Propulsion Modulation Methods in People Post-Stroke during Resistive Ankle Exosuit Use

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***Abstract*— Locomotion requires careful coordination across the various joints and muscles of the body, which can be disrupted after neuromotor injuries such as stroke. People post-stroke often have weakness in their paretic, or more impaired, ankle plantarflexors and a corresponding reliance on the hip joint to generate sufficient forward propulsion. The field of robotic rehabilitation has developed wearable systems that provide joint- and task-specific training for survivors of stroke, and in turn, increase use of the ankle muscles. However, capturing ankle use at the plantarflexor level remains a challenge with conventional tools given the unknown relative contributions of the dorsiflexor muscles. Moreover, variability across individuals complicates the interpretation of user response to these robotic interventions. In this work, we used standard biomechanical analysis as well as shear wave tensiometry in five people post-stroke to gain insight into user-specific ankle and hip adaptations in response to three levels of targeted plantarflexion exosuit resistance. We show that at a group and individual-level, evidence suggests a shift in biomechanical strategy from relying on the hip to using the ankle to modulate propulsion, with a subset of participants completely shifting to the ankle by the end of training. This work represents a step towards exploring more individualized methods for characterizing user response during adaptation to wearable robotic training interventions.**

I. Introduction

Walking is the result of careful coordination across the numerous joints and muscles in the body, and is critical for individuals' independence. Unfortunately, neuromotor injuries such as stroke disrupt this coordination, leading to locomotor impairment [1]. In particular, people post-stroke present with hemiparetic and slow gait, largely due to weakened ankle muscles on the more impaired, or paretic limb. The inability to generate sufficient propulsive forces by the paretic calf muscles leads to propulsion asymmetry, which has been used to categorize impairment levels in people post-stroke [2]. Consequently, recent efforts in gait rehabilitation for survivors of stroke aim to restore gait function by increasing propulsive force generation in the paretic limb through increased ankle plantarflexor use [3]. However, as individuals can also modulate propulsion primarily through trailing limb angle (TLA) by relying more on the hip joint [4], there is a need for rehabilitation techniques that specifically target the ankle muscles towards increasing paretic propulsion generation.

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In recent years, numerous wearable, ankle-specific robotic systems have been proposed for facilitating post-stroke gait rehabilitation [5], [6]. These devices operate by increasing specificity to the target joint and allowing for increased repetition of functional exercises, both of which are linked to rehabilitation efficacy [7]. Our group has shown that a soft ankle exosuit that provides paretic plantarflexion assistance during the stance phase of gait and dorsiflexion assistance in the swing phase can improve propulsion symmetry [8]. Others have shown that ankle exoskeletons can improve ankle mechanics by increasing ankle torque and power with active assistance [9], [10]. More recent work has further demonstrated that an ankle exosuit that resists plantarflexion in stance can result in improvements in propulsion symmetry both during and immediately after short (2 min) bouts of training [11]. Resistive paradigms offer the additional benefit of increasing training intensity, which has been shown to improve rehabilitation outcomes [7]. While these results are promising at the group level, user-specific variability remains a challenge for characterizing human-robot interaction to inform individualized rehabilitation.

Specifically, the inherent redundancy of the musculoskeletal system allows for multiple viable responses to robotic systems. The heterogeneity of neuropathology after stroke further increases the variability across individuals' responses as muscle weakness can lead to the use of compensatory strategies [1]. Subgroup analyses have shown that robotic interventions are more effective for subsets of individuals that present with certain baseline characteristics, such as high or low walking speeds [12]. However, categorizations with high-level outcome measures, such as walking speed or clinical scores, do not provide insight to the biomechanical strategies used to modulate gait while wearing exosystems or the device's ability to target a specific impairment. An alternative is to categorize users based on kinematic changes in response to the external forces [13], [14], but ankle kinematics do not fully represent the underlying plantarflexor use.

A major challenge for investigating plantarflexor use comes from the lack of appropriate tools for measurement. Conventional non-invasive tools either capture the electrical input to a single muscle (e.g., electromyography (EMG)) or the net mechanical output from a group of agonist and antagonist muscles (e.g., joint torque), but not the force from a group of agonist muscles, such as the triceps surae [15]. EMG measures the neurological input to a muscle which, while related to force output, can be distorted after a stroke [16]. Moreover, with fatigue, generating the same force requires increasing levels of muscle activation [17], making EMG an unreliable measure of load at the muscle level. On the other hand, net ankle torque, computed from inverse dynamics, provides the net output from both the plantarflexors and dorsiflexors, and requires sophisticated instrumentation for data acquisition. However, as people post-stroke present with higher levels of co-contraction [16], the relationship between net ankle torque and agonist muscle loading are likely to be altered.

Shear wave tensiometry offers one potential solution for capturing plantarflexor-specific insight to a user's biomechanical strategy [18]. The tensiometer is a low-profile wearable device that measures axial stress in superficial tendons, such as the Achilles' tendon, from the speeds of waves traveling through the tissues. As the Achilles' tendon and triceps surae are positioned in-series, the load measured at the tendon is approximately equivalent to the load in the plantarflexors [19]. Prior work has shown that the tensiometer can capture tendon-level changes in able-bodied individuals walking with exosuit assistance at different speeds, and while carrying various loads [20]. Moreover, changes in tendon loading do not sh a 1:1 relationship with traditional "biological torque" measures, which estimate user-generated torque after accounting for the external exosuit-applied torque [21]. Thus, tensiometry may provide a more direct measure of changes at the plantarflexors during exosystem use. However, the feasibility of shear wave tensiometry has yet to be evaluated in post-stroke populations.

This work investigates individual-specific biomechanical strategies to modulate paretic propulsion in response to exosuit-applied plantarflexion resistance. We first evaluate the validity of tensiometry for people post-stroke through comparisons with net ankle torque during unperturbed walking. Then, we assess the changes in plantarflexor loading during and immediately after exposure to exosuit-applied stance-phase plantarflexion resistance. Finally, we study the relative use of ankle and hip-based strategies to modulate propulsion at a group and individual level. Based on prior literature and our prior work, we hypothesized that overall, ankle plantarflexor use would increase in response to plantarflexion resistance, but that individual strategies to modulate propulsion would vary more widely between using the ankle versus the hip.

II. Methods

A. Participants

We recruited five individuals post-stroke (5 male; 3 left paretic; 110 $\pm$ 46 months post-stroke (mean $\pm$ std); age: 56.0 $\pm$ 18.1 years; mass: 88.8 $\pm$ 12.4 kg; height: 1.85 $\pm$ 0.05 m) to participate in this single-session study. Across individuals, comfortable walking speeds ranged from 0.9-1.4 m s^{-1} and fast walking speeds ranged from 1.1-2.1 m s^{-1}. Baseline comfortable and fast walking speeds were calculated from the time to cover the middle 6m of a 10m walkway. All participants had one prior exposure to the resistive ankle exosuit, which occurred at least 14 months prior to this study. Individuals provided medical clearance and written informed consent prior to participation. The study was approved by the Harvard Longwood Medical Area Institutional Review Board, and all methods were conducted in accordance with the approved study protocol.

B. Experimental Protocol

In this study, we used a soft ankle exosuit designed for people post-stroke [22] that was previously modified to apply stance-phase plantarflexion resistance [23] (Fig. 1). Briefly, the exosuit consisted of an actuator located at the waist, with a Bowden cable routed down to the ankle to apply a dorsiflexion moment across the foot and shank segments. Similar to prior work, inertial measurement units (IMUs) (MTi-3, XSens, Enschede, Netherlands) located on both feet were used to identify key gait events and command a desired force profile, while a load cell (LSB200, Futek, Irvine, CA, USA) measured the cable tension to provide feedback to the force controller (see [11], [24] for details).

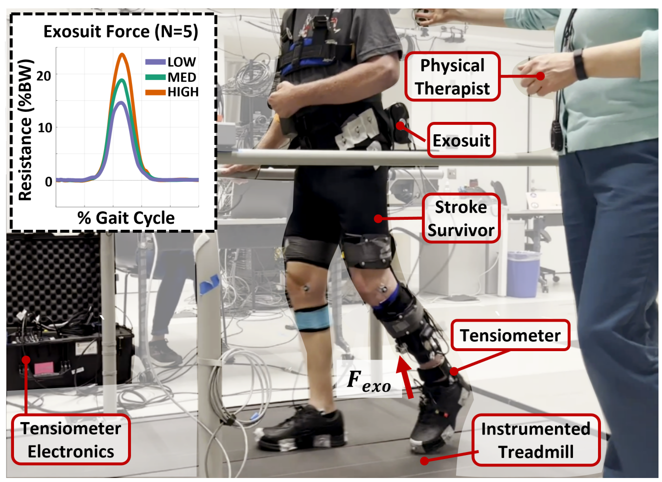

Fig. 1: **Experimental Setup.**

Participants walked on an instrumented split-belt treadmill (Bertec, Columbus, OH, USA) at a self-selected comfortable walking speed (0.7-1.0 m s^{-1}) for three 4-minute bouts. We used a constant speed to control for the known relationship between propulsion and walking speed [4]. The exosuit was inactive for the first and last minute of each bout, and applied no forces as the cable remained slack throughout the entire stride. In the middle two minutes, the exosuit provided active resistance, with a peak force magnitude of 15, 20, or 25 %bodyweight (%BW) depending on the bout, which we refer to as LOW, MED, and HIGH resistance, respectively, hereon. The order of force conditions was randomized. During active resistance, the commanded force was set to begin 30% between the onset of the paretic single support phase (i.e., non-paretic toe-off) and paretic toe-off, which corresponds to the mid-stance phase. The force was commanded to end by paretic toe-off. Peak force timing was set to align with the participant's baseline peak ankle torque timing, determined by an initial 2-minute walking trial without any device. For each participant, we used a fixed force profile for each trial to assess the relationship between resistance magnitude and user biomechanical response. Participants were instructed to "push off hard against the ground and spend time on their [paretic] leg" during the entire 4-minute bout given the known importance of task-specific instructions [25]. Participants rested for at least four minutes following each walking bout.

We collected lower limb motion capture data (Qualisys, Gothenburg, Sweden) at 200Hz and force plate data at 2000Hz from the treadmill. Markers were placed on bony landmarks that define the joint centers of the ankles, knees, and hips, with tracking markers for the feet, shanks, thighs, and pelvis. Two additional markers were placed at the proximal and distal ends of the exosuit cable to measure the exosuit moment arm relative to the ankle joint center. Joint kinetics and kinematics were computed through inverse dynamics using standard biomechanics software (Visual 3D, C-Motion, Germantown, MD, USA). A tensiometer was placed on the distal end of the paretic Achilles' tendon [18] and loading information was sampled at 100Hz. We also collected data from the exosuit sensors at 100Hz via Bluetooth.

C. Evaluation Metrics

We focused on the stance phase of walking for assessment as this corresponds to the period of forward propulsion generation and also reflects when the exosuit provides resistance. Data from strides in which the foot crossed over between treadmill belts were first removed from analysis. We computed paretic propulsion impulse for each stride as the integral of anterior ground reaction force. To assess changes in biomechanical strategies, we computed average tendon loading, average positive biological torque, average positive net torque, and average positive trailing limb angle (TLA) for each stance cycle. Kinetic data was normalized to participant body mass. Wave speed squared, which is proportional to axial stress in soft tissues [18], was used to quantify changes in tendon loading relative to unperturbed baseline walking. We normalized data within each subject and walking bout to the average peak tendon load measured during the initial baseline interval. To obtain biological ankle torque, we took the difference between exosuit-applied torque and net ankle torque computed from inverse dynamics [21]. During active resistance, exosuit-applied torque was always negative, and thus biological torque was always greater than or equal to net ankle torque. Finally, we chose to use average positive TLA during stance to isolate the period during which the limb is contributing to forward propulsion, i.e., in late stance.

We evaluated all metrics during four 30s intervals for each walking bout: the last 30s of the initial slack timepoint (BASE), the first and last 30s during active resistance (EARLY and LATE), and the first 30s after turning off resistance (POST). We used 30s intervals for analysis similar to prior studies that have investigated changes in paretic propulsion in people post-stroke [26], [27]. At each force level, we removed outliers, defined as more than three scaled median absolute deviations, from BASE before obtaining the average: $\overline{BASE}$. Then, for each participant, we removed outlier strides across all timepoints and force levels to obtain the proportional changes relative to $\overline{BASE}$. Finally, we analyzed proportional changes between each stride and $\overline{BASE}$ for each participant and force level. In this work, we considered changes in the magnitudes of these metrics over the different timepoints as representation of adaptation at the respective joints.

D. Statistical Analyses

1) Tensiometer feasibility: We evaluated the validity of shear wave tensiometry in individuals post-stroke through a series of participant-specific correlations between tendon loading and net ankle torque during BASE across all walking bouts. Correlations were categorized as weak ($R < 0.3$; $R^2 < 0.09$), moderate ($0.3 \leq R < 0.7$; $0.09 \leq R^2 < 0.49$), or strong ($R \geq 0.7$; $R^2 \geq 0.49$) [28].

2) Biomechanical response to plantarflexion resistance:

Group Level: We first tested for a main effect of the experimental condition on propulsion impulse during BASE using a linear mixed model with the condition order, force level and condition order $\times$ force level, as fixed effects. A lack of significance ($p \geq 0.064$) for each fixed effect confirmed washout of training effects between walking bouts, so all strides were used in subsequent analyses. We tested for differences in propulsion and biomechanical metrics (net ankle torque, biological ankle torque, tendon loading, TLA) between timepoints (BASE, EARLY, LATE, POST) at each force level using a series of linear mixed models. Differences in biological ankle torque were compared only during BASE, EARLY, and LATE as during POST and BASE, when no exosuit forces are applied, biological and net ankle torque are equivalent. For each biomechanical metric and force level, timepoint was a fixed factor and participant was a random factor. Significant main effects were explored using post-hoc pairwise comparison of estimated marginal means with a Sidak correction.

Participant Level: To further probe the individual-level strategy used to modulate propulsion, we first obtained a set of biomechanical metrics that were significantly correlated with changes in propulsion at each timepoint (EARLY, LATE, and POST) across all force conditions for each participant using a series of linear regressions. If more than one biomechanical metric was significantly correlated with changes in propulsion, we used multiple linear regression with stepwise selection with the identified biomechanical metrics as independent variables and propulsion as the dependent variable to account for collinearities between the metrics. If no biomechanical metrics were significantly correlated with changes in propulsion, we tested for force level-specific correlations for each timepoint and participant.

All statistical analyses were conducted in MATLAB 2022a (Mathworks, Natick, MA, USA) and SPSS 29 (IBM Corp, Chicago, IL, USA) with a significance level of 0.05.

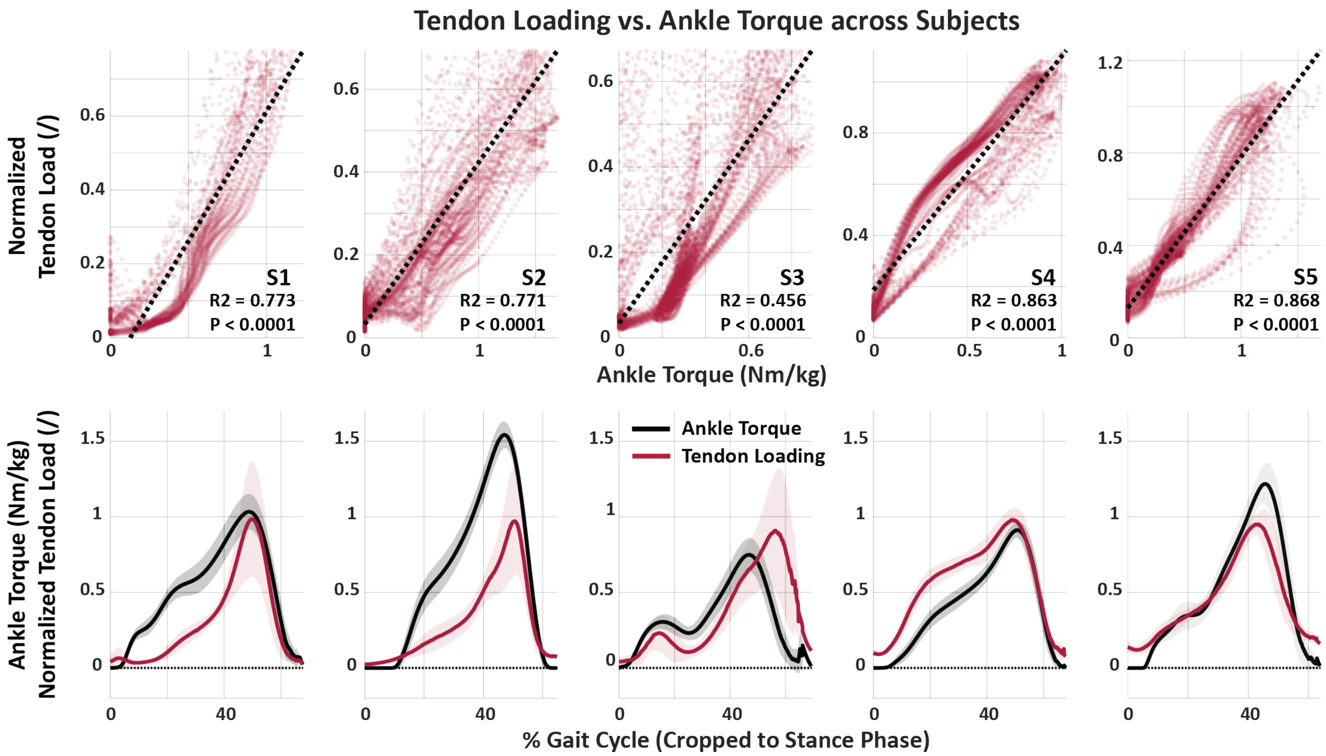

Fig. 2: **Tendon loading versus net ankle torque.** (Top) Individual correlation analyses for subjects S1 through S5. Statistical results represent data from the BASE conditions of all force magnitudes. Data used in the scatter plots are downsampled by a factor of 5 for ease of visualization. (Bottom) Mean timeseries data for tendon loading (maroon) and ankle torque (black) for participants S1 through S5 (from left to right). Shaded regions represent s.e.m.

III. Results

A. Shear wave tensiometry for use in people post-stroke

For all participants, net ankle torque and tendon loading across stance had significant moderate to strong positive correlations ($p < 0.0001$; $R^2 = 0.75 \pm 0.17$) (Fig. 2). Tendon loading across the stance phase also captured salient features of ankle torque such as the bimodal pattern seen in participant S3.

B. Group-level biomechanical response to ankle plantarflexion resistance

Compared to the corresponding BASE timepoint, paretic propulsion impulse was significantly higher during EARLY, LATE, and POST with HIGH resistance ($p \leq 0.030$) and during EARLY with MED resistance ($p = 0.001$) (Fig. 3). No significant differences were found between other timepoints at any force level ($p > 0.078$). On average, with HIGH resistance, propulsion impulse increased from $\overline{BASE}$ by 10.4%, 9.4%, and 6.2% during EARLY, LATE, and POST, respectively. These changes correspond to an absolute increase of 0.10–0.18 %BW s relative to $\overline{BASE}$. With MED resistance, propulsion impulse was, on average, 8.6% (0.17 %BW s) larger than $\overline{BASE}$ during EARLY. At the joint level, there were significant main effects of timepoint on net ankle torque across all force levels ($p < 0.001$). On average, net ankle torque increased in LATE and POST compared to BASE ($p < 0.027$), but was similar between EARLY and BASE ($p > 0.933$). Specifically, during LATE and POST, net ankle torque was greater than $\overline{BASE}$ by 7.1% and 8.3% with LOW, 4.9% and 4.8% with MED, and 6.4% and 5.6% with HIGH resistance, respectively. We found no significant differences between net ankle torque in LATE and POST ($p > 0.979$) at any force level. Similarly, there were significant main effects of timepoint on biological ankle torque across all force levels ($p < 0.001$) such that the biological ankle torque consistently increased from BASE to LATE ($p \leq 0.01$). On average, biological ankle torque in EARLY and LATE were greater than $\overline{BASE}$ by 9.4% and 18.1% with LOW, 12.1% and 18.3% with MED, and 18.4% and 23.5% with HIGH resistance.

Conversely, there were significant main effects of timepoint on tendon loading with MED and LOW resistance ($p \leq 0.001$), but not with HIGH resistance ($p = 0.211$). With MED resistance, tendon loading was greater during LATE and POST compared to BASE ($p \leq 0.002$) but not during EARLY ($p = 0.318$). On average, tendon loading in LATE and POST were 5.1% and 8.0% greater than $\overline{BASE}$, respectively. With LOW resistance, tendon loading only increased relative to BASE during POST ($p = 0.036$). Similar to propulsion and net ankle torque, tendon loading did not differ between LATE and POST with MED or LOW resistance ($p \geq 0.312$).

Finally, we investigated involvement of the hip by assessing changes in TLA. We found significant main effects of timepoint on TLA across force levels ($p < 0.001$). Most pairwise comparisons of TLA across timepoints were also significant ($p \leq 0.001$) (Fig. 3). On average, in LATE and POST, TLA was greater than $\overline{BASE}$ by 19.3% and 11.6% with LOW, 18.0% and 9.5% with MED, and 18.4% and 7.4%

with HIGH resistance, respectively.

C. Individual-level strategies to modulate propulsion with ankle plantarflexion resistance

As hypothesized, participant-specific strategies to modulate paretic propulsion in response to the applied resistance varied across individuals and timepoints (Table 1). Across the 15 individual-level regression models investigating changes during three timepoints (EARLY, LATE, and POST) for each of the five participants, two models identified no biomechanical measures that modulated propulsion ($R^2 \leq 0.046$; $p \geq 0.077$). The variance explained by the remaining 13 regression models ranged from 0.099 to 0.641 (mean = 0.318; $p \leq 0.022$). The majority (9/13) of the models indicated that one biomechanical measure modulated propulsion, with the remaining models using two to three measures. For most single-factor models (mean $R^2 = 0.231$; $p \leq 0.022$), TLA was the measure most associated with propulsion modulation, while the dominant measure for models with multiple predictors was more variable.

At the participant level, the first participant (S1) modulated propulsion with TLA, net ankle torque, and biological ankle torque during EARLY, with tendon load and net ankle torque during LATE, and with only net ankle torque during POST. Another participant (S2) started by modulating propulsion with TLA during EARLY, then by varying TLA and tendon load during LATE, and finally by only varying tendon load during POST. S3 presented a different trend and started by modulating propulsion with biological ankle torque during EARLY and ended with TLA during POST. None of the explored biomechanical metrics were related to changes in propulsion during LATE across all force levels ($p \geq 0.077$). Like S1 and S2, S4 started with a gait pattern that best correlated changes in TLA with changes in propulsion during EARLY. However, in POST, both TLA and net ankle torque modulated with propulsion. No explored measures were related to changes in propulsion during LATE for this participant ($p \geq 0.088$). Finally, S5 consistently modulated propulsion by varying TLA across all timepoints (mean $R^2 = 0.194$, $p \leq 0.022$).

IV. Discussion and Conclusion

In this work, we explored user-specific ankle and hip adaptations in response to short periods of targeted plantarflexion exosuit resistance in people post-stroke using standard biomechanical analysis and shear wave tensiometry. We showed that tendon loading and ankle torque were at least moderately correlated during baseline walking without exosuit resistance, demonstrating the feasibility of using shear wave tensiometry to estimate plantarflexor loading in individuals post-stroke, despite their altered tendon properties [29]. The moderate to strong correlations between tendon loading and ankle torque suggest that the triceps surae are not the sole contributors to net ankle torque in stance. This outcome may be due to higher dorsiflexor activity and co-contraction in this population [16] or contributions from peripheral stabilizing muscles that do not load the Achilles' tendon, such as the

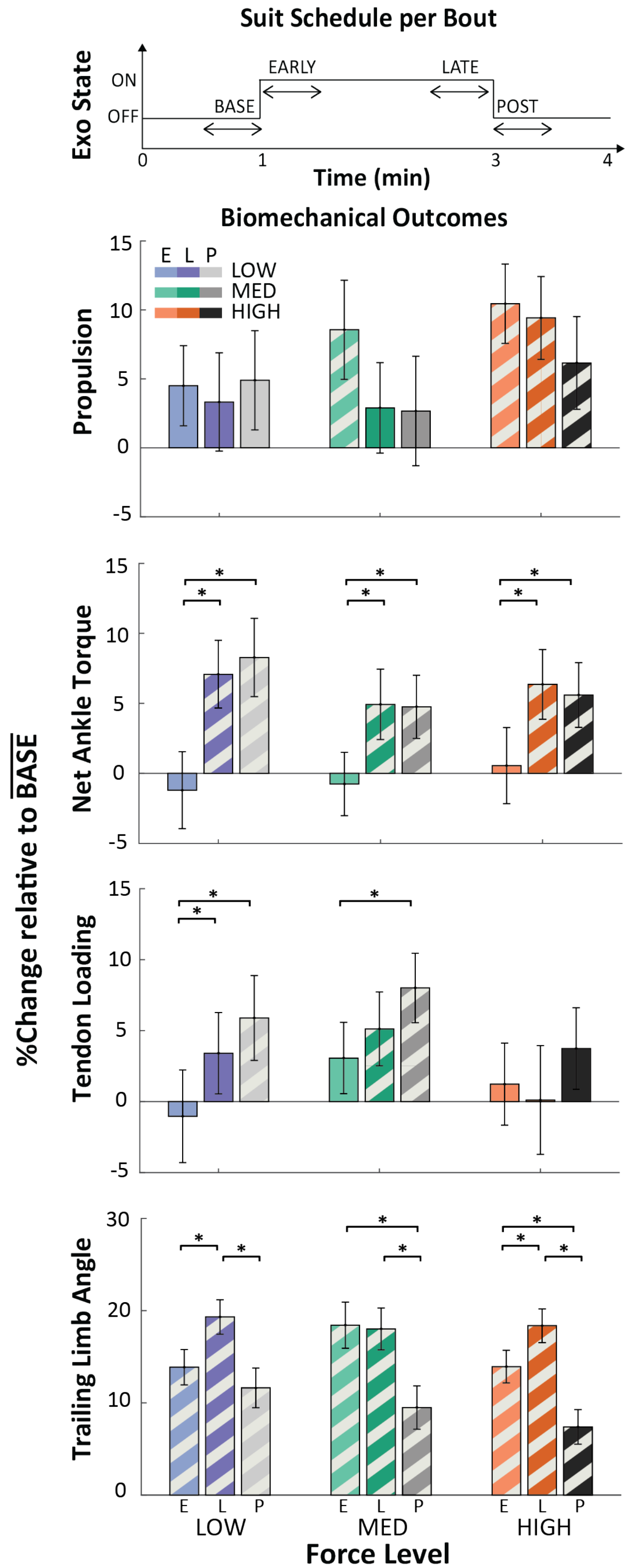

Fig. 3: Bar plots representing the average change relative to $\overline{BASE}$ for the dependent and independent biomechanical measures in this study. Changes during timepoints EARLY (E), LATE (L), and POST (P) are shown. Hatched bars indicate significant changes between each timepoint and BASE, and asterisks indicate significant differences between later timepoints. Error bars represent the 95% confidence interval. Across participants, propulsion increases relative to BASE using a combination of ankle and hip-based strategies.

				Biomechanical Measure				Final Model	
Subj ID	CWS (m/s)	Pp	Timepoint	Net Ankle Tor	Bio Ank Tor	Tendon Load	TLA	R^2	p-value
S1	1.19	0.5	EARLY	✔	✔	✓	✔	0.641	<0.001
			LATE	✔	✓	✔		0.509	<0.001
			POST	✔	NA	✓		0.292	0.004
S2	1.38	0.4	EARLY				✔	0.265	<0.001
			LATE			✔	✔	0.522	<0.001
			POST		NA	✔		0.229	<0.001
S3	1.18	0.4	EARLY	✓	✔		✓	0.104	0.004
			LATE					-	-
			POST		NA		✔	0.158	0.001
S4	0.91	0.3	EARLY	✓	✓		✔	0.452	<0.001
			LATE					-	-
			POST	✔	NA		✔	0.374	<0.001
S5	1.10	0.2	EARLY				✔	0.099	0.022
			LATE	✓	✓	✓	✔	0.280	<0.001
			POST		NA		✔	0.204	0.002

Table 1. Participant-specific changes in biomechanical measures relative to $\overline{BASE}$ that are significantly correlated to changes in propulsion impulse at different timepoints shown by dark red checkmarks. Gray checkmarks indicate biomechanical measures that were significantly correlated to propulsion independently but were not included in the stepwise selection due to redundancy with other measures. Biological ankle torque was not explored in POST (noted as NA), as no torque was applied by the exosuit. Participant self-selected overground comfortable walking speed (CWS) as measured by a 10m walk test and paretic propulsion percentage (Pp) [2], where 0.5 represents perfect symmetry, during baseline walking are also provided.

peroneus longus [30], [31], [32], both of which may decouple net ankle torque from the plantarflexor load. In analyzing group-level response to exosuit resistance, we found that propulsion impulse increases relative to baseline with exposure to MED and HIGH resistive forces. These changes were associated with increases in use of the targeted paretic ankle, measured by ankle torque and tendon loading, as well as use of the hip, measured by TLA. At the individual-level, we found that while 4/5 participants modulated propulsion with TLA immediately after resistance was initiated, three of these participants transitioned to using ankle-specific metrics, i.e., tendon loading and net ankle torque, over time. Thus, with functional exosuit-based resistance training, we observed a shift in user response from leveraging the hip joint towards engaging the ankle joint to generate paretic propulsion.

We found that adaptations at the ankle and hip occurred at different timescales by evaluating changes in gait biomechanics at multiple timepoints relative to the onset and offset of resistive force. Starting with the first 30s of resistance exposure, participants significantly increased propulsion impulse at the higher two force levels and sustained this increase through the first 30 seconds after exposure ended. These findings are consistent with our previous work in which propulsion changes at the end of exposure to exosuit-applied resistance were correlated with propulsion changes immediately after removing resistance [11], suggesting carryover effects. Participants also increased their TLA shortly upon onset of active resistance, but TLA differed between the end of active resistance and post-resistance, indicating that these changes were not sustained as strongly upon removal of the perturbation. Conversely, tendon loading and ankle torque only increased from baseline at the end of resistance exposure, but retained this increase in ankle mechanics during post-resistance, similar to propulsion. The faster change in TLA may reflect a reactionary response to plantarflexion resistance as individuals post-stroke have more control over proximal joints [4] and tend to respond to perturbations of balance through hip-based strategies [33]. On the other hand, the relatively longer timescale of ankle mechanics suggests that the changes were not a mechanical response to the external torque, but rather a fundamental alteration at the neuromotor level to the applied resistance. Moreover, increases in ankle torque were present even after LOW resistive forces, despite insignificant changes in propulsion. This outcome further supports a shift in the underlying gait strategy for generating propulsion following exposure to exosuit-applied resistance, even with similar propulsion magnitudes. Together, our group-level analyses indicate that functional plantarflexion resistance training has the potential to promote learning and retention of gait patterns that emphasize use of the ankle towards propulsion generation.

However, further investigation is needed to identify which individuals benefit most from ankle-targeted exosuit resistance training. While our previous work found no evidence of sustained changes in TLA at the group level [11], here, we found a strong reliance on the hip to modulate propulsion both during and immediately after exposure to resistance. One potential explanation may be from a difference in the central drive of this study's participants. Central drive refers to the capacity of an individual to voluntarily recruit the plantarflexors and has been shown to contribute to propulsion deficits in people post-stroke [34]. Qualitatively, we also found that our more impaired participants, those with slower self-selected walking speeds or with lower propulsion symmetries, tended to rely more on the hip to modulate propulsion even after resistance was removed. This is consistent with prior work suggesting that individuals shift to using proximal muscles for high intensity tasks [35], [36] and our past work showing increased involvement of proximal joints at higher resistance magnitudes [23]. Thus, perhaps tuning the resistance parameters to train at lower intensities would provide these individuals with the bandwidth to better modulate their gait strategies. Another explanation may be that individuals interpreted the instructions to "push off against the ground" differently. Future work may investigate the use of tools like biofeedback to provide more specific and quantitative feedback [37]. This variability across participant response also supports the need for tools that measure ankle use across environments and controller strategies such as the tensiometer used in this study. Through improved understanding of how gait biomechanics are altered in response to different robotic interventions, we can better design systems for addressing patient-specific characteristics.

Still, there are a number of limitations to be noted in this work, and opportunities for future research. First, this work was conducted with a small sample size in a heterogeneous population. Acknowledging this limitation, we conducted both group-level and participant-level analyses to understand the biomechanical adaptations associated with exposure to plantarflexion resistance in people post-stroke. In particular, we chose to explore the relationship between propulsion and biomechanical measures at the hip and ankle based on prior literature. However, two participants did not exhibit any significant correlations between the measures explored and propulsion during one timepoint, suggesting the presence of other mechanisms for modulating propulsion during resistance training that were not captured in our work. Thus, while our study provided valuable insights to the potential responses to exosuit resistance, further work with larger sample sizes is necessary to understand the relative prevalence of the strategies identified here. We also found evidence of a shift towards using the ankle to modulate propulsion, but did not characterize the timescale of this shift. Prior work in neuromotor learning has shown that intermittent exposure can accelerate the timescale of adaptation [38], and thus optimizing the scheduling of resistance may enable a faster shift to the ankle joint. Similarly, another possibility is that the individuals who only used TLA to modulate propulsion needed longer exposures to the resistive exosuit to learn how to use their ankle. We also found that the effect of condition order on baseline propulsion trended towards significance ($p < 0.10$), suggesting that the effects of resistance training may persist longer than evaluated in this work. A wearable, portable solution like the tensiometer may enable future work to use tendon load as a proxy for plantarflexor-level changes to track user response to ankle resistance over longer durations in overground or community-based environments [39]. Finally, although we interpret the use of TLA to modulate propulsion as an unintended response to the targeted resistance paradigm used in this work, we acknowledge that individuals likely need a combination of increased TLA and ankle torque to maximize propulsion generation [40].

In conclusion, this work explored user-robot interaction during targeted exosuit plantarflexion resistance in people post-stroke to increase paretic propulsion. Our findings show that the biomechanical strategy changes across the duration of resistance training. We further show that tensiometry can track plantarflexor loading during stance and is associated with changes in propulsion for a subset of users. This work represents an initial step towards exploring more portable methods for characterizing user response during adaptation to wearable robotic training interventions.

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