NEUROREHABILITATION

A soft robotic exosuit improves walking in patients after stroke

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Stroke-induced hemiparetic gait is characteristically slow and metabolically expensive. Passive assistive devices such as ankle-foot orthoses are often prescribed to increase function and independence after stroke; however, walking remains highly impaired despite—and perhaps because of—their use. We sought to determine whether a soft wearable robot (exosuit) designed to supplement the paretic limb's residual ability to generate both forward propulsion and ground clearance could facilitate more normal walking after stroke. Exosuits transmit mechanical power generated by actuators to a wearer through the interaction of garment-like, functional textile anchors and cable-based transmissions. We evaluated the immediate effects of an exosuit actively assisting the paretic limb of individuals in the chronic phase of stroke recovery during treadmill and overground walking. Using controlled, treadmill-based biomechanical investigation, we demonstrate that exosuits can function in synchrony with a wearer's paretic limb to facilitate an immediate 5.33 ± 0.91° increase in the paretic ankle's swing phase dorsiflexion and 11 ± 3% increase in the paretic limb's generation of forward propulsion ($P < 0.05$). These improvements in paretic limb function contributed to a 20 ± 4% reduction in forward propulsion interlimb asymmetry and a 10 ± 3% reduction in the energy cost of walking, which is equivalent to a 32 ± 9% reduction in the metabolic burden associated with poststroke walking. Relatively low assistance (~12% of biological torques) delivered with a lightweight and nonrestrictive exosuit was sufficient to facilitate more normal walking in ambulatory individuals after stroke. Future work will focus on understanding how exosuit-induced improvements in walking performance may be leveraged to improve mobility after stroke.

INTRODUCTION

Bipedal locomotion is a defining trait of the human lineage, with a key evolutionary advantage being a low energetic cost of transport (*1*). However, the economy of bipedal gait may be lost because of neurological injury with disabling consequences. Hemiparetic walking (*2–7*) is characterized by a slow and highly inefficient gait that is a major contributor to disability after stroke (*8*, *9*), which is a leading cause of disability among Americans (*10*). Despite rehabilitation, the vast majority of stroke survivors retain neuromotor deficits that prevent walking at speeds suitable for normal, economical, and safe community ambulation (*11*). Impaired motor coordination (*12*), muscle weakness and spasticity (*13*), and reduced ankle dorsiflexion (DF; drop foot) and knee flexion during walking are examples of typical deficits after stroke that limit walking speed and contribute to gait compensations such as hip circumduction and hiking (*14–18*), increase the risk of falls, and reduce fitness reserve and endurance (*3*, *4*, *9*, *12*, *19–21*). Even those able to achieve near-normal walking speeds present with gait deficits (*22*, *23*) that hinder community reintegration and limit participation to well below what is observed in even the most sedentary older adults (*24*, *25*), ultimately contributing to reduced health and quality of life (*10*, *26*, *27*).

Walking independence is an important short-term goal for survivors of a stroke; however, independence can be achieved via compensatory mechanisms. The persistence of neuromotor deficits after rehabilitation often necessitates the prescription of passive assistive devices such as canes, walkers, and orthoses to enable walking at home and in the community (*28–30*). Unfortunately, commonly prescribed devices compensate for poststroke neuromotor impairments in a manner that prevents normal gait function. For example, ankle-foot orthoses (AFOs) inhibit normal push-off during walking (*31*) and reduce gait adaptability (*32*). The stigma associated with the use of these devices is also important to consider, especially for the growing population of young adult survivors of stroke (*33*, *34*). The major personal and societal costs of stroke-induced walking difficulty and the limitations of the existing intervention paradigm motivate the development of rehabilitation interventions and technologies that enable the rapid attainment of more normal walking behavior.

Recent years have seen the development of powered exoskeletal devices designed to enable walking in individuals who are unable to walk (*35*, *36*). Central to this remarkable engineering achievement is a rigid structure that can support its own weight and provide high amounts of assistance; however, these powerful machines may not always be necessary to restore more normal gait function in individuals who retain the ability to walk after neurological injury, such as the majority of those after stroke. To address this opportunity, our team developed a lightweight, soft wearable robot (exosuit) that interfaces to the paretic limb of persons after stroke via garment-like, functional textile anchors. Exosuits produce gait-restorative joint torques by transmitting mechanical power from waist-mounted body-worn (*37*) or off-board (*38*, *39*) actuators to the wearer through the interaction of the textile anchors and a cable-based transmission.

Several factors, such as the compliance of the exosuit-human system (*40*), prevent exosuits from providing the assistance necessary to

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enable nonambulatory individuals to walk again (*41*); however, for ambulatory individuals, the lightweight and nonrestrictive nature of this technology has the potential to facilitate a more natural interaction with the wearer and minimize disruption of the natural dynamics of walking (*42*). Our first efforts developing exosuits led to the creation of systems that could comfortably deliver assistive forces to healthy users during walking (*39*, *40*, *43*–*47*). Recently, we demonstrated that assistive forces delivered through the exosuit interface produce marked reductions in the energy cost of healthy walking (*37*, *48*). Thus, although exosuits can only augment, not replace, a wearer's existing gait functions, we posit that they have the potential to work synergistically with the residual abilities of individuals with impaired gait to improve walking function.

The primary objective of this foundational study was to evaluate the potential of using the exosuit technology to restore healthy walking behavior in individuals after stroke. Toward this end, we evaluated the effects on hemiparetic gait of actively assisting the paretic limb during treadmill walking using a tethered, unilateral (worn on only one side of the body) exosuit designed to supplement the wearer's generation of paretic ankle plantarflexion (PF) during stance phase and DF during swing phase. We posited that this targeted assistance of the paretic ankle's gait functions would facilitate more symmetrical propulsive force generation by the paretic and nonparetic limbs and reduce the energetic burden associated with poststroke walking, which previous work has shown can be more than 60% more costly (*49*). Previous work on wearable assistive robots for persons after stroke has suggested that the timing of PF force delivery during walking could be an important contributor to positive outcomes in this heterogeneous population (*50*). Hence, we also evaluated different onset timings of PF force delivery for each individual, hypothesizing that this timing would need to be individualized to optimize outcomes.

Designed to be unobtrusive to the wearer when not powered, the exosuit's mass of ~0.9 kg is distributed along the length of the paretic limb similar to a pair of pants. Nonetheless, to understand the net effect of walking with an exosuit powered and assisting the paretic limb,

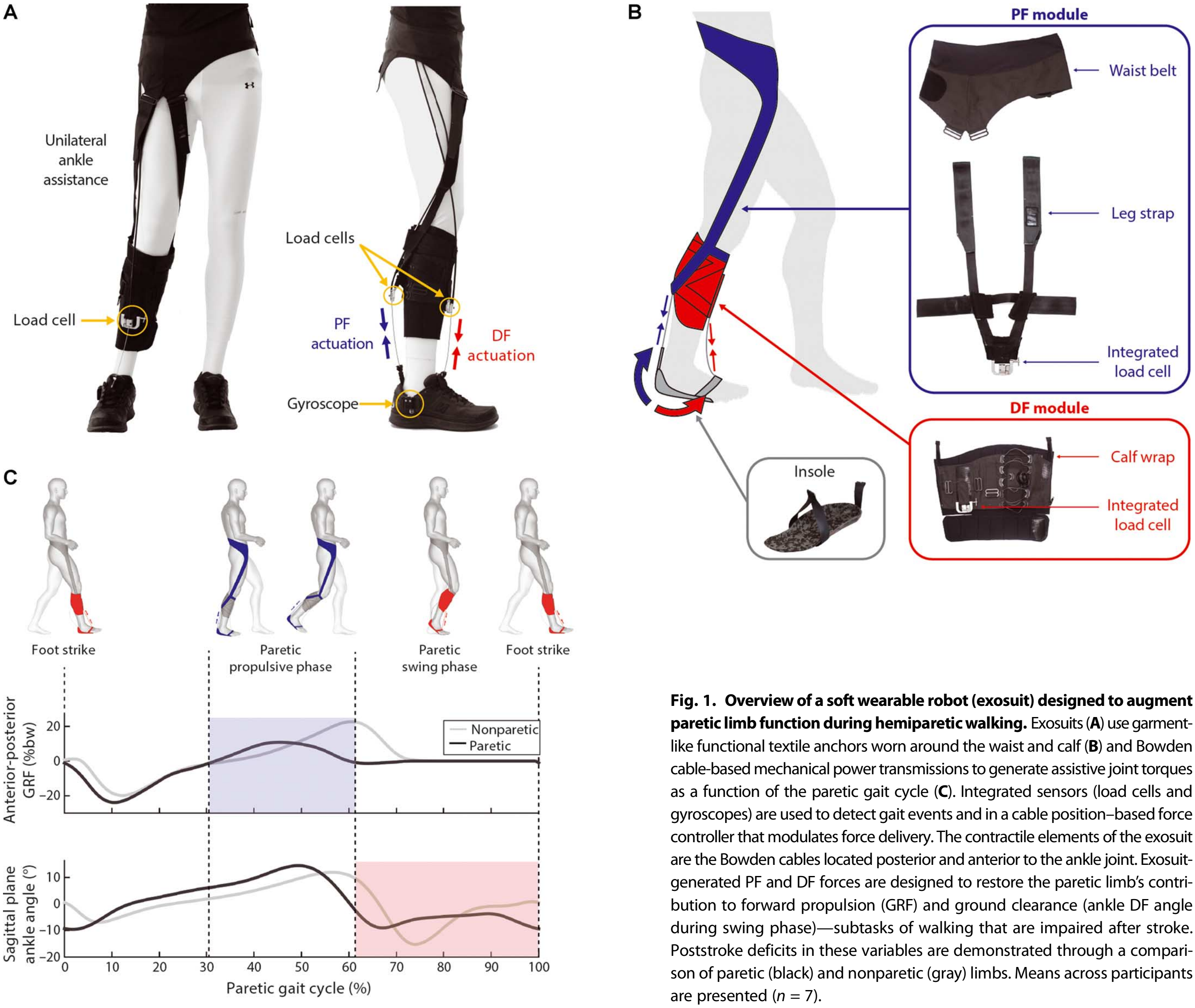

Fig. 1. Overview of a soft wearable robot (exosuit) designed to augment paretic limb function during hemiparetic walking. Exosuits (**A**) use garment-like functional textile anchors worn around the waist and calf (**B**) and Bowden cable-based mechanical power transmissions to generate assistive joint torques as a function of the paretic gait cycle (**C**). Integrated sensors (load cells and gyroscopes) are used to detect gait events and in a cable position–based force controller that modulates force delivery. The contractile elements of the exosuit are the Bowden cables located posterior and anterior to the ankle joint. Exosuit-generated PF and DF forces are designed to restore the paretic limb's contribution to forward propulsion (GRF) and ground clearance (ankle DF angle during swing phase)—subtasks of walking that are impaired after stroke. Poststroke deficits in these variables are demonstrated through a comparison of paretic (black) and nonparetic (gray) limbs. Means across participants are presented (n = 7).

it is necessary to evaluate whether there are effects because of simply wearing the exosuit passively (worn but unpowered). A secondary objective was thus to evaluate the effects of walking with the passive exosuit relative to walking with the exosuit not worn. Moreover, because one of the compelling aspects of soft wearable robots, such as exosuits, is their potential to provide gait assistance and, potentially, rehabilitation benefit during community-based walking activities, in addition to treadmill-based biomechanical investigation into the effects of a tethered exosuit, our final objective was to evaluate the effects of exosuit assistance delivered from a first-generation, body-worn (untethered) exosuit during overground walking. Ultimately, by investigating how individuals with poststroke hemiparesis respond to exosuit-generated active assistance of ankle PF and DF during treadmill and overground walking, this study serves to define the technology's potential for improving mobility and enabling more effective neurorehabilitation after stroke.

RESULTS

In brief, individuals in the chronic phase of stroke recovery participated in evaluating the feasibility of using a unilateral, ankle-assisting exosuit to restore more normal gait mechanics and energetics after stroke (see Fig. 1, A and B). Walking dysfunction after stroke may result from a variety of deficits, ranging from reduced balance (*51*, *52*) to atypical biomechanics. We expected exosuits to have an immediate effect on key biomechanical subtasks that are impaired after stroke, specifically ground clearance by assisting ankle DF during swing phase and propulsive force generation by assisting ankle push-off during stance phase (Fig. 1C). We focused on the immediate improvements in poststroke walking observed during exosuit assistance, not on long-term therapeutic changes that may arise from sustained use.

For our primary investigation of changes in poststroke mechanics and energetics during treadmill walking, two testing conditions were evaluated: walking with the exosuit tethered to an off-board actuation unit that delivered assistive forces to the wearer (powered) versus walking with the tethered exosuit not transmitting any forces (unpowered). This testing was conducted on an instrumented treadmill with the speed set to each participant's usual overground walking speed (Fig. 2). Two primary outcomes were considered: interlimb propulsion symmetry, computed from the mass-normalized peaks [% body weight (% bw)] of the anterior ground reaction force (GRF) recorded during each limb's stance phase (*53*), and the energy cost of walking, defined as mass-normalized oxygen consumption per meter ambulated (ml O_2/kg/m). Participants' propulsion impulse [% body weight seconds (% bw.s)] was also measured as a secondary propulsion outcome.

Participant baseline characteristics

Participants, on average, were not able to produce sufficient ankle DF to reach a neutral paretic ankle position during swing phase when walking with the exosuit unpowered, with the average peak ankle DF angle during swing phase measured at −1.85 ± 1.98° (1.85° of PF) for the nine participants recruited for this study. Forward propulsion data were not measurable for two participants who walked with a gait pattern that prevented independent measurement of GRFs for each limb by the independent treadmill force plates. The remaining seven participants presented with, on average, 46% less peak paretic propulsion (PP) (11.39 ± 2.31% bw) compared to peak non-PP (20.08 ± 2.03% bw), 52% less PP impulse (1.98 ± 0.38% bw.s) compared to non-PP impulse (4.23 ± 0.21% bw.s), and an energy cost of walking (0.22 ± 0.03 ml O_2/kg/m) 46% higher than what has been reported to be normal (0.151 ml O_2/kg/m) (*54*). See the "Participants and inclusion/exclusion criteria" section and Table 1 for greater detail.

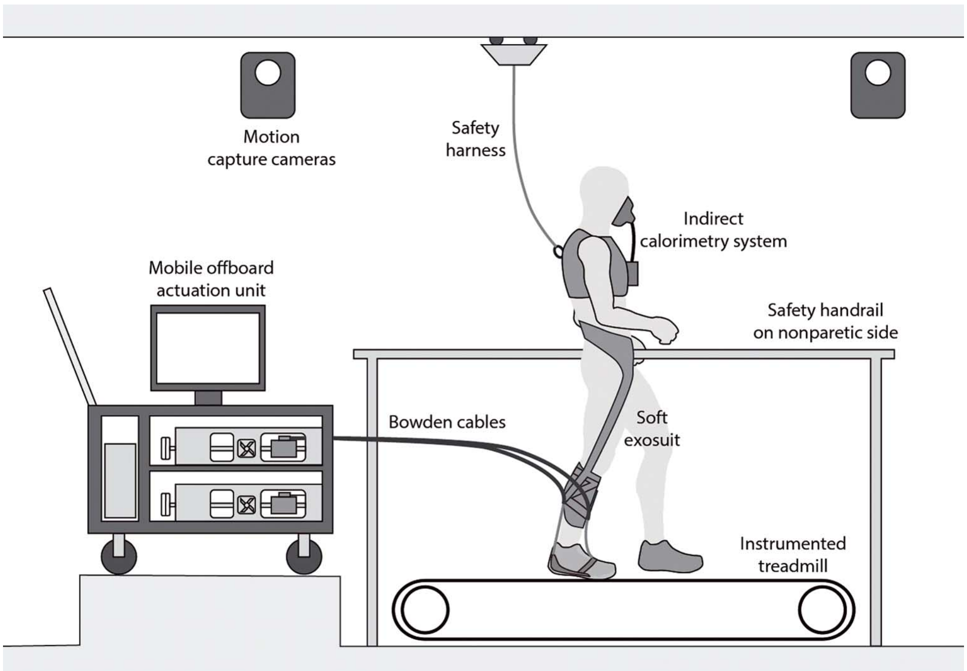

Fig. 2. Illustration of experimental setup. Ground reaction force and kinematic data were collected concurrently with metabolic data as participants walked on an instrumented treadmill with the exosuit worn either powered (delivering forces generated by an off-board actuation unit) or unpowered. Participants were harnessed, but no body weight was supported. Participants were allowed to hold a side-mounted handrail if necessary for safety.

Effective targeting of paretic limb ground clearance and forward propulsion deficits

On average, participants walked with 5.33 ± 0.91° more peak ankle DF angle during swing phase [$P < 0.001$, effect size (ES) = 1.96] with the tethered exosuit powered (3.49 ± 1.52°) versus unpowered (−1.85 ± 1.98°) (Fig. 3A). This increase was six times larger than the 0.90° minimal detectable change (MDC) score reported for this key metric of swing phase gait function (*55*). For all subsequent analyses, which are focused on the exosuit's effects on propulsion and walking economy, only data from when the onset timing of PF force delivery was tuned for each participant to their more effective timing were used (see the "PF assistance timing" section). Participants' peak PP increased by 11 ± 3% ($P = 0.009$, ES = 1.44) when walking with the exosuit powered (12.66 ± 2.35% bw) versus unpowered (11.39 ± 2.31% bw)—a change 1.6 times larger than the 0.80% bw MDC reported for this variable (*55*). Concurrently, participants' energy cost of walking was reduced by 10 ± 3% (P = 0.009, ES = 1.42) when walking with the exosuit powered (0.1979 ± 0.0218 ml O_2/kg/m) versus unpowered (0.2204 ± 0.0275 ml O_2/kg/m). These improvements ultimately contributed to a 20 ± 4% reduction in interlimb peak propulsion asymmetry (P = 0.002, ES = 1.93; Fig. 3B), a similar 19 ± 8% reduction in propulsion impulse asymmetry (P = 0.021, ES = 1.19), and a walking economy 32 ± 9% closer to the normal walking economy of 0.151 ml O_2/kg/m (P = 0.009, ES = 1.81; Fig. 3C). A subgroup analysis of exosuit-induced changes in propulsion asymmetry for individuals who used (n = 3) and individuals who did not use (n = 4) a handrail during testing revealed that both groups reduced propulsion asymmetry by a similar magnitude (handrail group %Δ, −22.8 ± 4.3%; no handrail group %Δ: −17.4 ± 5.5%; P = 0.45).

Across participants, relative improvements in walking economy were highly related to relative improvements in propulsion symmetry (R^2 = 0.89, $F_{1,5}$ = 41.09, P = 0.001; Fig. 3D). Although all participants presented with an improvement in propulsion symmetry, slower participants presented with the largest relative gains (R^2 = 0.63, $F_{1,5}$ = 8.50, P = 0.03; Fig. 3E).

Exosuit-generated assistive forces

An iterative, force-based, position controller was designed and implemented to control the exosuit's assistance of ankle PF and DF [see (*39*) and (*56*) and the "Soft exosuit design and operation" and "Exosuit-generated assistive forces" sections]. In brief, exosuit-generated assistive forces are a function of a commanded Bowden cable position, the wearer's gait kinematics, and the exosuit-human series stiffness—a parameter that accounts for the interaction between the commanded cable position, the material properties of the textile anchors and Bowden cables, and the compliance of the human tissue that supports the textile anchors (Fig. 4A) (*40*). To improve PP and ground clearance, user-defined features of the commanded ankle PF and DF cable position trajectory are adapted by the controller on a step-by-step basis. To increase PP, PF force is generated by the exosuit during the paretic stance phase, with the timing of force delivery defined as a percentage of the paretic gait cycle (% GC) and the amplitude of force delivery defined as a percentage of participants' body weight (% bw). More specifically, the onset timing of PF force delivery and peak amplitude of PF force delivered (constrained to 25% bw) were selected by the research team before testing and controlled on a step-to-step basis through adjustments in the commanded cable position based on measurements of delivered force made by a load cell integrated into the exosuit's textiles and located on the posterior shank (see the "PF assistance timing" section and Fig. 4B). The timing of peak PF force and PF force cessation is not directly controlled and varies with the wearer's kinematics (Fig. 4B). To improve paretic ground clearance, a DF cable position command profile is executed as a function of the paretic gait cycle with the onset and off timings of force delivery linked by the controller to the paretic swing phase. More specifically, the amplitude of the DF cable position command is selected by the research team before testing and is set to match the position of a neutral ankle angle, or the smallest ankle PF angle if neutral is not achievable, as identified through visual observation. With this approach, the amplitude of DF force that is delivered to the wearer during swing phase within and across steps varies with the wearer's gait in a manner that maintains the commanded cable position and thus ankle angle.

The reliable delivery of exosuit-generated forces was evidenced in low variability of the prescribed exosuit parameters within and across the 2-day testing protocol conducted to evaluate the differential effects of an early-onset versus late-onset timing of ankle PF force. Across the 2 days of testing, the average ± SE of the SD observed for PF onset, DF onset, and DF off timings were 1.51 ± 0.26% GC, 2.71 ± 0.16% GC, and 0.91 ± 0.20% GC, respectively. Similarly, the average ± SE of the SD in peak PF force was 1.15 ± 0.08% bw (Fig. 4C). As prescribed, across participants, exosuit-generated peak PF forces

Table 1. Participant baseline characteristics and gait performance. AD, assistive device; P, paretic; NP, nonparetic; AGRF, anterior GRF; F, female; M, male; Y, yes; N, no.

Participant	Side of paresis	Sex	Age (years)	Chronicity (years)	Regular orthosis	Regular AD	Handrail	Walking speed (m/s)	Peak P ankle angle (°)	Peak P AGRF (% bw)	Peak NP AGRF (% bw)	Cost of transport (ml O_2/kg/m)
01	Right	F	30	7.08	AFO	None	Y	1.05*	−0.02	12.80	24.57	0.28
02	Left	M	56	3.58	None	None	N	1.05	4.20	7.09	20.32	0.20
03	Left	F	52	0.75	None	Cane	Y	0.53	0.30	3.89	11.05	0.28
04	Left	M	51	2.83	AFO†	Cane	Y	0.93	−14.73	9.48	17.60	0.15
05	Left	F	37	1.08	AFO†	Cane	N	0.67	0.39	7.64	17.43	0.33
06	Right	M	44	2.33	None	None	N	1.29	−2.34	18.77	22.27	0.14
07	Right	F	46	4.25	None	None	N	1.3*	−0.80	20.06	27.34	0.18
08‡	Right	M	61	14.17	AFO	None	Y	0.92	−7.52	—	—	—
09‡	Left	M	67	3.33	None	Cane	Y	0.81	3.85	—	—	—

*The participant's actual 10-m overground walk test speed was higher than the speed tested on the treadmill. Participant #01's actual overground speed was 1.16 m/s, but this participant was not safe walking at this speed on the treadmill. Participant #07's speed was 1.72 m/s, but this speed was beyond the capabilities of the electromechanical actuator used for this study. †Participant #04 typically used a foot-up brace. Participant #05 used a custom brace that supported frontal plane motion. ‡GRF data unavailable.

averaged 25.02 ± 0.92% bw, with an SE across days of 0.13% bw (Fig. 4C). The peak PF assistive torque delivered to participants thus approximated 0.15 Nm/kg (table S1), which is 12% of the average peak paretic ankle moment that was observed across participants (1.26 ± 0.13 Nm/kg) and 9% of the average peak nonparetic ankle moment (1.68 ± 0.12 Nm/kg).

PF assistance timing

The only parameter manipulated across the 2 days of testing was the onset timing of PF force delivery. Across participants, a difference in baseline gait performance (walking with the exosuit unpowered) across these 2 days of testing was not observed for either primary variable of interest ($P > 0.05$). For each participant, the effect of an onset timing during terminal stance (late-onset) was compared to the effect of an onset timing about 10% earlier in the gait cycle, during midstance (early-onset). The exact timing varied across participants depending on the duration of each participant's paretic stance phase, with the actual commanded PF onset timings tested across participants averaging 36.9 ± 0.76% GC and 27.5 ± 1.94% GC for late- and early-onset timings, respectively (Fig. 4B). The order that these onset timings were tested was randomized across participants.

An early-onset timing of PF force delivery generated higher PF forces between 19% and 47% of the paretic gait cycle as compared to a late-onset timing ($P < 0.05$; Fig. 4B), demonstrating our ability to control the delivery of exosuit-generated forces. As previously described, exosuit-generated peak PF forces did not differ between days ($P > 0.05$; Fig. 4B). As hypothesized, the effect of exosuit-generated PF forces on participants' interlimb propulsion symmetry depended on whether participants received an early- or late-onset timing of PF force delivery (Fig. 4D). More specifically, four participants benefited more from a late-onset timing of PF force delivery and two benefited more from an early-onset timing. One participant benefited equally from both onset timings. Three of four who benefited more from a late onset of PF force delivery experienced greater propulsion asymmetry with an early-onset timing of PF force, whereas no participants experienced negative effects from a late-onset timing of PF force.

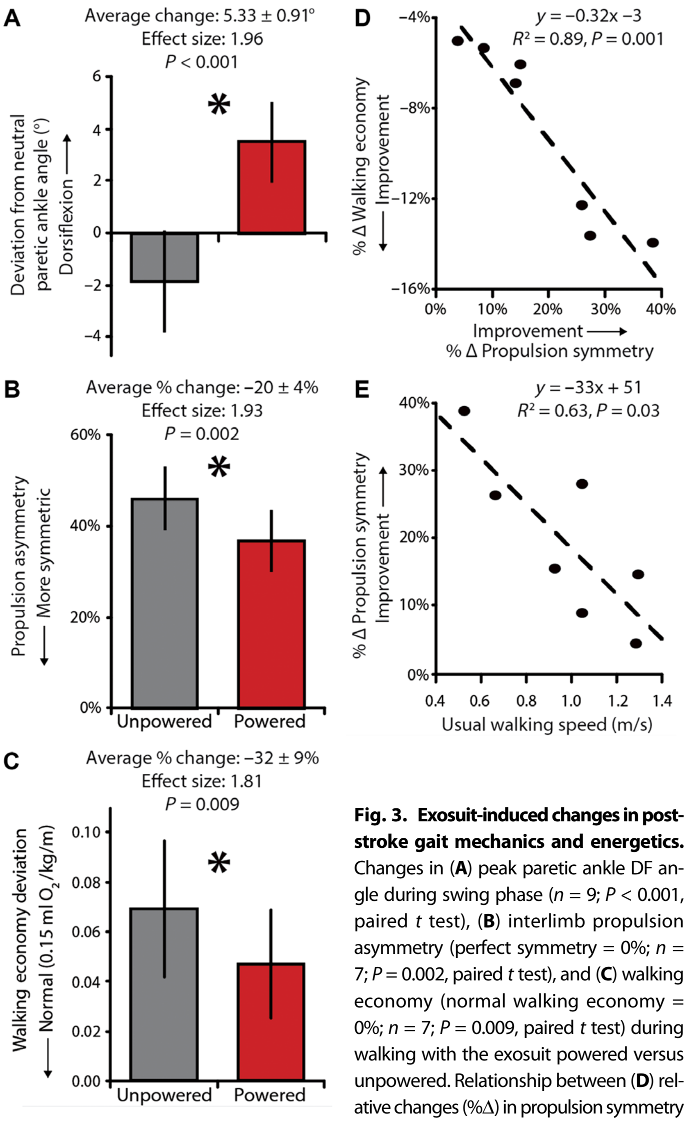

Fig. 3. Exosuit-induced changes in poststroke gait mechanics and energetics. Changes in (**A**) peak paretic ankle DF angle during swing phase ($n = 9$; $P < 0.001$, paired t test), (**B**) interlimb propulsion asymmetry (perfect symmetry = 0%; $n = 7$; $P = 0.002$, paired t test), and (**C**) walking economy (normal walking economy = 0%; $n = 7$; $P = 0.009$, paired t test) during walking with the exosuit powered versus unpowered. Relationship between (**D**) relative changes (%Δ) in propulsion symmetry and walking economy (correlation: $n = 7$, $P = 0.001$) and (**E**) participants' usual walking speed and relative change in propulsion symmetry (correlation: $n = 7$, $P = 0.03$). Means and SE are presented in (A) to (C). *$P < 0.05$.

Walking with a passive exosuit

Five individuals participated in evaluating the effects of walking with an exosuit worn, but not powered. As hypothesized, walking with the passive exosuit did not significantly modify participants' generation of propulsion from their paretic and nonparetic limbs or energy cost of walking (fig. S1). See the "Passive exosuit study" section of the Supplementary Materials for additional details regarding this secondary study.

Proof of principle: Overground walking with an untethered exosuit

Consistent with our findings on the treadmill, walking overground with an untethered exosuit (Fig. 5A) powered versus unpowered resulted in 4.9° more peak ankle DF angle during swing phase ($P < 0.002$; Fig. 5B), an increase from −4.13 ± 1.82° (4° of PF) to 0.74 ± 1.51° (0.74° of DF); 13% more peak PP, an increase from 10.3 ± 0.60% bw to 11.6 ± 0.60% bw ($P = 0.053$); and 14% more PP impulse, an increase from 2.0 ± 0.10% bw.s to 2.2 ± 0.10% bw.s ($P = 0.029$). Ultimately, a 16.3 ± 6.8% reduction in propulsion asymmetry was observed during overground walking with versus without exosuit assistance ($P = 0.045$; Fig. 5C). See the "Overground gait assistance with an untethered exosuit" section of the Supplementary Materials for more details regarding this secondary study.

DISCUSSION

Exosuits function in synchrony with the paretic limb of persons after stroke to overcome deficits in forward propulsion and ground clearance during hemiparetic walking, ultimately facilitating more economical locomotion. Using a tethered exosuit research platform, this study demonstrates the feasibility of unilateral assistance of paretic ankle function through a soft human-machine interface and the efficacy of minimal assistance (~12% of biological PF torques) delivered through this interface. Moreover, using a first-generation, untethered exosuit, this study demonstrates the potential for exosuits to provide gait assistance and training during overground walking. These findings support future research and development of gait-restorative exosuits, with studies using optimized body-worn actuators and evaluating clinic- and community-based outcomes especially warranted.

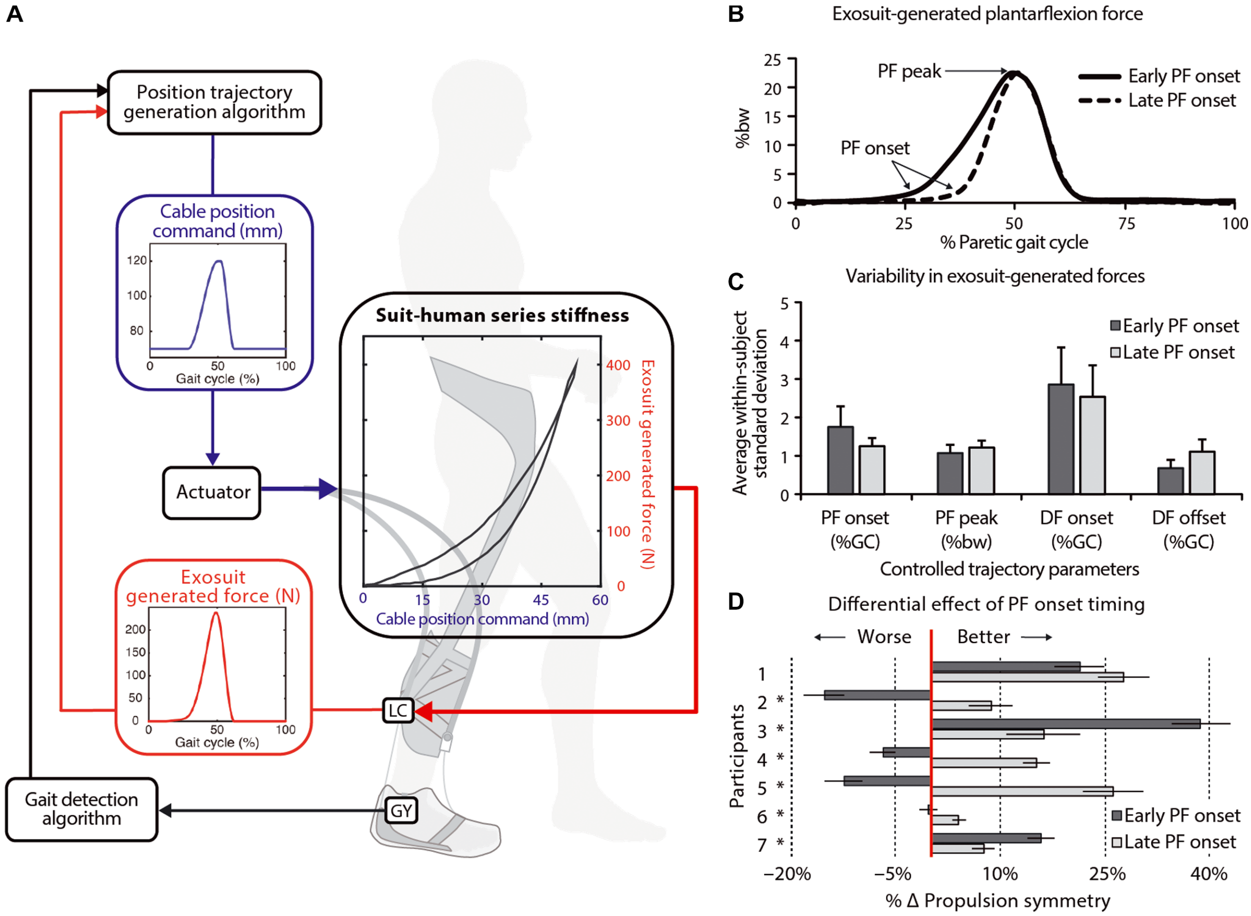

Fig. 4. Overview and assessment of exosuit controller. (**A**) Commanded position trajectories (blue) translate to exosuit-generated forces (red) based on the exosuit-human series stiffness. This parameter reflects the interaction of the cable position command with the compliance inherent to the textiles, cables, and human tissue. On a step-by-step basis, the timing of exosuit-generated force delivery is controlled as a function of gait subphases identified using shoe-mounted gyroscope sensors (GY). The amplitude of force produced by the exosuit is measured by load cells (LC) that measure the force delivered at the ankle—one for each PF and DF (only PF is shown). The delivered force is continuously monitored, and adjustments are made by the controller to maintain specific features of the delivered force profile. (**B**) Average exosuit-generated PF forces as a function of the paretic gait cycle (*x* axis). Two ankle PF force delivery onset timings were evaluated. PF forces were delivered during terminal stance or about 10% earlier in the paretic gait cycle, during midstance. (**C**) Variability in commanded force parameters (*x* axis). The force features prescribed by the controller on a step-by-step basis included the onset timing of PF force (PF onset), the peak amplitude of PF force (PF peak)—constrained to 25% bw, the onset timing of DF force (DF onset), and the off time of DF force (DF offset). (**D**) Relative changes (%Δ) in participants' interlimb propulsion symmetry based on the onset timing of exosuit-generated PF forces. Four participants benefited more from a late onset of PF force timing, two participants benefited more from an early onset, and one participant (#1) benefited equally from both timings. Means and SE are presented in (C) and (D). *$P < 0.05$.

Impaired forward propulsion is posited to be a major contributor to walking dysfunction after stroke (*14*, *23*, *57*–*63*), with the magnitude of interlimb propulsion asymmetry differentiating individuals as limited community versus community ambulators (*64*). Recent work has also linked gains in propulsion to improvements in the long-distance walking ability of persons after stroke (*65*), a factor identified by individuals in the chronic phase of recovery as limiting engagement at home and in the community (*8*). Commonly prescribed AFOs have been shown to restrict ankle range of motion and impair the generation of forward propulsion during walking (*31*), effects shown to increase the energetic demands of walking (*66*). A high energy cost of walking has been posited to be a primary contributor to physical inactivity in older adults (*67*) and persons with neurologically based walking deficits (*68*–*70*), as well as a predictor of impending declines in walking speed (*71*)—a powerful marker of reduced health and mobility (*72*, *73*) and mortality (*74*). Because it targets deficits in propulsion and improves walking economy, the exosuit technology is a promising alternative to the passive assistive devices typically used by ambulatory individuals with walking-related disabilities (*75*).

An existing alternative to AFOs is functional electrical stimulation (FES). Like exosuits, FES systems provide gait assistance through a lightweight and unobtrusive human-device interface; however, exosuits apply assistive joint torques in parallel with the underlying impaired musculature, whereas FES directly activates the impaired musculature. One limitation of FES that is directly addressed by exosuits is the high fatigability, weakness, and reduced neuromuscular capacity of the paretic limb musculature. By functioning in parallel with the paretic limb, exosuits provide supplemental assistive forces and may thus be better suited for use by individuals with larger impairments or during activities requiring longer walking durations such as community walking or gait training. Future study of how exosuits and FES may be optimally used to assist hemiparetic gait is needed.

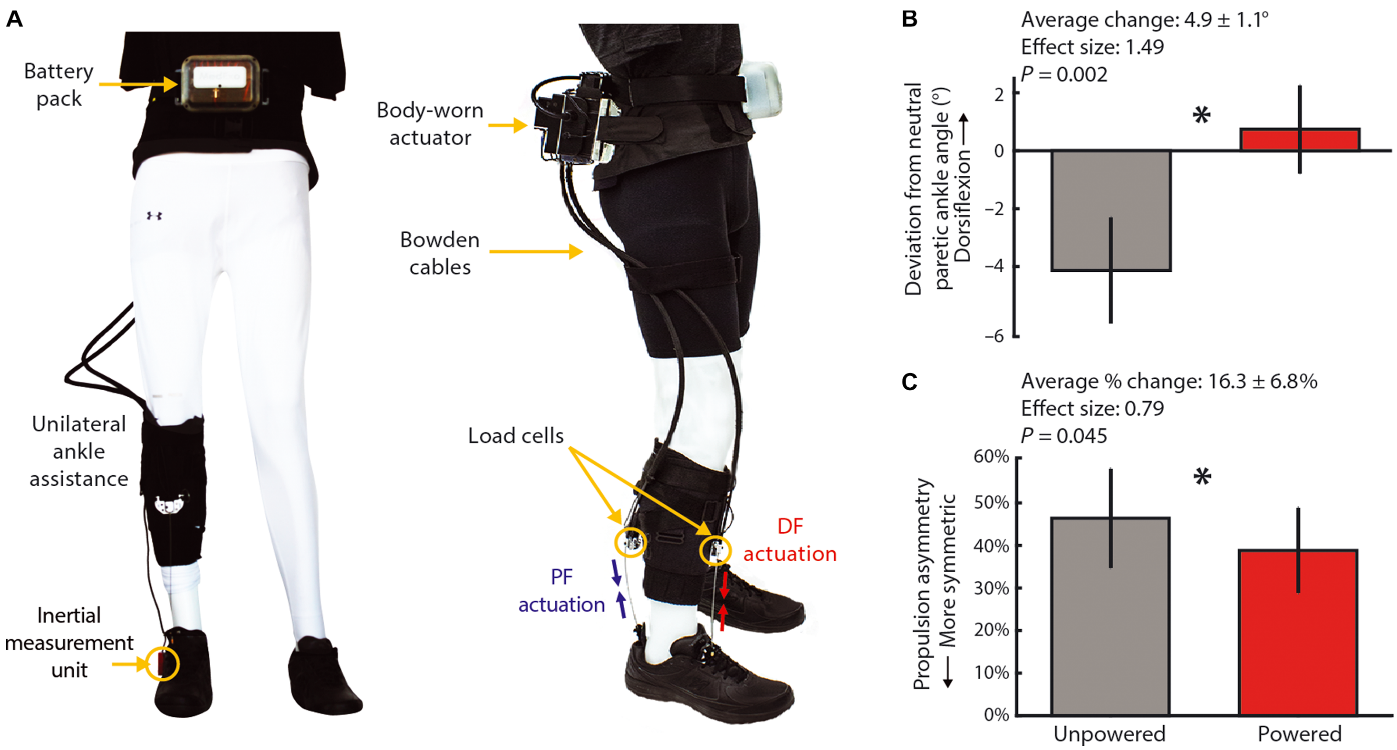

Fig. 5. Overview of an untethered, unilateral, ankle-assisting exosuit adapted for overground walking. (**A**) Mechanical power generated by a 2.63-kg actuator mounted posteriorly on the waist belt is transmitted to the wearer via cable-based transmissions. A 0.56-kg battery is attached anteriorly on the waist belt. The contractile elements of the exosuit are located anterior and posterior to the ankle joint and assist ankle DF and PF, respectively. Improvements in (**B**) peak paretic ankle DF during swing phase (n = 9; P = 0.002, paired t test) and (**C**) interlimb propulsion asymmetry (n = 9; P = 0.045, paired t test) during overground walking are presented. Means and SE are presented. *P < 0.05.

The results of this study suggest that successful translation of the exosuit technology may require refinement of selection criteria to identify appropriate candidates. For example, the findings of this study suggest that lower-functioning individuals may benefit more from exosuit intervention than higher-functioning individuals (Fig. 3E). Although this requires validation, it supports the potential use of the exosuit in more disabled cohorts after stroke. It is also not clear if higher-functioning participants could have seen added benefit with a higher amount of assistance, assistance at joints other than the ankle, or with directed training. Further investigation will be necessary.

A detailed analysis of the biomechanical mechanisms underlying exosuit-induced improvements in poststroke walking will be necessary to inform the development of future systems. Given the heterogeneity of poststroke motor impairment, customizable solutions to the particular biomechanical needs of wearers and the goals of rehabilitation are needed to enable widespread use (*76*). For example, the exosuit's effects appeared dependent, in part, on the onset timing of exosuit-generated PF force, with the optimal timing varying across participants (Fig. 4, B and D). Although this finding requires validation, it is consistent with the heterogeneous nature of poststroke motor impairment and highlights the importance of understanding the interaction between human and machine when designing and implementing controllers for gait-restorative wearable devices (*35*). On the basis of a biomechanical understanding of the role of the plantarflexors during walking, an earlier-onset timing of plantarflexor force generation would be beneficial to participants requiring assistance to control the anterior translation of the tibia over the foot, whereas a later-onset timing would be beneficial in those requiring assistance with push-off. It is likely that individuals after stroke will require some degree of assistance with both of these important functions of the plantarflexors; however, our finding that mistimed assistance of PF force delivery could negatively affect propulsive force generation highlights the need for individualization of timing. The personalization of wearable active assistive devices will require substantial investigation into biomechanical markers that can accurately predict optimal assistive strategies. On the basis of the aforementioned biomechanical framework, variables such as stance phase control of the tibia and ankle PF velocity may be promising indicators of effective PF assistance onset timing using our ankle exosuit. Further study and detailed biomechanical analyses are required.

Exosuit-induced improvements in poststroke gait were observed within minutes of powering the device and were comparable to, if not greater than, therapeutic gains observed after single-session (*77*, *78*) and multisession (*21*, *65*, *79*, *80*) clinical gait training programs. Although this study does not demonstrate a therapeutic effect, several benefits can be derived from such an immediate and substantial increase in poststroke walking performance. For example, an immediate improvement in walking capacity may increase the opportunity for walking practice of higher intensity and variability—key ingredients for effective neurorehabilitation (*70*, *81*, *82*). Improved walking capacity may improve self-efficacy and reduce barriers to community engagement (*8*, *51*, *83*). Together, these factors may facilitate better leveraging of key plasticity mechanisms such as salience, intensity, and repetition during gait training (*81*, *84*, *85*). Moreover, as observed with rehabilitation approaches centered on targeting ankle deficits during poststroke gait training (*49*, *86*, *87*), the exosuit technology's ability to facilitate walking practice of more normal gait mechanics also has the potential to promote locomotor restoration rather than compensation. For this proof-of-principle investigation, participants received no instruction on how to walk with the exosuit. Further study on how training may enhance exosuit-assisted walking is warranted.

Although this investigation's primary finding is an improvement in poststroke gait mechanics and energetics during exosuit-assisted walking versus walking with an exosuit unpowered, our secondary finding that wearing a passive exosuit did not significantly influence participants' walking (fig. S1) is equally important for a complete appreciation of the exosuit technology. Together, these findings demonstrate that exosuits are capable of providing targeted gait assistance through a human-machine interface that imposes minimal biomechanical and metabolic penalties when worn unpowered.

The tethered exosuit research platform used for this study provided a suitable test bed for demonstrating proof of principle of the exosuit concept; however, recognizing that there are differences between treadmill and overground walking after stroke, it was also necessary for us to evaluate the effects of exosuit assistance during overground walking with an untethered exosuit. Despite using a nonoptimized actuator pack, we observed improvements in PP and ground clearance that were comparable to what was observed during treadmill walking. The actuator data collected from this study will enable optimization of this body-worn system to make it suitable for both clinic and free-living settings. We estimate that the total mass of these systems will be <4 kg, including batteries, with the majority worn discretely around the waist to minimize its impact on natural walking behavior (*88*).

This study revealed the potential importance of individualizing the timing of exosuit-generated PF assistance. Because of logistical constraints, we could only evaluate two PF assistance onset timings. Although the two onset timings that were tested may not have been optimal for individual patients, they allowed us to evaluate the effects of PF assistance during two distinct phases of the gait cycle. In addition, we did not evaluate the effects of manipulating other exosuit parameters, such as the amplitude of assistance. Although future work elucidating optimal parameter tuning will be necessary to inform translation, the present study demonstrates the potential of tuning the exosuit technology to meet the needs of differengt gait presentations.

Considering the day-to-day variability inherent in the outcomes used in this study, a 2-day testing protocol to evaluate the differential effects of PF assistance timing is a potential limitation of this study. Nonetheless, this experimental approach was deemed necessary to minimize the burden on participants and the fatigue inherent to longer testing sessions. It is important to note that we made efforts to minimize the influence of participant day-to-day variability on this evaluation. For example, our between-day analyses compared the effects of the exosuit relative to each day's baseline performance, which we demonstrated did not significantly vary across participants across the 2 days of testing.

For safety, three participants required use of a handrail during treadmill testing. Although it is possible that these participants may have modulated the amount of body weight supported by the handrail between testing conditions, our demonstration that the exosuit-induced change in propulsion symmetry did not differ between individuals who did and did not use a handrail reduces this concern, as does our finding of improvements in PP during overground walking without a handrail. These findings are consistent with previous work that has shown that in individuals after stroke, changes in handrail forces are unrelated to the mechanisms used to increase forward propulsion (*89*). Nonetheless, future work quantifying changes in handrail support during exosuit-assisted walking may be revealing. In addition, studies that evaluate the long-term changes in poststroke gait because of extended use for gait training, on don/doff times, and on the robustness of the exosuit's effects to variability in exosuit placement are warranted before the utility of exosuits as assistive and rehabilitation robots is fully realized.

MATERIALS AND METHODS

Study design

The feasibility of influencing poststroke walking using a unilateral, soft wearable robot (exosuit) was evaluated through three experiments. The primary experiment focused on evaluating the effects of a tethered exosuit on poststroke gait mechanics and energetics during treadmill walking (video S1). This investigation is detailed in this section and focuses on nine individuals in the chronic phase of stroke recovery. In addition, we performed two complementary studies to better define the promise of the exosuit technology. The methods for these additional studies are detailed in two supplementary sections: "Passive exosuit study" and "Overground gait assistance with an untethered exosuit." Additional subject-level data are reported in table S2.

The exosuit was designed to support the paretic limb's generation of forward propulsion and ground clearance during walking by augmenting the paretic ankle's PF during stance phase and DF during swing phase (Fig. 1). We hypothesized that symmetry of the propulsive force output from the lower extremities would be increased, and the energy cost of walking reduced during exosuit-assisted walking compared to walking with the tethered exosuit unpowered. We used a 2-day testing protocol to evaluate the effects of two different onset timings of exosuit-generated PF force delivery on participants' forward propulsion, with the peak amplitude of PF force delivered constrained across testing days to 25% of each participant's body weight (Fig. 4). This allowed us to determine whether the exosuit's effects on poststroke walking depend on the timing of ankle PF assistance (*50*). Each day tested the effects of a different onset timing (randomized order) to minimize the influence of participant fatigue and after-effects on this evaluation. To minimize the influence of participant day-to-day variability, each testing day measured the effects of the exosuit relative to a daily baseline, with within-session changes from each day compared for each individual.

The two PF force delivery onset timings evaluated occurred during the paretic midstance (early-onset) or terminal stance (late-onset). The same DF assistance profile was used on both testing days. Each testing day included two 8-min testing bouts on a treadmill set to a speed comparable to each participant's usual overground walking speed (Table 1), with as much rest as needed provided between bouts. The first trial consisted of walking with the exosuit unpowered, and the other trial with the exosuit powered. Participants were allowed to hold a side-mounted handrail during treadmill testing if it was necessary for safety (Table 1). All participants who used a handrail used one during both bouts of testing and were instructed to use the smallest amount of handrail support needed. Data from the 8th minute of walking from each condition were used for all analyses.

Participants and inclusion/exclusion criteria

Participants (age, 49 ± 4 years; time since stroke, 4.38 ± 1.37 years; female, 44%; left hemiparetic, 56%; see Table 1) were recruited from rehabilitation clinics in the greater Boston area. Participant inclusion criteria included the following: age between 25 and 75 years, at least

6 months after stroke, able to walk independently for 6 min without stopping, sufficient passive ankle range of motion with the knee extended to reach a neutral position, and pass cognitive screening. Cognitive screening differed for those with versus without aphasia. Those without aphasia had to score ≥23 on the Mini Mental State Examination (MMSE). Those with aphasia had to score ≥19 on the MMSE, as well as ≥35 on the Auditory Verbal Comprehension section of the Western Aphasia Battery (WAB) and ≥10 on the Sequential Commands section of the WAB. Exclusion criteria included Botox injections within the past 6 months, substantial knee recurvatum during walking, serious comorbidities (including musculoskeletal, cardiac, and neuromuscular, skin, and vascular conditions), inability to communicate and/or be understood by investigators, a resting heart rate outside the range of 50 to 100 beats per minute or blood pressure outside the range of 90/60 to 200/110 mmHg, pain in the extremities or spine that limits walking, and experiencing more than two falls in the past month. Medical clearance and signed informed consent forms approved by the Harvard University Human Subjects Review Board were obtained for all participants.

Soft exosuit design and operation

In brief, the exosuit has a unilateral design and is worn on the paretic limb (Fig. 1). The exosuit is composed of garment-like, functional textiles that securely anchor to the body at the waist and paretic calf and interact with a low-profile shoe insole to generate assistive ankle torques through cable-based mechanical power transmission (Fig. 1 and Table 2). The exosuit consists of two separate textile modules (Fig. 1): the first is a monoarticular module that anchors at the paretic calf and is designed to generate ankle DF torque and the second is a multiarticular module designed to generate ankle PF torque. This PF module anchors distally at the paretic calf separately from the DF module. Its proximal anchor is at the waist. Straps attach to the waist and calf anchors and extend anteriorly over the proximal thigh, straddling the knee joint center. Lightweight laminates integrated into the modules provide reinforcement, create force transmission paths, and distribute pressure. Nonslip, breathable liners are also integrated into the modules to reduce their migration relative to the body. A total of two Bowden cables are integrated into the exosuit, one for each textile module. The proximal and distal attachments of these Bowden cables are the shank and foot, respectively. The DF module's Bowden cable is located anterior to the ankle joint, thus producing an ankle DF torque when retracted. The PF module's Bowden cable is located posterior to the ankle joint, thus producing an ankle PF torque when retracted. When retracted by an actuation unit, these Bowden cables transmit mechanical power to the wearer, producing the assistive joint torques depicted in Fig. 1. The overall mass of the exosuit's functional textile anchors, cables, and sensors is about 0.90 kg (waist textile anchor, 0.15 to 0.20 kg; thigh connecting straps, 0.20 to 0.22 kg; calf textile anchor and integrated sensors, 0.30 to 0.35 kg; cables and sheath, 0.12 to 0.15 kg).

A tethered, laboratory-based electromechanical actuation platform was created to enable evaluation of the exosuit technology (Fig. 2). Previous work from our laboratory describes in greater detail the controller and actuation platform used in both healthy (*38*) and poststroke (*39*) participants. Briefly, the actuation unit contains four linear actuators that allow simultaneous actuation of up to four Bowden cables; only two actuators were used in this study. Each actuator is equipped with a linear potentiometer (P3 America Inc.) that enables closed-loop control of the cable position and can deliver up to 300 N of force to the wearer. The closed-loop controller used position measurements from the linear potentiometers together with force measurements from load cells (Futek) integrated into the exosuit textiles, and foot rotational velocity measurements from a gyroscope (SparkFun) mounted to each shoe to iteratively adapt Bowden cable position trajectories and generate desired PF and DF assistive force profiles. The gyroscopes enabled real-time gait segmentation, and the linear potentiometers and load cells enabled iterative, force-based, position control. Together, these sensors enable exosuits to deliver appropriately timed assistive forces of adequate magnitude. The "Exosuit-generated assistive forces" section of the Supplementary Materials provides additional detail regarding the exosuit's controller.

Clinical evaluations

Treadmill speed was set based on participants' usual overground walking speed. This speed was measured using the 10-m walk test (10MWT). The 10MWT was also used to quantify participants' walking disability

Table 2. Participant measurements used for exosuit fitting. IC, iliac crest; W, wide; R, right; L, left.

Participant	Height (cm)	Weight (kg)	Paretic limb anthropometric measurements (cm)					Shoe size (U.S.)	Lateral support
			IC to ground	Hip circumference (at IC)	Thigh circumference	Below knee circumference	Calf circumference		
01	162	49	40.6	32.5	20.0	13.5	15.2	7W	Y
02	177	73	43.0	35.3	19.0	14.0	14.8	R-10.5 | L-11	N
03	158	90	36.3	48.0	22.0	14.5	15.5	9.5W	N
04	173	79	44.5	39.5	17.0	12.5	13.0	12W	N
05	172	73	—	—	—	—	—	9	Y
06	186	80	47.0	38.5	19.0	14.0	14.3	11W	N
07	167	60	40.5	38.3	18.3	14.0	14.3	9.5W	N
08	181	99	39.8	41.3	17.7	15.4	15.4	R-11.5 | L-11	N
09	182	97	47.0	40.0	23.5	16.0	17.0	R- 12.5W | L- 13W	N

(*73*) and evaluate their ability to safely walk without their regular orthoses, which could not be used with the exosuit because of their restriction of ankle PF. Participants who required the frontal plane ankle support provided by an ankle orthosis for safe ambulation were provided an optional exosuit module that passively controlled ankle inversion. This passive module was used by two participants (Table 1) during all testing. Participants were allowed to use their regular assistive device (cane) during the 10MWT if they typically used one for safety. A handrail was provided during treadmill testing for participants requiring one to safely walk on the treadmill (Table 1).

Motion analysis and metabolic data acquisition and analysis

Participants walked on an instrumented split-belt treadmill (Bertec) to measure GRFs independently from each limb. There was no barrier between the two treadmill belts, the small gap between the belts was not perceivable during walking, and participants were instructed to walk naturally. GRFs were used to compute PP and non-PP and this study's primary kinetic outcome: interlimb propulsion symmetry. PP was defined as the anterior GRF measured during the paretic limb's stance phase, normalized by body weight (% bw) (*65*, *79*, *86*, *89*–*91*). Interlimb propulsion symmetry was calculated as follows:

$$\text{Propulsion symmetry} = \text{PP}/\text{non-PP}$$

where non-PP represents the nonparetic limb's propulsive force during its stance phase. For this metric, 1 represents perfect symmetry. Using this metric, the magnitude of propulsion asymmetry was calculated as follows:

$$\begin{aligned}\text{Propulsion asymmetry} &= (\text{Non-PP} - \text{PP})/\text{Non-PP} \\ &= 1 - \text{Propulsion symmetry}\end{aligned}$$

Three-dimensional motion capture (VICON, Oxford Metrics) was used to measure participants' ankle motion during walking, with the peak DF angle during the paretic swing phase serving as this study's primary kinematic outcome. All marker and force data were filtered using a zero-lag, fourth-order, low-pass, Butterworth filter with a 5- to 9-Hz optimal cutoff frequency that was selected using a residual analysis algorithm (MATLAB, MathWorks Inc.) (*92*). Kinetic and kinematic analyses were performed using Visual 3D (C-Motion). The last 15 good strides from each testing condition were extracted for data analysis, with good strides defined as the absence of crossed force plate strikes. As depicted in Fig. 2, indirect calorimetry (K4b2, Cosmed) was concurrently used to measure participants' energy consumption (ml O_2/min) at rest (quiet standing) and during walking, with resting energy consumption subtracted from energy consumption during walking to yield participants' net energy consumption during walking. Participants' net energy cost of walking (EC_{net}) was computed by normalizing net energy consumption by body weight (kg) and walking speed (m/min). Similar to (*54*), participants' deviation from the normal energy cost of walking was computed as follows:

$$EC_{net} - N$$

where N represents the normal energy cost of walking (0.151 ml O_2/kg/m).

Statistical analyses

Interparticipant means and SEs are reported unless otherwise indicated. Peak PP, interlimb propulsion symmetry, peak paretic limb swing phase ankle DF angle, and mass- and speed-normalized energy consumption served as dependent variables because of their popularity and importance in the poststroke gait rehabilitation literature (*2*, *22*, *25*, *49*, *50*, *55*, *57*, *63*, *65*, *86*, *93*, *94*). All analyses were directed toward evaluating the exosuit's influence on poststroke gait. Given previous work positing that the optimal timing of PF force delivery may vary across participants because of poststroke heterogeneity, paired t tests on individual participant data were used to determine each participant's more effective PF force delivery onset timing in terms of the exosuit's influence on propulsion symmetry (see the "PF assistance timing" section). Paired two-tailed t tests subsequently evaluated group-level effects, with PF assistance onset timings individualized. For these analyses, $\alpha = 0.05$ and a Holm-Šídák correction was applied to adjust for multiple comparisons. To evaluate our hypothesis that improvements in propulsion symmetry would contribute to reductions in the energy cost of walking, linear regression measured the relationship between exosuit-induced changes in these two variables. To inform clinical translation, linear regression was also used to determine whether participants' baseline function (walking speed) influenced the effects of the exosuit. To evaluate differences in delivered PF force on the days testing an early-onset and late-onset of PF force delivery, we compared the magnitude of delivered force at each point in the gait cycle using paired t tests. As has previously been done (*95*), to minimize type 1 error, significance was concluded only if 40 consecutive pairs (4% of the gait cycle) indicated significance at $\alpha = 0.05$.

SUPPLEMENTARY MATERIALS

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Materials and Methods
Fig. S1. Effects of wearing a passive exosuit on poststroke propulsion and energy expenditure.
Table S1. Tethered exosuit ankle PF assistive forces.
Table S2. Additional individual subject-level data.
Video S1. Video demonstration of exosuit-assisted treadmill walking.

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A soft robotic exosuit improves walking in patients after stroke

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A softer recovery after stroke

Passive assistance devices such as canes and braces are often used by people after stroke, but mobility remains limited for some patients. Awad *et al*. studied the effects of active assistance (delivery of supportive force) during walking in nine patients in the chronic phase of stroke recovery. A soft robotic exosuit worn on the partially paralyzed lower limb reduced interlimb propulsion asymmetry, increased ankle dorsiflexion, and reduced the energy required to walk when powered on during treadmill and overground walking tests. The exosuit could be adjusted to deliver supportive force during the early or late phase of the gait cycle depending on the patient's needs. Although long-term therapeutic studies are necessary, the immediate improvement in walking performance observed using the powered exosuit makes this a promising approach for neurorehabilitation.

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