Slow walking limits the benefits of knee-ankle exoskeleton assistance for people with and without stroke

Russell M. Martin[1*], Rohan S. Punamiya[1], Thijs Tankink[2], Madison S. Lee[1], Maarten G. Lansberg[3], Steven H. Collins[1]

[1]Department of Mechanical Engineering, Stanford University, Stanford, CA, USA.
[2]Department of Human Movement Sciences, University Medical Center Groningen, University of Groningen, Groningen, The Netherlands.
[3]Department of Neurology and Neurological Sciences, Stanford University Medical Center, Stanford, CA, USA.

*Corresponding author. E-mail: rumartin@stanford.edu

Abstract

Background: Stroke often causes persistent gait impairments, reducing walking efficiency. Exoskeletons are a promising technology for improving the energy economy of walking, particularly when multiple joints are assisted, when the assistance is personalized, and when sufficient adaptation time is given. This study assessed how extended exposure and personalized knee-ankle assistance affect walking biomechanics at slower speeds in people with and without chronic stroke.

Methods: Six participants with chronic stroke and six healthy controls underwent a unilateral knee-ankle exoskeleton emulator assistance optimization protocol, walking for 180 cumulative minutes with assistance. Participants with stroke walked at their comfortable speed, while controls walked at the stroke group's mean speed. The ankle torque profile was personalized via human-in-the-loop optimization, while the knee trajectory was set based on walking speed. After optimization, validation trials compared no assistance, generic ankle, generic knee-ankle, and optimized knee-ankle assistance.

Results: Participants with stroke had strong overground walking ability, with an average 10 m walk test speed of 0.90 m/s, but selected a slower average treadmill speed of 0.54 m/s. Optimized knee-ankle assistance reduced metabolic cost by 6.4% in stroke participants (p=0.046) and 4% in controls (p=0.13). The metabolic cost reduction from generic assistance was slightly smaller in both groups. The difference in metabolic rate between stroke and control groups across conditions was not significant. Both groups also showed similar optimized ankle torque profiles, with stroke participants optimizing to an average peak torque of 0.39 Nm/kg. Re-testing at the end of the study showed the comfortable treadmill speed for participants with stroke increased to 0.87 m/s, suggesting a preference for faster walking developed with increased exposure.

Conclusions: The energy economy improvement in participants with stroke was modest despite personalizing ankle assistance, incorporating knee assistance, and offering more ankle torque and adaptation time than in prior stroke exoskeleton studies. Control participants walking at matched speed showed similarly small effects. These findings indicate that stroke-related impairments may constrain exoskeleton benefit primarily through a reduction in preferred walking speed.

Keywords: exoskeleton, stroke, hemiparesis, human-in-the-loop optimization, knee-ankle assistance, gait rehabilitation, walking energetics, assistive robotics

1 Introduction

Stroke-related gait impairments cause persistent mobility limitations that reduce function and quality of life. Stroke is the leading cause of disability in the US [1]; about 80% of stroke survivors experience persistent gait impairments beyond three months post-stroke [2]. These impairments compromise community ambulation and reduce quality of life, limit return to work, and increase long-term care costs [3–5]. People with chronic stroke typically have half the lower-limb strength of healthy individuals [6], leading to double the energy cost of walking [7] and half the self-selected walking speed [8]. Consistent with these limitations, gait improvement is the most commonly reported rehabilitation goal among people who have had a stroke [9, 10]. Improving mobility after stroke has downstream effects: interventions that improve walking speed [11] and endurance [3, 12] also improve overall function and quality of life in this population.

Exoskeletons are an emerging technology shown to improve the energy economy of walking in young healthy adults. Laboratory-based emulator systems combined with human-in-the-loop assistance personalization [13] provide unique insight into the maximum potential utility of exoskeletons. Studies using these systems have demonstrated that, relative to no assistance, optimized bilateral ankle assistance can reduce the energy cost of walking by 39% [14], while optimized bilateral hip-knee-ankle assistance can reduce energy cost by 50% [15]. These personalization approaches have also been extended beyond young healthy adults: an emulator study of older adults (average age 72 years) found that optimized assistance improved walking speed by 8% and reduced metabolic cost by 19% [16]. Insights from emulator studies have also informed real-world devices, including a portable ankle exoskeleton that increased walking speed by 9% and reduced cost of transport by 17% compared to walking in normal shoes [17].

Exoskeletons for augmenting the mobility of people with chronic stroke are nascent, but show potential. Soft ankle exoskeletons that provide plantarflexion and dorsiflexion assistance have demonstrated a range of biomechanical and functional benefits, including improved propulsion, better ground clearance, increased walking speed, and up to a 10% reduction in metabolic cost [18, 19]. These devices may be particularly beneficial for individuals with slower walking speeds after stroke, who have experienced greater percent gains in propulsion and walking speed with assistance [20]. Trailing limb angle has emerged as a proxy for exoskeleton benefit, as demonstrated using a myoelectric-controlled ankle exoskeleton across a range of walking speeds [21]. To elicit larger changes in trailing limb angle, vibrotactile-audio feedback has been used as a training tool, leading to increased walking speed with assistance [22]. A variety of control and assistance strategies have also been explored: a pilot study of unilateral ankle assistance with a low-torque impedance and pushoff controller reported more normative ankle kinematics during swing and heel strike, as well as improved knee kinematics in stance [23], while pilot testing of a unilateral knee exoskeleton found improved knee flexion during both stance and swing phases [24].

Several questions remain regarding how best to design, personalize, and evaluate exoskeleton assistance for people with stroke. First, the effect of systematic assistance personalization is under-explored, as most studies rely on hand-tuning or one-size-fits-all approaches. In healthy participants, personalization has been shown to substantially improve the exoskeleton metabolic rate reduction in able-bodied users [14, 17]; this benefit could be even greater for stroke participants given the heterogeneity of stroke impairments [25]. Second, many stroke exoskeletons are limited in the ankle torque they can apply - the stroke exoskeleton studies mentioned in the previous paragraph ranged from 0.14 Nm/kg [23] to 0.29 Nm/kg [21]. However, other studies of walking and running in healthy participants have shown a direct relationship between ankle torque and metabolic rate reduction [13, 16, 26], suggesting it may be useful to establish whether this relationship holds in people with stroke. Third, most stroke exoskeleton studies have assisted only at the ankle, and the effect of simultaneously assisting the ankle and knee is not known. However, prior work with healthy participants has shown increases in activation of the muscles that span the knee during ankle torque application [15, 27], possibly to stiffen the knee so that device plantarflexion torque propels the user forward instead of unnecessarily extending the knee. Furthermore, knee assistance for people with stroke has the potential to improve foot clearance in swing and thereby reduce hip circumduction. Fourth, adaptation time in stroke exoskeleton studies has been mostly limited to a few tens of minutes, despite evidence that training contributes to one-half of the overall exoskeleton metabolic rate reduction, and that healthy users require almost 2 hours of walking to fully adapt to ankle assistance [14]. Thus, a protocol that

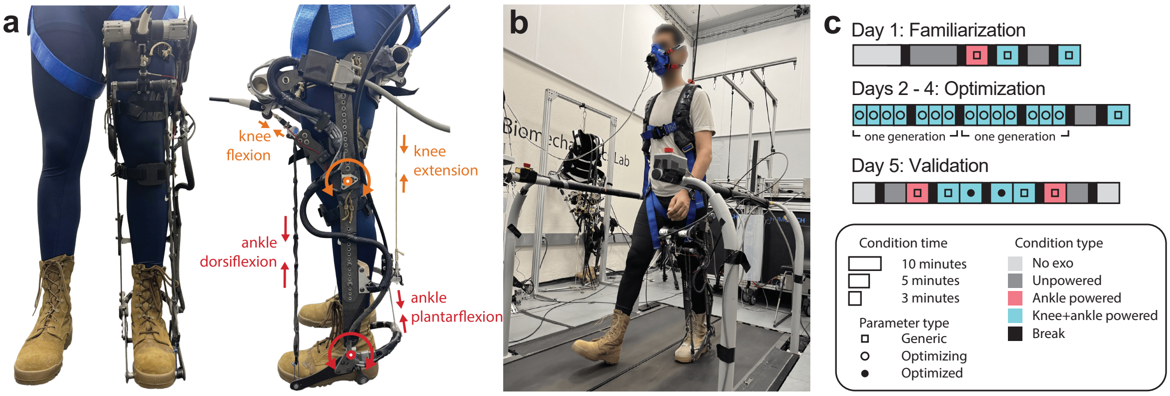

Fig. 1 Experimental overview. (a) Exoskeleton emulator front and side, directions of joint assistance denoted by arrows, indicating how shortening of a degree-of-freedom's rope would cause joint motion. (b) Experimental setup of a participant with chronic stroke walking in the exoskeleton emulator on an instrumented treadmill wearing a metabolics mask. The offboard motors can be seen directly behind the treadmill. (c) Protocol completed over the five-day series of visits by the participants with stroke. The control participants completed the same protocol, but did the conditions on Day 1–3 on their first visit day, and Day 4–5 on their second visit day.

emphasizes extended exposure could clarify the role of adaptation in this population. Finally, the effectiveness of exoskeleton assistance in stroke versus healthy individuals under matched conditions remains unclear; understanding this difference could inform the translation of exoskeleton design and control strategies from able-bodied to neurologically impaired users. Conducting these matched tests could also demonstrate the best-case scenario for a given exoskeleton scenario, since young, healthy participants may be able to garner a greater metabolic rate reduction from optimized assistance [16].

The aim of this study was to evaluate how personalized knee-ankle exoskeleton assistance affects walking energetics and biomechanics during extended exposure in individuals with and without chronic stroke. To accomplish these aims, we conducted human-in-the-loop optimization of exoskeleton assistance using a unilateral knee-ankle exoskeleton emulator with six people with chronic stroke and six speed-matched healthy controls. We then compared the effect of optimized assistance to generic assistance and knee-ankle assistance to ankle-only assistance. We expected that the optimizer would prefer larger ankle torque, and that optimized assistance would more beneficial than generic assistance. We also anticipated that the energy economy benefit of ankle assistance would be improved with the addition of knee assistance. Lastly, we predicted that the metabolic rate reduction from exoskeleton assistance would be larger in healthy participants, who could more easily adapt their gait strategy to maximize the benefit from the exoskeleton.

2 Methods

We recruited six participants with chronic stroke and six healthy control participants for this study. Participants wore a unilateral exoskeleton emulator (Fig. 1a, b) which provided a commanded torque at the ankle and a commanded position at the knee. The exoskeleton was worn on the affected side by participants with chronic stroke, and a randomly-assigned side for control participants. The protocol (Fig. 1c) included pre-study mobility tests, treadmill and exoskeleton familiarization, six generations of human-in-the-loop ankle assistance optimization, and a separate validation test comparing different assistance configurations. Each participant in this study walked in the powered device for at least 180 minutes, plus an additional 50 minutes on the treadmill walking either in the unpowered device or walking in their own shoes. Both groups completed the same protocol, which required two visits for controls and five for participants with chronic stroke to account for fatigue and recovery. Data collected included the metabolic cost of walking, exoskeleton-based kinematic and kinetic measurements, and ground reaction forces from the instrumented treadmill. All experimental procedures were approved by the Stanford University Institutional Review Board (IRB-44149) and all participants provided written informed consent prior to participation. A video of a participant with stroke walking in the different assisted and unassisted conditions is included in the Appendix.

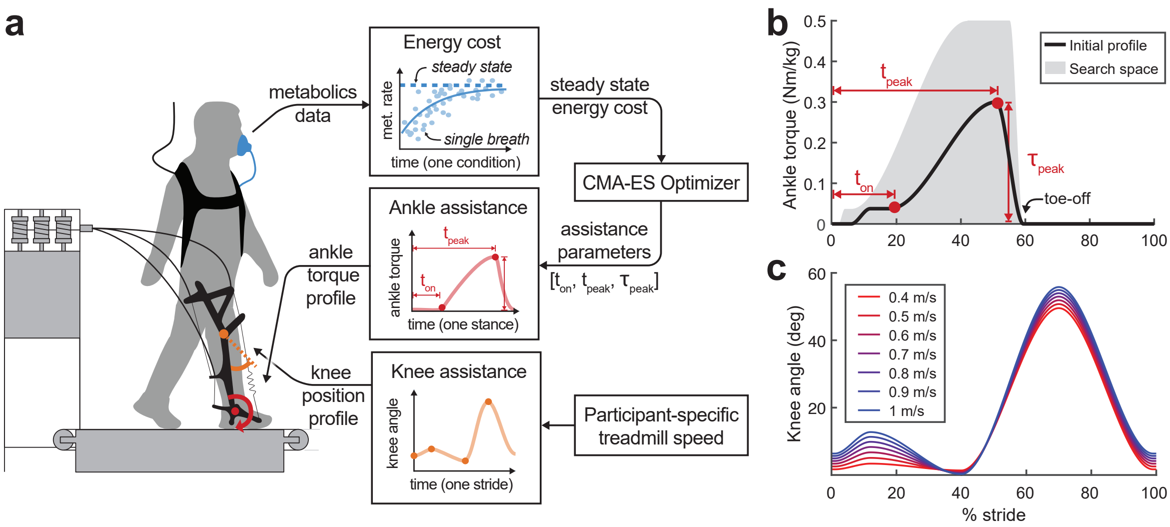

Fig. 2 Optimization and assistance overview. (a) The three parameters defining the ankle assistance profile were optimized by the CMA-ES optimization algorithm with a goal of minimizing the measured steady-state metabolic cost of walking. The knee position profile was not optimized, instead, it was fixed based on the participant's comfortable walking speed. (b) Ankle torque profile. The ankle torque parameterization was defined by three nodes: on time t_{on}, peak time t_{peak}, and peak torque τ_{peak}. The x-values of these nodes were defined as a function of percent stance (e.g. heel strike is 0% and toe-off is 100%). These nodes were splined together to produce the per-step torque profile. The allowable space searched for the ankle assistance, representing the possible splines that could be produced by the optimizer, is shown in gray; the optimizer was initialized to the spline shown in black. (c) Knee position profile. The knee profile was dependent on walking velocity, based on prior work relating walking speed and knee kinematics [28].

2.1 Participants

Twelve participants participated in the study: six with chronic stroke (6 male, height 177±4 cm, weight 88±26 kg, age 48±13 years, chronicity 4.7±3.9 years) and six healthy controls (6 male, height 179±8 cm, weight 71±8 kg, age 26±3 years) (Table 1). Inclusion criteria for stroke participants were deliberately broad to aid recruitment, requiring a single stroke that occurred more than three months prior, the ability to walk comfortably without a cane for at least ten minutes, and a minimum height of 162 cm. Control participants had no history of neurological conditions. Groups did not differ significantly in height (p=0.66) or weight (p=0.14), but age was significantly different (p=0.002).

2.2 Exoskeleton Emulator

The unilateral exoskeleton emulator used in this study assisted at the knee and ankle. This device is adapted from the bilateral hip-knee-ankle emulator described previously [29]. We commanded a position trajectory at the knee and a torque profile at the ankle. A complete description of the knee and ankle control strategies are given in Nguyen [30] and Bryan et al. [29].

The exoskeleton used powerful offboard motors that transmitted assistive torques to the lower limb through vectran ropes routed inside Bowden-cable sheaths to the wearable end effector mounted at the knee and ankle. The wearable frame was adjustable so that the device's knee and ankle axes coincided with the participant's anatomical joints, enabling torques to be applied in the sagittal plane to assist knee flexion/extension and ankle plantarflexion. Ankle dorsiflexion torque was provided passively by a chain of elastic bands, the total length of which was adjusted during device fitting so that users maintained toe clearance in swing.

Sensing was integrated at the exoskeleton end effector to enable measurement of joint kinematics and applied forces [30]. Encoders at the knee and ankle joints measured joint position; inline load cells attached to the exoskeleton measured the applied knee flexion and extension torque. At the ankle, a strain gauge measured the forces transmitted during plantarflexion assistance. Gait events, including heel strike and toe-off, were detected using an instrumented treadmill (Bertec).

Knee assistance (Fig. 2c) was delivered using a joint position trajectory as a function of stride time to promote reasonable joint kinematics and to limit gait maladaptation in response to ankle assistance. The desired knee angle trajectory was generated by spline interpolation through four nodes, which were placed at heel strike, stance flexion, stance extension, and swing flexion. The timing of these four nodes were set to approximately align with typical patterns of walking knee kinematics (0%, 12%, 40% and 70% of stride, respectively). The knee node's angles were selected as a function of walking speed [28].

Ankle assistance (Fig. 2b) was delivered using a stride-time based torque controller similar to that described in [29], but modified so that assistance was prescribed as a function of stance time rather than of stride time. The desired ankle torque profile was defined by three parameters: the onset time (t_{on}), the peak time (t_{peak}), and the peak torque magnitude (τ_{peak}). The torque curve was generated by spline interpolation through four nodes: heel strike (0, 0), onset (t_{on}, 0.1), peak (t_{peak}, τ_{peak}), and toe-off (98%, 0). Low-level control used proportional feedback control plus a feedforward term based on previous strides' error [31].

While walking in the exoskeleton, all participants wore a harness for fall protection, but this harness did not provide any bodyweight support. After the familiarization day, participants were strongly discouraged from using the treadmill handrails when walking at steady state.

2.3 Protocol

2.3.1 Familiarization

Before beginning tests with the exoskeleton system, participants first completed a set of clinical and functional assessments to quantify their mobility levels. These assessments included the Mini-Mental State Examination [32], the Timed Up and Go test [33], and the 10-meter walk test [34]. The 10-meter walk test was conducted twice at the participant's comfortable speed and twice at the participant's fast self-selected speed; the walking velocity reported was the average of the two tests at each speed. During overground assessments, participants were permitted to wear their prescribed ankle-foot orthosis (AFO) if desired; however, AFOs were not worn while walking with exoskeleton assistance since the exoskeleton provided rigid support in the inversion-eversion direction at the ankle.

Participants then completed a treadmill ramp test to determine a comfortable walking speed that could be sustained for up to ten minutes during the exoskeleton optimization protocol. The ramp test began by starting the treadmill at 0.30 m/s then slowly increasing the speed of the treadmill until the participant felt they had reached their maximum comfortable speed. The ramp test was conducted

Table 1 Participant demographics. Each row represents a participant, participants beginning with S and C represent members of the stroke and control group. The *side* column indicates whether the exoskeleton was worn on the left or right side; for participants with stroke, this indicates the side affected by hemiparesis. I and H in the *stroke type* column indicate ischemic and hemorrhagic stroke. TUG is Timed Up and Go. A star (*) indicates a significant ($p \leq 0.05$) difference between the stroke and control cohorts.

	Height (cm)	Weight (kg)	Age	Side	Chronicity (years)	Stroke Type	Mini Mental (out of 30)	TUG (s)	Overground comfortable speed (m/s)	Overground fast speed (m/s)	Treadmill preferred speed (m/s)
S1	175	72	54	R	11	H	29	21.2	0.62	1.10	0.40
S2	183	84	50	R	5	I	22	25.5	0.77	0.89	0.60
S3	172	59	23	R	3	I	30	13.1	1.14	1.30	0.65
S4	178	80	48	L	0.3	I	30	13.8	0.78	1.08	0.50
S5	178	104	62	L	7	I	30	13.3	1.00	1.11	0.50
S6	175	131	53	L	2	I	29	10.9	1.07	1.68	0.60
Avg	*176.8*	*88.3*	*48.3*		*4.7*		*28.3*	*16.3*	*0.90*	1.19	0.54
SD	*3.8*	*25.6*	*13.3*		*3.9*		*3.1*	*5.7*	*0.20*	0.27	0.09
C1	175	63	31	R							
C2	170	60	22	R							
C3	177	77	28	R							
C4	183	82	24	L							
C5	193	73	23	L							
C6	173	68	25	L							
Avg	*178.5*	*70.5*	*25.5**								
SD	*8.3*	*8.4*	*3.4*								

first without the exoskeleton in normal shoes and the participant's AFO if desired. Next, the ramp test was repeated while wearing the exoskeleton in zero torque mode, when the device was worn but not applying force at any of the joints. During the zero torque exoskeleton ramp test, we encouraged the participants to try to walk at the same treadmill velocity as they had when they did the ramp test without wearing the device; all participants successfully walked at this speed.

For the remainder of the study, we fixed the treadmill speed to participants' day one comfortable speed to limit the amount of changing variables during optimization, with the goal of helping the participants and the optimizer adapt more quickly. This decision was made out of concern that learning to walk with multijoint assistance while also learning to walk on a self-paced treadmill would be overwhelming, and also slow optimization, both of which would mean more visit days for the participants with stroke and decreased odds that the optimizer would converge by the end of the study.

Once a comfortable treadmill speed was set, we introduced assistance. Participants first walked with ankle-only assistance for about five minutes, gradually increasing ankle torque to match the initial profile (Fig. 2b). After a break, they walked in generic knee-ankle assistance for at least five minutes using the velocity-dependent knee position profile. The knee profile was adjusted as needed for participant comfort or to ensure proper foot clearance.

The final two trials on the familiarization day consisted of three minutes of generic knee-ankle assistance followed by three minutes of zero torque. These conditions were repeated on each subsequent visit to quantify learning effects, allowing changes in metabolic cost and biomechanics to be attributed to participant adaptation rather than changes in assistance profiles.

Participants with chronic stroke and control participants completed the familiarization protocol during their first laboratory visit (Fig. 1c). Controls were not assessed by the mobility tests (Mini-Mental State Examination, Timed Up and Go, and 10-meter walk test) or treadmill ramp tests, since they walked at the average speed of the stroke group.

2.3.2 Optimization

Human-in-the-loop optimization [13] was used to personalize ankle assistance by tuning the ankle assistance parameters with a goal of minimizing the metabolic cost. The ankle profile's timing parameters were initialized to t_{on}=30% stance and t_{peak}=85% stance based on prior unilateral ankle exoskeleton studies in healthy participants walking at 1.25 m/s [13]. The peak torque parameter was initialized to 0.30 Nm/kg, chosen based on the peak ankle torque applied in prior stroke exoskeleton studies [21]. We used covariance matrix algorithm-evolutionary strategy (CMA-ES) as the optimization algorithm [35] with an initial search space size of σ=0.3. Parameter search bounds (minimum, maximum) were set to (10, 65) percent stance for t_{on}, (70, 92) percent stance for t_{peak}, and (0, 0.5) Nm/kg for τ_{peak}. Seven conditions were evaluated per generation; the final condition of the generation was the mean from that generation. Each condition was three minutes of walking, during which the participant's steady-state metabolic cost of walking was estimated by fitting a first-order model to the breath-by-breath data [13]. We extended the evaluation time for each condition from two minutes [13, 14] to three minutes. This was done because the slower walking speeds likely result in a shallower metabolic cost landscape, and the longer duration reduces measurement noise. This allows the optimizer to more reliably differentiate the similar-costing candidate assistance profiles [36].

Stroke participants' optimization days (visits 2, 3, and 4; Fig 1c) included two generations of optimization (14 conditions total) followed by the two-trial learning test, identical to the familiarization day. 5-minute breaks were taken mid-generation and after each generation. Control participants completed four generations on their first visit and the final two on their second visit, with rest breaks after each generation. Control participants completed the two-condition learning test after the second and fourth generation.

2.3.3 Validation

After optimizing ankle assistance, we performed a final evaluation using validation tests. Participants completed five-minute trials for five conditions: normal shoes (with AFO if desired), exoskeleton in zero torque mode, generic ankle-only assistance, generic knee-ankle assistance, and optimized knee-ankle assistance. Each of these conditions was repeated twice in counterbalanced order (ABCCBA), then final results for each condition were found by averaging over each pair of trials. At the end of the

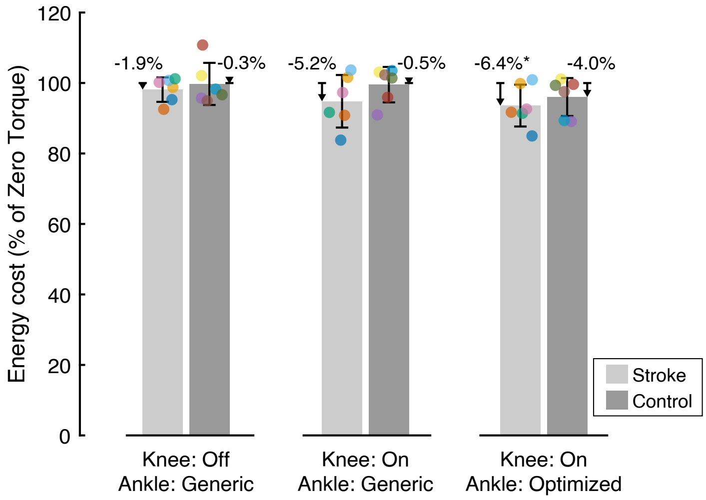

Fig. 3 Metabolic cost values for the powered exoskeleton conditions on the validation day. Values are given as a percentage of walking with the exoskeleton turned off; smaller values indicate improved energy economy. Significance ($p < 0.05$) with respect to zero torque condition is denoted by a star. There were no significant intra-condition differences (i.e. light gray vs. dark gray bar). Marker colors represent different participants, these markers are randomly scattered horizontally for readability.

validation day, we also repeated the walking speed ramp test once each in normal shoes, zero torque mode, and generic knee-ankle mode.

2.4 Data Collection and Analysis

Metabolic energy expenditure was measured breath-by-breath using a portable indirect calorimetry system (Quark CPET, COSMED). To reduce the thermal effect of feeding on metabolic measurements, participants were instructed to fast for three hours prior to the start of a data collection session. Ground reaction forces were recorded using an instrumented treadmill and low-pass filtered at 25 Hz with a third-order Butterworth filter. Exoskeleton knee and ankle joint angles were obtained from onboard encoders and filtered at 20 Hz using a third-order Butterworth filter. Knee flexion and extension torques were measured using inline load cells mounted in series between the exoskeleton and the rope transmission and filtered at 60 Hz. Ankle plantarflexion torque was measured using a strain gauge mounted on the heel spur connected to the rope transmission, also filtered at 60 Hz with a third-order Butterworth filter.

Metabolic cost was calculated on a breath-by-breath basis using established relationships between oxygen consumption (VO2), carbon dioxide production (VCO2), and metabolic power [37]. Steady-state metabolic cost for the 3-minute optimization conditions was estimated by fitting a first-order exponential model to the metabolic power data with a fixed time constant of 42 seconds, as described previously [13]. For the 5-minute validation conditions, steady-state cost was calculated by averaging metabolic measurements from the final 2.5 minutes of each trial. The metabolic cost of walking was computed by subtracting the cost of quiet standing from each walking condition. The cost values in the symmetric counterbalanced validation trials were obtained by averaging across the two trials. In this study, we report the metabolic cost of walking as a percentage of the cost of walking in the exoskeleton in zero torque mode to (1) isolate the effect of assistance independent of device-specific factors such as mass and emulator design and (2) allow for comparison between participants with chronic stroke and participants in the control group.

Kinematic, kinetic, and ground reaction force profiles were computed using the final 60 s of data from each trial in the double-reversal. Knee stance excursion was calculated as the maximum minus the minimum knee angle from 0–50% stride. Exoskeleton mechanical power was calculated as the product of the measured joint torque and the corresponding joint angular velocity. Total anteriorly-directed ground reaction force, used to quantify propulsive forces during push-off, was found by trapezoidal integration from 40–100% stance for the exoskeleton-worn leg. Individual strides were segmented, time-normalized from 0–100% of the stride at 0.1% increments, and then averaged across all strides and both trials to generate mean kinematic and kinetic profiles.

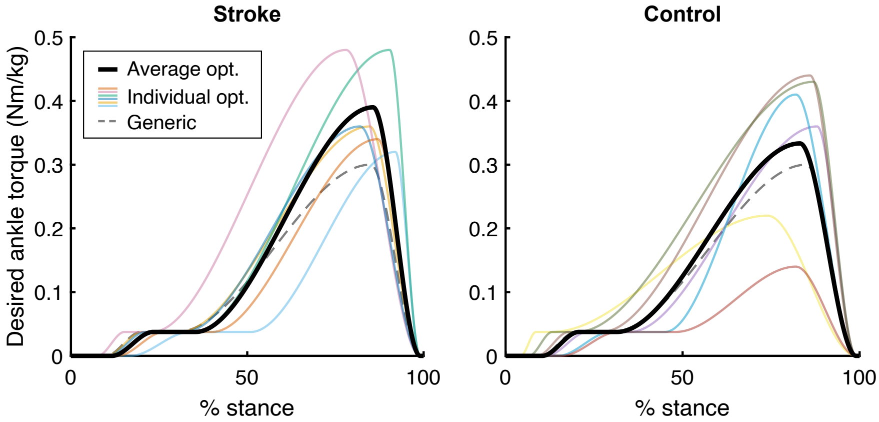

Fig. 4 Optimized torque profiles for participants with chronic stroke (left) and healthy controls (right). The average profile, found by creating a spline using the average of the participant's optimized parameters, is shown in black. Colored lines represent individual participants' optimized profiles. The dashed gray line represents the generic assistance, which was used to initialize optimization and was the same for all participants.

2.5 Statistical analysis

Paired t-tests (α=0.05) were used for within-subject comparisons between assistance conditions, and unpaired t-tests (α=0.05) were used for between-group comparisons within the same condition. To test whether the effect of exoskeleton assistance differed between participants with and without stroke, a linear mixed-effects model was fit with group (stroke vs. control) and condition as fixed effects and participant as a random intercept.

3 Results

3.1 Energy Cost

Optimized knee-ankle assistance reduced metabolic cost by 6.4% relative to zero torque walking in participants with stroke (p=0.046) and by 4.0% in control participants (p=0.13) (Fig. 3). Generic knee-ankle assistance yielded smaller nonsignificant reductions, decreasing metabolic cost by 5.2% in participants with stroke (p=0.15) and by 0.5% in control participants (p=0.83). With generic ankle assistance and the knee assistance off, metabolic cost was reduced by 1.9% in participants with stroke (p=0.24) and by 0.3% in control participants (p=0.92). Walking in normal shoes reduced metabolic cost by 17.0% (p=0.004) and 11.5% (p=0.01) relative to unpowered exoskeleton walking for participants with stroke and control participants, respectively. A linear mixed-effects model revealed no significant difference in the metabolic effect of assistance between stroke and control participants across conditions (F(1,32)=1.50, p=0.23).

3.2 Optimization

Human-in-the-loop optimization converged to similar ankle assistance profiles for participants with and without stroke (Fig. 4). For participants with stroke, optimized peak ankle torque was an average of 0.39±0.07 Nm/kg (mean±SD, range: 0.32–0.48); control participants optimized to a slightly lower average peak torque of 0.33±0.12 Nm/kg (range: 0.14–0.44 Nm/kg). The timing of assistance was generally similar across groups, preferring late peak times and moderate rise times. Peak torque occurred at an average of 86±5% of stance (range: 78–92%) for participants with stroke and 83±5% of stance (range: 74–88%) for control participants. The value for optimized torque onset time was 35±10% of stance on average (range: 23–51%) for participants with stroke and 31±14% of stance (range: 13–48%) for control participants.

3.3 Speed

Walking speed pre-tests of overground walking indicated that participants with stroke walked at a comfortable speed of 0.90±0.20 m/s and a fast speed of 1.19±0.27 m/s (mean±SD) (Fig. 5).

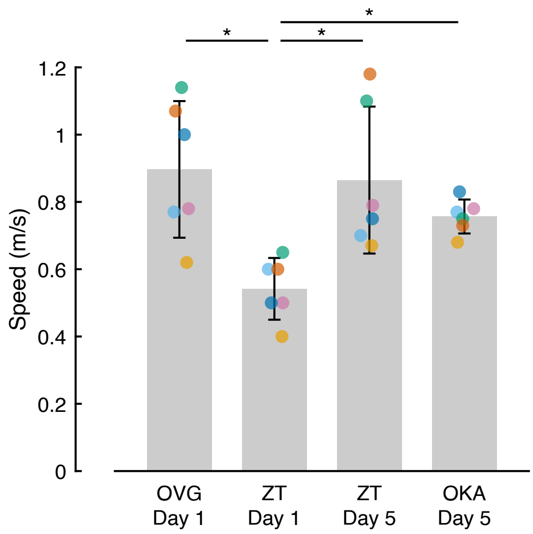

Fig. 5 Comfortable walking speed test results of the participants with stroke. Conditions shown are overground walking (OVG), exoskeleton with optimized knee-ankle on assistance (OKA), and exoskeleton in zero torque mode (ZT). The participants' comfortable walking speed in the exoskeleton on Day 1 was used for the subsequent optimization and validation protocol. Significant differences ($p < 0.05$) are shown with a star.

Based on comfortable overground walking speed, three participants were classified as "least limited community ambulators" (between 0.49-0.93 m/s), and three were classified as "unlimited community ambulators" (>0.93 m/s) [38]. Despite these overground capabilities, participants walked significantly slower on the treadmill. The familiarization day treadmill ramp speed test indicated these participants' comfortable treadmill speed was 0.54±0.09 m/s, which was significantly slower than their comfortable overground speed (p=0.002). This comfortable treadmill speed was the same for all participants in normal shoes and in the exoskeleton in zero torque mode. The second ramp test, conducted at the end of the final session on day 5, indicated that participants' comfortable treadmill speed had increased. When walking in the exoskeleton in zero torque mode, participants' comfortable speed increased to 0.87±0.22 m/s (p=0.009 vs. day 1 zero torque). In optimized knee-ankle mode, participants walked at a comfortable speed of 0.76±0.05 m/s (p=0.005 vs. day 1 zero torque, p=0.30 vs. day 5 zero torque).

3.4 Kinematics

Knee kinematics (Fig. 6, top row) of participants with stroke exhibited reduced knee range of motion in zero torque mode, characterized by a more flexed knee at heel strike (13±6° vs. 0±2° for control participants, p<0.001) and a nonsignificant trend towards a reduced peak knee flexion during swing (45±16° vs. 58±5°, p=0.10). When ankle torque was turned on but the knee was left unassisted, participants with stroke showed a significant increase in knee stance excursion compared to zero torque (17±7° vs. 11±4°, p=0.033), consistent with our expectation that ankle torque could induce compensatory or unintended knee motion. This effect was not observed in control participants (11±4° with ankle assistance vs. 12±2° in zero torque, p=0.68). When both ankle and knee assistance were provided, knee kinematics were closer to typical values, including heel strike angle (6±3° stroke, 3±0.3° control), stance excursion (10±2° stroke, 12±1° control), and swing peak flexion (53±2° stroke, 53±3° control). There were differences in knee kinematics between the stroke and control groups in knee assistance trials due to minor, participant-specific adjustments made on day one for comfort and foot clearance in participants with stroke. Knee angle tracking when the knee assistance was turned on was accurate for both groups, with RMS errors of 0.33° for participants with stroke and 0.32° for control participants.

Ankle kinematics (Fig. 6, bottom row) were affected by exoskeleton assistance in a similar manner in the stroke and control groups: ankle-only assistance reduced pushoff angular velocity compared to walking in zero-torque mode, while adding knee assistance reversed this effect. Control group participants exhibited high plantarflexion rates in zero torque (91±20°/s) and generic knee-ankle assistance conditions (99±25°/s), but showed reduced velocity with ankle-only assistance (48±13°/s, p=0.01 vs. zero torque, p<0.001 vs. generic knee-ankle). Participants with stroke had moderate

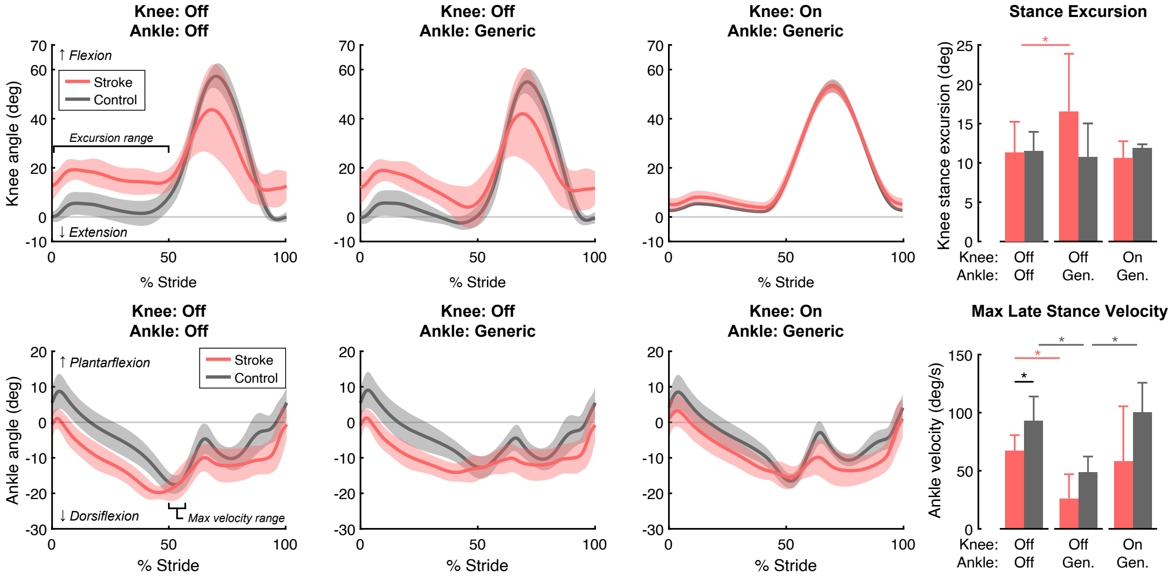

Fig. 6 Sagittal plane knee (top row) and ankle (bottom row) joint kinematics as measured by the exoskeleton for each condition, averaged by participant group. Notable kinematic changes are given in the bar plots in the rightmost column. Stance excursion is defined as the range in the knee position observed from 0 to 50% stride; late stance is defined as 50 to 60% stride. Clouds/whiskers represent ±1 standard deviation. Statistically significant ($p < 0.05$) differences between bars are reflected by a star.

angular velocities in zero torque (66±13°/s) and generic knee-ankle conditions (57±26°/s), and also displayed reduced velocity with ankle-only assistance (26±10°/s, p=0.002 vs. zero torque, p=0.057 vs. generic knee-ankle).

3.5 Kinetics

Applied torque was smaller at the knee than at the ankle, and peak knee torque did not differ significantly between stroke and control participants (Fig. 2 and 7). Average peak knee torque occurred during knee flexion, typically around 50% of stride, with values of 0.16 Nm/kg for control participants and 0.15 Nm/kg for participants with stroke. Stroke participants received a similar mean absolute torque (0.05 Nm/kg) as control participants (0.04 Nm/kg). The presence of knee assistance did not meaningfully change peak ankle mechanical power of generic assistance (Appendix Fig. A6). Participants with stroke had peak ankle power of 0.19±0.11 W/kg with the knee off versus 0.20±0.13 W/kg with the knee on (p=0.77). For control participants, peak ankle power was 0.15±0.06 W/kg versus 0.23±0.12 W/kg (p=0.06) with the knee off versus on. However, the addition of knee assistance altered the shape of the ankle mechanical power curve for the participants with stroke, particularly around 40% of the stride, where ankle power shifted from positive with knee assistance off to negative with knee assistance on. Ankle torque tracking was accurate, with stance-phase RMS errors of 0.023 Nm/kg for participants with stroke and 0.025 Nm/kg for control participants (Appendix Fig. A5).

Anteriorly-directed ground reaction forces during the propulsive phase of stance were significantly less for participants with chronic stroke versus healthy control participants across all conditions (p <0.05 for all intra-condition stroke vs. control comparisons, Appendix Fig. A7). Across all conditions, average anteriorly-directed ground reaction forces ranged from 0.12–0.15 Ns/kg for participants with stroke. For healthy control participants, this range was 0.26–0.29 Ns/kg.

3.6 Learning

Spatiotemporal measurements did not change systematically for either group in repeated evaluations of generic knee-ankle assistance conducted throughout the study (Fig. 8). Participants with stroke walked with a stride time of 1.52±0.23 s on the first day and 1.55±0.19 s on the final day (p=0.43). Control participants' stride time also did not change significantly, from 1.64±0.11 s in the first test to 1.72±0.14 s (p=0.17) in the final test. The difference in final-day stride time between groups was

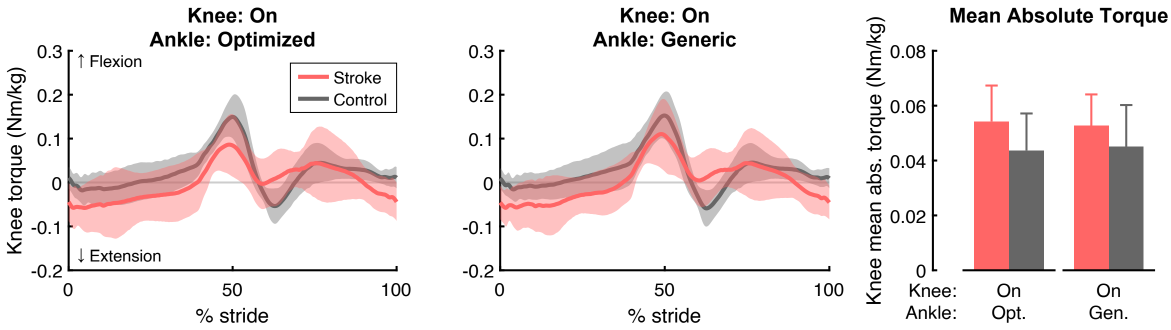

Fig. 7 Sagittal plane knee kinetics as measured by the exoskeleton for each condition with powered knee assistance, averaged by participant group. Clouds/whiskers represent ±1 standard deviation.

not significant (p=0.11). Stride time variability also did not significantly change across days. For participants with stroke, variability changed from 3.7±1.3% on day 1 to 3.0±1.2% on day 5 (p=0.11), while control participants changed from 2.6±1.2% to 1.9±0.6% (p=0.14). There was a trend toward a between-group difference in final-day stride time variability (p=0.058).

The energy cost of walking with generic knee-ankle assistance relative to zero torque changed inconsistently, showing no expected downward trend that would have suggested participants were learning to maximize exoskeleton benefits. Participants with chronic stroke had an average change in energy cost of -3.8%, 3.2%, -2.7%, -7.9%, and -5.2% from the first to the final evaluation of generic knee-ankle assistance, respectively (versus zero torque mode; negative values indicate metabolic rate reduction). For the control participants, these energy cost values were -6.5%, 1.8%, -6.7%, and -0.5% (Appendix Fig. A8).

4 Discussion

4.1 Effect of Walking Speed

The slow walking speed of 0.54 m/s had a strong effect on the healthy control participants, limiting the metabolic rate reduction of optimized unilateral knee-ankle assistance to only a 4% improvement versus zero torque mode. The outsized effect of walking speed on metabolic rate reduction in our study is illustrated by comparing to prior work which used a similar ankle exoskeleton configuration at faster speeds: [13] showed that a unilateral ankle exoskeleton emulator demonstrated a 24% improvement with optimized assistance at 1.25 m/s walking speed in 11 healthy participants. Our finding that very slow walking strongly limits exoskeleton metabolic rate reduction extends prior evidence of a direct relationship between speed and energy economy benefit. [29] showed that reducing speed from 1.25 to 1 m/s decreased metabolic rate reduction from 47% to 26% with optimized bilateral hip-knee-ankle assistance. A second study with one participant walking in optimized bilateral ankle assistance also found that walking at 1.25 m/s resulted in a 33% energy cost reduction, while at 0.75 m/s, this reduction was only 3% [13].

The presence of stroke did not introduce a meaningful difference in the metabolic rate reduction of exoskeleton assistance at the speeds tested in this study. There were no significant differences in metabolic rate reduction between the stroke and control groups (Fig. 3); these two groups also optimized to similar ankle torque profiles on average (Fig. 4) and had similar knee kinetics (Fig. 7). These results were unexpected; we anticipated that control participants would benefit more from exoskeleton assistance, that optimized ankle torque might differ between stroke and control groups, and that participants with stroke would need to rely more on the knee assistance and thereby experience larger knee torques.

Considering the relationship between speed and metabolic rate reduction and the similar responses of the stroke and control groups, greater benefits may have been achieved if participants had been allowed to walk at faster speeds. Control participants were capable of faster walking but were intentionally limited to the stroke group's average speed. Participants with stroke likely also could have walked faster. On day 1, these participants walked significantly slower on the treadmill than overground (0.54 vs. 0.90 m/s), likely reflecting a conservative speed choice in an unfamiliar setting and the need to sustain this speed throughout the study. However, their preferred treadmill speed increased by the end of the study to 0.76 m/s with assistance and 0.87 m/s without assistance (Fig. 5). These findings suggest that, if walking speed had allowed to vary, such as with a self-paced treadmill [16] or in overground walking with an untethered device [17], participants may have gradually increased their speed, potentially resulting in greater metabolic rate reductions.

In other studies, people with chronic stroke have shown the ability to walk faster in exoskeleton assistance than seen in our study, and this faster walking has led to improvements in the energy benefit of assistance. The largest energy cost reduction reported in people with stroke was a 10% improvement relative to unpowered walking in the device, achieved with unilateral ankle assistance at an average walking speed of 0.95 m/s [18]. Another study progressively increased treadmill speed during ankle exoskeleton walking, starting at 60% of overground speed and increasing by 10% each minute [21]. All participants were able to walk with assistance at their overground speed (average of 0.85 m/s) and half exceeded it, with higher speeds increasing joint power and exoskeleton energy delivery, though metabolic effects of each speed were not assessed due to the short condition durations. The difference between these studies and ours may reflect the duration of the walking tasks: during our familiarization day ramp test, participants were finding a speed to commit to for more than a total of 3 hours of walking, whereas in other studies, participants only walked with exoskeleton assistance for a handful of minutes. Because our participants knew that their comfortable speed would be used throughout the protocol, they may have chosen a more conservative pace to make the upcoming trials less strenuous.

Despite their slow comfortable treadmill speed, the participants with stroke in this study exhibited notably strong mobility compared to typical stroke populations. Based on their overground walking speed, three participants were classified as least limited community ambulators and three as unlimited ambulators [38]. While not formally surveyed, more than half of the participants mentioned engaging in some form of daily exercise or walking, suggesting that this cohort was more physically active and higher-functioning than the broader stroke population.

4.2 Effect of Human-in-the-Loop Optimization

Optimizing ankle assistance did not meaningfully improve energy cost beyond generic assistance at this study's walking speeds. Although the stroke group's optimized knee-ankle assistance gave a 6.4% (significant) energy cost reduction and generic knee-ankle gave a 5.2% (nonsignificant) reduction, there was no significant difference between the two conditions. This was unsurprising given the small metabolic effect size seen at these speeds relative to the larger variance seen in each condition. Prior work with healthy participants has shown that the effect size of optimization is about one-quarter of the total metabolic rate reduction delivered by the exoskeleton [14]. If the participants with stroke experienced a similar magnitude of benefit, detecting it will be challenging given the overall effect of assistance on metabolic rate at these speeds was small. Walking faster would increase the absolute metabolic rate reduction of both generic and optimized assistance, making the potential benefit of optimization easier to distinguish.

Although optimization yielded only modest gains beyond generic settings, the optimized parameters reveal that the underlying torque requirements for stroke participants are similar to those of healthy users. The optimizer trended towards higher torques and later peak times, which agrees with prior work [13, 14] that found a strong relationship between these parameters and metabolic rate. This suggests the sagittal plane exoskeleton torque requirements for stroke and healthy users may have similarities, and that knowledge from past studies with healthy participants could be used to initialize exoskeleton personalization for stroke participants.

4.3 Influence of Ankle Torque Magnitude

Optimization of ankle assistance consistently favored torques that were larger than those reported in most prior stroke exoskeleton studies, suggesting that additional energy economy improvements may be available in this direction. Participants with stroke in our study optimized to an average peak ankle torque of 0.39 Nm/kg (range: 0.32–0.48 Nm/kg), exceeding the peak torques used in

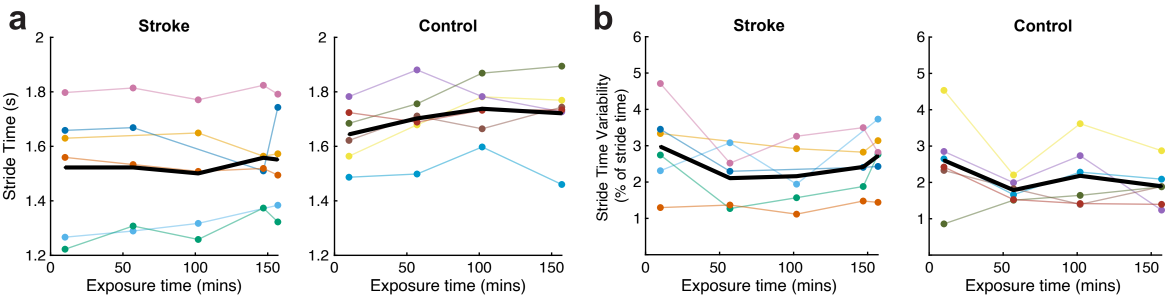

Fig. 8 Changes in exoskeleton-side (a) stride time and (b) stride time variability over the course of the experimental protocol for participants walking with generic knee-ankle assistance. The black line represents the average across all participants on each day, colored lines represent individual participants.

previous studies of ankle assistance for stroke, which have typically ranged from 0.14–0.29 Nm/kg [18, 19, 21–23, 39]. However, this preference for higher torques could be partly specific to the hardware, participants, and walking speeds involved in this study. For example, Awad et al. [18] reported the largest energy cost reduction in people with chronic stroke using an ankle exoskeleton, approximately 10% relative to unpowered walking, despite using substantially lower ankle torque of 0.15 Nm/kg. This larger metabolic rate reduction in spite of smaller torque could be due to differences in device mass, compliance, and envelope, since that study was done with a soft exoskeleton emulator (exosuit). The difference in metabolic rate reduction could also be explained by walking speed, as participants in that study walked considerably faster than in the present work (0.95 m/s versus 0.54 m/s).

Notably, the optimizer's preference for higher torques in our study emerged despite the relatively slow walking speeds tested. Prior work has found that faster walking speeds lead to greater peak torque magnitude: Bryan et al. [29] found optimized peak ankle torques of 0.56 Nm/kg at 1.0 m/s and 0.77 Nm/kg at 1.25 m/s. If such findings held true for people with stroke, we would expect that optimization of assistance at faster walking speeds might lead to even higher peak ankle torques.

4.4 Role of Knee Assistance

Adding knee assistance mitigated the maladaptive knee kinematics seen in ankle-only assistance. When participants with stroke walked in ankle-only assistance, late-stance knee extension was excessively large relative to the participants' preferred mid-stance knee angle. This large difference between mid-stance and late-stance knee angle was reflected in the significantly larger stance excursion value (Fig. 6). This effect was not observed in control participants, suggesting that the maladaptation was a consequence of reduced strength and coordination following stroke. These impairments could limit the ability to stiffen the knee joint, which may be necessary to transmit ankle torque into forward propulsion rather than knee extension. This increase in stance-phase knee excursion has also been seen in other ankle exoskeleton assistance pilot studies with participants with stroke [23, 30]. Adding knee assistance mitigated this maladaptation and helped maintain more typical stance-phase knee motion under ankle torque.

The addition of knee assistance also affected ankle joint mechanics, suggesting the potential for synergy between knee and ankle assistance. Ankle-only assistance decreased peak late-stance ankle velocity relative to walking unassisted. The addition of knee assistance removed this effect, returning joint velocity to a similar magnitude as seen when walking unassisted in both stroke and control groups. Adding knee assistance also shifted the average power at about 40% stride from positive to negative (Appendix Fig. A6). Humans typically apply negative biological power at this point in stride [40, 41], so the addition of knee assistance might help restore a more typical ankle power pattern during mid-stance by preventing premature ankle joint extension.

Future untethered exoskeletons for stroke could use smaller motors at the knee than at the ankle, as the knee torques required were much lower, with a maximum of 0.15±0.04 Nm/kg seen in this study. Although the absolute metabolic rate reduction from adding knee assistance is smaller than that of the ankle, it may be that the addition of knee assistance delivers meaningful benefit relative to the amount of added mass required to deliver such assistance.

4.5 Contribution of exposure time and learning to energy benefit

We could not confirm that stroke or control participants had completed learning by the end of the protocol despite the participants walking for a total of 180 minutes in powered exoskeleton conditions. Thus, it could be possible that more walking time would have led to greater metabolic rate reduction. Even when evaluated with the same assisted and zero-torque conditions each day, participants' spatiotemporal metrics and energy costs did not demonstrate any strong trends that might be indicative of changing expertise, confidence, or gait strategy (Fig. 8 and Appendix Fig. A8). In a different experimental setup, such as walking overground or on a self-paced treadmill, it may be easier to discern "expert" status using metabolics because faster walking enables a larger energy effect size. Furthermore, self-selected walking speed could be an additional proxy for expertise and confidence - participants might start walking at a slower speed, and gradually approach a faster speed as familiarity and proficiency with assistance increase.

4.6 Limitations

This study has several limitations that should be considered when interpreting the results. First, treadmill walking speed was fixed, which may have constrained natural gait patterns and limited metabolic rate reduction, as some participants may have preferred faster speeds as they adapted to assistance. Second, customization of knee assistance was limited: knee parameters were set only as a function of walking speed and were not optimized. This decision was motivated by the already time-consuming ankle optimization protocol and by pilot testing indicating that simultaneous adaptation to changing ankle and knee assistance significantly increased task complexity and learning demands. Third, due to the lengthy experimental protocol and multiple walking sessions, we only enrolled participants with stroke who demonstrated higher mobility levels compared to the general stroke population. As a result, these findings may not generalize to individuals with more severe impairments, and the study does not address which subpopulations of people with stroke may benefit most from exoskeleton assistance. Finally, all enrolled participants were male. This was not an intentional selection criterion; rather, it reflected the pool of referrals and enrollment constraints, as two screened female participants did not ultimately enroll.

5 Conclusion

In this study, we optimized knee-ankle exoskeleton assistance for participants with and without chronic stroke over three hours of cumulative treadmill walking. Three main points encapsulate the key findings of this study. First, the unexpectedly slow walking speeds had a strong influence on study outcomes in both groups and likely limited the metabolic rate reductions from assistance. Second, optimizing exoskeleton assistance did not meaningfully reduce energy cost beyond generic assistance, likely because the overall metabolic effect size of assistance at these speeds was small. Nonetheless, the optimized torque profiles revealed a preference for higher ankle torques in participants with stroke and showed notable similarity to optimized profiles previously identified in healthy individuals. Third, knee-ankle assistance mitigated the maladaptive knee kinematics induced by ankle-only assistance in participants with stroke. Although this knee assistance did not significantly change metabolic cost, it produced notable changes in ankle joint velocity and power.

Future work could explore several areas to build on these findings. First, optimizing knee assistance may further improve outcomes. While not every knee parameter may need optimization, those related to knee position during ankle torque application may provide the greatest metabolic rate improvement, whereas parameters such as swing-phase flexion could be hand-tuned based on observed foot clearance. Second, establishing a minimum walking speed in the exoskeleton as an inclusion criterion could help ensure participants are able to benefit from knee-ankle assistance. Based on our results, this minimum speed should likely be 0.6-0.8 m/s, with participants given ample time (e.g. 1 hour) to practice walking in the device before being evaluated for the walking speed criterion. Third, conducting a similar protocol with a self-paced treadmill or in overground walking may better capture the potential for energy economy improvements that can be delivered by exoskeleton assistance. Finally, future studies could investigate the effect of increased ankle torque for participants with stroke, particularly in combination with faster walking speeds.

Declarations

Ethics approval and consent to participate

All experiments were approved by the Stanford University Institutional Review Board under Protocol 44149. All participants provided written and informed consent to participate.

Consent for publication

All participants provided written and informed consent for the publication of these data, video, and images.

Availability of data and materials

Data generated in this study that was used to produce the figures in this manuscript will be made available at biomechatronics.stanford.edu upon final manuscript publication. Additional data for this study are available from the corresponding author on reasonable request.

Competing interests

The authors declare that they have no competing interests.

Funding

This work was supported by the National Science Foundation Graduate Research Fellowship Program (Grant number DGE-1656518).

Author Contribution

SHC and RMM designed the experimental protocol. RMM, RSP, and TT developed the exoskeleton controller. RMM, RSP, and MSL conducted the human subject experiments. RMM conducted data analysis with support from RSP and MSL. RMM, MGL, and SHC interpreted the data. RMM drafted and edited the manuscript. RSP, TT, MSL, MGL, and SHC edited and provided feedback on the manuscript. SHC and MGL conceived of the project. SHC oversaw the study. All authors read and approved the final manuscript.

Acknowledgements

We would like to thank Ziqin (Lisa) Zhu for assisting in the development of the exoskeleton controller and Gabrielle Walrath for assisting with experimental data collection. We also extend our appreciation to the participants with stroke for generously contributing their time and effort to this research.

References

[1] Khandelwal, P., Yavagal, D.R., Sacco, R.L.: Acute ischemic stroke intervention. J Am Coll Cardiol **67**(22), 2631–2644 (2016)

[2] Perry, J., Garrett, M., Gronley, J.K., Mulroy, S.J.: Classification of walking handicap in the stroke population. Stroke **26**(6), 982–989 (1995)

[3] Gordon, C.D., Wilks, R., McCaw-Binns, A.: Effect of aerobic exercise (walking) training on functional status and health-related quality of life in chronic stroke survivors. Stroke **44**(4), 1179–1181 (2013)

[4] Lord, S.E., McPherson, K., McNaughton, H.K., Rochester, L., Weatherall, M.: Community ambulation after stroke: how important and obtainable is it and what measures appear predictive? Arch Phys Med Rehabil **85**(2), 234–239 (2004)

[5] Barclay, R.E., Stevenson, T.J., Poluha, W., Ripat, J., Nett, C., Srikesavan, C.S.: Interventions for improving community ambulation in individuals with stroke. Cochrane Database Syst Rev (3) (2015)

[6] Bohannon, R.W., Andrews, A.W.: Limb muscle strength is impaired bilaterally after stroke. J Phys Ther Sci **7**(1), 1–7 (1995)

[7] Gersten, J.W., Orr, W.: External work of walking in hemiparetic patients. Scand J Rehabil Med **3**(1), 85–88 (1971)

[8] Chen, G., Patten, C., Kothari, D.H., Zajac, F.E.: Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds. Gait Posture **22**(1), 51–56 (2005)

[9] Bohannon, R.W., Andrews, A.W., Smith, M.B.: Rehabilitation goals of patients with hemiplegia. Int J Rehabil Res **11**(2), 181 (1988)

[10] Harris, J.E., Eng, J.J.: Goal priorities identified through client-centred measurement in individuals with chronic stroke. Physiother Can **56**(3), 171–176 (2004)

[11] Schmid, A., Duncan, P.W., Studenski, S., Lai, S.M., Richards, L., Perera, S., Wu, S.S.: Improvements in speed-based gait classifications are meaningful. Stroke **38**(7), 2096–2100 (2007)

[12] Klassen, T.D., Dukelow, S.P., Bayley, M.T., Benavente, O., Hill, M.D., Krassioukov, A., Liu-Ambrose, T., Pooyania, S., Poulin, M.J., Schneeberg, A., Yao, J., Eng, J.J.: Higher doses improve walking recovery during stroke inpatient rehabilitation. Stroke **51**(9), 2639–2648 (2020)

[13] Zhang, J., Fiers, P., Witte, K.A., Jackson, R.W., Poggensee, K.L., Atkeson, C.G., Collins, S.H.: Human-in-the-loop optimization of exoskeleton assistance during walking. Science **356**(6344), 1280–1284 (2017)

[14] Poggensee, K.L., Collins, S.H.: How adaptation, training, and customization contribute to benefits from exoskeleton assistance. Sci Robot **6**(58), 1078 (2021)

[15] Franks, P.W., Bryan, G.M., Martin, R.M., Reyes, R., Lakmazaheri, A.C., Collins, S.H.: Comparing optimized exoskeleton assistance of the hip, knee, and ankle in single and multi-joint configurations. Wear Technol **2** (2021)

[16] Lakmazaheri, A., Song, S., Vuong, B.B., Biskner, B., Kado, D.M., Collins, S.H.: Optimizing exoskeleton assistance to improve walking speed and energy economy for older adults. J NeuroEngineering Rehabil **21**(1), 1 (2024)

[17] Slade, P., Kochenderfer, M.J., Delp, S.L., Collins, S.H.: Personalizing exoskeleton assistance while walking in the real world. Nature **610**(7931), 277–282 (2022)

[18] Awad, L.N., Bae, J., O'Donnell, K., De Rossi, S.M.M., Hendron, K., Sloot, L.H., Kudzia, P., Allen, S., Holt, K.G., Ellis, T.D., Walsh, C.J.: A soft robotic exosuit improves walking in patients after stroke. Sci Transl Med **9**(400), 9084 (2017)

[19] Awad, L.N., Kudzia, P., Revi, D.A., Ellis, T.D., Walsh, C.J.: Walking faster and farther with a soft robotic exosuit: implications for post-stroke gait assistance and rehabilitation. IEEE Open J Eng Med Biol **1**, 108–115 (2020)

[20] Sloot, L.H., Baker, L.M., Bae, J., Porciuncula, F., Clément, B.F., Siviy, C., Nuckols, R.W., Baker, T., Sloutsky, R., Choe, D.K., O'Donnell, K., Ellis, T.D., Awad, L.N., Walsh, C.J.: Effects of a soft robotic exosuit on the quality and speed of overground walking depends on walking ability after stroke. J NeuroEngineering Rehabil **20**(1), 113 (2023)

[21] McCain, E.M., Dick, T.J.M., Giest, T.N., Nuckols, R.W., Lewek, M.D., Saul, K.R., Sawicki, G.S.: Mechanics and energetics of post-stroke walking aided by a powered ankle exoskeleton with speed-adaptive myoelectric control. J Neuroeng Rehabil **16**(1), 57 (2019)

[22] Herrin, K.R., Pan, Y.-T., Kesar, T.M., Sawicki, G.S., Young, A.J.: Robotic ankle exoskeleton and limb angle biofeedback for assisting stroke gait: a feasibility study. IEEE Robot Autom Lett **10**(2), 1011–1017 (2025)

[23] Pruyn, K., Murray, R., Gabert, L., Lenzi, T.: Autonomous powered ankle exoskeleton improves foot clearance and knee hyperextension after stroke: a case study. IEEE Trans Med Robot Bionics **7**(1), 51–58 (2025)

[24] Pergolini, A., Sanz-Morère, C.B., Livolsi, C., Fantozzi, M., Dell'Agnello, F., Ciapetti, T., Maselli, A., Baldoni, A., Trigili, E., Crea, S., Vitiello, N.: A unilateral active knee exoskeleton to assist individuals with hemiparesis – a pilot study. IEEE Trans Robot, 1–14 (2025)

[25] Sánchez, N., Schweighofer, N., Mulroy, S., Roemmich, R., Kesar, T., Torres-Oviedo, G., Fisher, B., Finley, J., Winstein, C.: Multi-site identification and generalization of clusters of walking impairment in individuals with chronic stroke. Biorxiv (2023)

[26] Miller, D.E., Tan, G.R., Farina, E.M., Sheets-Singer, A.L., Collins, S.H.: Characterizing the relationship between peak assistance torque and metabolic cost reduction during running with ankle exoskeletons. J NeuroEngineering Rehabil **19**(1), 46 (2022)

[27] Steele, K.M., Jackson, R.W., Shuman, B.R., Collins, S.H.: Muscle recruitment and coordination with an ankle exoskeleton. J Biomech **59**, 50–58 (2017)

[28] Lelas, J.L., Merriman, G.J., Riley, P.O., Kerrigan, D.C.: Predicting peak kinematic and kinetic parameters from gait speed. Gait Posture **17**(2), 106–112 (2003)

[29] Bryan, G.M., Franks, P.W., Klein, S.C., Peuchen, R.J., Collins, S.H.: A hip–knee–ankle exoskeleton emulator for studying gait assistance. Int J Robot Res **40**(4-5), 722–746 (2021)

[30] Nguyen, T.M.: Assisting walking in individuals with chronic stroke using exoskeletons. PhD thesis, Stanford University (2022)

[31] Zhang, J., Collins, S.H.: The iterative learning gain that optimizes real-time torque tracking for ankle exoskeletons in human walking under gait variations. Front Neurorobot **15**, 653409 (2021)

[32] Folstein, M.F., Folstein, S.E., McHugh, P.R.: "mini-mental state": a practical method for grading the cognitive state of patients for the clinician. J Psychiatr Res **12**(3), 189–198 (1975)

[33] Podsiadlo, D., Richardson, S.: The timed "up and go": a test of basic functional mobility for frail elderly persons. J Am Geriatr Soc **39**(2), 142–148 (1991)

[34] Bohannon, R.W.: Comfortable and maximum walking speed of adults aged 20—79 years: reference values and determinants. Age Ageing **26**(1), 15–19 (1997)

[35] Hansen, N.: The CMA evolution strategy: a tutorial. arXiv (2016)

[36] Martin, R.M., Collins, S.H.: Improving CMA-ES convergence speed, efficiency, and reliability in noisy robot optimization problems. Evol Comput, 1–35 (2026)

[37] Brockway, J.M.: Derivation of formulae used to calculate energy expenditure in man. Hum Nutr Clin Nutr **41**(6), 463–471 (1987)

[38] Fulk George D., He Ying, Boyne Pierce, Dunning Kari: Predicting home and community walking activity poststroke. Stroke **48**(2), 406–411 (2017)

[39] Takahashi, K.Z., Lewek, M.D., Sawicki, G.S.: A neuromechanics-based powered ankle exoskeleton to assist walking post-stroke: a feasibility study. J NeuroEngineering Rehabil **12**(1), 23 (2015)

[40] Fickey, S.N., Browne, M.G., Franz, J.R.: Biomechanical effects of augmented ankle power output during human walking. J Exp Biol **221**(22), 182113 (2018)

[41] Lee, S.J., Hidler, J.: Biomechanics of overground vs. treadmill walking in healthy individuals. J Appl Physiol (2008)

Appendix

Video 1 A multipanel video of participant with chronic stroke walking in normal shoes shoes, zero torque, generic ankle assistance, generic knee-ankle assistance, and optimized knee-ankle assistance. Link: https://drive.google.com/file/d/11tyHWpkMAOAkO-xkZwHb20B0ktzt4lvC/view?usp=sharing

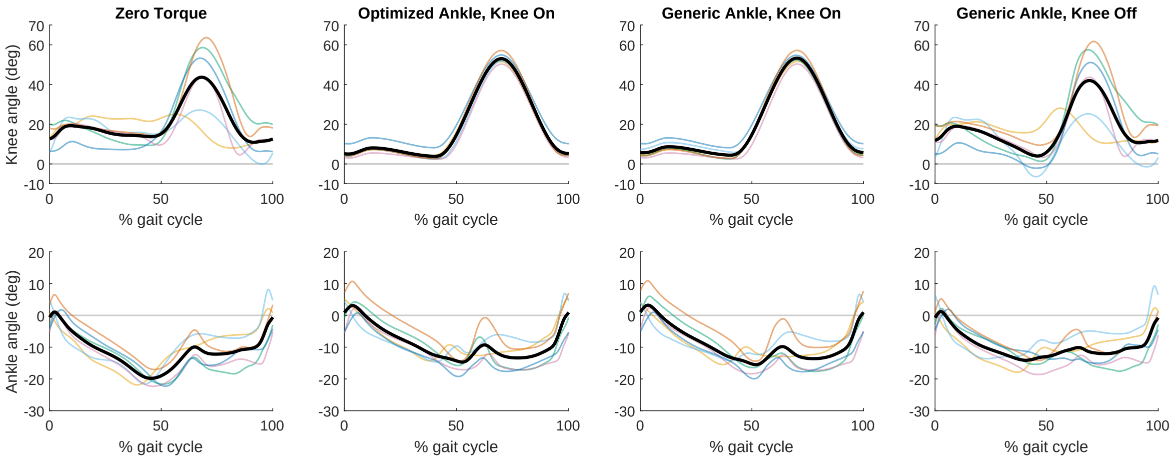

Fig. A1 Individual participant kinematics of participants with chronic stroke. Individual participants are represented by colored lines; average is represented by the black line.

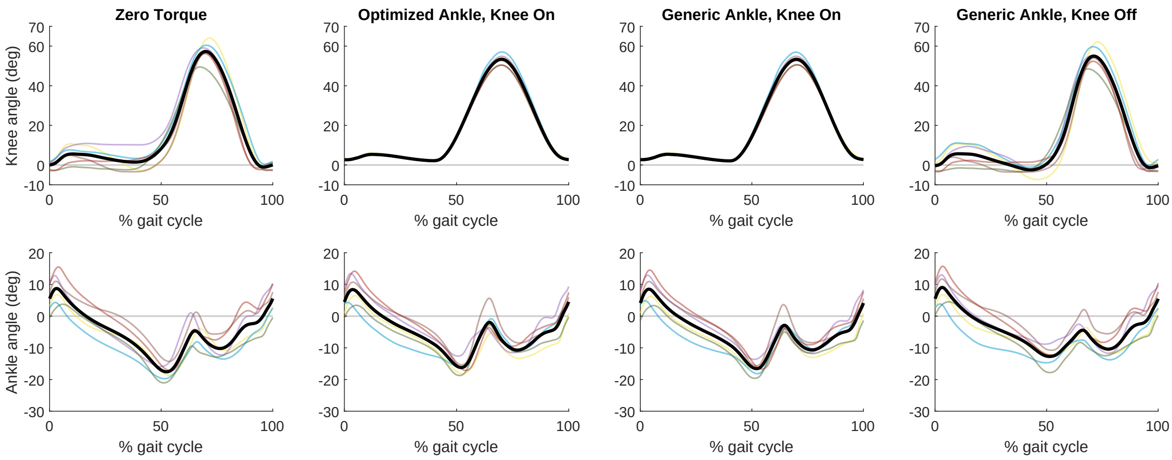

Fig. A2 Individual participant kinematics of healthy control participants. Individual participants are represented by colored lines; average is represented by the black line.

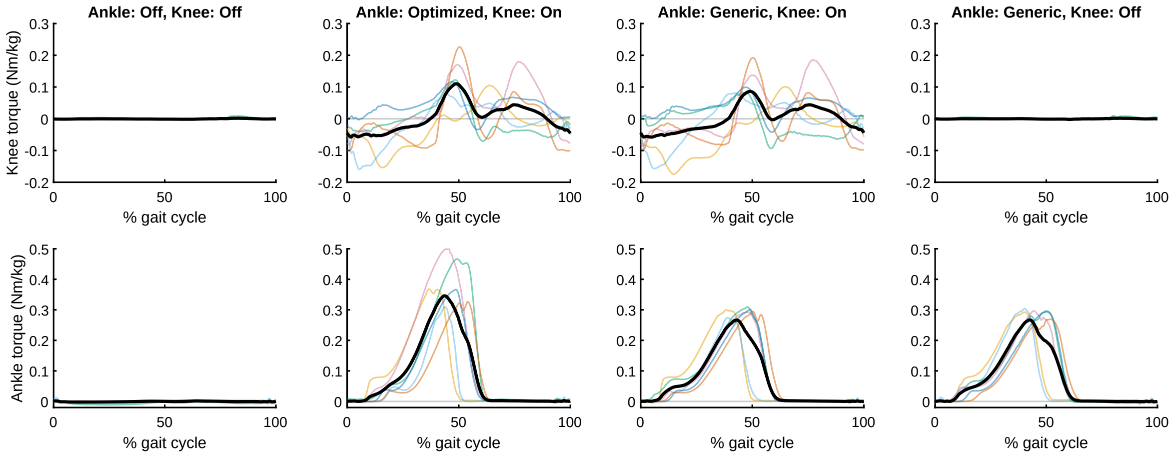

Fig. A3 Individual participant kinetics of participants with chronic stroke. Individual participants are represented by colored lines; average is represented by the black line.

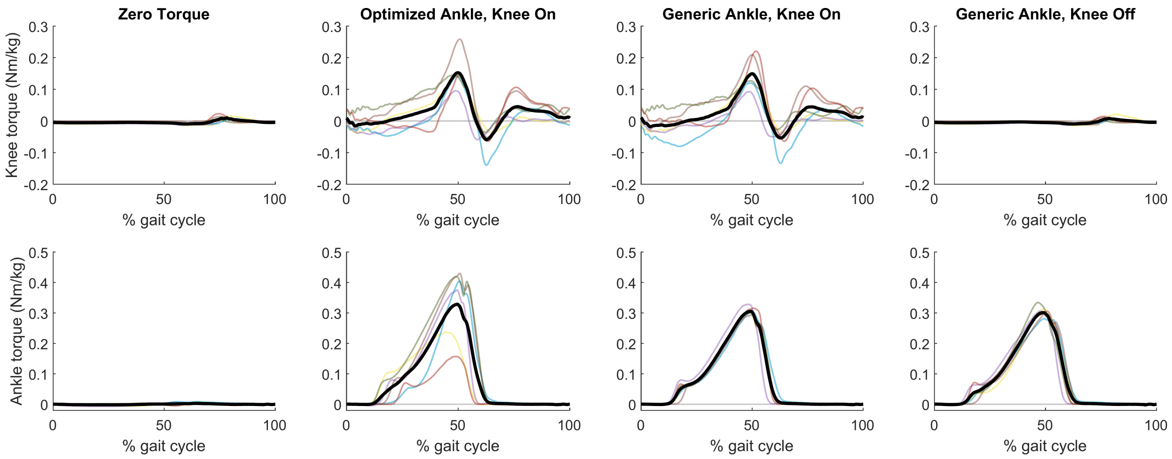

Fig. A4 Individual participant kinetics of healthy control participants. Individual participants are represented by colored lines; average is represented by the black line.

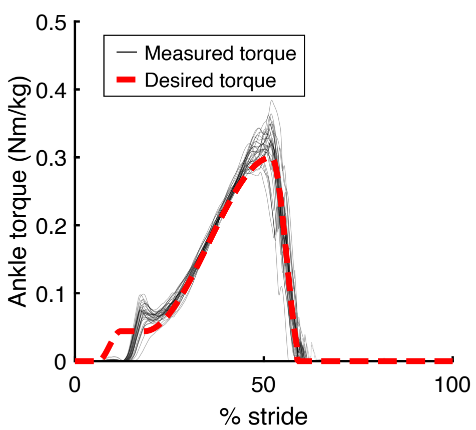

Fig. A5 Representative ankle torque tracking. The stance-phase root mean square error shown is 0.024 Nm/kg, the same as that across the entire study.

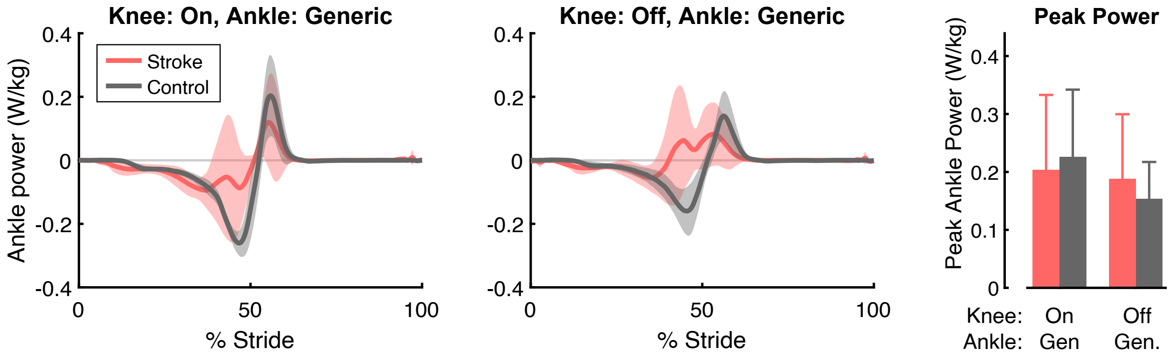

Fig. A6 Ankle joint mechanical power.

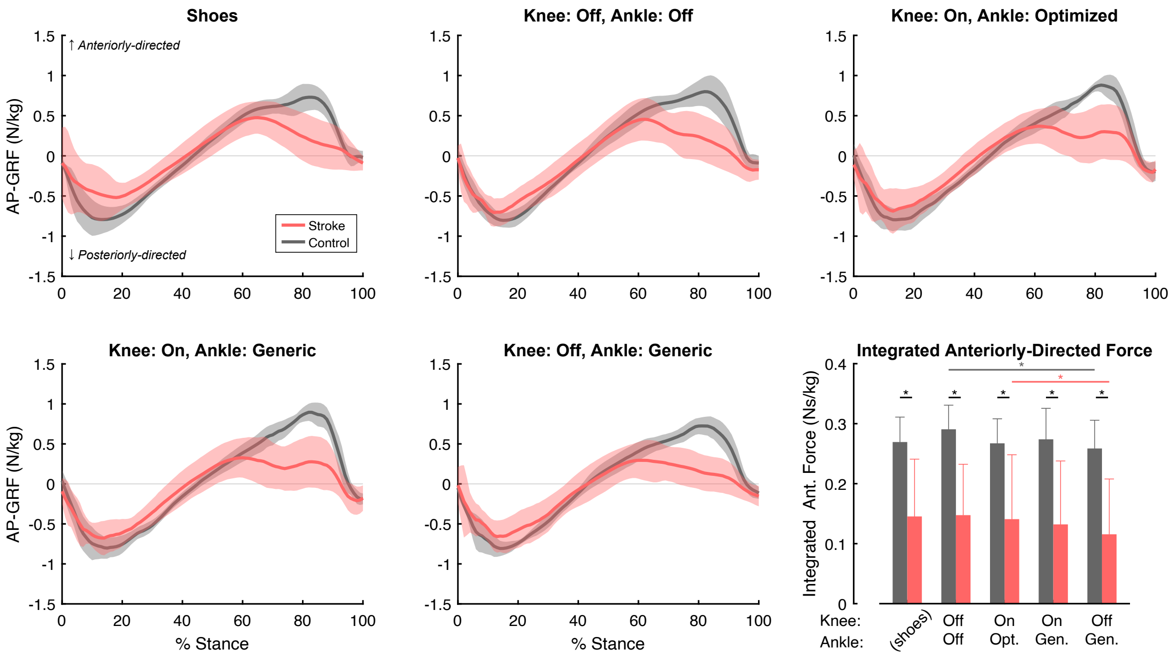

Fig. A7 Anteroposterior ground reaction forces and integrated push-off (anteriorly-directed) force from the exoskeleton-worn side.

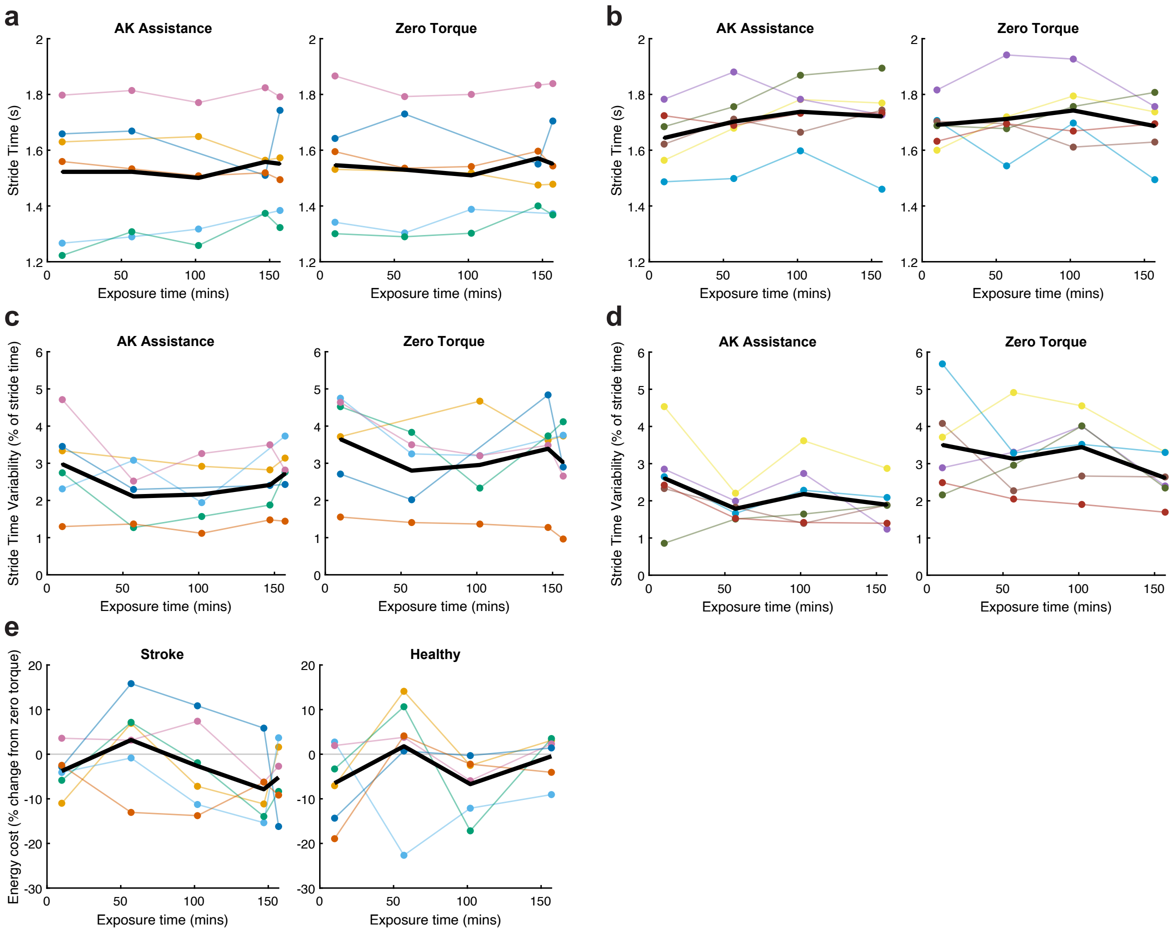

Fig. A8 Change of various gait parameters with exposure time when participants walked in generic ankle knee on mode and in zero torque mode. (a): stride time of participants with stroke; (b) stride time of control participants; (c) stride time variability of participants with stroke; (d) stride time variability of control participants; (e) energy cost of generic ankle knee on mode in both participant groups.