±0.6 deg, p = 0.042), AE-L (2.0±0.6 deg, p = 0.041). For flexion pulses, the only significant pairwise comparison was a decrease in HE relative to BL at P-E (-2.2±0.6 deg, p = 0.009). The significant three-way interaction in Model A of time point of measurement, phase, and direction was driven by six significant pairwise comparisons. For extension at late stance, HE increased significantly relative to BL at P-E (7.2±0.9 deg, $p < 0.001$) and P-L (7.7±0.9 deg, $p <$

0.001). Within the flexion direction and early stance phase, HE increased significantly relative to BL at P-L (3.3±0.9 deg, p = 0.044). Within the flexion direction and late stance phase, HE significantly changed relative to BL at P-E (-7.3±0.9 deg, $p <$ 0.001), P-L (-5.6±0.9 deg, $p <$ 0.001), and AE-E (3.7±0.9 deg, p = 0.008).

The main effect in Model B for HE of time point of measurement was significant in which P-L (17.0 $\pm$ 1.1 deg, p = 0.017), AE-E (17.1 $\pm$ 1.1 deg, p = 0.006), and AE-L (17.2 $\pm$ 1.1 deg, p = 0.002) all are of significantly greater HE than BL (16.0 $\pm$ 1.1 deg). The two-way effect in Model B of time point of measurement and direction pairwise comparison was an increase in HE within extension, from BL to P-L (2.2 $\pm$ 0.5 deg, p = 0.003). The three-way interaction between time point of measurement, joint & phase, and direction was driven by an increase of HE measured for knee & late stance extension pulses, relative to BL, during P-E (4.7±0.9 deg $p <$ 0.001) and P-L (5.3±0.9 deg $p <$ 0.001).

b) Pairwise Tests: The effect size of the change in HE between BL and post-BL phases are reported in Table V. Two out of the eight pulses significantly changed NPI at P-E, relative to BL in which both pulses (10 and 16 - both including late stance flexion torque) induced a positive change. One condition, pulse 8 (early flexion torque), increased NPI at AE-L, relative to BL.

2) NPI:

a) Linear Mixed Effect Models: The linear mixed effect Models A & B for NPI had an R^2 adjusted of 0.66 and 0.30, respectively. The statistically significant fixed effects tests for both NPI models are given in Table S3. The linear mixed model estimates of means with standard errors are given in Fig. 8; in which the first and second subplot columns correspond to Models A & B, respectively. In Model A for NPI, there were no significant between time point of measurement and within phase and direction pairwise comparisons to report. Similarly, in Model B for NPI, there were no significant between time point of measurement and within joint & phase and direction

Pulse #	HE				NPI			
	P-E	P-L	AE-E	AE-L	P-E	P-L	AE-E	AE-L
3	-0.774	-0.288	0.098	0.607	0.419	0.247	0.007	0.677
4	**1.484**	0.889	0.117	0.429	-0.377	-0.141	0.728	0.204
5	-0.815	-0.045	**1.562**	0.912	0.213	0.020	-0.087	0.230
8	0.944	**1.021**	0.007	0.208	-0.195	0.537	0.816	**1.080**
9	**2.326**	**1.721**	0.815	0.762	-0.382	-0.154	0.340	0.243
10	**-1.729**	**-1.057**	0.806	0.954	**1.561**	0.310	-0.068	0.163
13	**2.272**	**1.960**	0.889	**1.569**	-0.447	0.018	0.154	0.294
16	**-2.553**	**-1.908**	**1.333**	0.674	**1.566**	0.209	-0.277	0.419

TABLE V: The effect size for all pairwise comparisons between baseline and each following phase for all eight repeated-pulse conditions. Values are bolded if statistically significant at p = 0.05/32.

pairwise comparisons to report.

b) Pairwise Tests: The effect size of the change in NPI between BL and post-BL phases are reported in Table V. Five out of eight pulses significantly changed HE at P-E, relative to baseline, in which three pulses (4, 9, and 13 - all including early stance flexion or late stance extension torques) induced a positive change, and two pulses (10 and 16, all including late stance flexion torque) induced a negative change. The effects of these pulses were sustained at P-L for all pulses except pulse 4. Pulse 8 (early stance flexion torque) exhibited an increase in HE at P-L following a non-significant, but of similar magnitude, increased at P-E. Significant changes in HE were measured at AE-E for two pulses - 5 (positive increase in HE, in the opposite direction of non-significant effects measured during P conditions), and 16 (positive increase in HE, in the opposite direction of effects measured during P conditions). One pulse (13) showed a significant change in HE in AE-L, resulting from a positive increase in HE, in the same direction of effects measured during P conditions.

3) Responder Analysis:

a) HE: For the measure of HE, pulse conditions 3, 5, 10 and 16 had the dominant pattern of positive adaptation, with high proportions of the participants following the pattern:

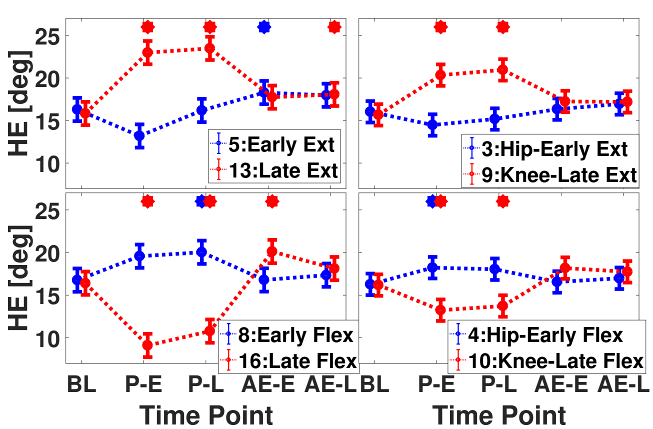

Fig. 7: Breakdown of HE by factor for the eight pulses tested in the repeated-pulse experiment. Diamonds indicate model-estimated means, whiskers indicate s.e.m., asterisks indicate statistically significant pairwise comparison to respective baseline estimate.

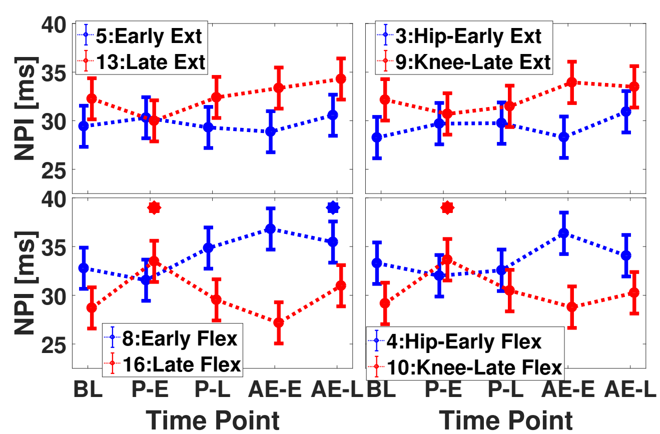

Fig. 8: Breakdown of NPI by factor for the eight pulses tested in the repeated-pulse experiment. Diamonds indicate model-estimated means, whiskers indicate s.e.m., asterisks indicate statistically significant pairwise comparison to respective baseline estimate.

Pulse #	HE				NPI			
	+Lrn	-Lrn	+Adpt	-Adpt	+Lrn	-Lrn	+Adpt	-Adpt
3	1	2	**12**	1	**7**	2	2	5
4	**9**	1	1	5	5	2	**9**	0
5	3	1	**12**	0	**5**	2	4	**5**
8	**9**	2	1	4	**8**	1	7	0
9	**13**	0	0	3	2	3	**8**	3
10	0	2	**14**	0	**8**	1	0	7
13	**11**	0	0	5	3	4	**6**	3
16	1	1	**14**	0	5	0	1	**10**

TABLE VI: Number of repeated-pulse responders for each behavioral pattern for each pulse condition and outcome measure. Bold entries indicate the dominant pattern for a specific pulse condition and outcome measure.

12, 12, 14, and 14 participants, respectively. The other four pulse conditions (4, 8, 9, and 13) had the dominant pattern of positive learning, with 9, 9, 12, and 11 participants following the pattern, respectively (Fig. 9 & Table VI).

b) NPI: For the measure of NPI, pulse conditions 3, 8, and 10 all had the dominant pattern of positive learning with 7, 8, and 8 participants following the pattern, respectively. Pulse conditions 4, 9, and 13 all had the dominant pattern of positive adaptation with 9, 8, and 6 participants following the pattern, respectively. The dominant pattern of pulse condition 16 was negative adaptation with a total of 10 participants following the pattern. As for pulse condition 5, there was an equal number of participants following the two dominant patterns of positive learning and negative adaptation with a total of 5 participants for each (Fig. 10 & Table VI)

IV. Discussion

In this work, we used a robotic exoskeleton to apply torque pulses to the hip and knee joints during stance with the intention of modulating propulsion dynamics. The selection of the set of pulses to test was informed by our prior work, where we investigated the differences in joint moments associated with modulation of push-off posture (a biomechanical component of propulsion) during walking. We conducted two experiments to measure the effects of torque pulses on biomechanical measures of propulsion, defined as normalized propulsive impulse (NPI) and hip extension (HE). In the first experiment (single-pulse experiment), we applied pulses only during single strides to quantify the intrinsic effects of the intervention - i.e., the effects measured under minimal changes in the participants's neuromotor coordination. In the second experiment (repeated-pulse experiment), we applied pulses continuously over multiple strides, to measure how participants would respond during and after the exposure to robotic training. Effects measured during the stride of pulse application were mostly in the same direction across the two experiments, while the effects measured after exposure to torque pulses largely differed across the two experiments. Analysis of data collected during the repeated-pulse experiment show the occurrence of either use dependent learning patterns or adaptation patterns for the selected outcome measures, depending on the specific pulse condition examined. We will discuss the effects during pulse application and after pulse application, for both experiments, separately, in the following sections.

A. Effects during pulse application

The changes in HE and NPI measured during pulse application in the repeated-pulse experiment generally aligned with those measured in the single-pulse experiment. In fact, the changes in both outcome measures for the single-pulse experiment were in the same direction as the changes measured in the first few strides during the repeated-pulse experiment for all conditions except one - HE associated to pulse 8 - whose effect size in the single-pulse experiment was close to zero (-0.057). Interestingly, the effects measured in NPI aligned with the expected modulation of TLA based on our previous biomechanical investigation [9], [10], while the effects measured in HE were usually in the opposite direction as NPI.

We found that the phase and torque direction were significant factors, as a given pulse condition often induced a reversed effect in HE or NPI with a change in phase or direction condition. For example, in both experiments, knee flexion and extension torques during late stance decreased and increased HE during pulse application, respectively. In the single pulse experiment, hip flexion and extension torques during

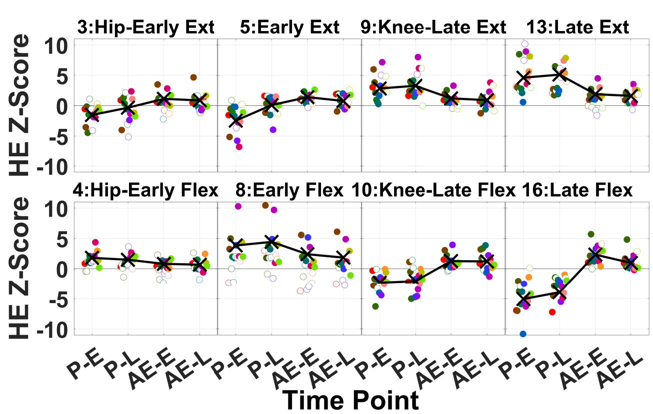

Fig. 9: Z-scores of responders to the dominant pattern for HE. Filled circles are individual participant z-scores, crosses are group mean Z-scores.

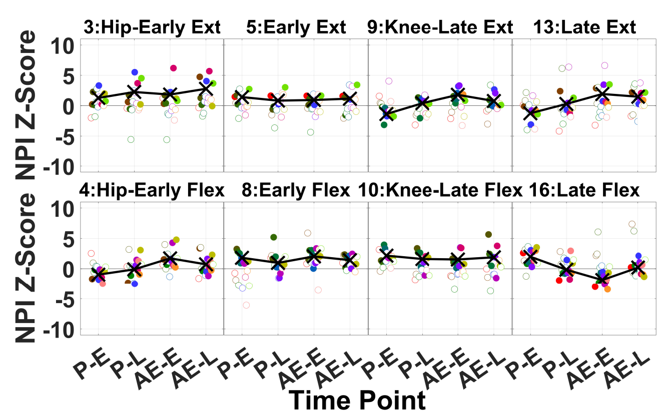

Fig. 10: Z-scores of responders to the dominant pattern for NPI. Filled circles are individual participant z-scores, crosses are group mean Z-scores.

late stance decreased and increased HE, respectively. Similarly, in the repeated pulse experiment, double joint flexion and extension torques during late stance decreased and increased HE, respectively. Conversely, early stance extension torques only decreased HE during pulse application significantly in the single pulse experiment. Early stance flexion torques significantly increased HE in the repeated pulse experiment only. In the single pulse experiment, early stance hip and and knee extension torques increased NPI, and late stance knee extension torque decreased NPI, during pulse application. Late stance knee flexion torque increased NPI during application and was confirmed in the repeated pulse experiment. Similarly, late stance double joint flexion torque increased NPI in the repeated pulse experiment.

A possible explanation for the alignment between the measured change in NPI and the expected change in TLA could be the fact that the applied torque pulses effectively modulated TLA, thus facilitating the generation of increased NPI with a roughly constant net ankle moment. The fact that the measured effect on HE was on the opposite direction as expected may arise from the fact that HE is only a surrogate for TLA, confounded by variables such as knee and ankle angles that may be crucially modulated during push-off. The agreement of the effects in HE and NPI between the two experiment suggests that effects during pulse application, especially in the first strides, are primarily due to biomechanics, and influenced by the participants' neuromotor response only to a small extent. Certainly, the active response of participants plays a larger role during later strides of pulse application, where some NPI effects cease to be significant (specifically pulses 10 and 16). The general decrease of significance of effects in NPI in later strides is certainly also influenced by physical constraints of the experimental setup, as discussed later.

To our knowledge, the effect of exoskeleton applied knee and hip torque during stance on propulsion mechanics have not been investigated prior to our work. Thus, a quantitative comparison between our results and previous work is not possible. Several previous studies investigated the effect of ankle torque during stance on joint kinematics, muscle activation, and metabolic rate, with consistent reports that ankle plantarflexion torque, as applied by an exoskeleton during late stance, can reduce metabolic cost and muscle activation, while subjects walked at fixed GS [14]–[16].

Previous studies using robotic intervention applied to the hip joint reported some effects on metrics related to propulsion. In [17], repeated application of 50% of nominal inverse-dynamics calculated hip torque lead to a significant reduction in HE. Our data with pulse conditions only applied to the hip joint do not support the previous finding, as the only purely assistive hip torque condition (pulse 11) resulted in an increase in HE in the single-pulse experiment. Moreover, when this torque condition was coupled with knee extension torque (pulse 13), such assistive hip torque condition resulted in a large increase in HE during pulse application. As such, our study is the first to demonstrate increases in HE induced via hip joint torque applied at late stance. Early stance hip extension torque applied repeatedly over multiple strides has been previously shown to support propulsive function as shown by reduction in metabolic cost [18], and in muscle activity [19]. In contrast, in our study, the effects of this condition on NPI were positive but not significant at the group level, and smaller than other conditions such as late stance flexion torque.

B. *Effects after pulse application*

In the single-pulse experiment, we did not measure any significant modulation in outcome measures in the strides following pulse application (Figs. 5 & 6). As for the repeated-pulse experiment, we measured significant changes in HE during after-effects, relative to BL, in three of the eight pulses applied. Significant after-effects in HE were only measured in the positive direction, as a result of the application of hip and knee extension torques in early or late stance, and as the result of the application of hip and knee flexion torques in late stance. Interestingly, only one condition (hip and knee extension torques at late stance), yielded after-effects in HE that persisted after the first few strides following training. This was the experimental condition that yielded the largest positive change in HE during pulse application both in the single-pulse and in the repeated-pulse experiment. One condition, early stance flexion torque, significantly increased NPI during late after-effects.

The evolution of the outcome measures appeared to follow patterns of learning or adaptation. Pulse condition 5 (early stance extension torques) exhibited significant positive after-effects in HE at the group level. The responder analysis classified the patterns for HE as positive adaptation pattern for HE, and - interestingly - learning pattern for NPI. Pulse condition 16 (late stance flexion torques) exhibited significant positive after-effects in HE; in which the responder analysis established a positive adaptation pattern for HE but a negative adaptation pattern for NPI. Pulse condition 13 (late stance extension torques) exhibited significant positive after-effects in HE, with responses classified as positive learning for both HE and NPI. Overall, the effects measured after pulse application are generally positive for both HE and NPI. However, there does not appear to be a clear pattern of association between the learning patterns and adaptation patterns across pulse conditions in the repeated pulse experiment. Also, the magnitude of the effects in HE are generally greater than those in NPI.

To our knowledge, the after-effects of hip and knee joint torques applied by an exoskeleton on propulsive measures have not yet been studied. However, previous work has examined the effects of exoskeleton applied force perturbations during the swing phase of gait. In two studies where an exoskeleton without pelvic constraints applied vertical perturbation forces during swing, adaptation patterns were induced in step height, with a positive effect during intervention followed by negative effects after intervention [20], [21]. A later study did not exhibit adaptation to vertical perturbations, likely due to a constrained pelvis configuration in the exoskeleton, but exhibited adaptation in step length to perturbations in the anterior/posterior direction [22]. As such, we believe that this study is the first to report after-effects in propulsion mechanics following the application of joint torque via a lower extremity exoskeleton.

C. Study limitations

The statistical model describing effects of single joint pulses for the repeated-pulse analysis of NPI was only able to account for a limited portion of the total variance in the dataset (R^2 adjusted = 0.3). This is likely due to the small effect of single joint pulses on NPI under the physical constraints of a fixed speed treadmill in conjunction with the limited anterior lunge of the robot (participant reduces propulsive effort to cease infringing upon the motion constraints of the setup), and a relatively noisy NPI measure yielding a low signal-to-noise ratio. However, it is important to note that this is the only model with a low R^2 adjusted, and despite the high model variability, the conservative Bonferroni corrected pairwise comparisons indicate the presence of significant modulation.

While it is known that push-off posture of the trailing limb is associated with propulsion, we were unable to directly measure TLA due to limitations in the experimental setup and utilized the surrogate measure of HE. Previous work evaluating the reliability of the ALEX II platform measures indicate that under comparable conditions, the anatomical hip range of motion is underestimated by 13-15% by the robot [23] compared to a measurement obtained with a goniometer. Yet, this mismatch is unlikely to affect our results, which are based on evaluation of the same measure across time points.

Only a subset of the 16 conditions were implemented in the repeated pulse protocol, based on single-pulse effects, due to practical limitations on session duration given participant fatigue. As a consequence, double joint pulse conditions with potentially strong effects were excluded, and a full factorial design was not possible thereby preventing a test of all interactions between factors, as done in the single-pulse analysis.

Given these limitations, we wish to implement an experiment in the future to better achieve and assess propulsion modulation. This experiment would involve an adaptive treadmill controller, which has the capacity to modulate treadmill belt speed in direct response to the real-time gait parameters of the participant [24], motion capture to directly measure TLA, and a full factorial design of pulse conditions.

D. Conclusion

In summary, joint torque pulses applied at the hip and knee by an exoskeleton during stance can modulate the propulsive measures of HE and NPI. While directional effects during application can be both negative and positive, the after-effects of these torque pulses are generally positive in direction. Most importantly, we have shown that in a specific condition (early stance flexion torque) sustained positive after-effects in NPI are achieved. These findings suggest that it would be feasible to develop and implement a propulsive training paradigm for individual post-stroke, for improved propulsion and ultimately GS.

Acknowledgment

This work was supported by the National Science Foundation (NSF-NRI-1638007) and by an Institutional Development Award (IDeA) from the National Institute of General Medical Sciences of the National Institutes of Health (NIH-U54-GM104941).

References

[1] J. Rosen and P. W. Ferguson, eds., *Wearable Robotics: Systems and Application*. Academic Press, 1st ed., 2020.

[2] B. H. Dobkin and P. W. Duncan, "Should Body Weight Supported Treadmill Training and Robotic Assistive Steppers for Locomotor Training Trot Back to the Starting Gate?," *Neurorehabil Neural Repair*, vol. 26, no. 4, pp. 308–317, 2014.

[3] J. Mehrholz, S. Thomas, C. Werner, J. Kugler, M. Pohl, and B. Elsner, "Electromechanical-assisted training for walking after stroke (Review)," *Cochrane Database of Systematic Reviews*, no. 5, 2017.

[4] A. Schmid *et al.*, "Improvements in Speed-Based Gait Classifications Are Meaningful," *Stroke*, vol. 38, pp. 2096–2101, 2007.

[5] M. G. Bowden, C. K. Balasubramanian, R. R. Neptune, and S. A. Kautz, "Anterior-Posterior Ground Reaction Forces as a Measure of Paretic Leg Contribution in Hemiparetic Walking," *Stroke*, vol. 37, pp. 872–877, 2006.

[6] C. L. Peterson, J. Cheng, S. A. Kautz, and R. R. Neptune, "Leg extension is an important predictor of paretic leg propulsion in hemiparetic walking," *Gait & Posture*, vol. 32, no. 4, pp. 451–456, 2010.

[7] H. Hsiao, B. A. Knarr, J. S. Higginson, and S. A. Binder-Macleod, "Mechanisms to increase propulsive force for individuals poststroke.," *Journal of Neuroengineering and Rehabilitation*, vol. 12, no. 40, 2015.

[8] H. Hsiao, B. A. Knarr, J. S. Higginson, and S. A. Binder-Macleod, "The Relative Contribution of Ankle Moment and Trailing Limb Angle to Propulsive Force during Gait," *Human Mov. Sci.*, pp. 212–221, 2015.

[9] R. L. McGrath, M. Pires-Fernandes, B. Knarr, J. S. Higginson, and F. Sergi, "Toward goal-oriented robotic gait training: The effect of gait speed and stride length on lower extremity joint torques," *IEEE ICORR*, pp. 270–275, 2017.

[10] R. L. McGrath, M. L. Ziegler, M. Pires-Fernandes, B. A. Knarr, J. S. Higginson, and F. Sergi, "The effect of stride length on lower extremity joint kinetics at various gait speeds," *PLOS ONE*, vol. 14, feb 2019.

[11] K. N. Winfree, P. Stegall, and S. K. Agrawal, "Design of a minimally constraining, passively supported gait training exoskeleton: ALEX II," *IEEE ICORR*, pp. 0–5, 2011.

[12] D. Zanotto, T. Lenzi, P. Stegall, and S. K. Agrawal, "Improving transparency of powered exoskeletons using force/torque sensors on the supporting cuffs," in *IEEE ICORR*, pp. 0–5, 2013.

[13] R. L. McGrath and F. Sergi, "Single-stride exposure to pulse torque assistance provided by a robotic exoskeleton at the hip and knee joints," *IEEE ICORR*, vol. 2019-June, pp. 874–879, 2019.

[14] A. T. Asbeck, S. M. De Rossi, K. G. Holt, and C. J. Walsh, "A biologically inspired soft exosuit for walking assistance," *International Journal of Robotics Research*, vol. 34, no. 6, pp. 744–762, 2015.

[15] P. Malcolm, W. Derave, S. Galle, and D. D. Clercq, "A Simple Exoskeleton That Assists Plantarflexion Can Reduce the Metabolic Cost of Human Walking," *Plos One*, vol. 8, no. 2, pp. 1–7, 2013.

[16] J. Zhang *et al.*, "Human-in-the-loop optimization of exoskeleton assistance walking," *Science*, vol. 356, no. June, pp. 1280–1284, 2017.

[17] T. Lenzi, M. C. Carrozza, and S. K. Agrawal, "Powered Hip Exoskeletons Can Reduce the User's Hip and Ankle Muscle Activations During Walking," *IEEE TNSRE*, vol. 21, no. 6, pp. 938–948, 2013.

[18] Y. Ding *et al.*, "Effect of timing of hip extension assistance during loaded walking with a soft exosuit," *Journal of neuroengineering and rehabilitation*, vol. 13, no. 1, p. 87, 2016.

[19] Y. Ding *et al.*, "Biomechanical and physiological evaluation of multi-joint assistance with soft exosuits," *IEEE TNSRE*, vol. 25, no. 2, pp. 119–130, 2017.

[20] J. L. Emken and D. J. Reinkensmeyer, "Robot-enhanced motor learning: Accelerating internal model formation during locomotion by transient dynamic amplification," *IEEE TNSRE*, vol. 13, no. 1, pp. 33–39, 2005.

[21] E. H. Van Asseldonk, B. Koopman, and H. Van Der Kooij, "Locomotor adaptation and retention to gradual and sudden dynamic perturbations," *IEEE ICORR*, pp. 376–380, 2011.

[22] I. Cajigas, A. Koenig, G. Severini, M. Smith, and P. Bonato, "Robot-induced perturbations of human walking reveal a selective generation of motor adaptation," *Science Robotics*, vol. 2, no. 6, pp. 1–10, 2017.

[23] D. Zanotto, Y. Akiyama, P. Stegall, and S. K. Agrawal, "Knee Joint Misalignment in Exoskeletons for the Lower Extremities: Effects on User's Gait," *IEEE Trans. on Rob.*, vol. 31, no. 4, pp. 978–987, 2015.

[24] N. T. Ray, B. A. Knarr, and J. S. Higginson, "Walking speed changes in response to novel user-driven treadmill control," *Journal of Biomechanics*, vol. 78, pp. 143–149, 2018.