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A Propulsion Neuroprosthesis Improves Overground Walking in Community-Dwelling Individuals After Stroke

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ABSTRACT Functional electrical stimulation (FES) is a common neuromotor intervention whereby electrically evoked dorsiflexor muscle contractions assist foot clearance during walking. Plantarflexor neurostimulation has recently emerged to assist and retrain gait propulsion; however, safe and effective coordination of dorsiflexor and plantarflexor neurostimulation during overground walking has been elusive, restricting propulsion neuroprostheses to harnessed treadmill walking. We present an overground propulsion neuroprosthesis that adaptively coordinates, on a step-by-step basis, neurostimulation to the dorsiflexors and plantarflexors. In 10 individuals post-stroke, we evaluate the immediate effects of plantarflexor neurostimulation delivered with different onset timings, and retention to unassisted walking (NCT06459401). Preferred onset timing differed across individuals. Individualized tuning resulted in a significant 10% increase in paretic propulsion peak (Δ: 1.41 $\pm$ 1.52%BW) and an 8% increase in paretic plantarflexor power (Δ: 0.27 $\pm$ 0.23 W/kg), compared to unassisted walking. Post-session unassisted walking speed, paretic propulsion peak, and propulsion symmetry all significantly improved by 9% (0.14 $\pm$ 0.09 m/s), 28% (2.24 $\pm$ 3.00%BW), and 12% (4.5 $\pm$ 6.0%), respectively, compared to pre-session measurements. Here we show that an overground propulsion neuroprosthesis can improve overground walking speed and propulsion symmetry in the chronic phase of stroke recovery. Future studies should include a control group to examine the efficacy of gait training augmented by the propulsion neuroprosthesis compared to gait training alone.

INDEX TERMS Neuroprosthesis, functional electrical stimulation, propulsion, chronic stroke, overground walking.

IMPACT STATEMENT A wearable propulsion neuroprosthesis augments and retrains overground walking speed and propulsion in people with chronic post-stroke hemiparesis.

I. INTRODUCTION

Stroke-induced brain injury results in neuromuscular impairments that contribute to slow and effortful walking [1], [2], [3]. Impaired plantarflexor force generation during the stance phase of the gait cycle reduces the ability to propel the body forward, limiting overall walking speed [4], [5] and contributing to altered gait kinematics [5] that are associated with a higher energetic cost of walking [2], [3]. In addition, dorsiflexor impairments during the swing phase of the gait cycle reduce foot clearance, increase the risk of falls, and similarly lead to energetically inefficient gait compensations [1], [2]. Ankle-foot orthoses (AFOs) are commonly prescribed in the early stages of stroke rehabilitation to address dorsiflexor impairments and reduce the risk of falls [6], but the rigidity of

AFOs required to brace the ankle during swing phase can lead to muscle atrophy and weakness from learned disuse of the dorsiflexor and plantarflexor muscles [7]. That is, by restricting ankle movement, AFOs improve gait function through biomechanical compensation rather than through recovery of more natural gait biomechanics [8], [9]. Most importantly, AFOs do not address plantarflexor impairments and therefore overlook propulsion deficits altogether.

Exoskeleton [10], [11], [12] and exosuit [13], [14], [15], [16], [17] devices have recently been developed to target propulsion deficits. These wearable devices provide external mechanical assistance in parallel with the dorsiflexor and plantarflexor muscles, showing increased interlimb propulsion symmetry [14], [16] and paretic plantarflexor power [12], [16], [18] while also reducing the energetic cost of walking [14], [17], [19]. The immediate assistive benefits of the external mechanical assistance provided by such robotic devices is well-documented [20]; however, gait-assistive interventions that directly activate neuromotor pathways have potential to enhance the volitional capabilities of paretic muscles and may thus be more effective at enhancing functional recovery after neurological injuries, such as stroke [4], [21], [22].

In contrast to providing mechanical assistance in parallel with the underlying biology, functional electrical stimulation (FES) neuroprostheses augment or replace the force-generating ability of paretic muscles via electrically-evoked muscle contractions. FES neuroprostheses provide gait assistance by delivering electrical currents to peripheral nerves, which in turn, induces muscle contractions [23]. The rehabilitative potential of FES neuroprostheses is evident in the enhancement of muscle recruitment [24] and strength [25], [26] observed after neuroprosthesis-assisted walking.

Dropfoot FES stimulators are the most common type of FES neuroprosthesis. Designed to augment or replace the function of the paretic dorsiflexor muscles during walking, they are commonly prescribed as an alternative to AFOs [6]. Like AFOs, dropfoot stimulators reduce fall risk by addressing dorsiflexor impairments [27], [28], [29], but in contrast to AFOs, they do not restrict movement at the ankle, allowing for a more natural gait. However, similar to AFOs, dropfoot stimulators overlook the plantarflexor impairments that underlie deficits in propulsion.

A new class of FES neuroprosthesis has recently emerged to target the enhancement of plantarflexor function, in addition to dorsiflexor function. Propulsion neuroprostheses have demonstrated improved paretic propulsion peak and integral and swing phase foot clearance during fast treadmill walking [30], [31], [32], [33], [34]. Remarkably, a single session of treadmill walking augmented by propulsion FES improves paretic propulsion peak and integral [31], [35], walking speed [36], and interhemispheric symmetry of corticospinal input to the plantarflexor muscles [37]. The therapeutic potential of propulsion FES is evident in this ability to induce corticomotor plasticity that is associated with improved gait propulsion and clinically meaningful functional outcomes. Indeed, three-months of treadmill-based gait training augmented by propulsion FES resulted in durable changes in propulsion and walking speed that persisted at least three months after the training period ended [38].

Although promising, technological limitations have constrained propulsion FES to harnessed treadmill walking. Given the biomechanical differences between treadmill and overground walking [39], [40], [41], the immediate effects of propulsion FES on overground walking ability, and the therapeutic benefits of overground gait training with propulsion FES are unknown. The development of a fully wearable propulsion neuroprosthesis is essential to study the immediate effects of overground walking with FES-assisted propulsion following neurological injury, and the therapeutic retention of these effects after the FES assistance is removed. Given emerging efforts to extend rehabilitation paradigms from the clinic into the home and community [42], delivering safe and effective propulsion FES during overground walking, and understanding the biomechanical effects of this intervention, becomes critical.

The safe and effective delivery of propulsion FES during overground walking requires adaptively coordinating the plantarflexor neurostimulation required for propulsion enhancement with the dorsiflexor neurostimulation required for foot clearance. From a safety perspective, prior treadmill studies report that paretic plantarflexor neurostimulation during stance phase reduces the swing phase foot clearance needed for safe walking [34]. Imprecise control of the offset of stance-phase plantarflexor neurostimulation may result in continued activation of the plantarflexor muscles during the early swing phase when the ankle should be dorsiflexing to lift the foot. Similarly, imprecise control of the onset of swing-phase dorsiflexor neurostimulation may result in the plantarflexed ankle failing to dorsiflex during the stance-to-swing transition. In addition to the safe delivery of propulsion FES, imprecise control of the onset of stance-phase plantarflexor neurostimulation has the potential to disrupt body progression within the gait cycle and worsen propulsion symmetry. Indeed, our prior work with soft robotic exosuits [14] demonstrated the importance of individualizing the onset timing of exosuit-delivered plantarflexor assistance to maximize the improvement in propulsion symmetry across individuals and avoid impairing propulsion in some individuals. To best assist gait propulsion, an overground propulsion neuroprosthesis will likely similarly require individualized plantarflexor neurostimulation.

We present a fully wearable, unilateral, propulsion FES neuroprosthesis that coordinates the amplitude and timing of neuromotor stimulation to the paretic dorsiflexor and plantarflexor muscles during overground walking (see Fig. 1 and **Supplementary Materials**). Our first objective was to evaluate the immediate effects of overground propulsion FES on post-stroke walking speed, propulsion (i.e., paretic peak propulsion and interlimb propulsion symmetry), and foot clearance. We hypothesized that compared to unassisted walking, propulsion FES would increase overground walking speed and propulsion without hindering the foot clearance necessary for safe walking. We further hypothesized

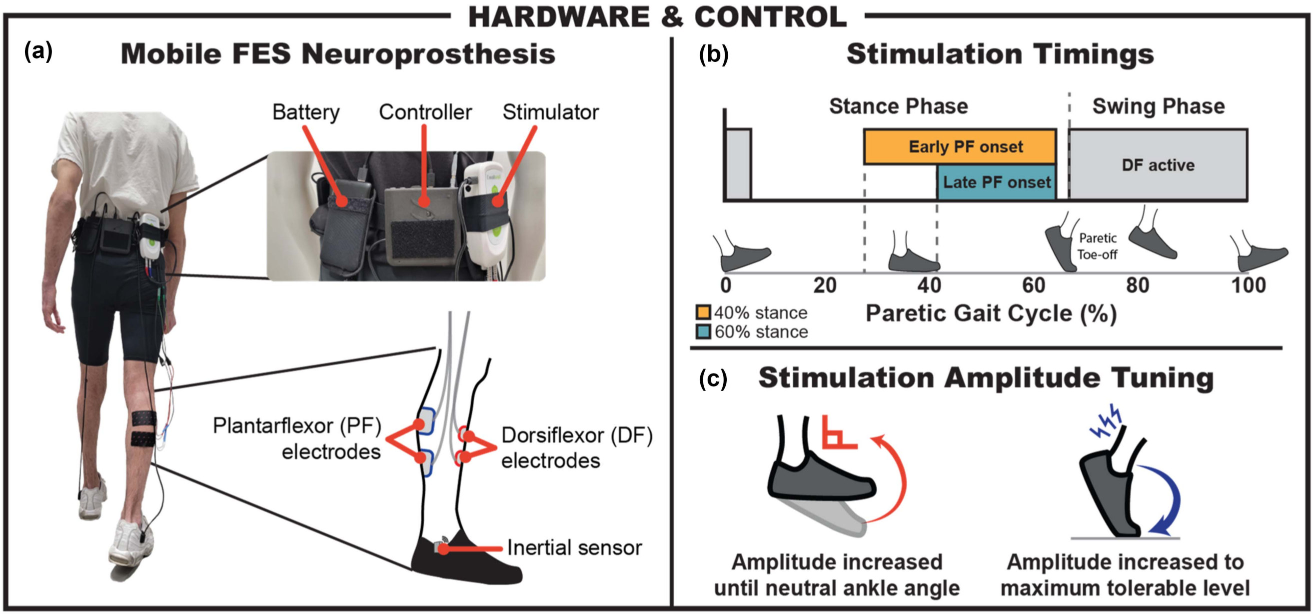

FIGURE 1. **Overview of the propulsion FES neuroprosthesis hardware and neurostimulation control. (a) The mobile neuroprosthesis consists of a waistbelt with the controller, battery, and stimulator and surface adhesive electrodes applied to the plantarflexor and dorsiflexor muscles. Inertial measurement units (IMU) are used to adaptively control, on a per-stride basis, the timing of applied neurostimulation. (b) Two onset timings of plantarflexor neurostimulation were evaluated in this study: Early stance (40% of stance phase) and late stance (60% of stance phase). All other neurostimulation timings were held constant as shown. (c) The amplitude of neurostimulation was tuned at the beginning of the session on an individual basis, with dorsiflexor neurostimulation set to a level that produced a neutral ankle angle, and plantarflexor neurostimulation set to the individual's maximum tolerable level.**

that individually-tuned plantarflexion neurostimulation onset timings would produce substantially larger propulsion improvements compared to generically applying an early or late plantarflexor assistance onset timing across individual subjects. Our second objective was to assess the rehabilitative potential of walking with propulsion FES. We hypothesized that compared to unassisted walking prior to the start of a walking session, overground walking speed, paretic propulsion peak, and propulsion symmetry would be improved after removal of the FES at the end of the session. From this study, we aim to better understand the immediate and rehabilitative effects of overground walking with a propulsion neuroprosthesis. This work will advance wearable neurostimulation technologies with potential to address intractable gait propulsion deficits that contribute to reduced walking function and community participation after stroke.

II. RESULTS

A. NEUROPROSTHESIS OVERVIEW

A fully wearable propulsion FES neuroprosthesis (Fig. 1) was developed to enable the individualization of the amplitude and timing of stance-phase neurostimulation delivered to the paretic plantarflexor muscles and swing-phase neurostimulation delivered to the paretic dorsiflexor muscles. The neuroprosthesis controller adaptively delivered neurostimulation as a function of specific gait subphases (see **Materials and Methods**). The effects of two different onset timings of plantarflexor neurostimulation were studied: an early onset timing set before paretic midstance (i.e., at 40% of stance phase) and a late onset timing set after paretic midstance (i.e., at 60% of stance phase). Relative to ground reaction forces, the *delivered* early onset timing was at an average 46.9 ± 18.9% of paretic stance phase and the late onset timing was at an average 63.7 ± 2.0% of paretic stance phase (**Table S1**). Other neurostimulation parameters, including the onset and offset timings of dorsiflexor neurostimulation and the amplitudes of both plantarflexor and dorsiflexor neurostimulation, were individually tuned for each study participant at the start of the session and held constant throughout the session, enabling this investigation to focus on the differential effects of the two plantarflexor neurostimulation onset timings (see **Materials and Methods**). Of note, the offset timing of plantarflexor neurostimulation was not tuned but held constant at 85% of paretic stance phase based on preliminary work showing a negative effect on swing-phase dorsiflexion when it was set to terminate closer to the paretic toe off gait event (**Fig. S1**).

B. STUDY PARTICIPANT CHARACTERISTICS

Study participants were adequately representative of community-dwelling individuals post-stroke. Their self-selected comfortable walking speeds were consistent with community-level ambulation [43], ranging from 0.50 to 1.36 m/s, and with an average of 0.93 ± 0.23 m/s. Baseline paretic propulsion peak similarly ranged from very low to near-normal, with peak anterior ground reaction force measurements ranging from 0.57 to 17.29% bodyweight (%BW), and with an average 8.50 ± 4.60%BW. Similarly, interlimb propulsion symmetry—where 50% indicates perfect symmetry—ranged from 2 to 42%, averaging 28.4 ± 12.2%. Though the cohort's propulsion and speed impairments were heterogeneous, study participants had overall mild dorsiflexor impairments, with an average peak dorsiflexion angle of 2.07 ± 5.04 degrees during the paretic swing phase. Individual study participant characteristics are reported in **Table S2.**

C. IMMEDIATE ASSISTIVE EFFECTS OF PROPULSION FES

When assessed at the group-level, the early and late plantarflexor neurostimulation onset timings did not produce significantly different changes in peak paretic propulsion (Δ: 0.37 ± 0.67%BW, $p = 0.579$). However, at the individual subject-level, the onset timing that was more effective in improving peak paretic propulsion (i.e., the preferred timing) differed across study participants. Five of the ten study participants benefited more from an early onset timing, whereas the other five benefited more from a late onset timing (Fig. 2(a)).

The preferred neurostimulation onset timing resulted in moderate-to-large differences in the FES-induced improvement in peak paretic propulsion, compared to either the early (Δ: 1.04 ± 0.44%BW, $p = 0.041$, ES = 0.752) or late (Δ: 0.67 ± 0.26%BW, $p = 0.030$, ES = 0.816) onset timings. Moreover, when compared to the non-preferred timing, the preferred timing resulted in large differences in the FES-induced improvement in peak paretic propulsion (Δ: 1.77 ± 1.09%BW, $p = 0.001$, ES = 1.623) and propulsion symmetry (Δ: 3.3 ± 3.6%, $p = 0.017$, ES = 0.923), respectively (Fig. 2(b)). When walking with their preferred plantarflexor neurostimulation timing, study participants demonstrated a median 10% increase in peak paretic propulsion (Δ: 1.41 ± 1.52%BW, $p = 0.017$, ES = 0.928) and an 8% increase in paretic plantarflexor power (Δ: 0.27 ± 0.23 W/kg, $p = 0.004$, ES = 1.199) compared to unassisted walking. In contrast, walking with their non-preferred timing did not significantly alter peak paretic propulsion or plantarflexor power ($p>0.05$) and furthermore resulted in a median 6% worsening of propulsion symmetry (Δ: −2.1 ± 2.5%, $p = 0.024$, ES = 0.859), compared to unassisted walking.

Of the ten study participants, seven did not have baseline swing-phase paretic dorsiflexion impairment. These participants presented with an average peak dorsiflexion angle of 4.50 ± 3.56 degrees during unassisted walking. For these individuals, as hypothesized, propulsion FES did not hinder swing-phase dorsiflexion for either of the two plantarflexor neurostimulation onset timings (preferred: 5.14 ± 5.25 degrees; non-preferred: 5.15 ± 5.52 degrees). In contrast, the three study participants who had baseline swing-phase paretic dorsiflexion impairment presented with an average peak dorsiflexion angle of -3.61 ± 2.70 degrees (i.e., 3.61 degrees of plantarflexion) during unassisted walking. As hypothesized, this dorsiflexion impairment was improved with both timings (preferred: −1.03 ± 3.40 degrees; non-preferred: 0.13 ± 4.22 degrees).

D. POST-SESSION REHABILITATIVE EFFECTS OF PROPULSION FES

Pre-to-post session evaluations of *unassisted* walking revealed a median 9% increase in fast walking speed (Δ: 0.14 ± 0.09 m/s, $p = 0.001$, ES = 1.508) that was accompanied by a 28% increase in peak paretic propulsion (Δ: 2.24 ± 3.00%BW, $p = 0.043$, ES = 0.746) and a 12% increase in propulsion symmetry (Δ: 4.5 ± 6.0%, $p = 0.041$, ES = 0.752) after the FES was removed (Fig. 3). Similarly, we observed a median 14% increase in comfortable walking speed (Δ: 0.12 ± 0.12 m/s, $p = 0.013$, ES = 0.983) that was accompanied by a 13% increase in peak paretic propulsion (Δ: 1.61 ± 1.64%BW, $p = 0.012$, ES = 0.985) and a 10% increase in propulsion symmetry (Δ: 4.8 ± 6.1%, $p = 0.036$, ES = 0.779). Post-session reductions in paretic peak dorsiflexion angle were observed during both comfortable-speed walking (Δ: −1.80 ± 2.55 degrees, $p = 0.052$, ES = 0.706) and fast-speed walking (Δ: −2.19 ± 2.28 degrees, $p = 0.014$, ES = 0.962); however, only the change during fast-speed walking reached statistical significance.

III. DISCUSSION

We present a propulsion neuroprosthesis that safely and effectively coordinates plantarflexor and dorsiflexor neurostimulation during *overground* walking to produce meaningful immediate improvements in peak paretic propulsion and walking speed that carry-over to unassisted walking after removal of the neuroprosthesis. This study builds on previously studied, *treadmill-based* propulsion interventions [24], which have demonstrated comparable retained improvements in unassisted overground walking speed and peak paretic propulsion measured during treadmill walking. An important distinction is that our study demonstrates that delivering *overground* propulsion FES intervention produces concurrent improvements in *overground* walking speed and *overground* peak paretic propulsion. By delivering the intervention overground and measuring outcomes overground, the findings of this study may be more representative of the benefits that can be observed with real-world community walking enhanced by the propulsion neuroprosthesis. Advancing propulsion-targeting interventions currently restricted to harnessed treadmill walking [32], [33], [34], [35], [36], [37] to overground walking can open new opportunities for community-based rehabilitation paradigms that leverage the unique neuromotor intervention enabled by propulsion FES.

The heterogeneity of post-stroke gait deficits necessitates individualized interventions. For unimpaired individuals, the coordinated activation of the plantarflexor muscles with the numerous other muscles responsible for stable, efficient, and fast locomotion is well-documented, and the importance of well-timed plantarflexor neurostimulation during walking on the treadmill has been previously established [44]. However, the effect of modulating the timing of plantarflexor neurostimulation during overground walking post-stroke has yet to be

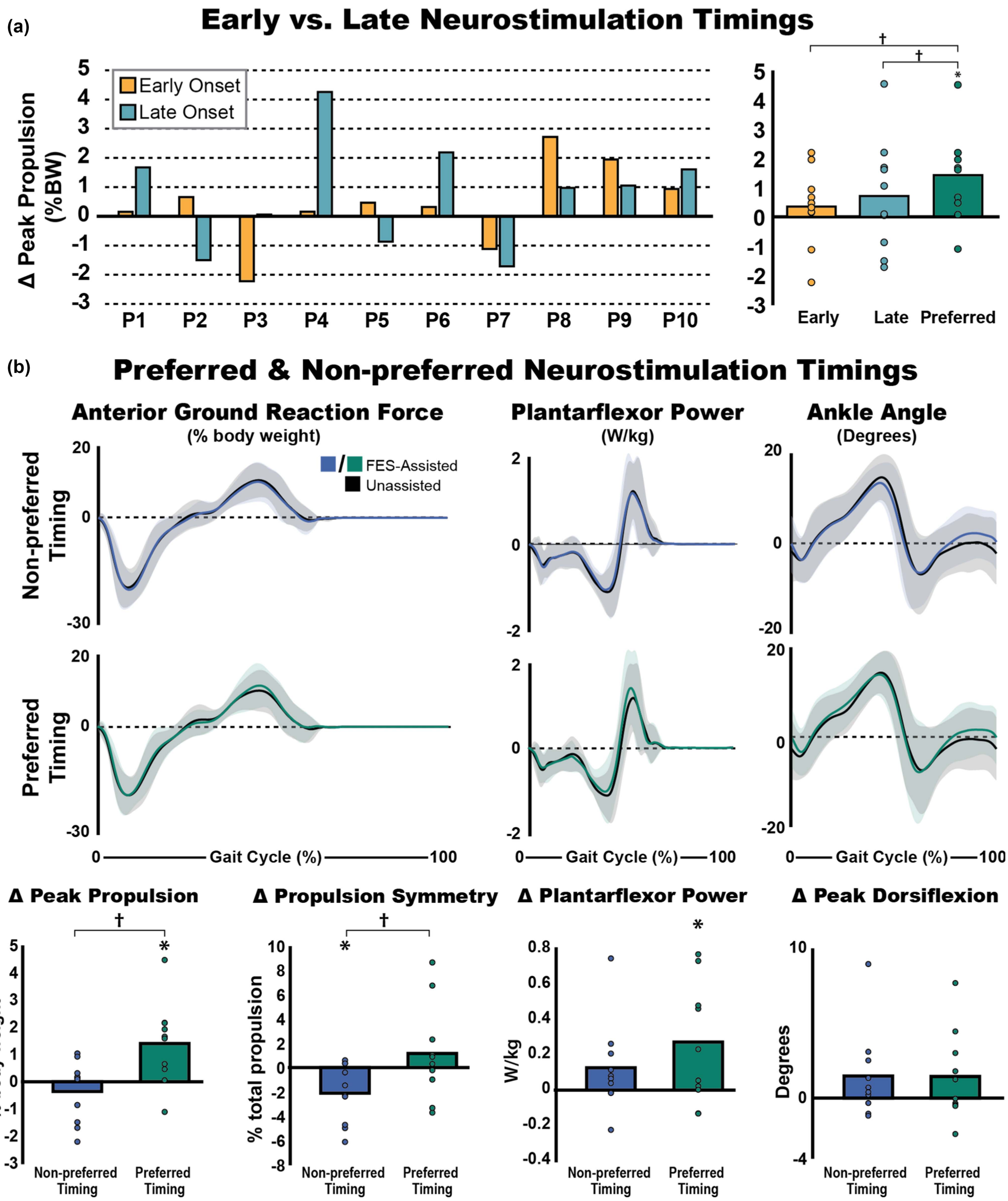

FIGURE 2. **Immediate changes in propulsion biomechanics while using propulsion FES. (a) Preferred onset timing was selected as the plantarflexor neurostimulation onset timing that produced the greater change in paretic propulsion peak from unassisted to assisted walking. (b) Aggregate stride data across all study participants demonstrates the difference between unassisted to assisted walking for each plantarflexor neurostimulation onset timing. Bar plots show group-level changes with overlaid scatterplot of individual subject differences, computed as the difference between assisted and unassisted walking, for each plantarflexor neurostimulation onset timing. Data are reported as mean ± standard error. * Indicates a significant change from unassisted to assisted walking ($p < 0.05$). † Indicates a significant difference between the changes produced by the timing conditions.**

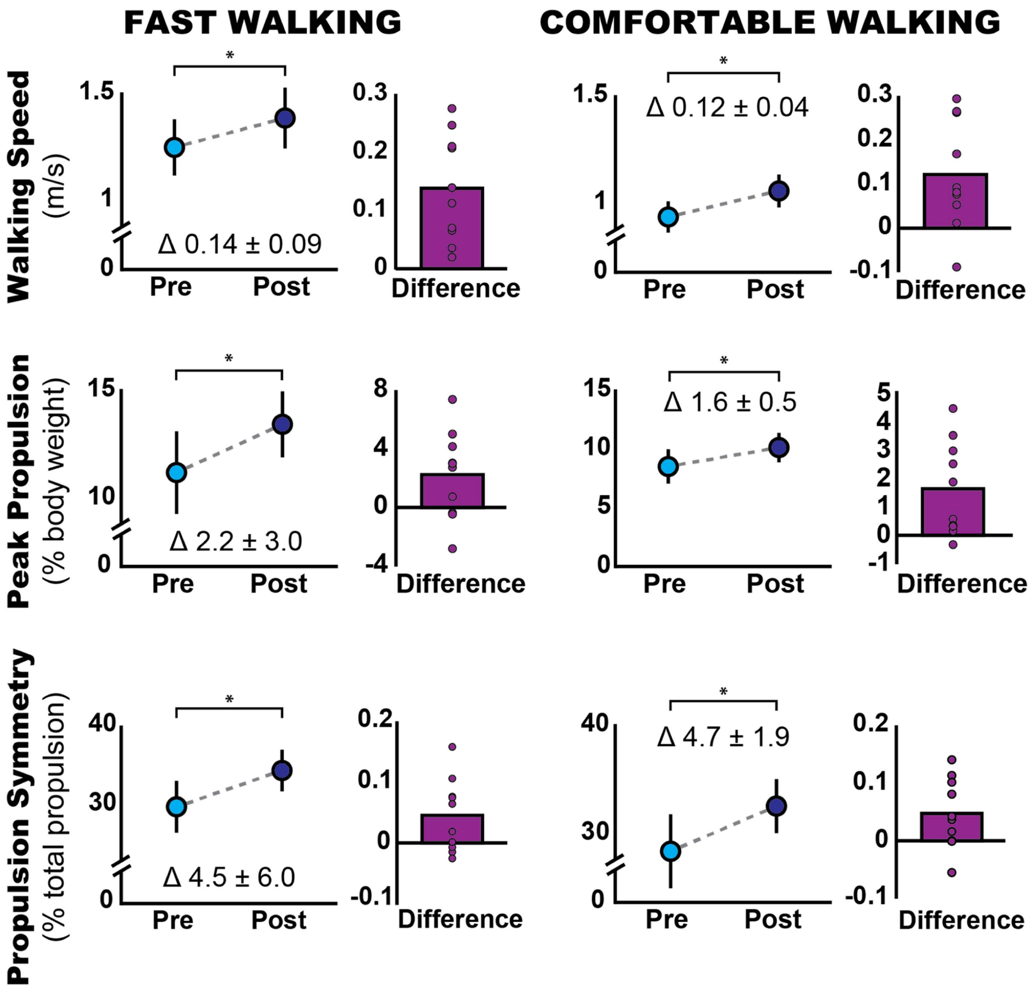

FIGURE 3. Post-session effects of a single session of walking with the propulsion neuroprosthesis. Changes in unassisted walking speed, paretic propulsion peak, and propulsion symmetry after a single session of propulsion FES at fast and comfortable walking speeds. Data are reported as means with error bars as standard error. Percent changes are reported as medians. * Indicates a significant change from pre-session to post-session unassisted walking ($p < 0.05$).

studied, partly due to an absence of propulsion FES technology capable of providing the precision of neurostimulation timing necessary for the greater step-to-step variability of overground walking. Kesar et al. [34] assessed different plantarflexor neurostimulation timings using a foot-switch based control system; however, the FES system was limited to the treadmill and the study only reported on the timing that resulted in the largest improvement, rather than presenting the differential effect between the timings, which our study specifically explored. Interestingly, they found an 18% improvement in paretic propulsion peak, which was larger than our median 10% increase, possibly due to biomechanical advantages afforded by the treadmill, such as better positioning of the trailing limb for generating propulsion.

A. PROPULSION NEUROSTIMULATION ADDRESSES WALKING QUALITY DEFICITS PRESENT IN FAST WALKING INDIVIDUALS

Despite the fast average walking speed of this study population, deficits in interlimb propulsion symmetry were still moderate to severe (i.e., < 36% propulsion symmetry) [43]. It is notable that despite their high baseline walking speed, the FES neuroprosthesis was able to improve their walking quality by increasing peak paretic propulsion and interlimb propulsion symmetry. Given the relationship between paretic propulsion and other aspects of gait quality, such as stability and energy efficiency, this finding is especially important. With a more heterogeneous sample spanning slower walking speeds, the rehabilitative benefit from the neuroprosthesis may vary, warranting further study.

B. DIFFERENTIAL EFFECTS OF PLANTARFLEXOR NEUROSTIMULATION ONSET TIMING ON IMMEDIATE PROPULSION BENEFITS

The importance of patient-tailored plantarflexor neurostimulation during propulsion FES is especially evident in our finding that mis-timed plantarflexor neurostimulation results in a worsening of an individual's propulsion ability. At a group level, the difference in paretic propulsion peak improvement between the two timings was 1.8%BW, which matches the minimum detectable change (MDC) threshold reported in the literature [45] and suggests that the effect of these two timings is meaningfully different. At an individual level, five out of ten participants had between-timing differences that exceeded this MDC, and an additional two people approached the MDC (i.e., $\Delta > 70\%$ MDC). If this study had looked at a wider range of neurostimulation timings, similar to other studies targeting the optimization of assistance with wearable devices in both healthy [46] and clinical populations [47], we expect more substantial differences between preferred and non-preferred timings, further highlighting the importance of tailoring neurostimulation timing for each individual.

C. MAINTAINING FOOT CLEARANCE WITH PROPULSION FES ASSISTANCE

The safe delivery of propulsion assistance is critical for everyday walking and training with a propulsion FES neuroprosthesis. Indeed, among patient populations with impaired swing-phase dorsiflexion, poor coordination of the dorsiflexor neurostimulation required for foot clearance assistance and the plantarflexion neurostimulation required for propulsion assistance creates substantial risk for tripping and falls. Prior work shows that the addition of stance-phase plantarflexor neurostimulation to swing-phase dorsiflexor neurostimulation can significantly decrease the swing-phase dorsiflexion angle [34]. Similarly, in pilot testing we conducted before this study (described in **Fig. S1**), we observed that a plantarflexor neurostimulation offset timing at the paretic toe-off event reduced foot clearance during swing phase, with this negative effect remedied when the plantarflexor neurostimulation offset timing was set earlier. This negative effect was further remedied by also shifting the dorsiflexor onset timing to be earlier. These modifications are thought to be necessary to account for electromechanical delays [48] that cause a latency from the delivery of neurostimulation to the onset of the muscle contraction. Together, these findings illustrate a need to precisely tune, on a step-by-step basis, the timing and amplitude of plantarflexor and dorsiflexor neurostimulation when both are delivered together. Indeed, for the present study, the group-level change in peak swing phase dorsiflexion angle from unassisted walking for both the preferred and non-preferred timings exceeded the treadmill MDC of 0.9 degrees [49]. This finding was not statistically significant ($p > 0.05$), indicating that there was not a significant negative interaction between plantarflexor and dorsiflexor neurostimulation. At an individual level, with the non-preferred timing, four out of ten individuals increased peak dorsiflexion angle by the MDC, compared to unassisted walking, whereas two participants had reductions that exceeded the MDC. Similarly, for the preferred timing, five out of ten individuals increased dorsiflexion angle beyond the MDC, compared to unassisted walking, whereas one individual had a reduction in dorsiflexion angle by the MDC. The MDC used for this comparison of dorsiflexion angle is based on a treadmill dataset, whereas this study assesses changes during overground walking. The applicability of treadmill MDCs to overground walking should be considered. Future studies are needed to investigate methods to systematically tune neurostimulation during the stance-to-swing transition to maximize exposure to propulsion assistance while avoiding unsafe reductions in foot clearance.

D. PROPULSION FES LEADS TO CARRYOVER IMPROVEMENTS IN OVERGROUND SPEED AND PROPULSION

An important finding of this study is that a single session of walking with propulsion FES resulted in retained improvements in walking speed and gait propulsion that met or surpassed previously reported MDC thresholds [45], [49], [50]. The magnitudes of these single-session changes were comparable to the improvements in paretic propulsion peak (median Δ: 3.28%BW) and speed (comfortable Δ: 0.18 $\pm$ 0.07 m/s; fast Δ: 0.18 $\pm$ 0.09 m/s) reported after a 3-month neurorehabilitation clinical trial of the FastFES intervention, which combined propulsion FES with fast treadmill walking [38]. These findings motivate further study of the time-course of improvements over multiple training sessions, which was beyond the scope of this foundational study.

E. LIMITATIONS

Although the study sample size provided sufficient power for the large (ES > 0.990) and clinically meaningful [45], [49], [50] effects observed in this study, the generalizability of these findings is limited to community dwelling individuals with fast walking speeds [43].

Control group comparison of the post-session rehabilitative effects was beyond the scope of this study. Thus, we are not able to distinguish whether post-session changes are due to the combination of walking and neurostimulation or due to the neurostimulation alone. Regardless, the immediate effects of using the propulsion neuroprosthesis demonstrate that walking with neurostimulation does differ from unassisted walking.

The two plantarflexor neurostimulation onset timings used in this study were chosen to represent key biomechanical functions of the plantarflexors: i) an early onset of plantarflexor assistance (i.e., before midstance) supports progression of the limb during midstance; ii) a late onset of plantarflexor assistance (i.e., after midstance) provides push-off assistance during the step-to-step transition. However, these two timings do not provide a comprehensive assessment of all possible neurostimulation timings. Future work may consider a sweep of timing profiles to precisely individualize the delivery of propulsion FES.

Each neurostimulation timing was assessed once to mitigate fatigue-induced changes in gait from repeated testing. Gait variability across repeated measurements is thus a potential limitation mitigated by contextualizing the magnitude of the changes in our outcomes using previously reported MDC values [45], [49], [50] that account for variability across repeated measurements. The context of previously reported MDC values should be considered when applying to new studies or specific populations. Future studies that evaluate reproducibility

FIGURE 4. Experimental protocol. A pre-session and post-session evaluation of unassisted walking consisted of three trials of an instrumented 10-meter walk test (10 mWT) to assess comfortable walking speed (CWS) and three trials to assess fast walking speed (FWS). FES tuning consisted of a seated tuning phase to adjust electrode placement and minimize ankle inversion/eversion and a walking tuning phase to calibrate the neurostimulation amplitudes needed for walking. Dorsiflexor neurostimulation amplitude was calibrated to induce neutral ankle angle and plantarflexor neurostimulation amplitude was calibrated to the maximum tolerable level for each participant. The FES-assisted walking session consisted of treadmill and overground walking. Treadmill exposure included two five-minute walks at CWS. Each minute of the treadmill walks alternated between unassisted walking (No FES) and FES-assisted walking at each of the timing conditions in a randomized order. Overground training included five sets of eight 10 mWTs at FWS. Each set alternated between unassisted walking (No FES) and FES-assisted walking at each of the timing conditions in a randomized order.

of the effects of different plantarflexor neurostimulation onset timings are warranted.

IV. CONCLUSION

We developed a propulsion FES neuroprosthesis that coordinates neurostimulation to the paretic dorsiflexors and plantarflexors during overground walking, resulting in both immediate and retained improvements in post-stroke walking speed, peak paretic propulsion, and propulsion symmetry. This study demonstrates the importance of individualized tuning of plantarflexor neurostimulation timing and motivates the advance of real-time methods for propulsion estimation. This study highlights the potential for propulsion FES to target intractable post-stroke gait deficits and contribute to enhancing the quality of everyday living for individuals post-stroke. Further research with a control group is necessary to investigate the efficacy and long-term effects of propulsion FES interventions. Further development of the neuroprosthesis technology to incorporate automated neurostimulation tuning is warranted to maximize the immediate and therapeutic benefits of walking with propulsion FES.

V. MATERIALS AND METHODS

Ten individuals in the chronic phase of stroke recovery (2 female; 53 ± 11 years old; 8 ± 1 years poststroke) completed this study (see **Table S2** for complete participant characteristics) (NCT06459401). Each study participant completed one session that consisted of i) neurostimulation tuning and exposure, ii) overground gait training with propulsion FES, and iii) pre-session and post-session evaluations (see Fig. 4 for an overview of the experimental protocol). Mean ± standard error (SE) are reported for all conditions and effect sizes (ES) are calculated using the pooled standard deviation (SD), unless otherwise specified. Medians are reported for all percent changes to account for outliers that may occur during the calculation of the percent change from baseline for those with small baseline values.

All study data were processed using commercial motion analysis software (Visual 3D, C-Motion Inc., Boyds, MD) and a computing platform (MATLAB, The MathWorks Inc., Natick, MA). Kinetic and kinematic data were filtered using a fourth-order Butterworth lowpass filter with a cutoff frequency at 10 Hz. Pre- and post-session walking speeds were calculated as the average of three 10mWT trials. Timeseries data were stride-segmented by initial foot contact and normalized to the gait cycle. Point metrics from the timeseries data were calculated in MATLAB (see **Supplementary Materials** for a complete description of the study's **Materials and Methods**).

SUPPLEMENTARY MATERIALS

Table S1 presents the measured amplitude and timing values of dorsiflexor and plantarflexor neurostimulation. **Fig. S1** shows when neurostimulation was active within the gait cycle and illustrates the need for adjusted neurostimulation timings based on pilot work. **Table S2** summarizes participant baseline characteristics.

AUTHOR CONTRIBUTIONS

D.C., A.A., J.S., C.W., and L.A. formed the concept and methodology of the study. D.C., A.A., and J.S. conducted the investigation. L.A. and C.W. supervised the project and acquired funding. D.C. and A.A. wrote the initial manuscript draft and created data visualizations with input from all authors. All authors contributed to and approved the final manuscript.

CONFLICT OF INTEREST

The authors declare that they have no competing interests.

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