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Interactions Between Posture And Locomotion In Nonimpaired And Poststroke Walking

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INTERACTIONS BETWEEN POSTURE AND LOCOMOTION IN NONIMPAIRED AND POSTSTROKE WALKING

by

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PHD IN REHABILITATION SCIENCE

ABSTRACT

Background: Inappropriate paretic-limb force generation is a hallmark of impaired walking function poststroke. In addition to weakness, the paretic limb misdirects foot forces during stance, which may be due to inappropriate postural influence over weakened locomotor control. Purpose: Investigate neural control interactions between posture and locomotor functions by manipulating postural influence during walking of nonimpaired and individuals poststroke. I present four studies investigating measures associated with locomotor function and foot-force direction under varying postural demands during walking. Methods: Study 1 compared comfortable walking speed (CWS) outcomes following two body-weight-support (BWS) training protocols for N=29 participants with chronic (≥5 months) poststroke hemiparesis and slow baseline CWS (<1.1 m/s). Study 2 tested the ability of a novel support apparatus to fully minimize postural demands of walking for N=20 nonimpaired participants. Study 3 used the support apparatus to characterize fore-aft (Fy) to vertical (Fz) ground reaction force (GRF) ratios (Fy/Fz ratios), joint moments, and muscle activity during propulsion of N = 16 nonimpaired participants walking at 1.0 and 0.5 m/s under 0 to 30% BWS versus within the support apparatus. Study 4 compared Fy/Fz ratios of N=17 individuals ≥ 6 months poststroke with residual hemiparesis at their CWS under 0 to 30% BWS versus within the support apparatus. Results: Study 1 showed significant pre-post improvement of CWS that, on average, did not reach the 0.16 m/s minimal clinically important difference regardless of intervention group. Study 2 showed that the support apparatus reduced trunk motion and negative mechanical work for nonimpaired individuals, and enabled matching of vertical GRFs to those of typical walking. Study 3 showed nonimpaired Fy/Fz ratios decreased across force targets while walking externally stabilized in the support apparatus, demonstrating a relative decoupling of fore-aft and vertical GRF components. Study 4 showed the paretic-limb Fy/Fz ratio was not better directed during externally stabilized walking in the support apparatus. Conclusions: Although nonimpaired individuals used intact locomotor control strategies when the support apparatus minimized postural demands, minimizing postural demands did not facilitate more appropriate paretic-limb foot-force direction. Individuals poststroke may rely on postural control mechanisms to compensate for loss of voluntary locomotor control.

Keywords: poststroke; walking; posture; locomotor control; body-weight support

DEDICATION

This dissertation work is dedicated to my husband, Dustin, for his willingness to put our lives on hold through seven years of graduate school for me to pursue my dreams. Thank you for understanding and supporting my vision and never giving up on me;

My dogs, Maddie and Arnold, who have been with me through every pivotal experience of my adult life and given me a reason to smile every single day;

My parents, David and Audrey Kirtland, who raised me to be curious, tenacious, and possess a sense of purpose. I would never have had these opportunities without your guidance and support;

My sister, for setting an example of what it is to be a strong, career-driven female who embraces all facets of life. I believe that the impossible is possible because of you;

And to my dissertation committee members, particularly my mentor, David Brown, for providing me with such exceptional training and helping me to develop into a scientist.

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TABLE OF CONTENTS

LIST OF TABLES

LOCOMOTOR FORCE GENERATION DURING WALKING POSTSTROKE WITH AND WITHOUT POSTURAL CONSTRAINT

LIST OF FIGURES

Figure *Page*

INTRODUCTION

WALKING AND BALANCE OUTCOMES FOR STROKE SURVIVORS: A RANDOMIZED CLINICAL TRIAL COMPARING BODY-WEIGHT-SUPPORTED TREADMILL TRAINING WITH VERSUS WITHOUT CHALLENGING MOBILITY SKILLS

MINIMIZING POSTURAL DEMANDS OF WALKING WHILE STILL EMPHASIZING LOCOMOTOR FORCE GENERATION FOR NONIMPAIRED INDIVIDUALS

LOCOMOTOR FORCE GENERATION DURING NONIMPAIRED WALKING WITH AND WITHOUT COODINATED POSTURAL CONTROL

LOCOMOTOR FORCE GENERATION DURING WALKING POSTSTROKE WITH AND WITHOUT POSTURAL CONSTRAINT

SUMMARY

LIST OF ABBREVIATIONS

ABC	Activities Specific Balance Confidence
BBS	Berg Balance Scale
BF	Biceps Femoris
BWS	Body Weight Support
COM	Center of Mass
CWS	Comfortable walk speed
DGI	Dynamic Gait Index
FWS	Fast walk speed
Fy	Fore-aft Force
Fz	Vertical Force
GAS	Gastrocnemius
GDS	Geriatric Depression Scale
GRF	Ground Reaction Force
HF	Hands free
HF+C	Hands free + Challenge
SIS	Stroke Impact Scale
SOL	Soleus
TA	Tibialis Anterior
VM	Vastus Medialis

QUOTES

In his seminal 1906 publication, The Integrative Action of the Nervous System, the highly regarded neurophysiologist Charles Sherrington said, “posture accompanies movement like its shadow.”

So, I figured out how to make it rain.

INTRODUCTION

"Nothing epitomizes a level of independence and our perception of a good quality of life more than the ability to travel independently under our own power from one place to another." – A.E. Patla Adaptability of Human Gait, Volume 78

Background

This section introduces the complex interactions between postural and locomotor functions that are required for functional walking ability. I describe the clinical, personal, and societal problem of impaired walking function poststroke, and one of the hallmarks of poststroke walking impairments known as misdirected foot forces. I summarize the current knowledge regarding the importance of appropriately directed foot forces for functional walking ability and the consequences of misdirecting these foot forces poststroke. I follow with proposed theoretical underpinnings for the approach I took to explore mechanisms underlying walking behavior of nonimpaired and individuals poststroke during walking with requirements to coordinate postural and locomotor functions versus walking with minimal postural demands. I propose an explanation for misdirected foot forces poststroke and introduce my novel experimental approach for testing my hypotheses. Finally, I describe the subsequent chapters in this dissertation that reflect these efforts.

Stroke survivors exhibit slow walking speeds that limit performance of activities of daily living and community engagement

Stroke is the leading neurological cause of chronic disability in the U.S., with an estimated 7,000,000 current stroke survivors and an incidence of 795,000 new cases per year (1). The societal burden of chronic, stroke-related impairments is projected to reach $240 billion in total costs by the year 2030 (2). Recovery of walking function is among the most important goals reported by affected individuals (3); however, many of these individuals are unable to regain fully functional gait patterns, even following walking rehabilitation (4–10). Limited walking ability restricts activities and participation of stroke survivors (11) and may lead to a sequela of secondary health conditions due to limb disuse (12–17). Better strategies to improve walking function are essential in order to mitigate the societal and personal impacts of stroke.

Appropriately directed ground reaction forces (GRFs) are required for functional walking

Functional walking requires progressing the body forward while maintaining stability of the body's center of mass (COM) (18,19). Control over the net angular momentum of the body can be observed through the magnitude and direction of sagittal plane GRFs as the foot interacts with the ground (20,21). These forces include the fore-aft (Fy) and vertical (Fz) GRF components (Figure 1, black vectors). The Fz GRF magnitude is proportional to body weight and the resultant sagittal direction is dictated by the Fy GRF. Appropriately directed GRFs are required to coordinate posture and locomotion throughout the stance phase during walking (Figure 1, blue vectors). In late stance, the direction of the sagittal GRF vector represents the combined role of body support and forward propulsion (Figure 1, blue vectors) (20). Misdirected GRFs, particularly during late stance (Figure 1, red vector), are a major factor limiting walking function of stroke survivors (22–25). These misdirected forces have both muscular and neural origins (26); however, exact mechanisms are still being actively explored.

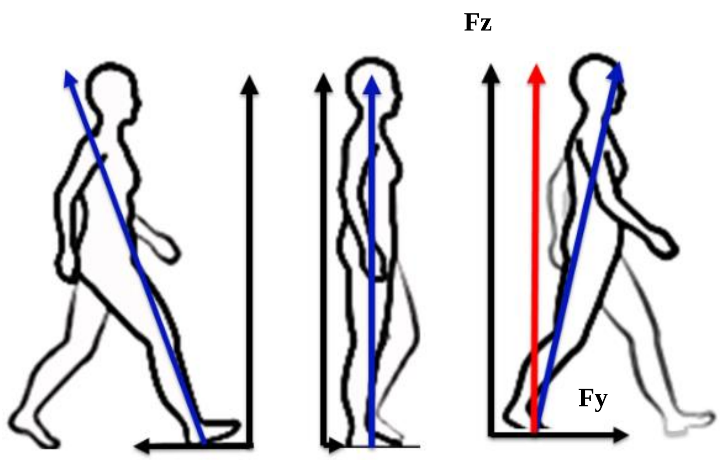

Figure 1. Example of Fy and Fz GRF components (black) following foot strike, at mid stance, and during push off. Appropriately directed sagittal resultant forces (blue) and misdirected sagittal resultant forces (red) in late stance.

Functional gait patterns of nonimpaired individuals are characterized by symmetry in GRFs between limbs on a step-by-step basis, with appropriately directed Fy GRFs during the stance phase (Figure 2, blue) (27). The Fy GRF is negative in early

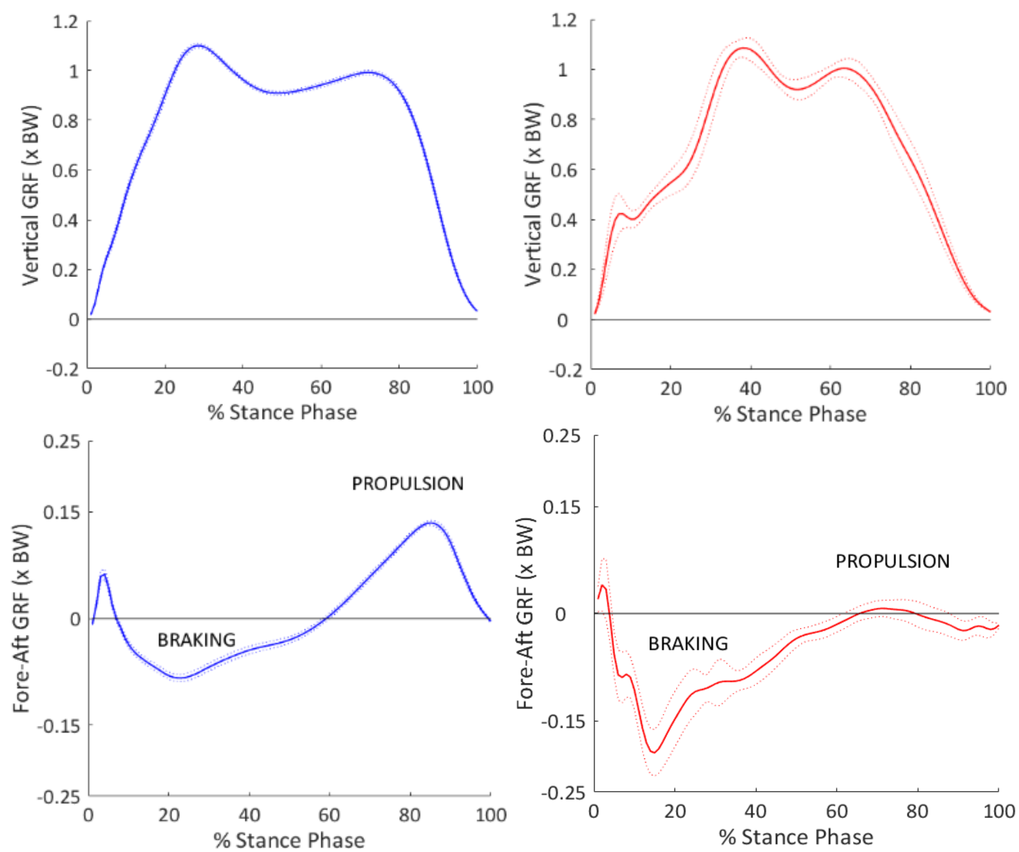

Figure 2. Fz GRFs (top) and Fy GRFs (bottom) of a nonimpaired individual (left column; blue) and paretic limb of an individual poststroke (right column; red).

stance, indicating a braking force, and results from underlying joint moments performing postural functions to decelerate and maintain upright orientation of the body and accept weight. The Fy GRF becomes positive in late stance, indicating propulsion, due to underlying joint moments acting to perform the locomotor function of propelling the body forward (18,28,29). At constant, comfortable walking speeds, braking and propulsive forces are roughly equal in magnitude and opposite in direction, resulting in a net impulse close to 0 N·s over a stride.

In contrast to appropriately directed GRFs, individuals poststroke exhibit interlimb asymmetry in Fy GRF direction (23,24,30). The paretic limb acts mostly as a brake, generating little propulsive force (Figure 2, red), and the nonparetic limb compensates with greater propulsive forces to maintain or increase walking speed. These inappropriate Fy GRFs cause the sagittal resultant GRF to be misdirected, which causes a net angular rotation of the body that must be corrected by compensatory strategies to avoid loss of balance.

Individuals with the greatest hemiparetic severity exhibit the most asymmetry in Fy GRFs, with the paretic limb contributing the least to forward propulsion (23,24). This imbalance in force generation limits walking speed (23–25), reduces efficiency (31–34), and may predispose stroke survivors to secondary health conditions like osteopenia and sarcopenia, particularly in the paretic limb, due to disuse (12,15,16). A better understanding of the neurological causes underlying misdirected GRFs can allow the development of strategies that encourage appropriate force generation patterns from the paretic limb, to better address gait impairments during walking rehabilitation poststroke and engender long-term improvements in walking function.

Spatiotemporal measures of poststroke walking behavior are related to misdirected GRFs

Able-bodied individuals increase propulsion largely through increasing trailing limb angle (35). In contrast, individuals poststroke have smaller trailing limb angles at foot off and are therefore unable to use this strategy as effectively to modulate speed (36–

38). Paretic trailing limb angle is a significant predictor and positively related to the propulsive GRF impulse from mid to late stance, highlighting the importance of lower-limb position in achieving adequate propulsion (36). Reduced trailing limb angle is accompanied by increased nonparetic and decreased paretic-limb stance duration during walking poststroke (39). The double support phase occupies a greater percentage of the gait cycle, particularly for slower walkers (40). An unwillingness to load the paretic limb appears to underlie shortened single-limb support (41). Vertical loading profiles are altered poststroke, with prolonged duration of paretic load acceptance and quicker rates of offloading in comparison to the nonparetic limb, which is loaded more quickly and offloaded more slowly (41). Strategies to encourage greater trailing limb angle and vertical loading may, therefore, also serve to promote greater paretic-limb propulsion.

Altered lower-limb joint moments underlie misdirected GRFs during walking poststroke

Braking and propulsive forces are the result of underlying joint moments acting to decelerate and accelerate the body over the course of a step (42). The uniarticular knee and hip extensors are the dominant contributors to the Fz GRF in early stance, with vasti muscles decelerating the body and the combined effort of lower-limb extensor muscles creating a support moment to maintain upright orientation and avoid collapse under gravity (42,43). While eccentric knee activity is critical for supporting the leg and trunk in early stance, individuals with severe impairments following stroke have excessive paretic vastus lateralis activity and increased braking (24) reflecting a greater postural role of the paretic limb.

In late stance, the ankle plantarflexors primarily contribute to the Fz GRF (42,44). Decreased paretic-limb plantarflexor muscle activity and increased hip and knee flexor activity acting to offload the limb results in a reduced plantarflexor moment, which is associated with low propulsion (45). Even in stroke survivors with symmetric gait patterns, joint moments are altered to compensate for lower plantarflexor moments compared to nonimpaired individuals (46). The compensatory positive work requirements of the nonparetic limb are exacerbated by a larger amount of negative mechanical work performed by the paretic limb at foot strike. Farris et al. (2015) (33) noted that reduced paretic propulsion and increased hip flexor moments resulted in 52% greater average rate of mechanical work during walking on a treadmill compared to nonimpaired individuals. Thus, improving the ability of the paretic limb to generate appropriate locomotor forces in terminal stance may reduce overall muscular work and improve functional outcomes of poststroke walking.

Single session and training interventions have limited success in increasing paretic propulsion

Interventions designed to increase paretic propulsion during walking within a single session, or over the course of a training program, include body-weight support (BWS), robotic devices (e.g., lower-limb exoskeleton), fore-aft resistive force environments, treadmill training at fast speeds to increase trailing limb angle, and functional electrical stimulation (FES) among others (9,10,47–52). Approaches that engender a high degree of active participant involvement in training may be preferable to those that provide passive assistance in generating appropriate locomotor patterns (53). Robotic devices like lower-limb exoskeletons and FES operate under the premise that limb weakness, particularly of the ankle plantarflexors, is the limiting factor in paretic-limb propulsion (50,52,54). For this reason, they passively manipulate force generation characteristics of the paretic limb through guided foot placement and artificial activation of nerves. The reduced active force production by the participant may be one reason why these types of interventions show limited transfer to overground walking environments. Improved functional walking status is related to plantarflexor muscle contribution to propulsion (55,56); however, weakness may not be the primary cause of inappropriate paretic-limb force generation. Individuals poststroke appear to be capable of generating increasing amounts of paretic propulsion given the right environmental context (48,49,57,58).

Theoretical premise underlying the investigations in this dissertation

Prior research in our lab by Liang and Brown (58) used a pedaling paradigm with minimal postural demands to isolate locomotor behavior of individuals poststroke. They observed inappropriately directed shear-to-normal crank forces against the pedal during a nonseated, “posturally loaded” task, but more appropriate foot-force direction during a seated, “minimal posture” task. They proposed that this finding reflected an inappropriate coordination between posture and locomotor control systems, where the paretic limb had difficulty expressing weakened locomotor control when locomotor tasks also required postural coordination.

In neurologically nonimpaired individuals, separate, central controllers for posture and locomotion act in concert to maintain upright posture and allow expression of symmetric locomotor patterns, producing functional walking patterns (Figure 3) (59–62). The descending pathways involved in locomotion consist of parallel pathways derived from the cerebral cortex and brainstem that make connections with various regions in the brain (e.g., mesencephalic locomotor region, subthalamic nucleus, pontine locomotor

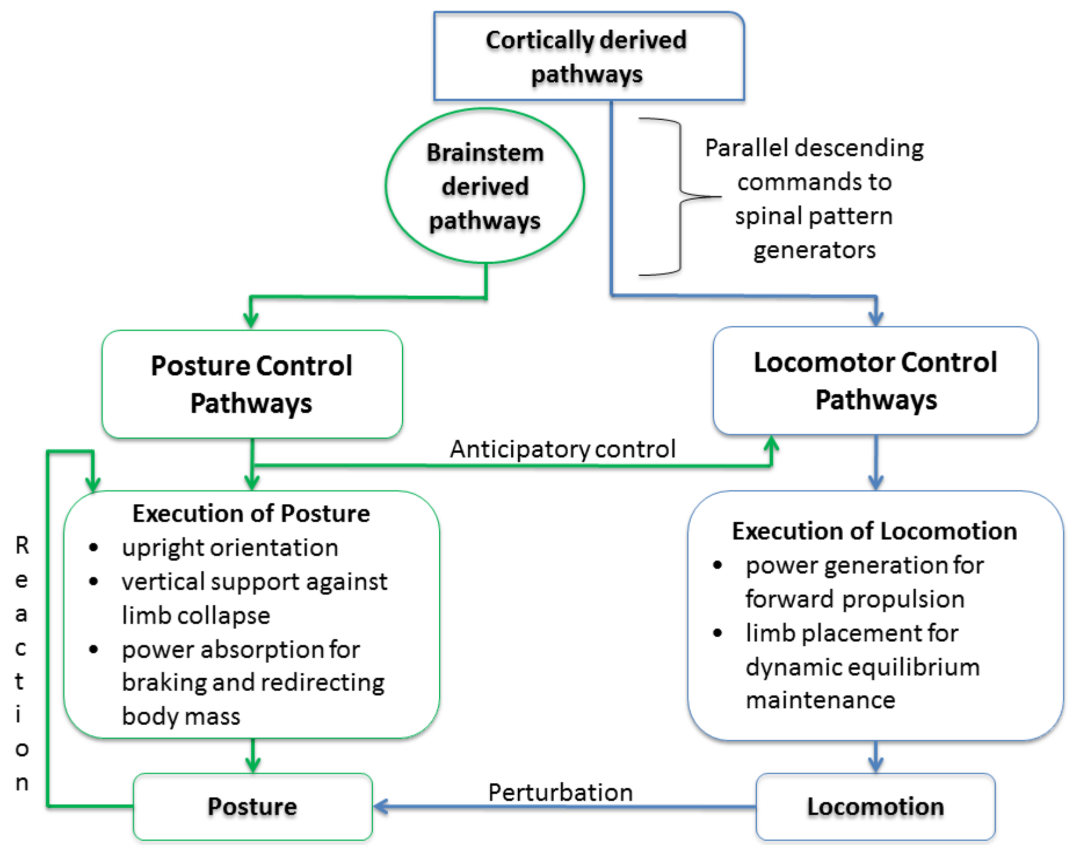

Figure 3. Postural-locomotor interaction model during unperturbed walking describing parallel descending commands facilitating primary postural (green) and locomotor (blue) functions. Stepping movements perturb posture, and feedback from load receptors during the stance phase is used through reactive mechanisms to initiate gait transitions. Anticipatory mechanisms allow for adjustments in limb trajectory to maintain dynamic equilibrium on a step-by-step basis.

region, cerebellum) before terminating at various levels of the spinal cord and influencing spinal neuronal networks (61,63–65). Brainstem-derived descending pathways are thought to affect axial musculature and play a role in postural orientation during walking (63,65), while cortically derived pathways are more involved in the control of appendicular musculature producing locomotor forces (64,66,67).

Following stroke, overexcitation of operational brainstem-derived pathways, particularly pathways controlling postural and mechanically coupled muscle groupings, may interfere with weakened cortically derived locomotor control pathways (Figure 4)

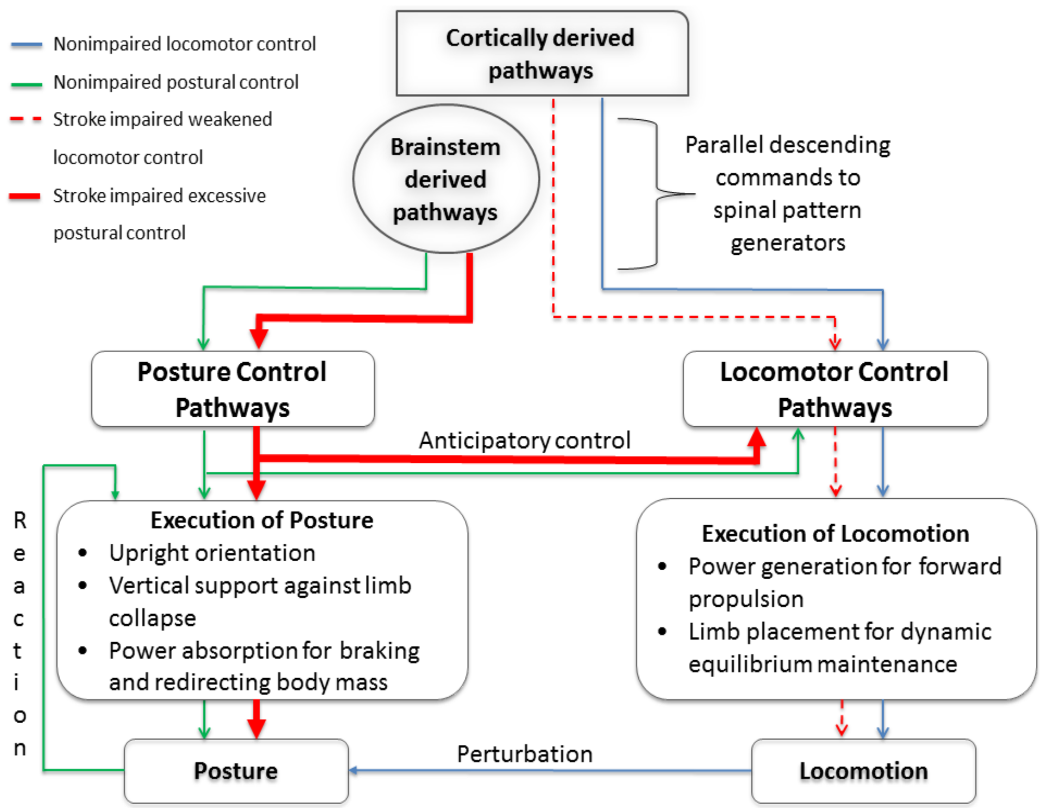

Figure 4. Nonimpaired postural-locomotor interaction model during typical walking (green and blue lines) and proposed inappropriate postural influence over weakened locomotor control poststroke (thick and dashed red lines, respectively).

(58,68–72). The misdirected GRFs of poststroke walking may be partly due to this undue influence of postural control mechanisms over locomotor control. If the coordination of posture and stepping during walking indeed reflects separate control systems (Figure 3) (59,60), we should theoretically be able to decouple stepping functions from postural functions during the gait cycle. We therefore proposed that minimizing postural demands of walking may reduce excessive postural influence and allow for more appropriate locomotor behaviors of the paretic limb during walking. For my dissertation research, I fabricated a novel postural support apparatus that allowed investigation of locomotor force generation with minimal requirements for postural coordination.

Overview of the following chapters presented in this dissertation

In *chapter 2*, I present the results of a 6-week randomized clinical trial that I participated in during the first two years of my dissertation research investigating effects of BWS training on improving walking speed of stroke survivors. BWS partially addresses postural demands during locomotion through providing vertical offloading of the limbs and reducing accelerations acting on the body (73,74). This study also relied on principles of active participant engagement in training and did not provide any passive assistance to individuals or allow use of assistive devices or handrail support. We used the primary outcome measure of comfortable walking speed (CWS) pre to post training to determine the effectiveness of the BWS training protocols for improving walking function of stroke survivors.

However, BWS only provided partial postural support and not all BWS studies demonstrate improvements in paretic propulsion (9). There remain postural demands imposed on the nervous system that may prevent individuals poststroke from generating appropriately directed GRFs. Thus, *chapter 3* of this dissertation describes the development of a postural support apparatus that enabled a fully externally stabilized walking environment that served three purposes, (1) externally stabilized the trunk (restricted trunk movement), 2) fully offloaded the trunk and upper body mass from the lower limbs, but still allowed participants to make contact with the treadmill surface with their feet to generate forces, and 3) minimized the need to control body COM accelerations by holding participants in place. I report our findings regarding the ability of this apparatus to minimize postural demands of walking for nonimpaired individuals.

Externally stabilized walking was a novel environment; therefore, in *Chapter 4* we further examined the neuromechanical strategy underlying locomotor force generation during walking conditions that required posture and locomotor coordination versus walking conditions where the novel support apparatus mechanically decoupled locomotor force generation from postural behaviors of nonimpaired individuals. We manipulated postural and locomotor force generation requirements of walking under two experimental conditions, 1) walking under increasing amounts of BWS, which partially reduced postural demands, but did not fully minimize them relative to locomotor demands and 2) externally stabilized walking within the novel support apparatus that minimized postural demands relative to locomotor requirements while manipulating force generation through providing a range of vertical force targets. I report further evidence resulting from this experiment that we fabricated a walking environment of reduced anticipatory and reactive postural control influence over locomotor force generation behaviors.

Following characterization of nonimpaired locomotor behaviors during typical versus externally stabilized walking, we investigated our primary question in *Chapter 5* regarding whether minimizing postural demands of walking could encourage more appropriate control of locomotor force generation behaviors poststroke, as characterized primarily by more appropriate Fy/Fz GRF ratios during the second half of the stance phase. I describe the locomotor responses of individuals poststroke under BWS versus externally stabilized walking and discuss the implications of our findings in terms of the neural control strategy governing the paretic limb during walking.

In Chapter 6, I summarize the key findings from each manuscript presented in this dissertation document, discuss implications of these findings for rehabilitation practices, and recommend future research directions.

WALKING AND BALANCE OUTCOMES FOR STROKE SURVIVORS: A RANDOMIZED CLINICAL TRIAL COMPARING BODY-WEIGHT-SUPPORTED TREADMILL TRAINING WITH VERSUS WITHOUT CHALLENGING MOBILITY SKILLS

by

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ABSTRACT

Background

Treadmill training, with or without body-weight support (BWSTT), typically involves high step count, faster walking speed, and higher heart-rate intensity than overground walking training. The addition of challenging mobility skill practice may offer increased opportunities to improve walking and balance skills. Here we compare walking and balance outcomes of chronic stroke survivors performing BWSTT with BWSTT including challenging mobility skills.

Methods

Single-blind randomized clinical trial comparing two BWSTT interventions performed in a rehabilitation research laboratory facility over six weeks. Participants were 18+ years of age with chronic (≥5 months) poststroke hemiparesis due to a cortical or subcortical ischemic or hemorrhagic stroke and walking speeds <1.1 m/s at baseline. A hands-free group (HF; n=15) performed BWSTT without assistance from handrails or assistive devices, and a hands-free plus challenge group (HF+C; n=14) performed the same protocol while additionally practicing challenging mobility skills. The primary outcome was change in comfortable walking speed (CWS), with secondary outcomes of fast walk speed (FWS), six-minute walk distance, Berg Balance Scale (BBS) scores, and Activities Specific Balance Confidence (ABC) scores.

Results

Significant pre-post improvement of CWS ($Z=-4.2$, $p\leq 0.0001$) from a median of 0.35 m/s (range 0.10 to 1.09) to a median of 0.54 m/s (range 0.1 to 1.17), but no difference observed between groups ($U=96.0$, $p=0.69$). Pre-post improvements across all participants resulted in reclassified baseline ambulation status from sixteen to ten household ambulators, three to seven limited community ambulators, and ten to twelve community ambulators. Secondary outcomes showed similar pre-post improvements with no between-group differences.

Conclusions

The addition of challenging mobility skills to a hands-free BWSTT protocol did not lead to greater improvements in CWS following six weeks of training. One reason for lack of group differences may be that both groups were adequately challenged by walking in an active, self-driven treadmill environment without use of handrails or assistive devices.

Key words: poststroke rehabilitation; walking; balance; body-weight-support treadmill training; robotics; hemiparesis; mobility skills

Trial registration #: NCT02787759 Falls-based Training for Walking Post-Stroke (FBT); retrospectively registered June 1st, 2016

INTRODUCTION

A major focus of stroke rehabilitation is to improve walking function to a safe level for community ambulation. In particular, body-weight-supported treadmill training (BWSTT) is a technique that yields moderate improvements in walking function for individuals in both subacute and chronic phases poststroke.[1][2][3][4][5] The most recent Cochrane Review[6] determined that BWSTT increased walking velocity (i.e., 0.06 m/s on the 10-m walk test) and endurance (i.e., 14.19 m on the six-minute walk test); however, it did not improve walking function to a greater extent than other interventions. One essential requirement for functional walking is successful navigation of everyday walking challenges, which require mobility skills that allow a person to avoid, or recover from, balance disruptions.[7][8] However, few BWSTT protocols include opportunities to navigate walking challenges (e.g., obstacles, distractions, etc.). Protocols that do include challenges generally incorporate them during an overground training component. For example, the LEAPS trial[9] and other smaller studies[3][10][11] incorporated overground mobility skills into their protocols and saw large improvements in walking speed (i.e., >0.2 m/s).

Mobility skill practice may be essential for walking improvements; however, combining BWSTT with overground training makes it difficult to discern whether improvements in walking speed should be attributed to the BWSTT or overground component. BWSTT protocols also commonly allow handrail use and involve manual assistance from a therapist or robot[9][10][12][13] further making it difficult to determine which components of a training protocol are critical for improvements. Thus, we investigated the efficacy of challenging skill practice during walking rehabilitation by incorporating it into one of two comparable protocols and further isolated the effects of skill practice through *eliminating use of handrails or assistance from both protocols*. We focused solely on a treadmill paradigm since it offers advantages over training overground for such challenging skill practice, including safety, minimal space requirements, and the ability to repetitively perform a high number of skill repetitions without interruption.[14][15]

The purpose of this single-blind, randomized controlled clinical trial was to compare walking and balance outcomes of two BWSTT protocols over a six-week period. A hands-free (HF) protocol involved only BWSTT, and a hands-free plus challenge (HF+C) protocol was equivalent except for the addition of challenging mobility skills. The selected skills encompassed seven of eight dimensions of community mobility proposed by Patla and Shumway-Cook. [7][8] We expected both groups to improve but expected that performing challenging mobility skills would improve overground CWS for the HF+C group to a greater extent than the HF group. We also expected that improvements would be greater for the HF+C group after a six-month follow-up period. We expected these findings because HF+C training offered opportunities for participants to develop new and improved balance and gait-control strategies and increased confidence in responding to balance disturbances, as compared with HF walking. We secondarily explored changes in fast walk speeds (FWS), six-minute walk distance, Berg Balance Scale (BBS) scores, and Activities-Specific Balance Confidence (ABC) scores.

METHODS

This study was a single-blind randomized clinical trial comparing walking and balance outcomes of two BWSTT interventions, HF versus HF+C, performed in a rehabilitation research laboratory facility over 6 weeks. We selected walking and balance outcomes of CWS, FWS, BBS, and ABC scores due to their strong clinical relevance, association with ambulation status after stroke, and expert panel recommendations.[16][17][18][19][20]

Power calculations

We performed a power calculation for the primary outcome variable, change in CWS, based on a repeated measures ANCOVA (covariate baseline walking speed) with two groups, $p \leq 0.05$ to reject the null hypothesis, 80% power, and 0.5 effect size to detect a gait velocity difference of at least 0.16 m/s - the proposed minimal clinically important difference.[21] Given these parameters the estimated sample size was 16 individuals per group.

Participants

We recruited participants from June 2012 through January 2015 until the end of our funding period and we completed data collection through follow-up in January 2016. We randomized thirty-nine individuals over 18 years of age with chronic (≥5 months) poststroke hemiparesis due to cortical or subcortical ischemic or hemorrhagic stroke, confirmed by computed tomography, magnetic resonance imaging, or clinical criteria (Figure 1). If participants were unable to complete the first week of training we removed them from subsequent analyses, because they had not experienced more than three sessions of the training protocol. This decision yielded N=29 individuals for our primary analysis; n=15 in the HF and n=14 in the HF+C group. We will refer to these numbers from this point on throughout the text.

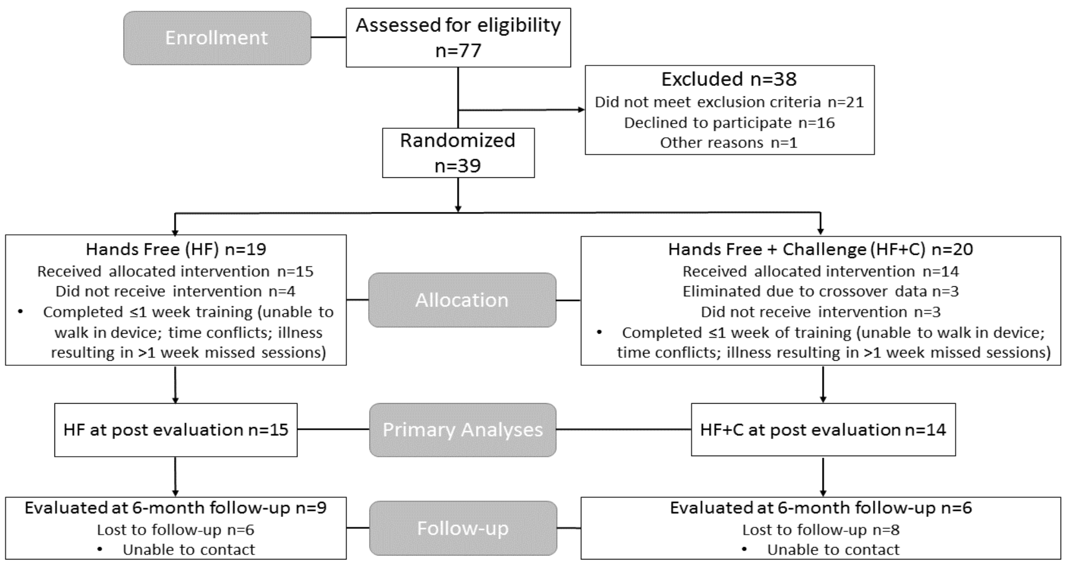

Figure 1. Consort diagram for participant flow through study.

Participants were medically approved for exercise, able to ambulate ≥14 m with or without an assistive and/or orthotic device but had slow CWS <1.1 m/s at baseline. Exclusion criteria were serious medical conditions; resting systolic blood pressure >180 mmHg and/or diastolic blood pressure >110 mmHg; resting heart rate >100 bpm; spasticity management including botulinum toxin injection (<4 months) or phenol block (<12 months) to the affected lower extremity; intrathecal or oral baclofen (<30 days); Mini-Mental State Exam score <24; currently undergoing lower-limb physical therapy; participation (<6 months) in long-term (>4 weeks) BWSTT, limb-loaded pedaling, or lower-extremity strengthening; plans to move out of area; and transportation barriers to study site.

Assessments

All participants gave informed consent as approved by the Institutional Review Board of the University of Alabama at Birmingham prior to initial assessment (protocol #: F120425008). We contacted a participant's physician regarding any concerns raised to determine appropriateness for study participation and to clarify any specific exercise precautions.

It was not practical for research staff overseeing training to be blinded to intervention group, but a blinded physical therapist conducted all assessments. The physical therapist assessed outcomes for CWS, FWS, six-minute walk distance, BBS, and ABC scores at baseline, mid study (i.e., 3 weeks; all except ABC), immediately post intervention, and six-months post intervention. Participants performed the 10-m walk in a straight hallway with no obstructions and instructions to "walk at the speed that feels most comfortable to you" for CWS and "walk at the fastest speed that you feel you can safely attain" for FWS. We determined CWS as the average of three trials and FWS as the fastest time achieved out of three attempts, with sufficient rest provided as necessary between trials. Participants performed the six-minute walk around an oval 85-ft walkway while the therapist followed with a Stanley distance wheel to record distance covered. We allowed participants to use an assistive device and/or ankle-foot orthosis as necessary. We considered the following changes in outcome measures to be clinically meaningful: 10-m walk 0.16 m/s;[21] six-minute distance 34.4 m;[22] BBS 4.13 points;[23] and ABC increase above 67% (fall-risk threshold).[24] We used the Stroke Impact Scale (SIS), Dynamic Gait Index (DGI), and Geriatric Depression Scale (GDS) to further characterize participants.

Randomization to study groups

We stratified randomization by baseline CWS (severe <0.5 m/s; moderate ≥0.5 m/s). We used a random number generator (www.random.org) to generate two lists of "0" and "1" sequences, one for each stratification. As each sequential participant entered the study, we allocated the next "0" or "1" value to that person. We designated "0" for the HF group and "1" for the HF + C group. The lab coordinator sequentially enrolled participants and the principal investigator assigned qualified participants to groups prior to initiating training. Using this procedure, we successfully achieved balanced groups on baseline walking behaviors (e.g., speed classification, use of overground assistive devices, orthoses, etc. (Table 1)). We scheduled training sessions such that one participant performed training at a time, minimizing exposure between groups. We did not inform participants about the a priori expectation of greater walking improvements for participants who performed mobility skills during training.

Table 1. Classification of baseline walking measures for each group

Walking Behavior	Hand Free n=15	Hands Free + Challenge n=14	p value
Walking speed severity (severe <0.5; moderate≥ 0.5)	7 moderate 8 severe	6 moderate 8 severe	0.84
Community Walking Status	8 HH; 2 LC; 5 C	8 HH; 1 LC; 5 C	0.86
Assistive Devices used for overground assessments	n=6 none n=2 single-point cane n=5 quad cane n=1 hemi walker n=1 other	n=5 none n=5 single-point cane n=2 quad cane n=2 hemi walker	0.41
Ankle-Foot Orthosis	n=6 none n=5 rigid plastic, no joint n=3 rigid plastic, joint n=1 external ankle support	n=11 none n=2 rigid plastic, joint n=1 metal double upright with joint	0.07
Functional Ambulation Classification	n=3 Dependent, Level II n=3 Independent, level surfaces only n=9 Independent, level and non-level surfaces	n=1, Dependent, supervision n=1, Independent, level surfaces only n=12, Independent, level and non-level surfaces	0.14

**p values for chi square comparisons between groups; household (HH); limited community (LC); community (C))*

Training Environment

All walking training occurred at the University of Alabama at Birmingham Locomotor Control and Rehabilitation Robotics Laboratory over a Bertec treadmill while supported by a self-driven robotic device called the KineAssist.[25][26] The KineAssist provided different levels of BWS at the approximate body center of mass through a hip/pelvis interface and maintained prescribed BWS during all phases of the gait cycle. The KineAssist operates via “cobotics”, which is software that senses human movement and allows devices to take direction from this movement. Participants interacted with the

KineAssist via a pelvic mechanism equipped with force transducers that sense forces being applied by the individual. The cobotics software sent these forces to the treadmill control panel, which in turn commanded the belts to run at a given speed based on a predetermined force-velocity relationship.

In this manner, the KineAssist allowed "self-driven" or intentional movement in six degrees of freedom, including translational movement in three perpendicular axes: surge (forward/backward movement over the treadmill), heave (vertical center of mass motion), and sway (side-to-side translation); and rotational movement about three perpendicular axes: roll (hip hiking), pitch (forward/backward tilting), and yaw (left/right rotation). A torso harness provided additional support of upright orientation by preventing extreme forward lean in the event of a loss of balance. Both training groups walked without the use of handrails or assistive devices, which were unnecessary given the KineAssist's safety mechanisms. When participants lost their balance, the device caught them after a short descent, and the clinician/researcher assisted them into a standing position to continue training with little interruption. We also discouraged participants from using the clinician/researcher for support.

Intervention Protocols

Intervention protocols adhered to physical activity recommendations (i.e., 20 – 60 minutes of aerobic exercise ≥3 days/week) of the American Heart Association for stroke survivors.[27] Interventions comprised eighteen sessions of 30-minute walking over six weeks. Participants received alternate rest days to minimize excessive fatigue (training

Monday, Wednesday, and Friday). We rescheduled missed sessions to occur on one of the two free days in the weekly schedule.

Two trained research staff oversaw each training session. Sessions began with participants performing a BWS test over the treadmill that involved timed 10-m walks at three or four levels of BWS (0, 10, 20, and/or 30%) depending on height constraints and/or comfort with higher BWS. We selected the lowest BWS per session that resulted in the participant walking ≥0.08 m/s (minimal detectable change)[21] faster than with 0% support. In some cases, 0% BWS facilitated the fastest training speed. This approach differed from other BWSTT protocols that generally started training with high BWS (30 - 40%) and progressed to lower levels.[1][5][9][10][13] However, the approach used in this study allowed participants to train with BWS that best facilitated their fastest CWS on a session-by-session basis.

We asked all participants to walk at a target intensity of 60 – 80% heart rate reserve based on the Karvonen formula.[28] We calculated max heart rate as 220 – age and then used the Karvonen formula (target heart rate = ((max heart rate − resting heart rate) × % intensity) + resting heart rate) to determine the training range for each participant. We adhered to this target range in order to ensure comparable intensity between training groups. We monitored heart rate and Borg Scale rating of perceived exertion (RPE) each minute (heart rate) and every other minute (RPE) to document participants' actual and perceived exertion, respectively. Participants walked with their selected BWS at a speed that maintained their heart rate within the target range for six, five-minute bouts. We required participants to take brief standing breaks if their heart rate exceeded 80% heart rate reserve and allowed them to take voluntary breaks if necessary;

however, encouraged them to continue walking as soon as possible. The clock continued running for both heart rate and voluntary breaks. Every five minutes we offered participants a seated break; however, as participants progressed with training we encouraged them to walk continuously and some individuals attained 30 continuous minutes of walking by the end of six weeks. We additionally tracked number of steps with a step watch (Orthocare Innovations) worn on the nonparetic limb and distance traveled with a Stanley distance wheel placed over the edge of the treadmill.

During each *HF+C protocol session*, participants walked for the same 30-minute training period but in addition to walking performed three of nine mobility skills for 10 minutes per skill (Table 2) that we randomized such that participants practiced all nine skills over three successive sessions. We adjusted the level of challenge for each participant to allow for failed attempts. These challenging mobility skills encouraged participants to learn how to avoid losing balance or required individuals to adapt to changing conditions. All skills encompassed Patla & Shumway-Cook's proposed mobility dimension of minimum walking distance,[7][8] as participants were encouraged to take as many steps as possible while performing each skill. We did not include one of the dimensions (i.e., traffic considerations) for practical reasons related to the training environment. We have provided the other dimensions of mobility addressed by each skill in Table 2.

Table 2. Nine walking skills experienced by participants in the HF+C group.

Task	Delivery mechanism	Mobility dimension
Hurdles	A pliable hurdle positioned in front of participant's paretic & nonparetic limb for 5 min each; challenge increased through raising hurdle height	terrain characteristics
Foam Shoes	3- or 6-in. foam blocks strapped to the bottom of participants' shoes; challenge increased by using thicker foam or walking faster	terrain characteristics
Backward Walking	Participants walked backward; challenge increased by faster speeds	postural transitions
Head Turns	Participants asked to look in 4 different directions (i.e., up, down, left, or right) for 10s each; increased challenge through nodding head up/down & left/right for 10s each	attentional demands
Backward Perturbation	Participants exposed to sudden brief accelerations of the treadmill belt, of sufficient magnitude to disturb forward progression; challenge increased through larger accelerations	postural transitions; external physical loads
Speed Up/ Slow Down	Participants walked at double their CWS for 10s & then recovered at their CWS for 20s; challenge increased if CWS increased	time constraints; postural transitions
Variable Speeds	Participants walked at a variety of randomly selected speeds between 0.2 m/s < daily CWS and 0.2 m/s > daily CWS; challenge increased if CWS increased	postural transitions
Narrow Stepping	Participants walked in ambient lighting conditions while keeping their feet within a narrow area of the treadmill belt designated by laser light beams; challenge increased through narrowing lights	terrain characteristics; ambient lighting
Long Stepping	Participants walked in ambient lighting conditions while taking long steps with both feet over a laser beam placed in front of them; challenge increased by moving laser farther away	terrain characteristics; ambient lighting

Statistical analyses

We used SPSS version 24 to conduct all statistical tests and used $p \leq 0.05$ to define statistical significance. We compared baseline characteristics with unpaired t tests (continuous variables) and chi square tests (categorical variables). We examined outcome data for completeness and discovered that 13.3% (n=2) of our post measurement outcomes were missing for the HF group and 14.2% (n=2) for the HF+C group. We therefore used a last measurement carried forward imputation approach and used the mid-assessment data point for these participants (noted in Figure 2).

We next examined correlations between variables to determine if baseline CWS was significantly related to change in CWS. We did not find a significant relationship between these variables for either group (HF $r=0.28$, $p=0.45$; HF+C $r=0.12$, $p=0.57$); thus, we did not include baseline CWS as a covariate in analyses.

We conducted Shapiro-Wilks tests for each group for change in CWS and discovered a significant departure from normality for the HF group ($p=0.004$); thus, we conducted a Mann-Whitney U test to determine if there were differences between groups. In the absence of group differences, we collapsed participants into a single group and checked pre and post CWS for normality. Both analyses revealed significant departures from normality ($p=0.001$ & $p=0.004$ respectively); thus, we conducted a Wilcoxon Signed Rank test to determine if participants significantly improved CWS pre to post assessment. We evaluated changes in community walking classification with a chi square analysis.

At six-month follow-up, we were missing 40% (n=6) of CWS data for the HF group and 57% (n=8) for the HF+C group. We therefore chose to analyze only complete data to determine if groups differed in CWS from post measurement to six-month follow-up. The complete data set was normally distributed (Shapiro Wilk $p>0.05$); thus, we used a repeated measures ANOVA to test for main effects of group and time.

For exploratory analyses on our secondary outcome variables, we conducted a repeated measure MANOVA to determine if there were main effects of group or time (pre - post) for FWS, six-minute distance, BBS, or ABC scores.

RESULTS

Characterization of participants at baseline

Although we randomly allocated participants to each intervention, groups significantly differed in age (HF+C was younger) and geriatric depression scale (HF exhibited greater depression) (Table 3). However, upon further analysis neither of these variables were significantly related to change in CWS for both groups. We found no significant differences between groups for any other measures.

Table 3. Participant baseline characteristics

Participant Characteristic	HF (n=15)	HF+C (n=14)	p value
Age (years), mean (SD)	60.3 (12.8)	48.9 (14.4)	0.03*
Gender, # males	7	8	0.57
Time since stroke (months), mean (SD)	47.7 (64.7)	52 (71.4)	0.87
Side of hemiparesis, # Left	9	11	0.28
Fugl-Meyer, mean (SD)	18.9 (5.1)	18.9 (6.5)	0.99
ABC (%), mean (SD)	56 (30)	73 (15)	0.09
GDS, mean (SD)	8.4 (6.7)	3.9 (3.5)	0.03*
DGI, mean (SD)	13.7 (6.0)	16.6 (4.5)	0.16
SIS mobility, mean (SD)	35.8 (8.5)	39.8 (4.4)	0.13
SIS ADL, mean (SD)	36.9 (10.3)	40.3 (6.2)	0.30
SIS participation, mean (SD)	28.8 (8.0)	30.4 (5.1)	0.52
CWS, mean (SD)	0.53 (0.38)	0.52 (0.32)	0.95
FWS, mean (SD)	0.71 (0.46)	0.77 (0.51)	0.76
6-minute distance, mean (SD)	182.2 (131.9)	193.9 (113.7)	0.80
BBS, mean (SD)	42.5 (10.7)	45.9 (9.3)	0.37

$p<0.05$; ABC=Activities Specific Balance; GDS = Geriatric Depression Scale; DGI=Dynamic Gait Index; SIS=Stroke Impact Scale; CWS=Comfortable Walk Speed; FWS=Fast Walk Speed; BBS=Berg Balance Scale

Participants did not experience any adverse effects related to the training protocols. Results of the Mann-Whitney U test for change in CWS between groups revealed that groups were not significantly different (*U=96.0, p=0.69)*. However, collapsed into a single group (N=29) participants significantly improved (*Z=-4.2, p≤0.0001*) CWS pre to post assessment with an effect size *r=0.55*, from a median of 0.35 m/s (range 0.10 to 1.09) to a median of 0.54 m/s (range 0.1 to 1.17), as indicated by the Wilcoxon signed-rank test (Figure 2). Using the minimal clinically important difference of 0.16 m/s to classify participants as either "responders" or "nonresponders" as others have done,[10] there were five participants classified as responders (i.e., two HF and three HF+C).

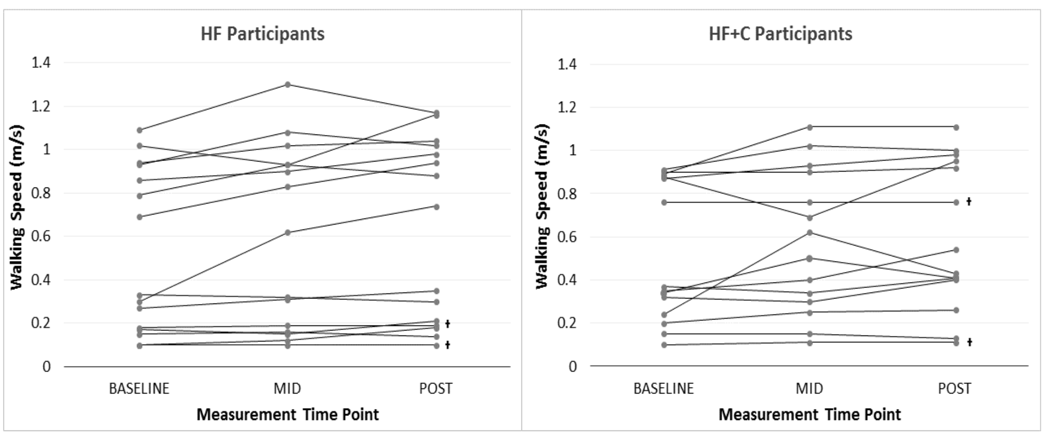

Figure 2. Baseline, mid, and post CWS measurements for n=15 participants in the HF group (left) and n=14 HF+C group (right). Participants with the midpoint measurement carried forward are marked with a † at post.

Improvements in CWS resulted in reclassification of walking ability post intervention. At baseline, there were sixteen participants with ambulation level classified as household (<0.4 m/s), three as limited community (0.4 – 0.8 m/s), and ten as community (>0.8 m/s). Post intervention, these numbers changed significantly (χ^2 *(4, N = 29) = 27.2, p <0.0001)*, to ten household, seven limited community, and twelve community ambulators.

For participants with complete data (n=9 HF; n=6 HF+C), the repeated measures ANOVA revealed that groups did not significantly differ in CWS from post assessment to six-month follow-up (*p=0.58*). We also did not detect a change in CWS from post to follow-up assessment (*p=0.76*), with a HF group (post *M=0.75 m/s; 95% CI 0.47 to 1.03* vs. follow-up *M=0.78 m/s*; *95% CI 0.49 to 1.06)* and a HF+C group (post *M=0.67 m/s*; *95% CI 0.33 to 1.01* vs. follow-up *M=0.63 m/s; 95% CI 0.28 to 0.97*).

The exploratory repeated measure MANOVA for secondary outcomes (n=15 HF; n=14 HF+C) revealed similar findings to that of CWS. Although participants collectively improved performance in these outcomes pre to post assessment (Table 4), we did not detect a main effect of group for any variable (*p=0.08 to 0.87*).

Table 4. Changes in secondary outcome measures pre to post assessment.

Measure	Group	Pre Mean (95% CI)	Post Mean (95% CI)	Effect size η^2	Sig. change pre to post
FWS (m/s)	HF (n=15) HF+C (n=14)	0.68 (0.42 to 0.93) 0.77 (0.50 to 1.04)	0.78 (0.49 to 1.07) 0.88 (0.58 to 1.19)	0.46	p≤0.0001
6-minute distance (m)	HF (n=15) HF+C (n=14)	182.2 (116.8 to 247.6) 193.9 (126.2 to 261.6)	221.1 (146.6 to 295.5) 225.6 (148.5 to 302.7)	0.43	p≤0.0001
BBS (points)	HF (n=15) HF+C (n=14)	42.5 (37.2 to 47.9) 45.9 (40.4 to 51.5)	45.1 (39.6 to 50.6) 47.6 (41.9 to 53.3)	0.21	p=0.01
ABC (%)	HF (n=15) HF+C (n=14)	56.1 (43.7 to 68.6) 71.9 (59.0 to 84.7)	61.8 (51.6 to 72.0) 74.3 (63.7 to 84.9)	0.14	p=0.05

FWS=fast walk speed; BBS=Berg balance scale; ABC=activities specific balance

DISCUSSION

We sought to determine whether HF+C BWSTT was superior as compared to HF BWSTT in improving walking outcomes for individuals with chronic poststroke hemiparesis. Contrary to our primary hypothesis there were no differences in walking outcomes between groups. It is possible that the challenge of walking hands free was a strong enough stimulus to elicit walking and balance improvements, regardless of group assignment. Nine of fifteen participants in the HF group and nine of fourteen in the

HF+C group required assistive devices to perform overground walking assessments. However, we did not permit participants to use any form of assistive device during training. Thus, it is likely that walking without handrails over the self-driven treadmill sufficiently challenged dynamic balance of both HF+C and HF participants leading to improvements in overground walking outcomes.

Given that participants in both groups improved walking function, our findings demonstrated that stroke survivors can benefit from training involving challenging walking conditions. Recent recommendations for neurorehabilitation included training that involves many repetitions and continues to challenge the patient.[29] Training environments that incorporated skilled movements appeared to induce better and longer lasting improvements in motor function following cerebrovascular injury.[29][30][31][32][33][34][35][36] The nine mobility skills used in this study offered participants this experience; however, practicing these skills was not essential to engender improvements in walking function for these chronic stroke survivors.

While this finding could be considered surprising, both of our training protocols relied on the principle of active participant involvement. Winstein and Kay (2015)[37] authored an elegant review of learning-dependent neuroplasticity following neurological injury. They highlighted that repetitive tasks alone are insufficient for motor learning to occur and that instead training must involve problem solving and be goal-directed, task-specific, and challenging, but doable. Both of our protocols adhered to these principles. The research staff did not assist participants to move their limbs through the gait cycle, did not intervene to prevent losses of balance, and did not provide cues as to the “best” way to walk or perform a task. We instead allowed participants to develop their own movement strategies, learn from unexpected events, and problem-solve on their own to prevent similar occurrences. Both training groups walked in an active “intention-driven” environment with the same goal of maintaining their heart rate within the prescribed zone. Thus, all participants were provided with opportunities to make active adaptations to their walking pattern.

While groups improved CWS, average gains did not reach the minimal clinically important difference of 0.16 m/s. [21] However, 0.16 m/s is used as criteria for the subacute phase after stroke and there is no agreed upon clinically important difference for walking speed in the chronic phase.[20] One reason why we did not observe larger increases using these protocols may be that we allowed participants to adopt their CWS during training. Studies with greater improvements in overground walking speed have instead used “maximum tolerable speed”.[5][19] Additionally, our protocols were only six weeks and studies that noted larger improvements were generally longer in duration (e.g., 12 to 16 weeks).[9][10] Our observed speed improvements did, however, change the community ambulation status of many participants. We saw more limited community as compared to household ambulators following training.

Limitations

We acknowledge several limitations to the present study, including small sample sizes short of a priori power estimates and a large amount of data missing at follow-up. Additionally, our research question was designed to evaluate the effects of challenging mobility skills during treadmill training. Thus, we cannot address whether overground training plus mobility skills could have led to greater increases in overground walking speeds. The experimental design may also have been strengthened by an additional group that performed only challenging mobility skills without walking practice.

Regarding our lack of group differences in CWS, the HF+C group performed each mobility skill once per week for 10 minutes, which might have been insufficient dosing to yield greater improvements. While we ensured that each individual was challenged, we did not adapt skills to meet each individual's unique requirements; thus, an individualized approach might have elicited greater gains. We must also consider that CWS might not be the best outcome measure to detect the types of improvements the HF+C group may have experienced. All participants were in the chronic phase after stroke; however, treadmill training is shown to be effective in helping these individuals past "plateaus" in walking function.[5] While we had a wide range of time since stroke represented, our groups were well-balanced on this characteristic. Finally, we did not specifically exclude participants with minor musculoskeletal conditions commonly experienced poststroke (e.g., heel spur, plantar fasciitis, mild arthritis) so long as these conditions did not prevent engagement in training. While including these individuals yielded better ecological validity, it is possible that these comorbidities limited gains in walking function.

Conclusions

The addition of challenging mobility skills to one of two comparable BWSTT protocols did not lead to greater improvements in walking and balance outcomes for individuals in the chronic phase poststroke. However, participants collectively improved CWS pre to post intervention resulting in changes in community ambulation status. These findings suggest that stroke survivors are capable of performing, and responding to, walking training that offers no assistance via handrails or therapist and may additionally incorporate essential skills related to competent walking function. In order to engender larger improvements in walking and balance outcomes poststroke future studies may consider incorporating challenges into walking interventions in combination with other promising strategies like task-specific training, individualized exercise prescription, and high-intensity training.

List of abbreviations

Body-weight-support treadmill training (BWSTT); Hands free (HF); Hands free + Challenge (HF+C); Comfortable walk speed (CWS); Fast walk speed (FWS); Berg Balance Scale (BBS); Activities Specific Balance Confidence (ABC); Stroke Impact Scale (SIS); Dynamic Gait Index (DGI); Geriatric Depression Scale (GDS)

Ethics approval and consent to participate

All participants gave informed consent as approved by the Institutional Review Board of the University of Alabama at Birmingham prior to initial assessment (protocol #: F120425008).

Consent for publication

All participants gave informed consent to the following statement, “Your information from this research may be published for scientific purposes; however, your identity will not be given out.”

Availability of data and materials

The datasets generated and/or analyzed during the current study are available in the Mendeley repository “Dataset for: Walking and balance outcomes for stroke survivors: A randomized clinical trial comparing body-weight-supported treadmill training with versus without challenging mobility skills", [http://dx.doi.org/10.17632/r9f657nhsp.1].

Competing interests

David A. Brown receives royalties on sales of the KineAssist device. Sarah A. Graham and Elliot J. Roth declare no conflicts of interest.

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Author's contributions

ER and DB designed this clinical trial and secured funding. SG collected, analyzed, and interpreted all outcome data related to this study and composed the majority of the manuscript. ER and DB contributed content to the manuscript. All authors read and approved the final manuscript.

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MINIMIZING POSTURAL DEMANDS OF WALKING WHILE STILL EMPHASIZING LOCOMOTOR FORCE GENERATION FOR NONIMPAIRED INDIVIDUALS

by

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ABSTRACT

In motor control studies the interdependent nature of the neural controllers for posture and locomotion makes it difficult to separate components of stepping control from postural maintenance functions. To better understand the separate influences of postural versus locomotor control during walking, we fabricated a novel postural support apparatus. This apparatus was intended to minimize postural demands of walking, but allow for matched locomotor force generation, thus isolating the control of stepping. We tested the ability of this support apparatus to minimize postural demands of walking tasks for nonimpaired participants (N=20) and characterized the behavior of these participants when walking in this environment. We demonstrated that the apparatus reduced trunk motion in flexion/extension, lateral flexion, and transverse rotation, minimized peak vertical ground reaction forces to 15.8% body weight, and reduced total positive and negative work compared to walking with typical postural demands. Additionally, using visual feedback, participants were able to successfully match vertical forces during supported walking to those of walking with typical postural demands. We plan to use this apparatus to design future experiments exploring mechanisms underlying postural and locomotor control in both nonimpaired walking and of individuals with impaired coordination of posture and stepping.

Index Terms— walking, posture, robotics, biomechanics, nonimpaired

I. INTRODUCTION

Functional walking is complex from a control perspective, as it requires stabilizing posture and moving the limbs while maintaining dynamic equilibrium [1]. Posture and locomotor control mechanisms must maintain upright orientation and support body segments against gravity and accelerations while also facilitating rhythmic movements of stepping [1][2]. Massion (1992) [2] proposed that these tasks are enabled by separate neural controllers for posture and movement that act interdependently to influence spinal neuronal networks and facilitate coordinated locomotor output. In neurologically nonimpaired individuals these separate, central controllers for posture and locomotion act in concert to produce functional walking patterns. However, the very interdependent nature of these postural and locomotor control systems makes it difficult to investigate the control of stepping without the confounding effects of posture.

If the integration of posture and stepping during walking indeed reflects separate control systems, we should theoretically be able to isolate stepping functions from postural functions during the gait cycle. In the direction of progression (sagittal plane) the output of the postural and locomotor controllers is reflected in lower-limb power absorption to decelerate the body center of mass (COM) and generation to accelerate the body COM [3]. Power absorption at the ankle and knee in early stance and the hip in late stance functions to control forward momentum of the body and provide support against limb collapse (postural functions) [4]. Power generation also has a role in postural support, particularly at the hip during early stance; however, the largest power generation occurs at the ankle in late stance to propel the body COM forward (locomotor function) [5]. As muscles and other soft tissues act to both absorb and generate power during walking this lower-limb work results in ground reaction forces (GRFs) during the stance phase [6]. The stereotypical bimodal shape of the vertical GRF and alternating negative (braking) and positive (propulsive) components of the fore-aft GRF also reflect components of support and progression [7][8][9]. Theoretically, if we could minimize the need to control posture during walking we could better isolate the control of stepping, which should be reflected in a reduced need to perform the aforementioned postural functions.

There have been previous attempts to minimize postural demands of locomotor tasks in order to investigate the control of locomotion without the confounding effect of posture [10]. Liang and Brown (2013) [10] used a pedaling paradigm with minimal postural demands to isolate locomotor behavior of individuals poststroke. They observed inappropriately directed shear forces against the pedal during a nonseated, "posturally loaded" task, but improved foot-force direction during a seated, "minimal posture" task. These results have yet to be replicated in a more functionally relevant walking paradigm. Methods for minimizing postural demands of walking relative to stepping demands include body-weight support and air stepping [11][12][13][14][15][16], which only partially address aspects of postural control. While body-weight support indeed minimizes the need to support the body against gravity, it does not provide full support of upright trunk orientation. Some air stepping paradigms do offer support of body orientation by fully supporting the trunk in an upright [14][15] or side-lying [15][16] position; however, the absence of ground contact forces in such configurations alters kinematics of walking in unpredictable ways [14] and feedback from load-sensing afferents is critical in facilitating gait transitions [17]. A paradigm that minimizes postural demands relative to locomotor requirements should provide support of upright orientation against body COM accelerations and vertically support body weight against collapse, while still allowing for ground contact forces that closely mimic those experienced during typical walking.

We fabricated a postural support apparatus intended to fulfill these requirements. In this first demonstration our aim was to test the ability of this support apparatus to minimize the postural demands of walking for nonimpaired individuals, which we operationally define as the need to maintain upright orientation, support body weight, and control body COM accelerations through power absorption and generation. We expected that the support apparatus would achieve these objectives as evidenced by:

1) Reduced trunk motion during walking within the apparatus compared to typical walking;
2) Vertical GRFs minimized to no more than limb weight (~18 - 20% body weight) [18] when walking in the support apparatus, and vertical GRFs similar to those that occur during typical walking when participants were provided with specific instructions to generate vertical force using physical effort rather than body weight during the stance phase; and
3) Reduced total lower-limb power absorption and generation when walking in the support apparatus compared to typical walking, and similar power generation, but minimal power absorption when generating matched vertical forces.

II. METHODS

Twenty healthy, nonimpaired individuals (11 females, 9 males) mean age 26.8 years (SD=4.9) participated in this study. Participants filled out a Physical Activity Readiness Questionnaire (PAR-Q) [19] to ensure that they were safe to participate in physical activity. Exclusion criteria were history of neurologic and/or musculoskeletal disorders that could affect postural control or walking function. All participants signed informed consent as approved by the Institutional Review Board of the University of Alabama at Birmingham.

A. Typical treadmill and robotic walking environments

Participants walked in one treadmill condition considered “typical” (typical requirements for controlling upright orientation, COM accelerations, and supporting body weight) over a motorized, dual-belt, instrumented treadmill (BERTEC, Columbus, OH, USA). This condition served as a control. Next, participants walked while supported only by the KineAssist™ robotic device (HDT Global, Solon OH) (Fig. 1A) [20]. We used this device as a base apparatus upon which we built the postural support apparatus. We felt that it was essential to demonstrate how this robotic device impacted walking prior to the additional support apparatus features. Individuals interacted with the KineAssist via a pelvic mechanism that can allow all six degrees of freedom while walking. However, for this experiment we allowed three degrees of freedom, including surge (relative forward/backward movement over the treadmill), heave (vertical COM motion), and pitch (forward/backward trunk/pelvic tilting). With a locking mechanism we prevented sway (side-to-side pelvic translation), roll (hip hiking), and yaw (left/right pelvic rotation).

B. Minimal postural demand (supported) walking environment

We fabricated a postural support apparatus (Fig. 1B) that served three purposes, (1) externally stabilized the trunk (restricted trunk movement), 2) fully offloaded the trunk and upper body mass from the lower limbs, but still allowed the participant to make contact with the treadmill surface with their feet, and 3) minimized the need to control body COM accelerations by holding the participant in place. The pelvic mechanism of the KineAssist device completely supported the backboard component of the apparatus with no weight transmitted to the participant. We framed the backboard with 1.5-inch t-slotted aluminum framing with an additional support bar running down the center. A ⅜-inch stainless steel plate connected the frame and center bar to prevent rotations of the board. Two 180° pivot brackets with L-handles that allowed locking further connected the backboard to a 90° (with 45° support) t-slotted aluminum frame. We adjusted the vertical position of the brackets along the t-slotted framing to accommodate participants of different heights and we further used these brackets to lock the backboard in a small degree of forward inclination (5°) to accommodate natural trunk lean during walking [21]. Straight-strut steel channel connectors and six, 1-inch long socket-cap screws connected the horizontal portion of the 90° frame to the pelvic mechanism of the KineAssist. We fabricated the shoulder pads and surface that participants rested against out of 2.5-inch thick high-density foam over ¼-inch plywood encased in synthetic vinyl fabric. We adjusted the shoulder pad height via rigid 1-inch diameter aluminum tubing

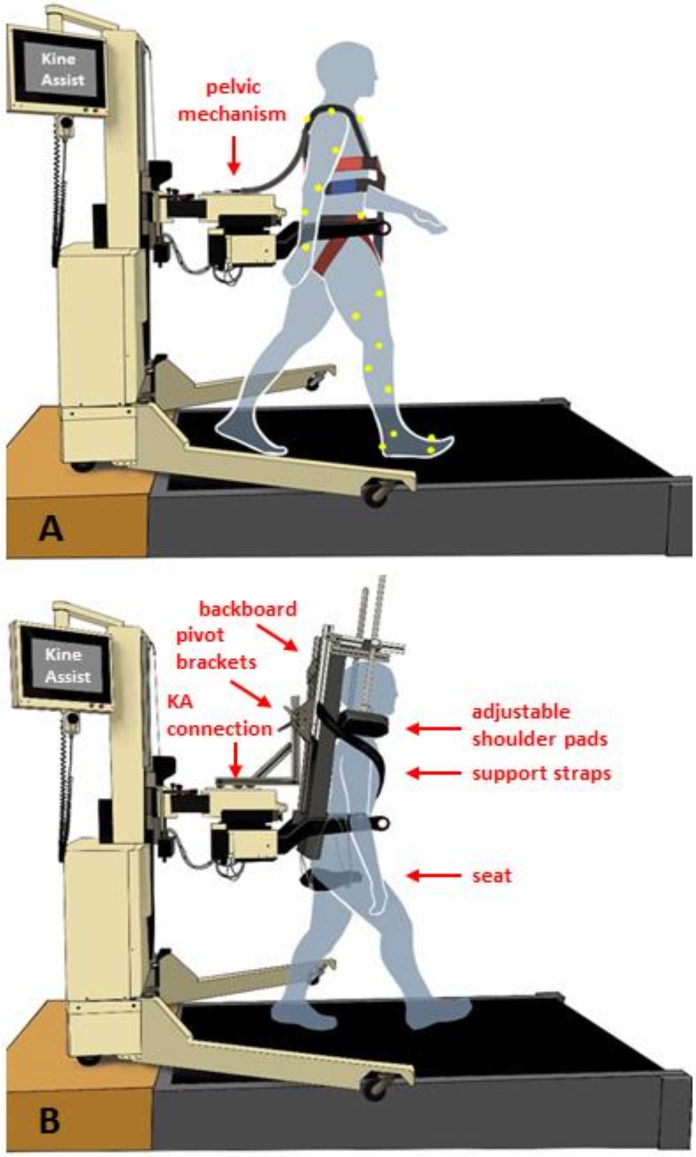

Fig. 1. Walking environment in A. the KineAssist robotic device and B. the support apparatus. Example kinematic marker set-up shown in yellow on right side of body in A; note sacral marker not visible.

held in tube holders with locking handles, and the anterior-posterior position of the tube holders to accommodate participants of varying chest and shoulder depth.

During supported walking trials the participant leaned against the padded backboard with support straps across their torso to hold them in place and minimize their need to actively control upright trunk orientation. A narrow seat provided full offloading of the trunk and upper body mass from the limbs without impeding limb movements. To achieve this configuration, we first raised participants into the air via the pelvic mechanism of the KineAssist so that their feet were not in contact with the treadmill. We then lowered them until the full surface of both feet just made contact with the surface of the treadmill. The support apparatus minimized linear and angular accelerations of the trunk and pelvis by locking all degrees of freedom of movement, including the shoulder pads to restrict vertical movement. It was essential to restrict vertical movement so that participants did not have to control vertical excursions of their body COM or support body weight with their lower limbs during supported walking.

The support apparatus enabled two unique walking conditions wherein participants were upright and in contact with the treadmill surface, yet did not need to control upright postural orientation, support body weight, or maintain body position in relation to the moving treadmill belt. Participants could walk either by stepping with minimal contact forces applied to the treadmill or be instructed to exert stance phase effort by instead pushing their foot against the moving treadmill belt during stance using the shoulder pads as leverage. In the latter case, we presented participants with a visual force target via an oscilloscope suspended at eye level that enabled matching of vertical forces to the typical and robotic walking conditions. We instructed participants to hit the force target during the stance phase of each step. We considered steps with a vertical GRF magnitude falling within ± 5% of this target successful. Participants performed two supported walking conditions, 1) supported walking without effort and 2) supported walking with effort.

C. Experimental set-up

Participants completed a total of twelve, 30-second walking trials at 1.0 m/s across the four walking conditions (three per condition). In order to ensure comparable spatiotemporal characteristics across trials we obtained the comfortable walking cadence for each participant during their walking trials over the treadmill and instructed participants to maintain this cadence, via auditory feedback from a metronome, during all walking trials. In order to match vertical GRFs for the support apparatus with effort condition we obtained the average second peak (push-off peak) vertical GRF from the dominant limb of participants during walking trials in the KineAssist device. We assessed limb dominance through asking participants which limb they would kick a ball with. We displayed this target vertical GRF value ± 5% on an oscilloscope for the dominant limb only. We instructed participants to hit the target by pushing their dominant foot against the moving belt beginning when their foot was directly beneath them during the stance phase. For the support apparatus without effort condition we instructed participants to take light steps (no pushing) against the moving treadmill belt. For all conditions we collected GRFs at 1000 Hz via the Bertec, dual-belt, force-instrumented treadmill and kinematic data at 100 Hz via an eight-camera, Qualisys motion capture system (Qualisys Inc., Gothenburg, Sweden) with 33, 1-cm passive-reflective markers placed bilaterally over anatomical landmarks of the trunk, arms, legs, and feet, with three markers defining each segment (example marker locations shown in Fig. 1A).

D. Data processing

We calculated all variables over either a full gait cycle (foot strike to ipsilateral foot strike) or stance phase (foot strike to ipsilateral foot off). We defined these gait events with a vertical GRF threshold of 1.5% body weight. All reported values are averages across either the full gait cycle or stance phase. We included an average of 69 steps (SE=1) per participant in analyses for the treadmill condition, 70 steps (SE=1) for the KineAssist condition, and 66 steps (SE=1) for supported walking without effort. During supported walking with effort we only included steps in analyses that fell within ± 5% of the vertical force target, which yielded an average of 32 steps (SE=2) per participant. We used Visual 3D (C-Motion, Germantown, MD, USA) to obtain joint angles and powers and performed all post processing of the data in MATLAB (Mathworks®, version R2016a).

E. Kinematic (spatiotemporal) variables

We calculated stride and stance durations over the full gait cycle and stance phase, respectively. Stride duration indicated the time elapsed from foot strike to ipsilateral foot strike and stance duration was the time elapsed from foot strike to ipsilateral foot off. We calculated stance phase limb angle (leading and trailing limb) as the angle between the vector connecting the dominant-limb lateral toe and ASIS marker and the laboratory’s vertical axis. Negative values indicate leading limb position at foot strike and positive values trailing limb position at foot off.

F. Kinematic (joint angle) variables

We calculated trunk range of motion relative to the pelvis in all three planes (flexion/extension, lateral flexion, and transverse rotation) for the full gait cycle using Visual 3D. We also present ensemble average angles for the ankle, knee, and hip joints for the full gait cycle of each walking condition.

G. Kinetic variables

We present a comparison of peak vertical GRF values and ensemble average vertical and fore-aft GRFs, normalized to body weight, for the full gait cycle. Visual 3D calculates joint powers as the product of the net muscle moment and joint angular velocity ($P = M \times \omega$). We normalized joint powers to body mass (W/kg) and integrated the area under the power-time curve to obtain mechanical work ($W = \int P \times dt$) (J/kg) where negative work indicates power absorption and positive work indicates power generation. We present positive, negative, and net work for each individual joint and conducted statistical comparisons on the total positive, negative, and net work summed across the joints.

H. Statistical analyses

We conducted statistical tests using SPSS version 24. We conducted three separate one-way, repeated measure MANOVAs (repeated across walking conditions) to compare spatiotemporal variables, trunk range of motion in all three planes, and total positive, negative, and net work. We conducted a one-way, repeated measure ANOVA for peak vertical GRF values. We employed Greenhouse Geiser corrections, as necessary, for violations of sphericity, a priori alphas of 0.05, and Bonferroni corrections for multiple comparisons. We provide figures for ensemble average joint angles and powers to aid in interpretation of work calculations.

III. RESULTS

A. Spatiotemporal characteristics across walking conditions

Participants maintained walking cadence across trials as evidenced by no significant differences ($p>0.05$) in average stride duration across walking conditions (Table I). We observed a main effect of condition for stance duration ($F(1.3, 25.0) = 99.6$, $p<0.0001$), average trailing limb angle at foot off ($F(1.5, 29.0) = 75.8$, $p<0.0001$), and leading limb angle at foot strike (F (1.9, 37.0 = 128.1, $p<0.0001$). Post hoc comparisons revealed that trailing limb angle was significantly smaller for the supported without effort condition compared to the other walking conditions (Table I). Leading limb angle was not significantly different between the two supported walking conditions.

TABLE I. Spatiotemporal characteristics across walking conditions. Values are mean [95% CI].

	Treadmill[1]	KineAssist[2]	Supported[3] with effort	Supported[4] without effort
Stride duration (s)	1.18	1.18	1.18	1.19
	[1.16 to 1.20]	[1.16 to 1.20]	[1.16 to 1.20]	[1.18 to 1.21]
Stance duration (s)	0.77 (2,3,4)	0.78 (1,3,4)	0.72 (1,2,4)	0.55 (1,2,3)
	[0.75 to 0.78]	[0.76 to 0.80]	[0.71 to 0.74]	[0.51 to 0.60]
Trailing limb angle (°)	23.2 (4)	23.0 (4)	23.2 (4)	14.8 (1,2,3)
	[22.3 to 24.1]	[22.1 to 24.0]	[22.0 to 24.5]	[12.8 to 16.7]
Leading limb angle (°)	-21.8 (2,3,4)	-19.9 (1,3,4)	-13.2 (1,2)	-14.4 (1,2)
	[-22.6 to -21.0]	[-20.7 to -19.2]	[-14.0 to -12.4]	[-15.8 to -13.0]

Note: Each condition is labeled with a superscript 1 to 4 in the top row. Superscripts next to means indicate significantly different pairwise comparisons at the Bonferroni corrected p<0.05 for each variable.

B. Support apparatus effect on trunk motion during walking

Trunk range of motion was reduced in all directions when walking in the support apparatus as compared to treadmill and KineAssist walking (Table II). We detected main effects of walking condition for trunk flexion/extension (F (1.4, 25.7) = 72.8, p≤0.0001), lateral flexion (F (1.6, 31.2) = 146.4, p≤0.0001), and rotation (F (1.7, 33.2) = 98.0, p≤0.0001).

TABLE II. Trunk range of motion across walking conditions. Values are mean [95% CI].

	Treadmill[1]	**KineAssist**[2]	**Supported**[3] **with effort**	**Supported**[4] **without effort**
Flexion/extension (°)	4.2 (2,3,4)	8.5 (1,3,4)	2.7 (1,2,4)	1.0 (1,2,3)
	[3.9 to 4.6]	[7.0 to 10.0]	[2.1 to 3.2]	[0.8 to 1.1]
Lateral flexion (°)	8.6 (2,3,4)	11.0 (1,3,4)	2.0 (1,2,4)	0.6 (1,2,3)
	[7.4 to 9.8]	[9.7 to 12.3]	[1.4 to 2.6]	[0.5 to 0.7]
Transverse rotation (°)	8.5 (2,3,4)	5.2 (1,3,4)	2.0 (1,2,4)	0.7 (1,2,3)
	[7.1 to 9.9]	[4.4 to 5.9]	[1.3 to 2.7]	[0.4 to 1.0]

Note: Each condition is labeled with a superscript 1 to 4 in the top row. Superscripts next to means indicate significantly different pairwise comparisons at the Bonferroni corrected p<0.05 for each variable.

C. Support apparatus effect on vertical GRFs

The support apparatus enabled participants to walk with minimal or matched vertical GRFs against the moving treadmill belt. We detected a main effect of condition as expected (F (1.4, 27.4) = 1712.6, p≤0.0001). The support apparatus minimized vertical GRFs to 15.8% body weight, 95% CI [12.7 to 19.0] when walking without effort (Fig. 2A).

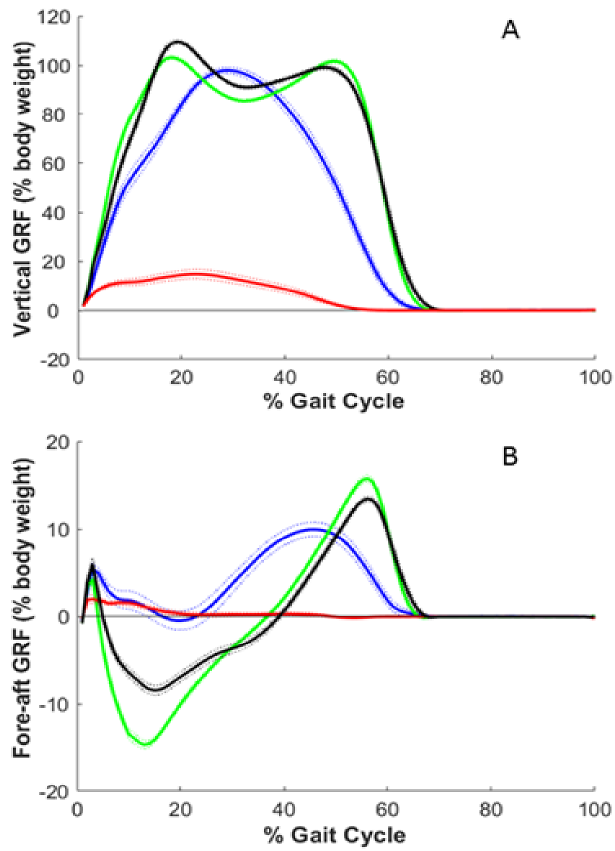

Fig. 2. Ensemble average A. vertical GRFs and B. fore-aft GRFs for each of the four walking conditions. Values are mean (SE). Treadmill (green); KineAssist (black); support apparatus with effort (blue); support apparatus without effort (red).

For the supported with effort condition participants took an average of 32 steps (SE = 2) within the required ± 5% range of their vertical GRF target. In this manner the support apparatus enabled participants to successfully match vertical GRFs between the support apparatus with effort condition and treadmill and KineAssist walking, as vertical GRF values (push-off peaks for treadmill and KineAssist walking) were not statistically different across conditions (treadmill 102.6% body weight, 95% CI [101.0 to 104.3], KineAssist 101.4% body weight, 95% CI [99.7 to 103.1], and supported with effort 101.0% body weight, 95% CI [98.8 to 103.2]). The profile of the vertical GRF, however, was different between the treadmill and KineAssist versus supported walking conditions.

The characteristic valley in midstance (visible for treadmill and KineAssist walking in Fig. 2A) that corresponds to the body COM reaching its highest position was not present. Additionally, the rate of loading and offloading to and from peak vertical force during supported walking with effort was appreciably longer than treadmill or KineAssist walking. Notably, for both supported walking conditions the fore-aft component of the GRF did not reflect braking force during the first half of stance (Fig. 2B). During the second half of stance the supported without effort condition also did not have propulsive force; however, the supported walking with effort condition did have a propulsive force component.

D. Joint angles and power trajectories during walking conditions

Power absorption and generation and accompanying range of motion at each lower-limb joint were generally smaller during supported without effort walking compared to treadmill and KineAssist walking (Fig. 3A-F). Peak ankle plantarflexion

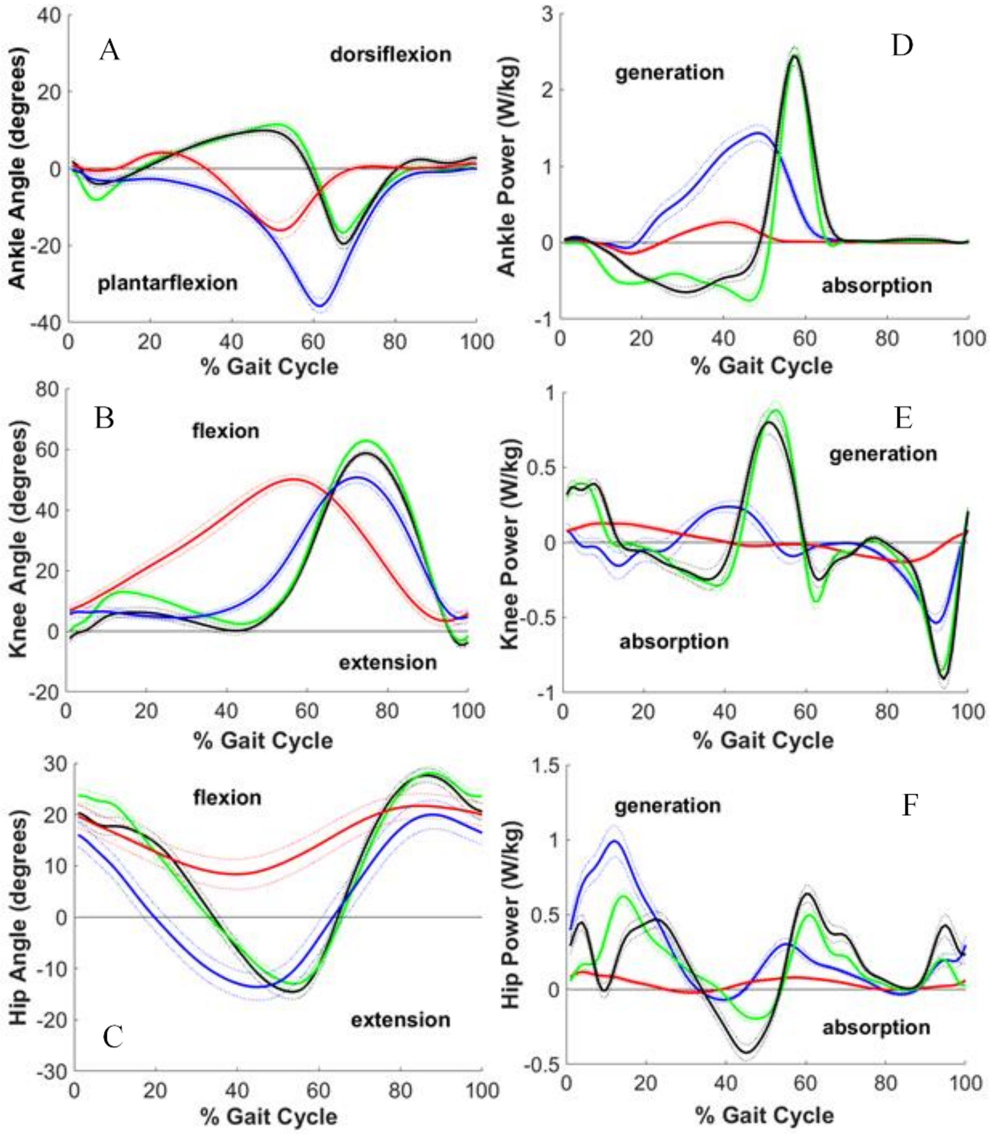

Fig. 3. Left column ensemble average joint angles for A.) ankle, B.) knee, and C.) hip joints, normalized to 100% of the full gait cycle. Right column ensemble average power curves for D.) ankle, E.) knee, and F.) hip joints, normalized to 100% of the full gait cycle. Values are mean (SE). Treadmill (green); KineAssist (black); support apparatus with effort (blue); support apparatus without effort (red).

was much larger in the supported with effort condition as compared to treadmill and KineAssist walking (Fig. 3A); however, the range of motion at the ankle joint was similar for these conditions. The ankle dorsiflexed throughout the first half of the stance phase during treadmill and KineAssist walking and moved from a position of peak dorsiflexion to peak plantarflexion in late stance, necessitating fast angular velocity and a corresponding large peak in ankle power generation (Fig. 3D). In contrast, the ankle started in a plantarflexed position during supported with effort walking and continued to plantarflex throughout the stance phase, allowing a slower angular velocity and lower peak power production.

E. Total work performed during walking conditions

Total positive and negative work were reduced during supported walking without effort as compared to the other three walking conditions (Fig. 4). We detected main effects of walking condition for total positive $F(1.6, 31.1) = 105.6$, $p \leq 0.0001$, negative $F(3, 57) = 366.7$, $p \leq 0.0001$, and net work $F(1.7, 32.2) = 71.8$, $p \leq 0.0001$. Total positive work was greatest when walking in the supported with effort condition (0.90 J/kg, 95% CI [0.78 to 1.02]) and least in the supported walking without effort condition (0.17 J/kg, 95% CI [0.13 to 0.20]). Positive work was also greater when walking in the KineAssist (0.78 J/kg, 95% CI [0.72 to 0.83]) as compared to over the treadmill (0.62 J/kg, 95% CI [0.59 to 0.66]).

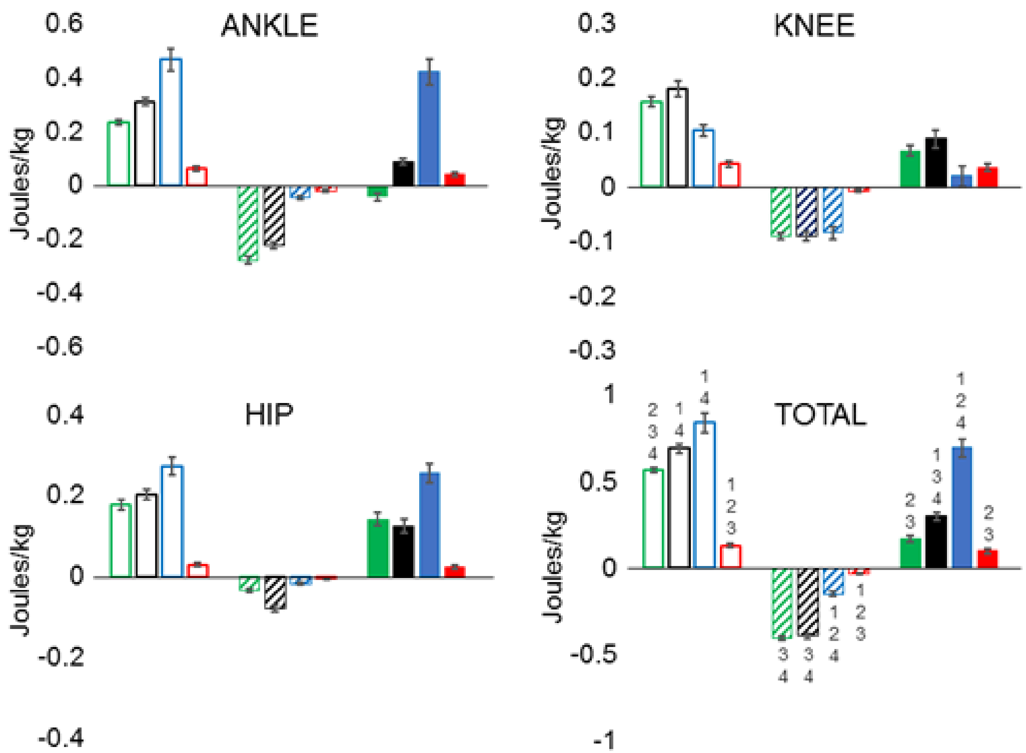

Fig. 4. Total positive (open bars), negative (striped bars), and net (solid bars) work across the lower-limb joints (ANKLE, KNEE, HIP) and the sum of all joints (TOTAL) across walking conditions. Values are mean (SE). Significant pairwise comparisons are reflected as numbers above or below each set of bar graphs for total work. All significant at Bonferroni corrected p≤0.001. Treadmill (green)[1]; KineAssist (black)[2]; support apparatus with effort (blue)[3]; support apparatus without effort (red)[4].

The supported walking with effort (-0.24 J/kg, 95% CI [-0.20 to -0.27]) and without effort (-0.07 J/kg, 95% CI [-0.05 to -0.09]) conditions revealed less total negative work compared to walking over the treadmill (-0.51 J/kg, 95% CI [-0.47 to -0.54]) and in the KineAssist (-0.49 J/kg, 95% CI [-0.46 to -0.53]). Pairwise comparisons of total negative work between treadmill and KineAssist walking were not statistically different; however, negative work in the supported conditions was significantly reduced as compared to both treadmill and KineAssist walking.

Finally, there were no differences in total net work between the treadmill (0.12 J/kg, 95% CI [0.08 to 0.16]) and supported walking without effort (0.10 J/kg, 95% CI [0.07 to 0.12]) conditions. There was significantly higher total net work in the KineAssist (0.29 J/kg, 95% CI [0.24 to 0.34]) and supported with effort (0.67 J/kg, 95% CI [0.55 to 0.78]) walking conditions.

IV. DISCUSSION

Our primary goal was to fabricate a walking environment that minimized postural demands (defined as the control of upright orientation, support of body weight, and control of body COM accelerations) in relation to stepping demands. As expected the support apparatus used to achieve this goal indeed minimized trunk motion in all three planes, allowed for minimal vertical GRFs in the supported without effort walking condition, and enabled reduced total lower-limb power absorption in both supported walking conditions. The support apparatus also enabled participants, via visual force feedback, to generate similar magnitudes of vertical GRF during walking using physical effort rather than body weight during the stance phase.

A. Reduced trunk movements during supported walking reflected reduced need to control upright orientation.

Trunk movements during walking complement movements of the pelvis and correspond to the body's effort to optimize COM excursions in order to conserve energy [22][23]. The swinging limb also perturbs stability of the trunk during typical walking necessitating compensatory movements. The head, arms, and trunk segments account for a large portion of the body's mass; thus, controlling these segments is critical for maintaining body stability during walking [24]. The reduced range of motion of the trunk while walking in the support apparatus is evidence that participants were no longer required to regulate the position of their trunk as carefully as during treadmill or KineAssist walking. The slightly greater range of motion observed during supported walking with effort as opposed to without effort was likely due to the compliant nature of the support apparatus materials, as participants actively pushed against the shoulder pads while targeting vertical forces.

B. Leading limb angles reflected minimal need to control forward momentum during walking in the support apparatus.

Trailing limb angles were very similar between the treadmill, KineAssist, and supported walking with effort conditions. This finding is particularly important because it allows comparisons of gait mechanics when the foot is in the same place in relation to the body's COM. In contrast, both supported walking conditions had smaller leading limb angles compared to the treadmill and KineAssist walking conditions. Walking typically requires dynamic control over a moving COM that must be stabilized on a step-by-step basis [25]. Dynamic control of the COM is partly reflected in the distance an individual must place their foot forward to briefly stabilize their COM within their base of support at foot strike. During supported walking the body's COM remained fixed in place; thus, the foot did not have to be placed at a sufficient distance forward to stabilize a dynamically changing COM (relative movement on a treadmill). This finding is consistent with investigations of the effects of body-weight support on spatiotemporal characteristics of walking that reported decreased step lengths with increased levels of support [26].

C. The support apparatus enabled participants to take light steps with no more than limb weight or to match vertical GRFs to those that occurred during walking without the apparatus.

The peak vertical GRF values were different, as expected, between treadmill and KineAssist and supported walking without effort. The support apparatus enabled participants to take light steps with no more than limb weight on the treadmill surface when walking without exerting effortful pushes during the stance phase. This condition demonstrated that the apparatus indeed provided full offloading of the head, arms, and trunk mass from the limbs. Also of interest was the fact that the fore-aft GRF component remained fairly neutral, indicating no braking or propulsion while walking without effort. This finding is consistent with the notion that the body COM did not have to be accelerated in any direction in order to maintain walking speed or upright orientation when supported.

Participants successfully targeted peak vertical GRFs during supported walking with effort, allowing the comparison of other walking behaviors (e.g., joints powers) under walking conditions with matched vertical loads, but very different postural control requirements. Even under these matched vertical load conditions there were minimal requirements to control accelerations of the body's COM, as evidenced by the characteristic bimodal shape of the vertical GRF that corresponds to body COM accelerations [6] being replaced with a single peak that occurred during midstance, and the noticeably different rate of loading and offloading in the shape of the supported with effort vertical GRF.

D. Power absorption typically related to braking and redirecting the body COM was minimal during supported walking.

Power absorption in early stance, particularly at the ankle joint, functions to decelerate the body's forward momentum and provide vertical support [27]. The ankle dorsiflexes under eccentric contraction of the plantarflexors through the first half of stance and plantarflexor power absorption modulates the associated braking impulse [28]. Both braking and power absorption were minimal during walking in the support apparatus, particularly at the ankle. The altered ankle joint trajectory (minimal dorsiflexion) during the supported walking conditions also substantiates that the support apparatus minimized the need to control forward momentum.

Power absorption at the knee also plays a critical postural role in weight acceptance as the knee extensors eccentrically control knee flexion preventing limb collapse [29]. Given that the knee joint was flexed at foot strike during supported walking with effort and did not continue to flex throughout the stance phase, it is unlikely that the negative work observed at the knee in this condition related to weight acceptance. Instead, if considered along with the relatively larger positive work at the hip, it is likely that the limb was quickly moved into a position directly underneath the body and stiffened to hit the vertical GRF target. Further exploration of underlying muscle activity will be necessary to determine whether eccentric activity of the hamstrings or quadriceps led to the negative work at the knee or if some degree of co-contraction influenced work calculations in the negative direction.

There was minimal negative work at the hip in late stance prior to foot off. This period of the gait cycle typically involves power absorption as the hip flexors eccentrically contract. Eccentric activity of the hip flexors has been shown to be functionally involved in forward acceleration of the trunk [29] [30]. Thus, the reduced negative work at the hip during supported walking also corroborates reduced need to control forward acceleration of the body COM.

Despite the lower total negative work during supported walking with effort, total positive work was increased as compared to typical treadmill walking. This increase in positive work appeared to occur primarily at the ankle. We plan to explore the underlying reasons for increased positive work when walking with minimal postural, but matched vertical GRF requirements in future studies.

E. Limitations

We only investigated these novel walking behaviors at one speed and at a single vertical force requirement in the present study. Future investigations should explore whether these findings are consistent across speeds and vertical force levels. For example, will there be a proportional decrease of vertical and fore-aft GRF components similar to that observed during simulated reduced gravity [31], such that the resultant GRF vector retains a similar orientation? We also investigated only the dominant limb of nonimpaired individuals in the present study. In future investigations it would be interesting to observe how the limbs interact during walking with minimal postural demands, but matched vertical forces. Additionally, we do not present muscle activity in the present study; however, we plan to explore muscle activity during walking in these different conditions because certain muscles (e.g., soleus and gastrocnemius) have been shown to play different roles in body support versus forward progression during walking [27][29].

F. Conclusions

We demonstrated that the novel postural support apparatus reduced trunk motion in flexion/extension, lateral flexion, and transverse rotation and reduced total positive and negative work during supported walking compared to walking with typical postural demands. Additionally, we demonstrated that supported walking minimized total lower-limb power absorption related to postural control even under matched vertical force generation requirements. This novel supported walking environment requires further investigation to better understand the motor strategy underlying walking with minimal postural demands and we plan to further characterize these behaviors. The present findings enable us to design future experiments to explore neuromechanical mechanisms underlying postural and locomotor control in both nonimpaired walking and in the control of walking for individuals with posturokinetic disorders.

V. REFERENCES

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LOCOMOTOR FORCE GENERATION DURING NONIMPAIRED WALKING WITH AND WITHOUT COORDINATED POSTURAL CONTROL

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ABSTRACT

Background: Upright bipedal walking requires coordinated interaction between postural and locomotor control systems. We previously demonstrated that we could mechanically decouple postural functions from locomotor functions during the gait cycle, using a novel postural support apparatus, in order to investigate locomotor force generation under minimal postural influence. We propose that if postural demands are fully minimized relative to locomotor requirements during walking, then postural functions should have negligible influence over locomotor force generation behaviors. Methods: To investigate this prediction, we manipulated locomotor force generation requirements of walking for N = 16 nonimpaired individuals walking at two speeds (1.0 m/s and 0.5 m/s) under two experimental conditions: 1) under increasing amounts of BWS (0 to 30% body weight), which partially reduced postural demands, but did not fully minimize them relative to locomotor demands and 2) externally stabilized walking within the novel support apparatus that minimized postural demands relative to locomotor requirements while allowing participants to generate matched vertical forces to those of BWS walking using visual force feedback. Results: The ratio of fore-aft to vertical GRF (Fy/Fz ratio) remained invariant and lower-limb joints moments scaled across BWS levels reflecting a similar neuromechanical strategy. Average muscle activity of muscles directly involved in vertical weight support like the plantarflexors decreased with increasing BWS; however, other lower-limb muscles did not change their magnitude of activity. In contrast, we demonstrated that when the support apparatus minimized postural requirements during externally stabilized walking, the motor strategy underlying locomotor force generation was dependent instead on the magnitude of vertical force

generated. The Fy/Fz ratio decreased demonstrating a relative decoupling of Fy and Fz GRF components as force demands decreased. We further demonstrated that the Fy/Fz ratio was not speed dependent when participants had no requirement to maintain dynamic equilibrium during externally stabilized walking. Finally, muscle activity across all muscles decreased in response to decreasing force generation across targets. Conclusions: We posit that BWS reduces vertical loading requirements placed on the limbs, but still imposes postural demands through requirements to maintain upright orientation, control body accelerations, and support remaining body weight, which is reflected by the similar but scaled neuromechanical strategy we observed across BWS levels. When the postural support apparatus mechanically decoupled posture from locomotion, locomotor force generation reflected muscular effort without activity associated with anticipatory or reactive postural influence. Future directions: Mechanically decoupling posture and locomotor control functions during walking offers a unique opportunity to probe locomotor control of populations that have difficulty coordinating posture with locomotion (i.e., those with posturokinetic disorders) in future studies.

INTRODUCTION

Upright bipedal walking requires coordinated interaction between postural and locomotor control systems. There are two primary biomechanical goals during functional walking, 1) postural maintenance against gravity and internal and external disturbances and 2) progression of the body in a desired direction at a desired speed (1–7). Massion (3,4) proposed separate central controllers for posture and movement that act interdependently to accomplish a motor act. This neurophysiological model described parallel descending pathways originating from the cerebral cortex and brainstem that allow the execution of movement while maintaining posture. Here we apply this model to the motor task of walking (Figure 1), where these pathways synapse on locomotor networks within the spinal cord to facilitate simultaneous postural control and coordinated stepping (1,6–8). We describe the interaction of postural adjustments and responses with locomotor control, where postural control pathways maintain upright orientation and vertical weight support and are involved in power absorption for braking and redirecting body mass (Figure 1, green) (9) and locomotor control pathways generate power for forward propulsion and position the limbs to maintain dynamic equilibrium (Figure 1 (blue) (2,10).

We previously suggested that if the dual output of posture and stepping reflects coordination of separate control systems, as described by this model, we should be able to mechanically decouple stepping functions from postural functions during the gait cycle in order to determine the impact of each system. Recently, we demonstrated that it is possible to minimize mechanical postural requirements of walking in order to investigate locomotor force generation with negligible postural influence (11). We used a novel

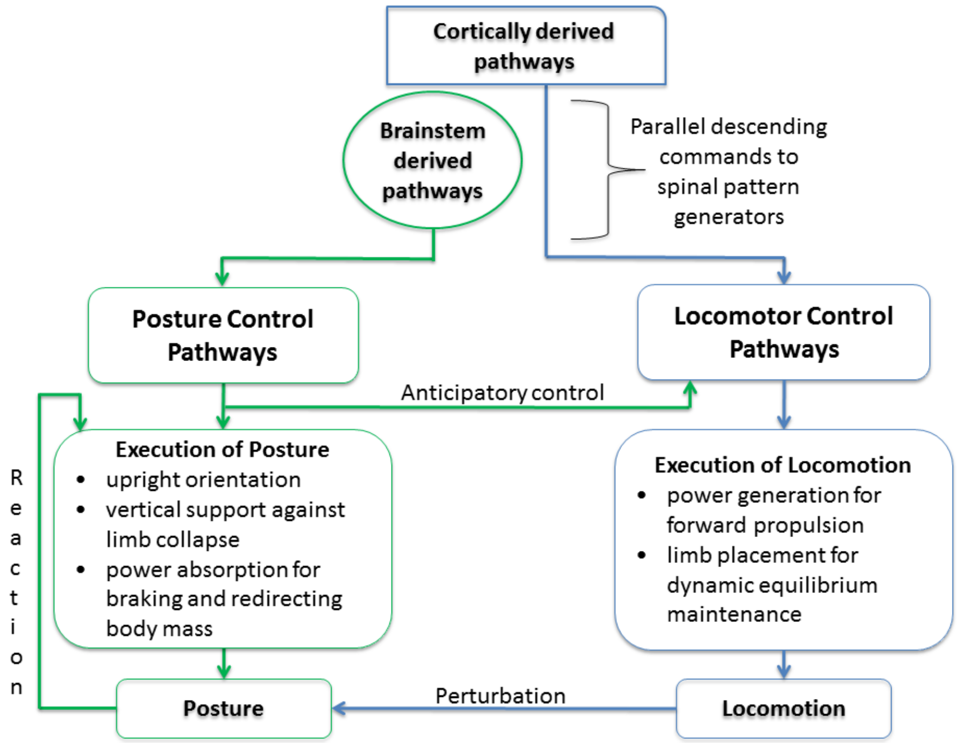

Figure 1. Postural-locomotor interaction model during unperturbed walking describing the parallel descending commands facilitating primary postural (green) and locomotor (blue) functions. Stepping movements perturb posture, and feedback from load receptors during the stance phase is used through reactive mechanisms to initiate gait transitions. Anticipatory mechanisms allow for adjustments in limb trajectory to maintain dynamic equilibrium on a step-by-step basis.

postural support apparatus that provided an externally stabilized walking environment that reduced vertical support to limb weight, but enabled participants to replicate locomotor force generation when provided with vertical ground reaction force (GRF) targets and instructions to generate these forces during the stance phase of walking. Nonimpaired participants exhibited reduced trunk motion in flexion/extension, lateral flexion, and transverse rotation, and reduced lower-limb power absorption related to postural control, even under matched locomotor force generation requirements when walking within this apparatus (11). We additionally observed reduced leading limb angles at foot strike and little to no braking during the first half of the stance phase. Despite these altered postural behaviors participants exhibited locomotor functions of matched trailing limb angles and large power generation during the second half of the stance phase.

Here we sought to further examine the neuromechanical strategy underlying locomotor force generation where posture and locomotor control coordination is required versus where locomotor force generation is decoupled from postural behaviors. We propose that a walking environment with minimal mechanical postural demands not only minimizes upright orientation and vertical support requirements of walking, but also reduces anticipatory control influence of posture on locomotor output (3,12–14) through removing the need to maintain dynamic equilibrium. Our adapted postural-locomotor interaction model predicts that if postural demands are present (required to support body weight and maintain upright orientation) and locomotor demands are held constant (speed, cadence, and dynamic equilibrium maintained) the neuromechanical output of the system should reflect the same strategy as long as biomechanical configurations of the body and limbs remain similar.

In support of this prediction, proportional decreases of the vertical and fore-aft components of the GRF have been observed in simulated reduced gravity allowing the resultant GRF vector to retain a similar orientation (15). The direction of sagittal plane resultant GRFs during stance reflects control over the net angular momentum of the whole body (16,17) making this neuromechanical variable an important characteristic of coordinated posture and locomotor control. During the propulsive portion of the stance phase the direction of sagittal resultant GRFs reflects the combined acts of vertical support and forward propulsion of body mass (18). Additionally, while certain muscles appear to have specific antigravity functions (e.g., plantarflexors) and are affected by altered vertical loading (e.g., under body-weight support (BWS)), muscle activity of the lower limbs is highly variable across studies if speed is held constant, with some studies reporting increases in activity and others decreases (19–22). We propose that the variable nature of muscle activity reflects the coordination of posture and locomotor demands. If postural demands are instead fully minimized relative to locomotor requirements, as in our externally stabilized walking environment, then postural influence should have negligible influence over locomotor force generation behaviors.

To investigate these predictions, we manipulated locomotor force generation requirements of walking under two experimental conditions, 1) walking under increasing amounts of BWS, which partially reduced postural demands, but did not fully minimize them relative to locomotor demands (19,23) and 2) externally stabilized walking within the novel support apparatus that minimized postural demands relative to locomotor requirements while manipulating force generation through providing a range of vertical force targets. We conducted walking trials under these conditions at two speeds, reflecting different locomotor demands. At each speed, we expected that under increasing BWS levels where locomotor demands of speed and cadence were held constant, the ratio of propulsive to vertical GRF (Fy/Fz ratio) would be invariant due to proportional scaling of vertical support demands and requirements to accelerate the upper body and trunk mass forward. We further expected lower-limb joint moments at the ankle, knee, and hip to scale proportionally with the provision of BWS, reflecting the same strategy underlying coordinated postural and locomotor control at each level of support. Finally, we expected that average muscle activity of muscles most sensitive to changes in vertical loading (i.e., plantarflexors) would decrease activity during propulsion as BWS increased, but that activity of other muscles would not change.

In contrast, we expected that the Fy/Fz ratio would not remain consistent across force targets during externally stabilized walking due to a decoupling of postural demands from locomotor requirements. We further expected lower-limb joint moments directly involved in locomotor force generation (i.e., ankle and knee) would decrease with decreasing force target. Finally, we expected that muscle activity would decrease during propulsion across all muscles with decreasing locomotor force generation requirements since this activity reflected muscular effort directed only to locomotor force generation without coordination of postural control.

METHODS

Sixteen non-neurologically impaired individuals (8 females, 8 males) mean age 49.3 years (range 26 to 71 years) participated in this study. Exclusion criteria were history of neurologic and/or musculoskeletal disorders that could affect postural control or walking function. Participants filled out a Physical Activity Readiness Questionnaire (PAR-Q) (24) to ensure that they were safe to participate in physical activity. All participants signed informed consent as approved by the Institutional Review Board of the University of Alabama at Birmingham.

Body-weight supported walking environment

Participants first walked over a motorized, dual-belt, instrumented treadmill (BERTEC, Columbus, OH, USA) at two speeds, 0.5 m/s and 1.0 m/s, while connected to the KineAssist™ robotic device (HDT Global, Solon OH). Individuals interacted with the KineAssist via a pelvic mechanism that allowed all six degrees of freedom around the pelvis/hip complex while walking (25). However, for this experiment we only allowed the three sagittal degrees of freedom: surge (relative forward/backward movement over the treadmill), heave (vertical center of mass (COM) motion), and pitch (forward/backward trunk/pelvic tilting). With a locking mechanism we prevented sway (side-to-side pelvic translation), roll (hip hiking), and yaw (left/right pelvic rotation). We imposed these limitations to ensure greater similarity between BWS and externally stabilized walking (described below) in that both environments emphasized control of only sagittal plane walking mechanics. Sagittal plane kinematics and kinetics during walking in this BWS environment are not affected by these limitations (11). The pelvic mechanism of the KineAssist is equipped with bilateral force sensors that measure vertical forces applied to the harness through the pelvis and uses them within a closed-loop controller to provide a near constant level of BWS at the approximate body COM. The KineAssist provides a safe environment during walking and catches a participant in the event of a loss of balance after a preprogramed descent distance (e.g., 4 inches).

Externally stabilized walking environment

Participants next walked at the same two speeds, 0.5 m/s and 1.0 m/s, in a support apparatus that minimized postural demands of walking as detailed fully in Graham et al. 2018 (Figure 2) (11). Briefly, the apparatus served three purposes, (1) externally stabilized the trunk (restricted trunk movement), 2) fully offloaded the trunk and upper body mass from the lower limbs but allowed the participant to make contact with the

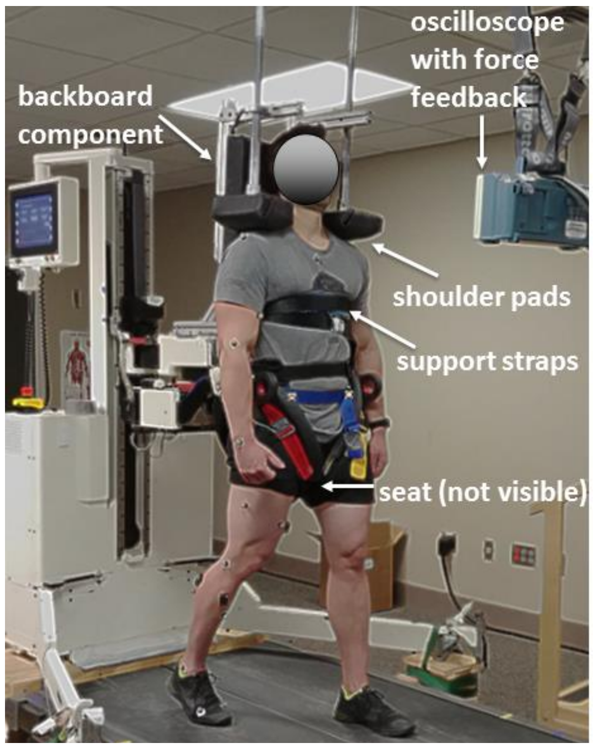

Figure 2. Externally stabilized walking environment with example marker locations shown on right side of body (sacral and back markers not visible).

treadmill surface with their feet, and 3) minimized the need to control body COM accelerations by holding the participant in place and preventing forward/backward travel on the treadmill belt. The apparatus is equipped with a bicycle saddle seat that allows full offloading of the limbs without impeding limb movements. Shoulder pads offer stabilization so that the participant can push their foot against the surface of the treadmill during the stance phase of walking to hit a visual force target of a prescribed magnitude of vertical GRF.

Experimental protocol

Participants walked under four levels of BWS (0, 10, 20, and 30%) for one-minute at each of two speeds for a total of eight trials. We allotted participants approximately one minute to acclimate to each level of support prior to collecting data (26,27). To ensure comparable spatiotemporal characteristics, we measured the comfortable walking cadence of each participant during their 0% walking trial for each speed respectively and instructed participants to maintain this cadence, via auditory feedback from a metronome, during all walking trials at each speed. We removed BWS and provided participants with several minutes of rest between walking trials at each level of support. In order to ensure similar peak magnitudes of vertical GRFs for externally stabilized walking trials, we measured the average push-off peak vertical GRF from the dominant limb of participants during BWS walking trials at each level of support in the KineAssist device. We assessed limb dominance by asking participants to identify the limb they would kick a ball with.

Participants performed externally stabilized walking at each speed under the same respective cadence requirements within the support apparatus while targeting four different levels of vertical GRF. Participants performed two 30-second trials at each target level for a total of sixteen trials (eight per each of two speeds). We displayed target vertical GRF values ± 5 % on an oscilloscope for the dominant limb only. We instructed participants to hit the target by pushing their dominant foot against the moving belt beginning when their foot was directly beneath them during the stance phase.

Data acquisition

We collected GRFs at 1000 Hz via the Bertec, dual-belt, force-instrumented treadmill and kinematic data at 100 Hz via an eight-camera, Qualisys motion capture system (Qualisys Inc., Gothenburg, Sweden) with 33, 1-cm passive-reflective markers placed bilaterally over anatomical landmarks of the trunk, arms, legs, and feet, with three markers defining each segment (example marker locations shown in Fig. 2).We shaved (if necessary) and cleaned the skin with rubbing alcohol before collecting muscle activity at 2000 Hz from the tibialis anterior (TA), soleus (SOL), lateral gastrocnemius (GAS), vastus medialis (VM), and biceps femoris (BF) with a Delsys Trigno Wireless EMG system (Trigno™ Wireless System, DelSys, Inc, Boston, MA, USA). EMG sensor placement followed SENIAM surface electromyography recommendations (28). We calculated all variables over either a full gait cycle (foot strike to ipsilateral foot strike) or stance phase (foot strike to ipsilateral foot off). We defined these gait events with a vertical GRF threshold value of 2.0% body weight. We used Visual 3D (C-Motion, Germantown, MD, USA) to obtain marker trajectories for limb angle calculations and joint moments and performed all post processing of the data in MATLAB (Mathworks®, version R2016a).

Spatiotemporal measures

We calculated stride duration as the time elapsed from foot strike to ipsilateral foot strike and stance duration as the time elapsed from foot strike to ipsilateral foot off. We calculated stance phase limb angle (leading and trailing limb) as the angle between the vector connecting the dominant-limb lateral toe and ASIS marker and the laboratory's vertical axis (29). Negative values indicate leading limb position at foot strike and positive values trailing limb position at foot off.

Kinetic measures

We low-pass filtered GRFs with a cutoff of 20 Hz and calculated average vertical and fore-aft GRF values normalized to body weight (% BW) during the stance phase. We previously demonstrated that nonimpaired participants can successfully match peak vertical forces when provided with visual vertical GRF feedback during the stance phase. Here, we report the average vertical force resulting from the applied BWS and target force levels because we felt that it was important to reflect differences in average force production resulting from the different rates of loading and offloading of the limb that occur between BWS versus externally stabilized walking. Since participants had varying accuracy in attaining vertical force targets, we eliminated steps where the vertical GRF value did not fall within ± 5% of the trial average only for externally stabilized trials, in order to ensure distinct force levels across target categories.

We provide group ensemble average fore-aft and vertical GRF curves for walking trials in each condition at 1.0 m/s for comparison of loading characteristics. We characterized GRF direction during the propulsive period of the stance phase with the ratio of positive fore-aft to respective time period vertical GRF (Fy/Fz ratio). Visual 3D calculates joint moments using an inverse dynamics approach. We provide group ensemble average joint moments for the ankle, knee, and hip during the stance phase. We normalized joint moments to body mass (Nm/kg) and calculated average joint moments at the ankle, knee, and hip that occurred during the same time period of positive fore-aft GRF during the propulsive portion of the stance phase.

Muscle activity measures

We bandpass filtered muscle activity between 20 and 400 Hz, full-wave rectified it, and calculated each participant's average activity for all muscles at each force level and speed for BWS and externally stabilized trials respectively during the propulsive period of the stance phase. We removed steps containing noisy artifacts greater than 3 SD from the average peak EMG value across steps within a trial.

Statistical analyses

We conducted statistical tests using SPSS version 24 with separate statistical tests for each speed (1.0 and 0.5 m/s). We used 2 (condition; BWS vs. externally stabilized) x 4 (force level) repeated measures ANOVAS to make comparisons between walking conditions and across force levels when we were interested in possible condition x level interactions and/or post hoc comparisons between levels. We analyzed temporal variables, average vertical forces, and ankle, knee, and hip joint moments in this manner.

For all repeated measures analyses we employed Greenhouse Geiser corrections, as necessary, for violations of sphericity, a priori alphas $p \leq 0.05$, and Bonferroni corrections for multiple comparisons.

For variables where we were only interested in whether there was a decreasing or increasing relationship across force levels within each walking condition, we calculated the slope of the regression line across force levels for each participant. We used one sample *t* tests or Wilcoxon Signed Rank tests to determine if the group average slope was significantly different from zero. We analyzed limb angles, force ratios, and muscle activity in this manner, with one sample *t* tests for limb angles and force ratios, and Wilcoxon Signed Rank tests for muscle activity due to a few instances of deviations from normality. We used an a priori alpha $p \leq 0.05$ for these comparisons.

RESULTS

We first present results demonstrating control over walking conditions (temporal and average vertical GRF comparisons) as well as results confirming our previous findings regarding leading and trailing limb angles during each walking condition. We follow with results that address our primary research question regarding force ratios, joint moments, and muscle activity measures.

Temporal characteristics across walking conditions

Temporal variables were well controlled due to matched cadences, similar to our previous findings (11). Stride time was not significantly different across levels for BWS or externally stabilized walking at 1.0 m/s with no condition x level interaction ($p = 0.38$), main effect of condition ($p = 0.27$), or main effect of level ($p = 0.33$) (Table 1). Stride time was also not significantly different across levels for each condition at 0.5 m/s with no significant condition x level interaction ($p = 0.11$), main effect of level ($p = 0.21$), or main effect of condition ($p = 0.30$).

Table 1. Stride and stance duration [95% CI) across BWS and targets for each walking condition.

		BWS 0%	Target 1	BWS 10%	Target 2	BWS 20%	Target 3	BWS 30%	Target 4
1.0 m/s	stride time (s)	1.15 [1.10 to 1.19]	1.18 [1.10 to 1.26]	1.14 [1.10 to 1.18]	1.17 [1.10 to 1.23]	1.14 [1.10 to 1.18]	1.15 [1.10 to 1.20]	1.14 [1.11 to 1.18]	1.14 [1.08 to 1.21]
	stance time (s)	0.76 [0.73 to 0.79]	0.69 [0.65 to 0.74]	0.74 [0.71 to 0.77]	0.70 [0.66 to 0.73]	0.72 [0.69 to 0.76]	0.68 [0.65 to 0.72]	0.72 [0.69 to 0.75]	0.66 [0.62 to 0.71]
0.5 m/s	stride time (s)	1.72 [1.62 to 1.81]	1.86 [1.61 to 2.11]	1.72 [1.63 to 1.81]	1.78 [1.63 to 1.92]	1.73 [1.64 to 1.81]	1.76 [1.63 to 1.89]	1.74 [1.65 to 1.82]	1.72 [1.60 to 1.83]
	stance time (s)	1.24 [1.16 to 1.33]	1.19 [1.09 to 1.28]	1.22 [1.14 to 1.29]	1.17 [1.09 to 1.26]	1.21 [1.14 to 1.28]	1.16 [1.08 to 1.24]	1.20 [1.13 to 1.26]	1.13 [1.04 to 1.21]

We did not detect a condition x level interaction for stance time at 1.0 m/s ($p = 0.11$), but we did observe a significant main effect of level ($F(1.9, 28.5) = 13.2$, $p \leq 0.0001$), indicating a decrease in stance duration with either increasing BWS or decreasing force target and a significant main effect of condition ($F(1, 15) = 47.8$, $p \leq 0.0001$) where stance duration was shorter for externally stabilized walking as compared to BWS walking (Table 1).

We similarly did not detect a condition x level interaction for stance time at 0.5 m/s ($p = 0.78$) but did observe a significant main effect of level ($F(2.1, 31.4) = 8.6$, p =

0.001), indicating a decrease in stance duration with either increasing BWS or decreasing force target and a significant main effect of condition ($F(1, 15) = 9.0$, $p \leq 0.01$) where stance duration was shorter for externally stabilized walking compared to BWS walking.

Limb angle characteristics across walking conditions

Leading limb angle decreased across BWS levels at both 1.0 m/s (slope = 0.11, $p \leq 0.0001$) and 0.5 m/s (slope = 0.05, $p \leq 0.01$), indicating that foot strike occurred closer to the body as BWS increased (Table 2). In contrast, leading limb angle increased from high (1) to low (4) targets during walking at 1.0 m/s (slope = -0.58, $p = 0.03$), indicating that foot strike occurred farther from the body as target decreased, but did not change across targets at 0.5 m/s (slope = -0.25, $p = 0.23$).

Table 2. Leading and trailing limb angles [95% CI] across BWS and targets for each walking condition.

		BWS 0%	Target 1	BWS 10%	Target 2	BWS 20%	Target 3	BWS 30%	Target 4
1.0 m/s	Leading Limb Angle (°)	-20.2 [-21.5 to -18.8]	-12.4 [-14.2 to -10.6]	-18.4 [-19.7 to -17.0]	-13.2 [-14.7 to -11.7]	-17.7 [-19.1 to -16.4]	-13.3 [-14.9 to -11.7]	-16.5 [-18.1 to -15.0]	-14.3 [-16.0 to -12.6]
	Trailing Limb Angle (°)	23.7 [22.5 to 25.0]	25.1 [23.9 to 26.3]	23.2 [22.0 to 24.4]	24.5 [23.4 to 25.7]	23.1 [21.9 to 24.3]	23.8 [22.7 to 24.9]	23.6 [22.4 to 24.9]	23.4 [21.8 to 24.9]
0.5 m/s	Leading Limb Angle (°)	-16.4 [-17.8 to -15.1]	-10.7 [-12.6 to -8.8]	-15.3 [-16.6 to -13.9]	-11.1 [-13.1 to -9.0]	-14.9 [-16.6 to -13.3]	-11.4 [-13.2 to -9.6]	-14.9 [-16.6 to -13.1]	-11.4 [-13.6 to -9.3]
	Trailing Limb Angle (°)	19.8 [17.9 to 21.7]	23.1 [20.7 to 25.4]	19.5 [17.9 to 21.2]	22.5 [20.2 to 24.7]	19.5 [18.0 to 21.1]	21.7 [19.8 to 23.7]	19.6 [18.1 to 21.2]	21.3 [19.3 to 23.3]

Trailing limb angle was not statistically different across BWS levels at both 1.0 m/s (slope = -0.005, $p = 0.76$) and 0.5 m/s (slope = -0.01, $p = 0.82$) (Table 2). In contrast, trailing limb angle decreased from high to low targets at 1.0 m/s (slope = -0.59, $p \leq 0.0001$) and at 0.5 m/s (slope = -0.59, $p = 0.01$).

Vertical ground reaction force characteristics across walking conditions

Individuals achieved target force levels with a variety of success, so, based on measured vertical force values achieved, we grouped the data into four distinct categories of vertical force for each walking condition, confirmed by a significant main effect of level for each condition at both 1.0 m/s (BWS; $F(3, 45) = 1217.8$, $p \leq 0.0001$; externally stabilized $F(1.5, 22.8) = 129.9$, $p \leq 0.0001$) and 0.5 m/s (BWS; $F(3, 45) = 718.5$, $p \leq 0.0001$; externally stabilized $F(1.2, 17.9) = 100.4$, $p \leq 0.0001$) and significant pairwise comparisons between all levels for post hoc comparisons ($p \leq 0.0001$) (Table 3). There were significant main effects of condition for both speeds ($F(1, 15) = 116.0$, $p \leq 0.0001$ at 1.0 m/s) and ($F(1, 15) = 56.6$, $p \leq 0.0001$) at 0.5 m/s, indicating that average forces were higher for BWS compared to externally stabilized walking.

Table 3. Average vertical force across BWS and targets [95% CI] for each walking condition.

		0% BWS	Target 1	10% BWS	Target 2	20% BWS	Target 3	30% BWS	Target 4
1.0 m/s	Fz (% BW)	78.0 [76.9 to 79.1]	60.7 [57.9 to 63.5]	70.7 [69.3 to 72.2]	54.7 [51.6 to 57.7]	63.7 [62.1 to 65.3]	47.1 [42.5 to 51.7]	55.7 [54.4 to 57.1]	39.0 [34.2 to 43.8]
0.5 m/s	Fz (% BW)	71.3 [70.2 to 72.4]	55.4 [50.9 to 60.0]	64.7 [63.0 to 66.3]	50.5 [46.4 to 54.6]	57.5 [56.1 to 58.9]	46.1 [42.3 to 49.9]	51.0 [49.5 to 52.4]	40.4 [36.5 to 44.2]

*Note all pairwise comparisons between levels within each condition (BWS or externally stabilized) significantly different at $p \leq 0.0001$

Consistent with our previous findings, the profile of the vertical GRF reflected single peaks associated with hitting force targets along with longer rates of loading and offloading during externally stabilized walking (Fig. 3B), as opposed to bimodal peaks

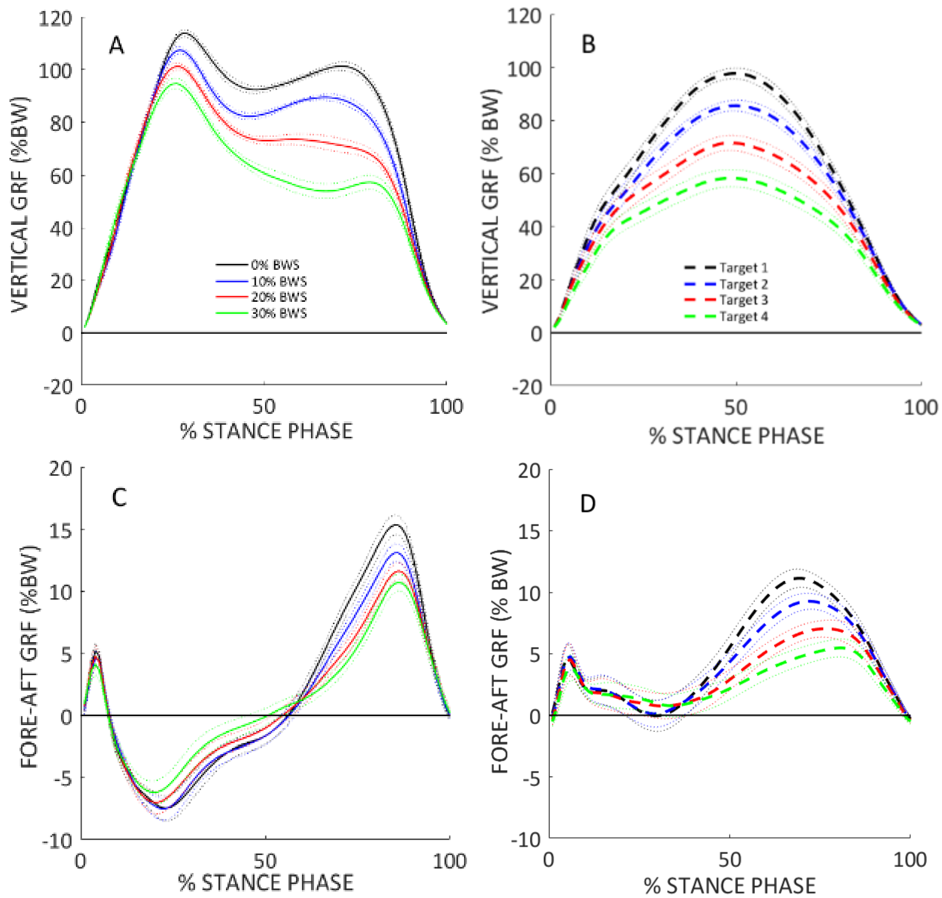

Figure 3. Group ensemble average vertical (a & b) and fore-aft (c & d) GRFs during walking at 1.0 m/s across BWS levels (left column; smooth lines) and vertical force targets (right column; dashed lines).

with steeper loading and offloading during BWS walking (Fig. 3A). The braking component of the fore-aft force was also negligible during externally stabilized walking (Fig. 3D) compared to BWS walking (Fig, 3C) consistent with our previous findings.

Force ratios characteristics for each walking condition

The Fy/Fz ratio during the propulsive portion of the stance phase did not significantly change across BWS levels during walking at either 1.0 m/s (slope = 0.0001, p = 0.49) or 0.5 m/s (slope = 0.0006, p = 0.38) (Figure 4; left column). In contrast, the force ratios significantly decreased from highest (1) to lowest (4) target for externally stabilized walking at both 1.0 m/s (slope = -0.01, p = 0.01) and 0.5 m/s (slope =-0.01, p = 0.05) (Figure 4; right column).

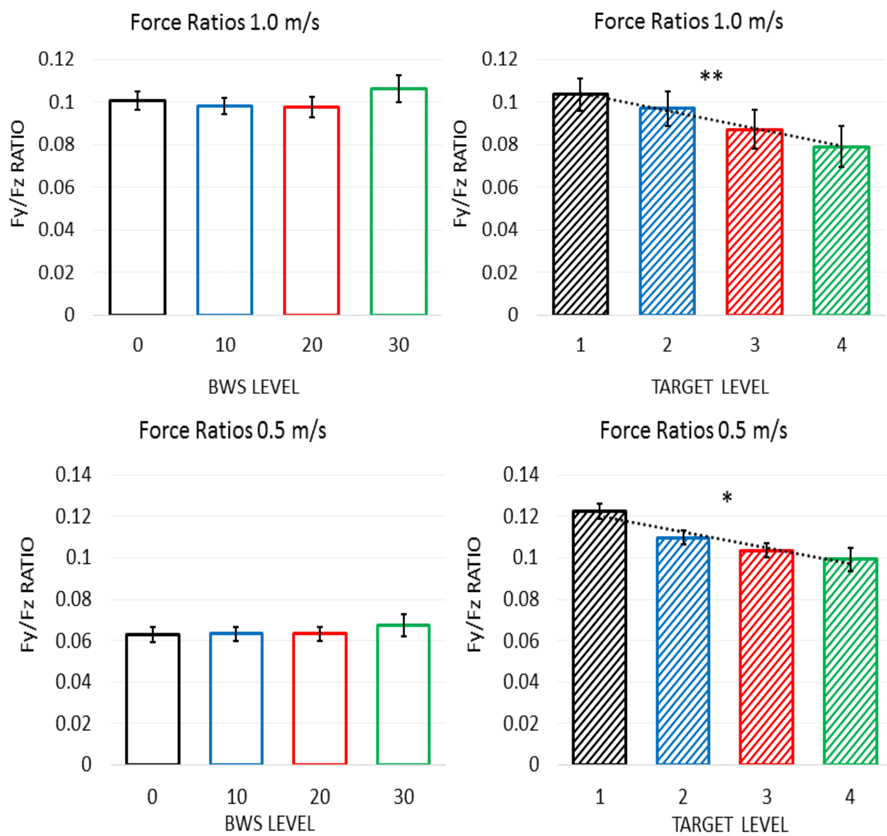

Figure 4. Force ratios across BWS levels (open bars; 0, 10, 20, 30) and targets (dashed bars; 1, 2, 3, 4) for walking at 1.0m/s (top) and 0.5 m/s (bottom). * sig slope at p≤0.05; ** sig slope at p≤0.01.

Lower limb joint moment characteristics for each walking condition

Joint moment profiles at the ankle, knee, and hip displayed similar trajectories for BWS and externally stabilized walking (Figure 5). We observed a significant condition x level interaction for ankle moment at 1.0 m/s (F (2.2, 32.5) = 3.3, p = 0.05), but not at 0.5 m/s (p = 0.26). Main effects demonstrated larger ankle plantarflexor moments during

BWS compared to externally stabilized walking at 1.0 m/s (F (1, 15) = 191.5, p ≤ 0.0001), but not at 0.5 m/s (p = 0.96). Post hoc comparisons revealed that ankle moment decreased across force levels within each condition (p ≤ 0.05) (Figure 6).

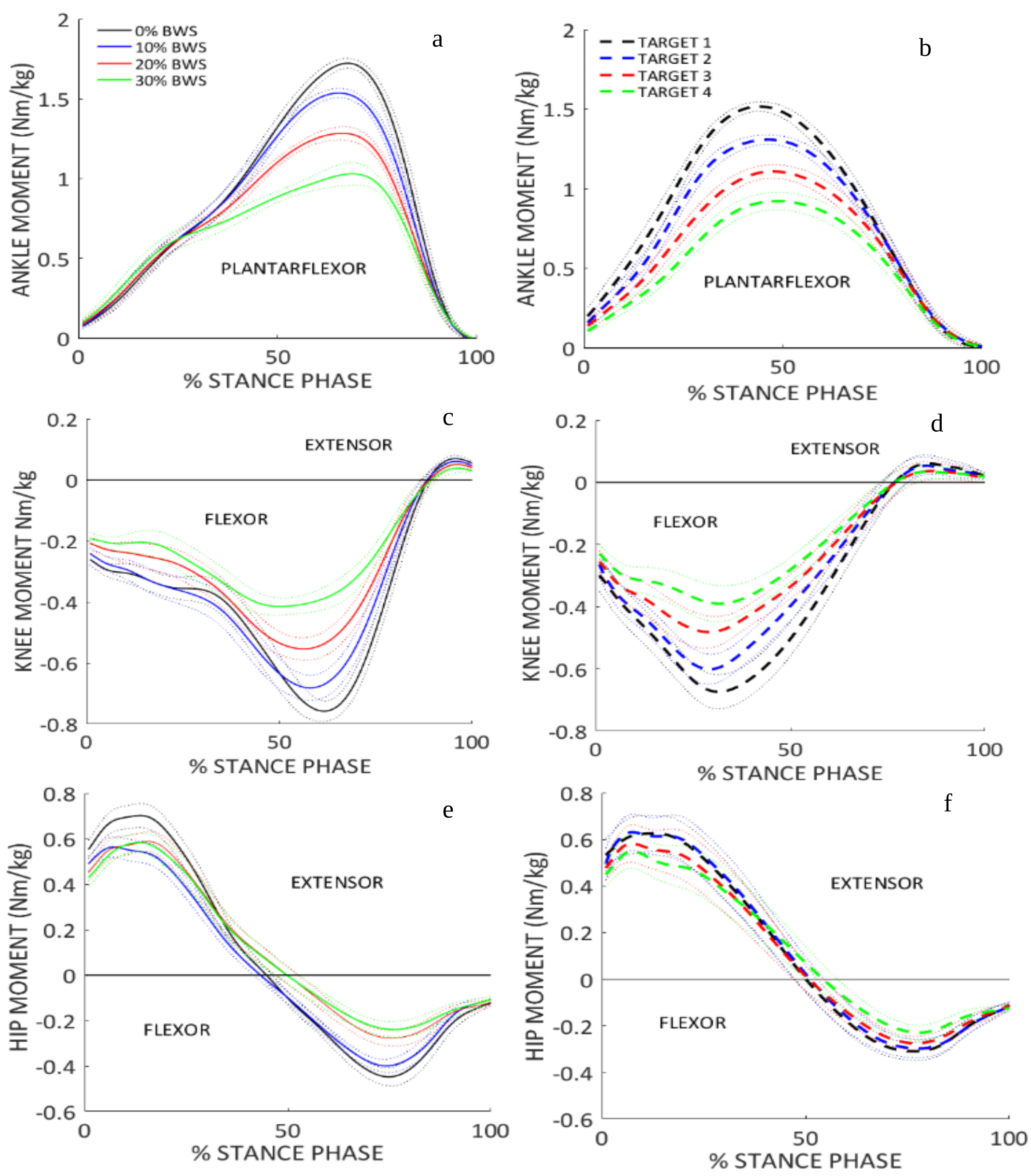

Figure 5. Group ensemble average joint moment profiles at the ankle (a & b), knee (c & d), and hip (e & f) for BWS walking (left column; smooth lines) and externally stabilized walking (right column; dashed lines).

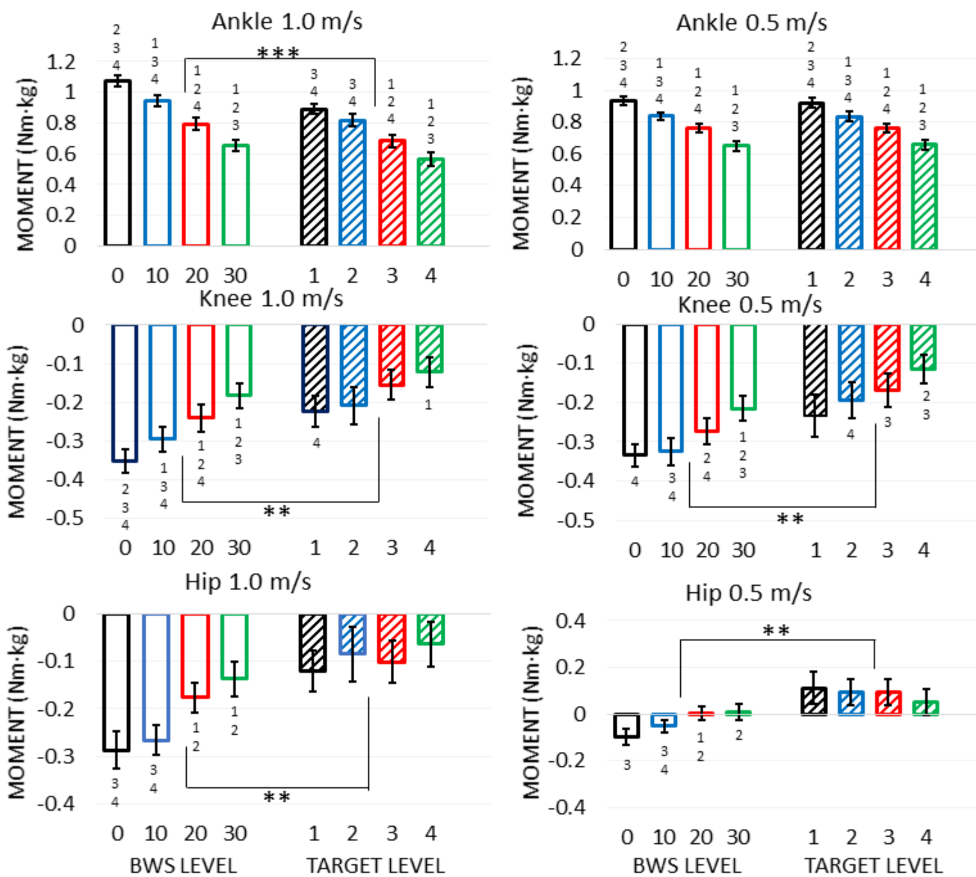

Figure 6. Average joint moments during propulsion for the ankle, knee, and hip for BWS walking (open bars; 0, 10, 20, 30) and externally stabilized walking (dashed bars; 1, 2, 3, 4). Main effects of condition *** p≤0.001; ** ≤0.01. Post hoc comparisons between levels within conditions significant at p≤0.05 represented as superscript numbers.

We did not detect a significant condition x level interaction at either speed for knee moments ($p = 0.84$ at 1.0 m/s; $p = 0.64$ at 0.5 m/s). We observed main effects of condition at 1.0 m/s ($F(1,15) = 18.5$, $p \leq 0.01$) and 0.5 m/s at ($F(1,15) = 9.9$, $p \leq 0.01$) indicating larger flexor moments at the knee during BWS compared to externally stabilized walking. Post hoc comparisons revealed decreasing knee flexor moments across all force levels for BWS walking ($p \leq 0.05$), and only some significant pairwise comparisons between levels for externally stabilized walking ($p \leq 0.05$) (Figure 6).

We detected significant condition x level interactions for hip moments for both speeds ($F (3, 45) = 4.1$, $p = 0.01$) at 1.0 m/s and ($F (1.4, 21.5) = 4.6$, $p = 0.03$) at 0.5 m/s. We also observed main effects of condition for 1.0 m/s ($F (1, 15) = 8.3$, $p = 0.01$) and 0.5 m/s ($F (1, 15) = 9.5$, $p \leq 0.01$), indicating that the hip had a larger flexor moment during BWS compared to externally stabilized walking. We decomposed the interactions for the hip by running separate repeated measures ANOVAs for each walking condition. We detected main effects of level for BWS walking at both speeds ($F (1.8, 26.8) = 17.7$, $p \leq 0.001$) at 1.0 m/s and ($F (1.6, 3.9) = 7.8$, $p \leq 0.001$) at 0.5 m/s, indicating that the flexor moment at the hip decreased with increasing BWS level with significant pairwise post hoc comparisons between levels ($p \leq 0.05$) at both speeds (Figure 6). In contrast, we did not observe main effects of level for externally stabilized walking at either speed ($p = 0.41$ at 1.0 m/s; $p = 0.40$ at 0.5 m/s), indicating that the average moment at the hip joint did not significantly change with decreasing force target.

Muscle activity characteristics for each walking condition

Muscle activity did not significantly change across BWS levels for TA, VM, or BF at either speed (Table 4). GAS activity decreased across BWS levels at both speeds, and SOL activity decreased with increasing BWS only when walking at 0.5 m/s. In contrast, all muscles decreased in response to decreasing force target during externally stabilized walking. Figure 7 shows muscle activity profiles during the stance phase for a representative individual walking at 1.0 m/s under BWS compared to externally stabilized. While we did not perform an analysis of muscle activity profiles for this experiment, the observed phasing of muscle activity retained many similarities between BWS and externally stabilized walking.

Table 4. Median slopes for the relationship of average muscle activity across BWS levels or targets for each walking condition.

	Muscle	BWS levels	Targets
1.0 m/s	Tibialis anterior	slope = 0.01 $p = 0.28$	slope = -0.14 $p = 0.006^{*}$
	Soleus	slope = -0.03 $p = 0.64$	slope = -0.44 $p = 0.001^{*}$
	Gastrocnemius	slope = -0.11 $p = 0.01^{*}$	slope = -0.51 $p \leq 0.001^{*}$
	Vastus medialis	slope = 0.00 $p = 0.76$	slope = -0.23 $p < 0.01^{*}$
	Biceps femoris	slope = 0.00 $p = 0.92$	slope = -0.18 $p = 0.007^{*}$
0.5 m/s	Tibialis anterior	slope = 0.00 $p = 0.80$	slope = -0.15 $p = 0.002^{*}$
	Soleus	slope = -0.02 $p = 0.02^{*}$	slope = -0.39 $p \leq 0.001^{*}$
	Gastrocnemius	slope = -0.05 $p = 0.001^{*}$	slope = -0.37 $p = 0.001^{*}$
	Vastus medialis	slope = 0.00 $p = 0.78$	slope = -0.14 $p = 0.003^{*}$
	Biceps femoris	slope = 0.00 $p = 0.88$	slope = -0.08 $p = 0.01^{*}$

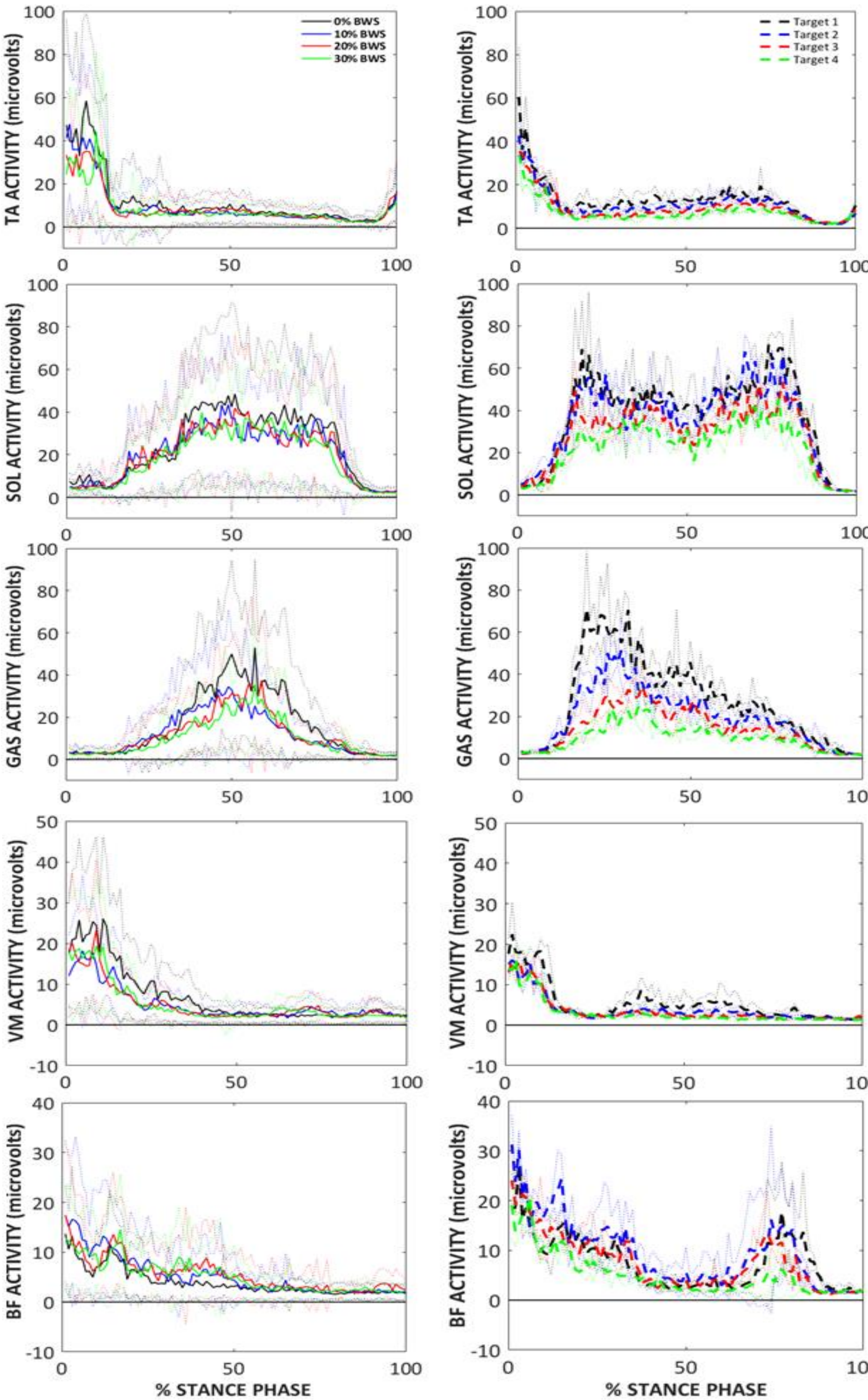

Figure 7. EMG ensembles for each muscle for a representative participant walking at 1.0 m/s for BWS walking (left column; smooth lines) and externally stabilized walking (right column; dashed lines).

DISCUSSION

We sought to characterize the neuromechanical strategy underlying force generation during walking with and without coordination of postural control by using walking conditions where we scaled locomotor force generation requirements through the provision of BWS versus through providing a range of vertical force targets during the stance phase of walking under negligible postural demands. In support of our hypotheses, we observed a proportional scaling of Fy and Fz GRFs and joint moments with increasing BWS, and changes in muscle activity for muscles that contribute to vertical support demands during propulsion. Also consistent with our expectations, we observed altered Fy/Fz ratios during externally stabilized walking, reflecting a decoupling of postural demands from locomotor requirements, and a different strategy across joints as compared to BWS walking. Additionally, activity of all muscles decreased as we reduced locomotor force generation requirements.

Temporal characteristics were well matched, but stance duration reflected dependence on vertical loading

Similar to our previous findings (11), stride duration was well matched between walking conditions and across force levels at both speeds. This control over walking cadence was important for these manipulations because it ensured that this aspect of locomotor demands remained constant as we altered force generation requirements under walking conditions with and without postural demands. Interestingly, stance duration became progressively shorter as vertical loading on the limbs decreased in both walking environments. This finding is consistent with those of Richter et al. 2017 (30) and likely reflects the importance of load receptors in regulating gait transitions (31,32). Stance duration was also somewhat shorter during externally stabilized compared to BWS walking, which is also likely explained by more gradual offloading of the limbs (and lower average vertical force) that occurred during externally stabilized walking. Indeed, earlier transitions into swing have been noted when the stance limb was prematurely unloaded (33,34).

Leading limb angle decreased with decreasing postural demands in both walking conditions while trailing limb angle only responded to changes in locomotor demands during externally stabilized walking

Leading limb angle decreased with increasing level of postural support via BWS at both walking speeds. It also remained close to the body for all levels of externally stabilized walking, similar to our previous findings (11). Leading limb angle reflects placement of the limb at foot strike to briefly restore the body COM within the base of support and regain body stability (10,35). Providing BWS minimizes accelerations of the body COM allowing the foot to be placed closer to the body to restore stability (23,36). Leading limb position did not change across target force levels at 0.5 m/s, but increased as participants generated less force at 1.0 m/s. This positive slope was driven mostly by the difference between the highest and lowest force targets and may suggest that leading limb angle was less precisely regulated when force generating demands were low.

Trailing limb angle did not change with increasing BWS at both walking speeds. This consistency in push-off position of the limb reflects the fact that we held locomotor demands constant via matched cadence requirements, and force generation had to be sufficient to maintain walking speed and dynamic equilibrium. Trailing limb angle is strongly related to locomotor demands during typical walking, evidenced by trailing limb position contributing almost twice as much as ankle moment to propulsion generation in response to increases in walking speed (29). In contrast, trailing limb angle decreased at both walking speeds when we instead varied locomotor force generating demands in the externally stabilized walking environment. This finding may reflect the relationship between trailing limb angle and locomotor force generation without coordinated postural control. There was no need to accelerate the body COM in relation to the moving treadmill; thus, changing force generation demands did not require participants to maintain the same trailing limb position.

Force ratios were invariant across BWS levels, but decreased across force targets when externally stabilized

The ratio of fore-aft to vertical force during the propulsive portion of the stance phase did not change with the provision of BWS at either walking speed. This ratio reflects the proportion of GRF that is directed to propel the body forward (numerator) and to support body weight (denominator) and is a product of the coordination between posture and locomotor control. Indeed, Gruben & Boehm (2012) (16) refer to the sagittal GRF direction as summarizing the output of the control system and defining net body motion during walking. While we did not compare directly between speeds, it is obvious that the ratio was very different at 1.0 m/s versus 0.5 m/s, reflecting reduced requirement for propulsive force generation in relation to vertical support when walking slower. Our findings of invariant force ratios across BWS levels are similar to those of simulated gravity experiments that demonstrated the fore-aft and vertical force components both scaled in response to reduced vertical support demands (15,20,37,38). This strategy has been proposed to be useful in keeping the GRF vector directed toward the hip joint (i.e., body COM), which has important energetic and stability implications (16,39).

We expected the relationship between fore-aft and vertical GRFs to be similar in magnitude during externally stabilized walking due to similarities in trailing limb angle between walking conditions. However, we did not expect Fy/Fz ratios to remain the same during externally stabilized walking under decreasing locomotor force generation demands due to decoupling of postural demands from locomotor force generation. We observed a decrease in Fy/Fz ratios with decreasing force targets indicating a disproportionate decrease in fore-aft propulsion in relation to vertical GRF. This finding accompanied decreasing trailing limb angles that we observed across force targets. As there was no requirement to accelerate the body COM or retain upright posture in this walking condition, participants altered the control strategy underlying force generation. The decoupling of fore-aft and vertical forces is not characteristic of typical walking and is likely due to the fact that participants were not constrained to coordinate posture and locomotion in this stabilized environment (40).

We observed an earlier onset of propulsion during externally stabilized walking compared to BWS walking. This finding complements our previous results of reduced power absorption when nonimpaired individuals walked in the postural support apparatus (11). A period of power absorption usually occurs just after mid-single support during typical walking that tends to slow forward advancement of the body (41), which can be interpreted as a postural role. This period during mid stance was likely replaced by active locomotor force generation in the present study.

Unexpectedly, we observed disproportionately high Fy/Fz ratios during externally stabilized walking at 0.5 m/s that looked more similar to the ratios at 1.0 m/s as opposed to the corresponding BWS condition at 0.5 m/s. While somewhat surprising, this finding supports the concept that participants did not have to maintain dynamic equilibrium in response to the moving treadmill belt when externally stabilized. The magnitude of Fy/Fz ratio was likely related the biomechanical configuration of the body being more similar to BWS and externally stabilized walking at 1.0 m/s than BWS walking at 0.5 m/s (e.g., slightly more forward trunk inclination). Further investigation of the kinematics of body position would be necessary to confirm this explanation.

Joint moments at the ankle and knee responded similarly to decreases in locomotor force generation requirements between conditions, but differed at the hip

We observed decreases in average ankle, knee, and hip joint moments during propulsion with increasing BWS level, consistent with previous findings of the effects of BWS on lower-limb joint moments (20,27,37). This finding also suggests that the lower-limb joints responded similarly to increases in BWS reflecting a similar control strategy underlying walking across force levels within this condition. The average ankle and knee joint moments similarly decreased with decreasing force target during externally stabilized walking, reflecting their role in generating locomotor force production (42,43). However, the hip joint responded differently across force levels between BWS and externally stabilized walking. The average hip flexor moment decreased with increasing BWS. In addition to accelerating the leg into swing, the hip flexor moment has been proposed to also accelerate the trunk forward through redistribution of power across joints (10,43,44). As BWS increased there was less need to accelerate the body (23), which explains the decrease in hip flexor moment. However, there was no need to accelerate the body during externally stabilized walking; thus, the observed hip flexor moment was smaller compared to BWS walking, likely only related to accelerating the leg into swing. This small hip flexor moment did not change across force targets, because there were no changing demands for accelerating the trunk.

Gastrocnemius and soleus responded to the provision of BWS, but all muscles decreased activity with decreasing locomotor force generation requirements during externally stabilized walking

The roles of lower-limb muscles in support versus progression have been well characterized using dynamic simulations (45–52). Muscles like the vasti and gluteus maximus both support the body in early stance as well as decelerate the body COM (45,50). Gastrocnemius and soleus are responsible for nearly all vertical support in late stance and also propel the body COM forward (45,47). Thus, the same muscles that support the body also modulate forward progression making it difficult to isolate their respective roles during typical walking. At 1.0 m/s we observed a decrease in average muscle activity of the gastrocnemius in response to increasing BWS. This finding is consistent with the contention that gastrocnemius plays an integral role in postural support and is sensitive to changes in vertical support requirements (19) and reduced need to counteract gravity (18,53). At 0.5 m/s both gastrocnemius and soleus decreased average activity with increasing BWS. This finding could be viewed in one of two ways. First, it is possible that soleus plays a larger role than gastrocnemius when locomotor demands of propulsion are high and that lowering locomotor demands via reducing walking speed allows both gastrocnemius and soleus to respond to reduced support requirements. Second, faster walking speeds induce larger accelerations on the trunk, and soleus has been suggested to be responsible for accelerating the trunk in late stance (54). In either case, these findings support previous work that suggests SOL and GAS both contribute to body support, but SOL is the primary contributor to forward acceleration of the trunk (55).

TA, VM, and BF activity were highly variable across participants resulting in no clear relationship to increasing BWS. This finding is consistent with other investigations that observed variable effects of body weight unloading on lower-limb muscle activity (19,21,53,56). These muscles are largely involved in generating angular momentum of the body in early stance and would therefore be more likely to be affected during the first half of the step (57). Also, given that these muscles have combined roles in postural maintenance and forward progression (58,59) we would not necessarily expect relatively small amounts of BWS (i.e., 0 to 30%) to engender changes in the activity of these muscles because BWS still required postural and locomotor coordination.

In contrast, all muscles decreased activity across force targets at both walking speeds during externally stabilized walking. Speed requirements represent one way to manipulate locomotor demands of walking (60). Lower-limbs muscles have been shown to be more responsive to increasing locomotor demands as opposed to scaling of postural demands via BWS, as amplitude of muscle activity responds greatly to the need for larger muscular force output (19,61,62). This is likely for reasons that we outlined in our posture-locomotor interaction model presented in the introduction. As long as postural requirements of walking are present, a certain amount of muscle activity must be delegated to weight support and upright orientation and the pattern of activity reflects the same strategy. However, if the force generation demands of the task change significantly then muscular effort and muscle activity reorganizes accordingly. Externally stabilized walking imposed minimal postural demands. In this unique environment we demonstrated that muscle activity indeed scales with changes in muscular effort via reduced locomotor force generating demands even at the same walking speed. This is likely because walking speed had less influence on muscle activity in this externally stabilized environment due to reduced need to maintain dynamic equilibrium.

Revised postural-locomotor interaction model without postural coordination

The externally stabilized walking environment enabled a decoupling of postural behaviors from locomotor force generation (Figure 8). Results of this study support our

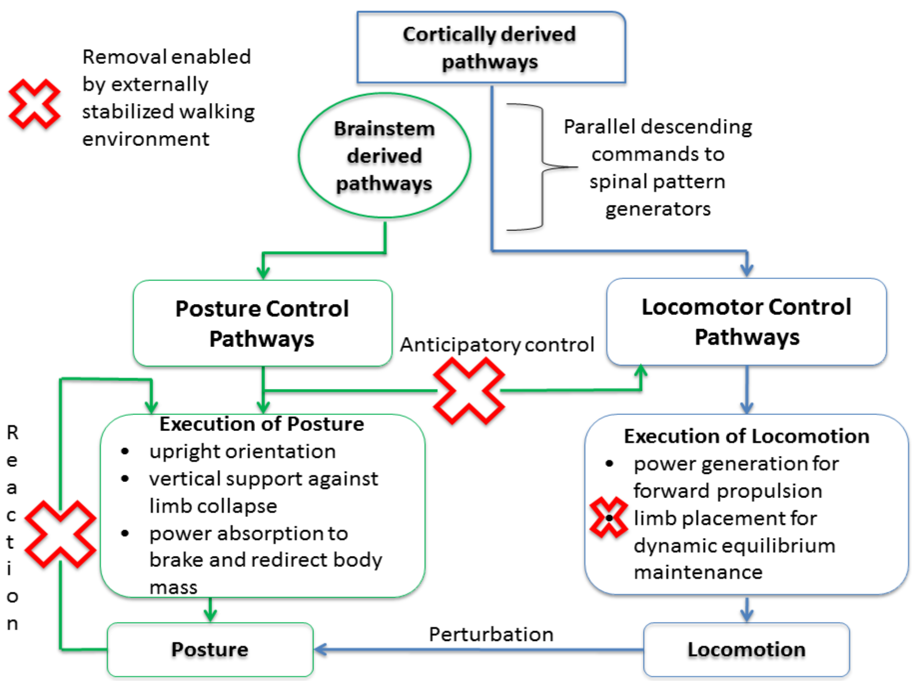

Figure 8. Revised postural-locomotor interaction model to reflect decoupling of postural influence from locomotor force generation behaviors during externally stabilized walking. Red Xs reflect decoupling of postural behaviors from locomotor output.

argument that externally stabilized walking reduced or eliminated anticipatory control influence of posture on locomotor behaviors through removing the need to maintain dynamic equilibrium during walking. Additionally, the moving limbs no longer perturbed posture, minimizing the need to make continual adjustments to the walking pattern to preserve postural orientation. We cannot suggest that the neural control of posture has been entirely removed from this model, as we have no direct measurement of the influence of these control pathways. However, we do suggest that such an externally stabilized environment allows the investigation of locomotor force generation behaviors under minimal postural influence.

Limitations

It is important to recognize the unique nature of the externally stabilized walking environment used in this study. We do not mean to suggest that this environment represents typical walking, but instead argue that it allowed us to mechanically decouple force generation aspects of locomotion from requirements to coordinate postural control. However, one limitation of this approach is that externally stabilized walking did not allow for normal inverted pendulum behavior of the body during the stance phase (63). In this stabilized environment there was no longer a forward “fall” of the body COM in response to gravity, which we aimed to achieve. However, this means that participants could not take advantage of inverted pendulum mechanisms of energy transfer, or elastic energy storage, and thus forces generated by the body were likely replaced through purely active, as opposed to a mix of active and passive, mechanisms.

We only investigated the dominant limb of nonimpaired individuals in the present study. Thus, while interactions between limbs remained intact for BWS walking, externally stabilized walking only emphasized one limb. Future studies observing interactions between limbs could offer important insight into the control of interlimb coordination as well as single to double-limb support transitions under minimal postural demands. We also did not include some potentially important EMG measurements of muscles involved in sagittal plane support and progression (e.g., gluteus maximus, rectus femoris) due to access limitations imposed by our walking environment and limited number of available sensors. More specific investigation of individual muscle roles in these walking environments including muscle activity profiles across the gait cycle could provide deeper insight into the regulation of locomotor force production with and without postural coordination.

Conclusions and future directions

We demonstrated that walking under conditions of decreasing postural requirements, facilitated by the provision of BWS, reflected a similar neuromechanical control strategy, as characterized by resultant force ratios, lower-limb joint moments, and muscle activity, as long as some postural demands (upright orientation and weight support) remained. We also showed that the plantarflexors were the only muscles to respond to changes in vertical loading. In contrast, we demonstrated that when walking under minimal postural requirements, the motor strategy underlying locomotor force generation was dependent instead on the magnitude of locomotor force generation without anticipatory or reactive postural influence. There was a relative decoupling of the fore-aft and vertical GRF components as force demands decreased. We further demonstrated that this strategy was not speed dependent when participants were externally stabilized with no requirement to maintain dynamic equilibrium.

This novel, externally stabilized environment offers a unique opportunity to probe locomotor control of populations that have difficulty coordinating posture and locomotion (i.e., those with posturokinetic disorders). Our findings of similar neuromechanical output during walking with BWS suggests one reason why this method of reducing postural demands of walking may be ineffective in improving stepping behaviors for individuals with posturokinetic disorders. For example, this could be one reason why BWS does not engender large improvements in walking behaviors poststroke (64,65). We plan to investigate the control of locomotor force generation using this externally stabilized walking environment for individuals poststroke who may have difficulty generating locomotor force with their paretic limb. This environment will allow us to determine whether minimizing postural influence during locomotor force generation has the potential to improve stepping control of these individuals.

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LOCOMOTOR FORCE GENERATION DURING WALKING POSTSTROKE WITH AND WITHOUT POSTURAL CONSTRAINT

by

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Prepared for Neurorehabilitation and Neural Repair

Format adapted for dissertation

ABSTRACT

Background: Inappropriate paretic-limb force generation is a hallmark of impaired walking function poststroke. In addition to weakness, the paretic limb misdirects foot forces during stance, due to inappropriately coordinated joint moments. One potential neuromechanical mechanism to explain misdirected foot forces has been described as an inappropriate neural interaction between feedforward postural control pathways and relatively intact spinal locomotor networks. Minimizing postural demands of walking may therefore be one way to encourage more appropriate paretic-limb foot-force direction. Methods: Here we compared locomotor force generation behaviors of N=17 individuals ≥ 6 months poststroke with residual hemiparesis walking at their comfortable speed in two environments: 1) under the provision of 0 to 30% BWS, which partially reduced postural demands of walking but did not fully minimize them relative to locomotor requirements and 2) under matched vertical forces but with full postural support provided through external stabilization from a novel postural support apparatus that mechanically decoupled locomotor force generation behaviors from postural influence by removing the need for reactive and anticipatory postural responses. Hypotheses: We expected that individuals poststroke would exhibit similar foot-force direction, characterized by a ratio of fore-aft to vertical GRF (Fy/Fz ratio), during the second half of the stance phase as we increased BWS due to remaining, yet reduced, requirements of coordinating posture with locomotion. In contrast, we expected that stroke survivors walking with external stabilization provided by the novel support apparatus would generate increased magnitude of Fy/Fz ratio compared to all levels of BWS walking due to the ability of weakened locomotor pathways to better express

themselves in the absence of excessive postural influence. Results: Our primary hypothesis regarding Fy/Fz ratios of the paretic limb was not supported. The paretic-limb Fy/Fz ratio was significantly smaller than the nonparetic limb during BWS walking and remained invariant across BWS levels; however, it was not more appropriately directed during externally stabilized walking. Conclusions: These findings suggest that individuals poststroke may rely on postural control mechanisms to compensate for loss of voluntary locomotor control. The paretic limb appeared to rely on postural functions in an environment where they were not required for successful performance of the task.

INTRODUCTION

Inappropriate paretic-limb force generation is a hallmark of impaired walking function poststroke (1–4). In addition to weakness, the paretic limb misdirects foot forces during stance, due to inappropriately coordinated joint moments (2,3,5–9). Misdirected forces manifest as reduced propulsion and relatively longer and improperly timed braking ground reaction forces (GRFs) (2,5,6). Reduced contribution from the plantarflexor muscles to the locomotor functions of forward propulsion and swing initiation leads to slow walking speeds, inefficient energy utilization, and greater fall risk (2,10–13).

One potential neuromechanical mechanism to explain misdirected foot forces has been described as an inappropriate neural interaction between feedforward postural control pathways and relatively intact spinal locomotor networks (14–18). We previously proposed that nonimpaired postural and locomotor coordination during functional walking is the result of separate, parallel descending pathways originating from the cerebral cortex and brainstem that synapse on locomotor networks within the spinal cord to facilitate simultaneous postural control and coordinated stepping (19) (Graham et al. 2018 unpublished) (20,21). However, following stroke cortical/subcortical damage may lead to subsequent reorganization of these postural and locomotor control systems (22–24).

Enhanced activity of alternate descending pathways, like operational postural networks (25,26), may exert excessive influence over weakened cortically derived locomotor control pathways involved in facilitating skilled locomotor patterns (24,27). Specifically, pathways like the reticulospinal pathway controlling postural and mechanically coupled muscle groupings may be disinhibited due to altered input from the corticospinal tract to the reticular formation (28). Excessive input from these brainstem-derived pathways can help explain abnormal coupling of joint moments (9,29) and reduced complexity of motor modules underlying the gait cycle poststroke (30,31).

Prior work from our laboratory demonstrated that individuals poststroke generated misdirected foot forces during a nonseated pedaling task when required to coordinate postural control (14). However, when performing a seated pedaling task, which imposed minimal postural task demands, the ratio of shear-to-normal crank force against the pedal was more appropriate. The pedaling task enabled constraint of task mechanics, but a critical remaining question was whether a similar mechanism occurs during less-constrained, yet more functionally relevant, upright walking.

Based on the proposed neurophysiological mechanisms discussed above and our previous work in pedaling, we propose a modification to our nonimpaired posture-locomotor interaction model that describes undue influence from brainstem-involved pathways poststroke preventing weakened cortical pathways from being appropriately expressed during walking (Figure 1 thick red vs. dashed lines). This undue influence of postural control mechanisms over locomotor control may result in postural functions (e.g., weight support, power absorption) predominating over locomotor functions (e.g., propulsion, power generation) (30) leading to a foot-force direction bias that favors support as opposed to progression.

There is similar evidence with upper limb movements that postural orientation influences movement capacity poststroke, evidenced by abnormal coupling of elbow flexion and shoulder extension during reaching that was exacerbated in sitting as opposed to supine positions (32). Other work has demonstrated a negative impact of actively

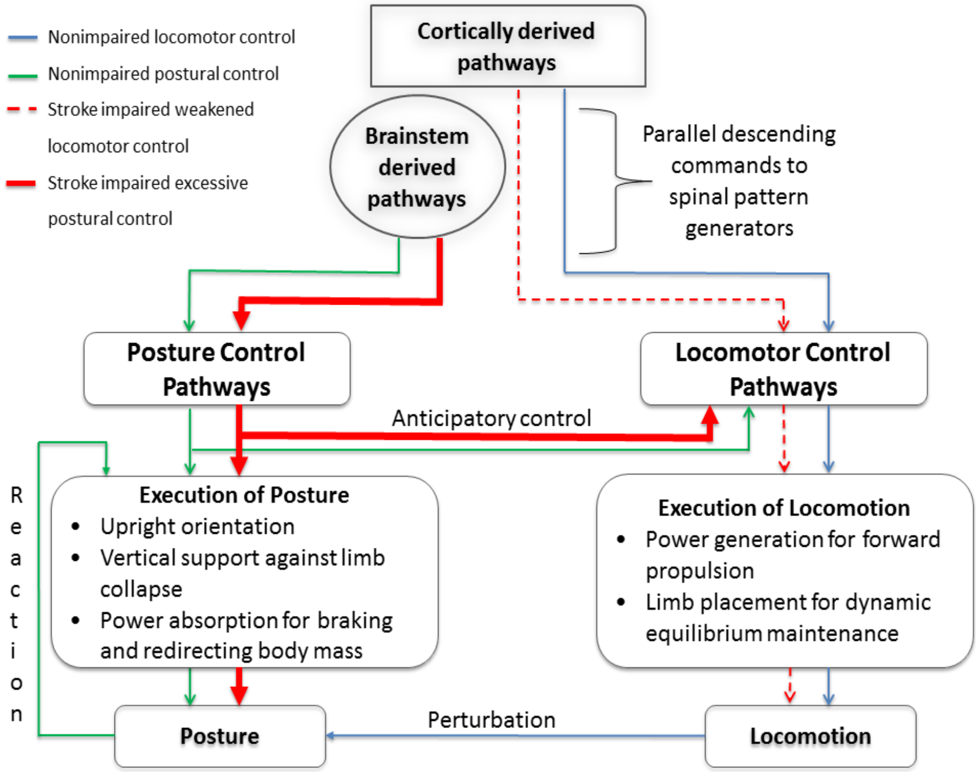

Figure 1. Nonimpaired postural-locomotor interaction model during typical walking as described in Graham et al 2018 (green and blue lines) and proposed inappropriate postural influence over weakened locomotor control poststroke (thick and dashed red lines, respectively).

supporting the limb against gravity on speed and range of elbow extension movements that was mitigated through externally supporting the limb (15,16). One way to offer support against gravity during walking is to provide body-weight support (BWS) (33–36). BWS decreases postural demands of walking through reducing requirements to produce vertical support and control trunk accelerations (37–40). However, standard BWS protocols do not fully offload the limbs or support the trunk in an upright position; thus, walking with BWS still requires coordination between posture and locomotor functions. Indeed, the neuromechanical control strategy underlying walking remains similar under the provision of 0 to 30% BWS, evidenced by invariant ratios of fore-aft to vertical (Fy/Fz ratios) GRFs (19,41) (Graham et al. 2018 unpublished). The Fy/Fz ratio during propulsion reflects the proportion of the GRF performing forward progression (numerator) versus support (denominator), which reflects control over the net angular momentum of the whole body during stance and is essential for coordinated posture and locomotor control (42–45).

A critical question remains as to whether fully reducing postural demands of walking, like minimizing the need for reactive and anticipatory postural responses, has the potential to allow expression of appropriate locomotor control of individuals poststroke. We previously developed an externally stabilized walking environment that allowed us to mechanically decouple locomotor force generation behaviors from postural influence while providing matched vertical GRFs and similar trailing limb positions to those that occur during typical walking (19,46) (Graham et al. 2018 unpublished). This apparatus addressed all aspects of postural control by stabilizing an individual in an upright position, fully offloading upper body and trunk mass from the lower limbs, and minimizing accelerations acting on the body. It enabled nonimpaired individuals to perform locomotor force generation with negligible postural influence, through removing reactive and anticipatory postural influence from locomotor output. Here we used the same experimental apparatus to compare locomotor force generation behaviors of individuals poststroke between walking in two environments, 1) under the provision of BWS, which partially reduced postural demands of walking but did not fully minimize them relative to locomotor requirements and 2) in the externally stabilized walking environment that mechanically decoupled locomotor force generation behaviors from postural influence.

We expected that individuals poststroke would exhibit similar foot-force direction, characterized by the Fy/Fz ratio during the second half of the stance phase, as we increased the level of BWS due to remaining, yet reduced, requirements of coordinating posture with locomotion. In contrast, we expected that stroke survivors walking in the externally stabilized environment, with minimal anticipatory and reactive postural demands, would generate increased magnitude of Fy/Fz ratio compared to all levels of BWS walking due to the ability of weakened locomotor pathways to better express themselves through performing propulsion in an environment that did not require postural coordination. We additionally expected large asymmetry in power generation between the nonparetic and paretic limbs during BWS walking. However, we anticipated increased lower-limb power generation and decreased power absorption during externally stabilized walking compared to with BWS, particularly for the paretic limb, further reflecting an improved ability of the paretic limb to express remaining locomotor capacity under minimal postural influence.

As a secondary set of hypotheses, we expected reduced leading limb angles (i.e., similar to those observed with nonimpaired individuals) during externally stabilized compared with BWS walking, reflecting reduced requirement to control limb placement for dynamic equilibrium maintenance. We also expected similar trailing limb angles between BWS and externally stabilized walking, reflecting similar locomotor demands of maintaining walking speed and cadence.

METHODS

Seventeen individuals with poststroke hemiparesis but no other uncontrolled health conditions participated in this study (Table 1 for participant characteristics). Inclusion criteria were males and females who had sustained a unilateral cortical or subcortical stroke that resulted in hemiparesis (≥ 6 months post injury), older than 19 years of age, able to walk independently with or without an assistive device, medically stable (controlled hypertension; stable cardiovascular status), and able to provide written informed consent. Exclusion criteria were history of serious cardiac disease, uncontrolled blood pressure (systolic pressure >180 mmHg, diastolic blood pressure >110 mmHg), presence of cerebellar and brainstem deficits like gait ataxia, severe cognitive disorders, inability to follow simple commands, uncontrolled respiratory or metabolic disorders, and/or major or acute musculoskeletal problems. All participants signed informed consent as approved by the Institutional Review Board of the University of Alabama at Birmingham.

Table 1. Participant characteristics. Mean [range].

Age (years)	51.2 [26 to 69]
Gender	11 male; 6 female
Paretic limb	6 right; 11 left
Lower-limb Fugl Meyer score (max 34 points)	22 [12 to 33]
Overground comfortable walking speed (m/s)	0.60 [0.15 to 1.02]
Comfortable walking speed in device (m/s)	0.56 [0.2 to 0.98]

Experimental walking environments

Participants performed all walking trials over a motorized, dual-belt, instrumented treadmill (BERTEC, Columbus, OH, USA). We used the KineAssist™ robotic device (HDT Global, Solon OH) to deliver BWS and ensure a safe environment for participants. In the event of a loss of balance the device catches a participant after a preprogramed descent distance (e.g., 4 inches). The KineAssist connected to participants via a pelvic mechanism situated at their approximate center of mass (COM). The pelvic mechanism is equipped with a pelvic harness containing bilateral force sensors that sense vertical forces applied by the pelvis and sends this information via a closed loop controller to provide a near constant level of BWS. The pelvic mechanism allows all six degrees of freedom around the pelvis/hip complex while walking (47). However, to ensure greater similarity between BWS and externally stabilized walking (described below) we only allowed three degrees of freedom, including surge (relative forward/backward movement over the treadmill), heave (vertical COM motion), and pitch (forward/backward trunk/pelvic tilting). With a locking mechanism we prevented sway (side-to-side pelvic translation), roll (hip hiking), and yaw (left/right pelvic rotation). Thus, both walking environments emphasized control of only sagittal plane walking mechanics, which are not affected by these limitations (46).

The KineAssist further served as the base for the externally stabilized walking environment (Figure 2). This environment included the addition of a postural support apparatus composed of a backboard and seat complex as described fully in Graham et al. 2018 (46). Briefly, the backboard component externally supported participants in an upright position and reduced their need to actively maintain upright orientation during walking. The seat fully offloaded their upper body and trunk mass from their lower limbs without impeding hip movements, reducing vertical support requirements to no more than limb weight. In this configuration participant's feet were still in contact with the treadmill belt in order to generate GRFs. Shoulder pads additionally fixed participants in place by preventing vertical COM accelerations.

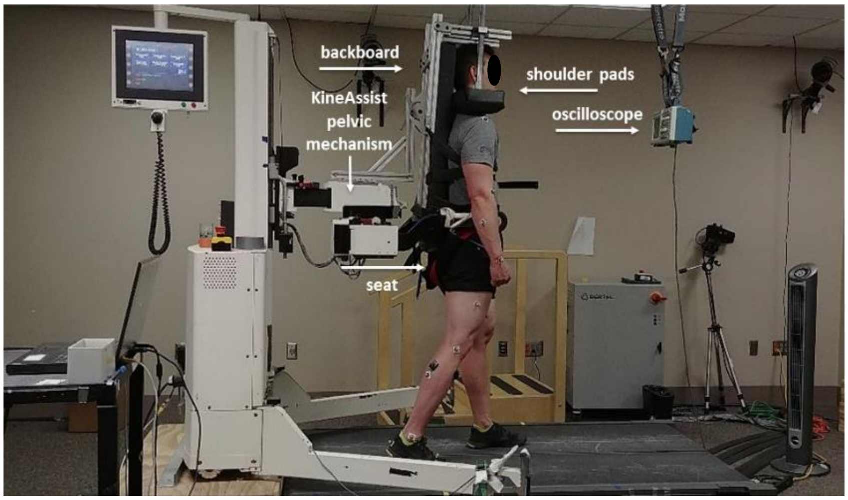

Figure 2. Externally stabilized walking environment

BWS walking protocol

Participants walked under four levels of BWS from 0 to 30% body weight. Prior to commencing trials, we provided participants time to acclimate to walking in the KineAssist device. Our goal was to have participants walk at their comfortable overground walking speed and we achieved this in most cases (see Table 1 for difference between overground and treadmill walking speeds). Once we determined the comfortable speed of each participant, we determined their walking cadence for their nonparetic limb only to accommodate temporal asymmetry. We provided this cadence via a metronome on all subsequent walking trials in an effort to encourage participants to maintain comparable spatiotemporal walking characteristics across trials.

We provided approximately one minute for participants to acclimate to each level of BWS before collecting data (48,49). Participants walked at each level of BWS for a goal of 90 seconds and no more than six minutes of walking across trials. We removed BWS and offered participants sufficient rest between each level of support. We initiated each trial when participants confirmed that they felt recovered from the previous trial. In order to obtain a range of vertical GRF targets for externally stabilized walking trials, following completion of BWS trials we calculated the average peak vertical GRF that occurred during push off for each level of BWS.

Externally stabilized walking protocol

We allotted participants approximately 20 minutes to recover following BWS walking trials while we processed their target forces and prepared the externally stabilized walking environment. Participants first performed externally stabilized walking while targeting vertical GRFs with their nonparetic limb. We made this decision in order to provide participants with the opportunity to first learn the task with their stronger limb. Participants performed up to 90 seconds of walking (no more than six minutes total) while targeting each of four decreasing levels of vertical GRF (target 1 (highest) to target 4 (lowest)) derived from their respective BWS trials. We displayed these target vertical GRF values ± 5 % on an oscilloscope for the nonparetic limb only. We instructed participants to hit the target by pushing their foot against the moving belt beginning when their foot was directly beneath them during the stance phase.

We assisted participants out of the apparatus and provided them with a ten-minute break following completion of their nonparetic-limb externally stabilized walking trials.

When participants were adequately recovered we assisted them back into the support apparatus to perform trials with their paretic limb. The protocol was identical to that of the nonparetic limb; however, in many cases participants were not able to achieve the higher target force values. In these instances, we provided participants with an additional force target of approximately 10% lower than their lowest target force (target 4) derived from the 30% BWS trial in order to measure steps with a range of vertical GRF generation.

Data acquisition

We collected GRFs at 1000 Hz via the Bertec, dual-belt, force-instrumented treadmill and kinematic data at 100 Hz via an eight-camera, Qualisys motion capture system (Qualisys Inc., Gothenburg, Sweden) with 33, 1-cm passive-reflective markers placed bilaterally with three reflective markers defining each segment as follows: acromion processes, manubrium of sternum, thoracic spine level with sternum, ASIS, sacrum, midline and lateral thighs, midline and lateral shanks, lateral malleoli, first and fifth digits, and calcaneus (example marker locations in Figure 2). We used Visual 3D (C-Motion, Germantown, MD, USA) to obtain marker trajectories for limb angle calculations and joint powers and performed all post processing of the data in MATLAB (Mathworks®, version R2016a).

Treatment of BWS versus externally stabilized data

For both walking conditions we calculated all variables over either a full gait cycle (foot strike to ipsilateral foot strike) or stance phase (foot strike to ipsilateral foot off). We defined these gait events with a vertical GRF threshold value of 2.0% body weight. We retained all data from BWS walking trials except in the event of a stumble or crossover step between force plates. However, for externally stabilized walking trials we retained steps that resulted in four distinct force categories. This process differed for the nonparetic and paretic limbs due to poorer force targeting accuracy of the paretic limb.

For the nonparetic limb we treated the data in the same manner as our previous investigations of nonimpaired individuals by eliminating steps where the vertical GRF value did not fall within ± 5% of the trial average for a given force target (19) (Graham et al. 2018 unpublished). In contrast, the paretic limb produced a variety of vertical force levels within a given trial preventing us from using this selection process. We instead rank ordered steps from highest to lowest average vertical GRF and then categorized steps based on force levels that corresponded to the trial averages for the nonparetic limb. These categories were 50 to 60% (target 1), 40 to 50% (target 2), 30 to 40% (target 3), and 20 to 30% (target 4) body weight. This approach yielded a different number of individuals that achieved each target category as indicated in our results.

Spatiotemporal variables

We calculated stride duration as the time elapsed from foot strike to ipsilateral foot strike and stance duration as the time elapsed from foot strike to ipsilateral foot off. We calculated stance phase limb angle (leading and trailing limb) as the angle between the vector connecting the dominant-limb lateral toe and ASIS marker and the laboratory’s vertical axis (50). Negative values indicate leading limb position at foot strike and positive values trailing limb position at foot off.

Kinetic variables

We low-pass filtered GRFs using a 4th-order Butterworth zero-lag filter with a cutoff of 20 Hz and calculated average vertical and fore-aft GRF values normalized to body weight (% BW) during the stance phase. We used the ratio of fore-aft to vertical GRF (Fy/Fz ratio) to characterize GRF direction during the second half of the stance phase when we would expect participants to be performing propulsion. For example, if the average fore-aft force was a positive value of 8% body weight and average vertical GRF was 60% body weight during the second half of stance, the Fy/Fz ratio would be 8/60 = 0.13.

Visual 3D calculates joint powers as the product of the net muscle moment and joint angular velocity ($P = M \times \omega$). We normalized joint powers to body mass (W/kg) and integrated the area under the power-time curve to obtain mechanical work ($W = \int P \times dt$) (J/kg) where negative work indicates power absorption and positive work indicates power generation. We present the breakdown of positive and negative work for each individual joint and conducted statistical comparisons on total positive and negative work summed across the joints.

Statistical analyses

We conducted statistical tests using SPSS version 24 (Chicago, Illinois, USA) with separate statistical tests to compare nonparetic versus paretic limbs during BWS walking, nonparetic limb during BWS walking versus externally stabilized walking, and paretic limb during BWS walking versus externally stabilized walking.

We used 2 (limb) x 4 (force level) repeated measures ANOVAs to compare stride and stance durations, average vertical GRFs, Fy/Fz ratios, total positive and negative work, and limb angles for the nonparetic versus paretic limbs during BWS walking. We used 2 (condition) x 4 (force level) repeated measures ANOVAs to compare these same variables for the nonparetic limb between BWS and externally stabilized walking. We employed Greenhouse Geiser corrections, as necessary, for violations of sphericity, a priori alphas $p \leq 0.05$, and Bonferroni corrections for multiple comparisons.

We addressed paretic limb comparisons between BWS and externally stabilized walking differently, as not all participants were represented in each of the four vertical force categories. We had n = 8 participants who achieved the highest force target 1, n = 12 achieved the next highest target 2, n = 15 achieved force target 3, and n = 17 achieved force target 4. We present descriptive data of stride and stance durations and average vertical GRFs for all individuals who hit each target.

In order to make statistical comparisons between BWS and externally stabilized walking for the paretic limb, we first investigated participants who attained all force targets (n = 8) with a 2 (condition) x 4 (force level) repeated measures ANOVA for the primary outcome measure of Fy/Fz force ratios. To characterize the performance of all participants we used the highest force target that a participant attained as a representation of their best effort and compared this value to their respective BWS condition. For example, if a participant's highest force target achieved was target 2, we compared variables at this value to their respective 10% BWS condition in order to make comparisons between conditions where similar peak vertical GRFs occurred. For these pairwise comparisons we used Wilcoxon signed rank tests for data that did not meet parametric assumptions and paired *t* tests for all other comparisons between BWS and externally stabilized walking with an alpha of $p \leq 0.05$. Finally, we used Spearman's rank correlations to determine if there were relationships between force target achieved and each variable investigated.

RESULTS

We first present comparisons between limbs for BWS walking (Section I). We follow with findings for our primary question of interest regarding the paretic limb during BWS versus externally stabilized walking (Section II). Finally, we present comparisons of the nonparetic limb during BWS versus externally stabilized walking (Section III).

Section I. BWS walking comparisons between nonparetic and paretic limbs

Temporal characteristics and vertical GRF comparisons demonstrated that the BWS walking condition was well controlled across BWS levels.

Stride and stance time during BWS walking

Participants maintained walking cadence while following the metronome cue during BWS walking, evidenced by no significant limb x level interaction ($p = 0.34$), main effect of level ($p = 0.22$), or main effect of limb ($p = 0.40$) for stride time. Average stride duration for each level of support ranged from 1.59 sec 95% CI [1.28 to 1.90] at 0% BWS to 1.68 sec 95% CI [1.37 to 1.98] at 30% BWS. We also did not detect a limb x level interaction for stance time ($p = 0.43$), or a main effect of level ($p = 0.08$). However, we observed a main effect of limb ($F(1, 16) = 34.8$, $p \leq 0.001$), indicating that the nonparetic limb spent longer in stance compared to the paretic limb (1.24 sec 95% CI [0.97 to 1.52] vs. 1.11 95% CI [0.84 to 1.38]).

Average vertical GRFs during BWS walking

We detected a significant limb x level interaction ($F(3, 48) = 3.0$, $p = 0.04$) during BWS walking where the paretic limb vertical GRF did not decrease as steeply in response to the provision of BWS as the nonparetic limb (Table 2). We also observed significant main effects of limb ($F(1, 16) = 592$, $p \leq 0.0001$) and level ($F(1.3, 21.2) = 363.4$, $p \leq 0.0001$), indicating higher average vertical GRFs produced by the nonparetic limb and decreasing average vertical GRFs with increasing BWS. Post hoc comparisons revealed significant differences ($p \leq 0.0001$) for both limbs between all levels of BWS demonstrating successful application of the BWS mechanism.

Table 2. Average vertical GRFs for the nonparetic and paretic limbs during BWS walking. Mean [95% CI].

	0%	10%	20%	30%
Nonparetic limb (% BW)	75.5 [72.4 to 78.6]	68.5 [65.8 to 71.1]	61.1 [58.6 to 63.7]	54.0 [51.9 to 56.1]
Paretic limb (% BW)	63.7 [59.2 to 68.1]	58.2 [54.1 to 62.4]	51.7 [48.5 to 54.8]	45.2 [42.5 to 48.0]

*All post hoc comparisons between levels for a limb significantly different at $p \leq 0.0001$

Results of our primary hypothesis regarding force ratios across BWS levels

The Fy/Fz ratio did not change across BWS levels for either limb, evidenced by no limb x level interaction ($p = 0.57$) or main effect of level ($p = 0.60$). The nonparetic limb had greater force ratios compared to the paretic limb, evidenced by a main effect of limb ($F(1, 16) = 32.8$, $p \leq 0.0001$) (Figure 3).

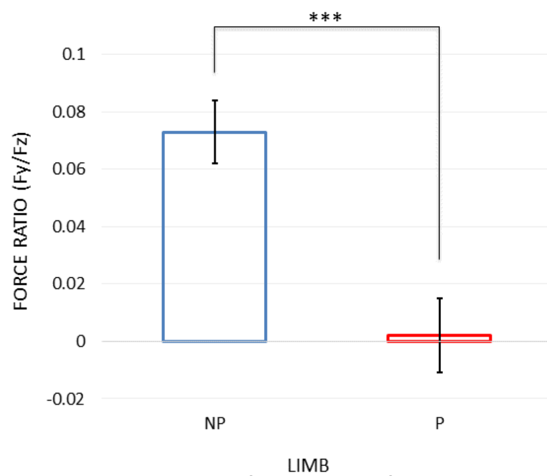

Figure. 3. Average force ratios for the nonparetic (NP; blue) and paretic (P; red) limbs during BWS walking. Error bars reflect SE. ***p≤0.0001

Results of our primary hypotheses regarding positive and negative work during BWS walking

We observed a reduction in total positive work across BWS levels along with larger positive work for the nonparetic compared to paretic limb (Figure 4), evidenced by main effects of level ($F(1.5, 24.2) = 19.1$, $p \leq 0.0001$) and limb ($F(1,16) = 54.7$, $p \leq 0.0001$). We also observed a significant limb x level interaction ($F(3, 48) = 9.4$, $p \leq 0.0001$), which we decomposed by running separate one-way repeated measures ANOVAs for each limb. We detected a main effect of level for both the nonparetic ($F(1.8, 28.5) = 24.0$, $p \leq 0.0001$) and paretic limbs ($F(1.6, 25.4) = 6.6$, $p = 0.008$), with post hoc comparisons revealing a greater number of significant decreases ($p \leq 0.01$) between levels for the nonparetic compared to paretic limb as indicated by superscript numbers above bars in Figure 4.

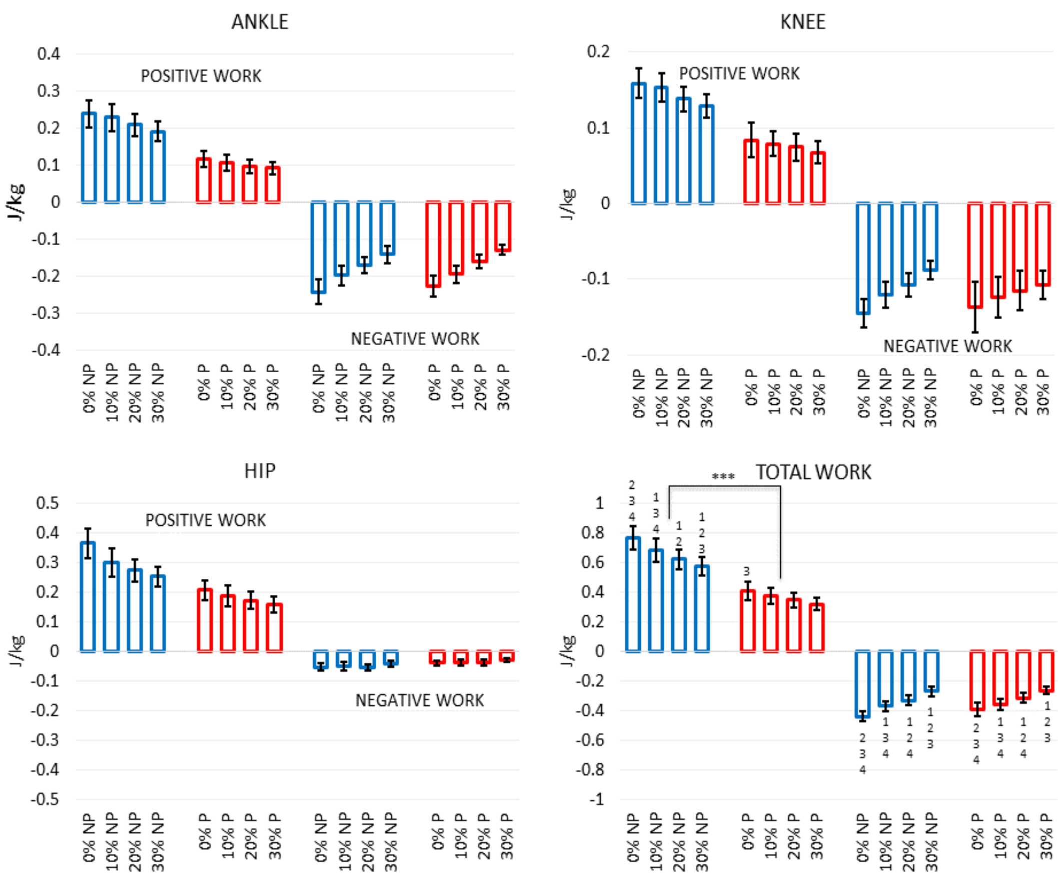

Figure 4. Positive and negative work across the lower-limb joints (ANKLE, KNEE, HIP) and sum of joints (TOTAL) for the nonparetic (blue) and paretic (red) limbs during BWS walking. Values are mean (SE). Significant pairwise comparisons reflected as numbers above or below each set of bar graphs for total work. All significant at Bonferroni corrected $p \leq 0.01$. *** indicates significant main effect of limb $p \leq 0.0001$.

Total negative work decreased across BWS levels ($F(1.4, 23.2) = 65.8$, $p \leq 0.0001$), but we did not observe differences between limbs ($p = 0.45$) or a limb x level interaction ($p = 0.13$). Post hoc comparisons revealed decreases in negative work between all levels of BWS for both limbs ($p \leq 0.01$) (Figure 4).

Leading and trailing limb angles during BWS walking

Leading limb angles did not significantly decrease across BWS levels for either limb ($p = 0.08$). We did not detect a significant limb x level interaction ($p = 0.45$);

however, we did observe a significant main effect of limb (F (1, 16) = 16.3, $p < 0.01$) indicating that the leading limb was placed further in front of the body on average for the nonparetic compared to paretic limb. Leading limb angles ranged from -15.9 degrees 95% CI [-18.3 to -13.5] at 0% BWS to -15.1 degrees 95% CI [-17.5 to -12.6] at 30% BWS for the nonparetic limb and -13.0 degrees 95% CI [-14.7 to -11.4] at 0% BWS to -11.8 degrees 95% CI [-13.5 to -10.0] at 30% BWS for the paretic limb.

Trailing limb angles did not change across BWS levels for either limb ($p = 0.77$). We did not detect a significant limb x level interaction ($p = 0.79$) or main effect of limb ($p = 0.08$). Trailing limb angles for the paretic limb ranged from 15.9 degrees 95% CI [12.8 to 19.0] at 0% BWS to 16.3 degrees 95% CI [13.2 to 19.4] at 30% BWS. Trailing limb angles for the nonparetic limb ranged 17.4 degrees 95% CI [14.9 to 20.0] at 0% BWS to 17.5 degrees 95% CI [14.8 to 20.2] at 30% BWS.

Section II. BWS versus externally stabilized walking for the paretic limb

Participants exhibited varying accuracy in achieving all four vertical force levels with their paretic limb during externally stabilized walking (Figure 5 for a participant who achieved all 4 targets and Figure 6 for a participant who only achieved the lowest 2). Table 3 shows descriptive statistics for variables reflecting control over the externally stabilized walking condition for the paretic limb. There was a different number of participants characterized at each level. Participants who achieved the highest force target 1 are reflected in all force categories. Participants who achieved target 2 are also represented at targets 3 and 4. Participants who achieved target 3 are also represented at target 4. Finally, we had two participants who only achieved the lowest force target.

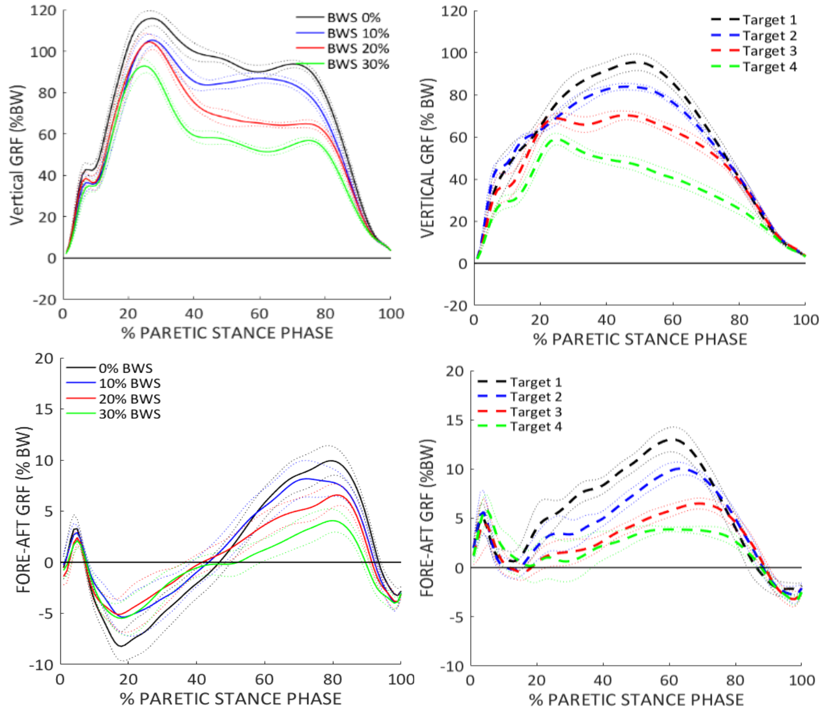

Figure 5. Average vertical (top) and fore-aft (bottom) GRFs for the paretic limb during BWS (left; solid) vs. externally stabilized (right; dashed) walking of a participant who attained all force targets.

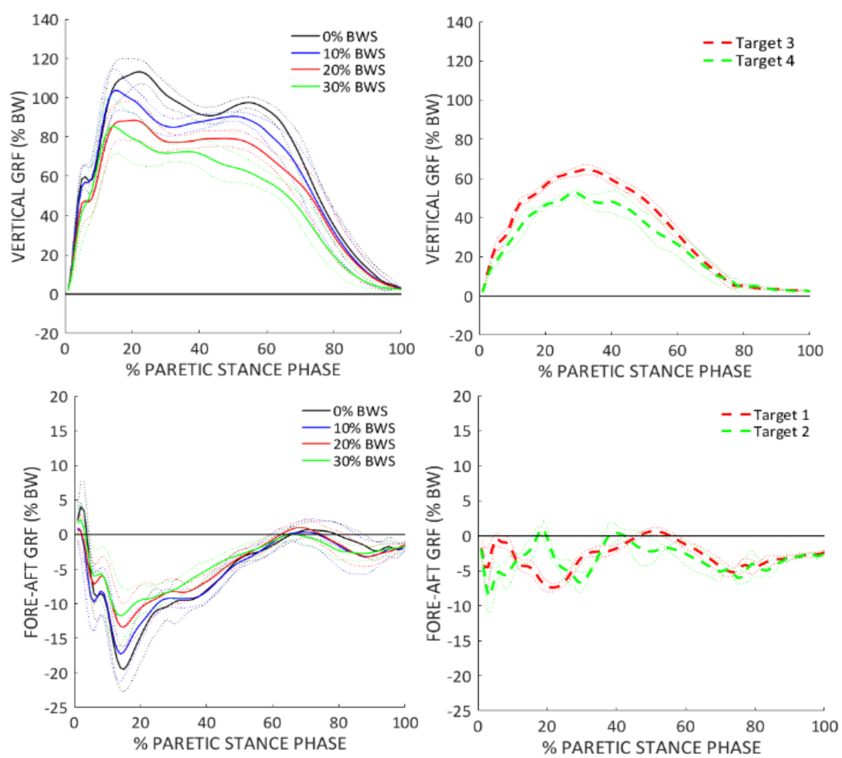

Figure 6. Average vertical (top) and fore-aft (bottom) GRFs for the paretic limb during BWS (left; solid) vs. externally stabilized (right; dashed) walking of a participant who achieved only targets 3 and 4.

Table 3. Variables reflecting control over the externally stabilized walking condition for the paretic limb. Mean (SD).

Vertical Force Target 1=highest	Participants achieving each target (n=17 possible)	Average Vertical GRF (%BW)	Stride Time (s)	Stance Time (s)
1	n=8	53.6 (2.3)	1.60 (0.40)	1.06 (0.36)
2	n=12	44.5 (2.7)	1.72 (0.61)	1.17 (0.47)
3	n=15	34.9 (1.7)	1.74 (0.56)	1.16 (0.47)
4	n=17	26.3 (1.8)	1.72 (0.50)	1.11 (0.44)

Results of our primary hypothesis regarding force ratios for the paretic limb during BWS vs. externally stabilized walking

For the eight participants who attained all force targets we detected a significant condition x level interaction ($F(3,21) = 17.0$, $p \leq 0.0001$) for force ratios between conditions, but no main effect of condition ($p = 0.51$). We conducted separate repeated measures ANOVAs for each condition. Force ratios were not significantly different across levels for the paretic limb during BWS walking ($p = 0.46$); however, they significantly decreased across force targets during externally stabilized walking ($F(1.2, 8.2) = 18.5$, $p = 0.002$) (Figure 7).

We did not observe significantly different Fy/Fz ratios ($p = 0.92$) between participant's best effort externally stabilized walking condition and their respective BWS condition for the paretic limb (Fy/Fz = -0.003, SE (0.02) during externally stabilized versus Fy/Fz = -0.001, SE (0.01) during BWS walking). We observed a modest but significant correlation between force target attained during externally stabilized walking and resulting force ratio ($\rho = 0.51$, $p = 0.04$).

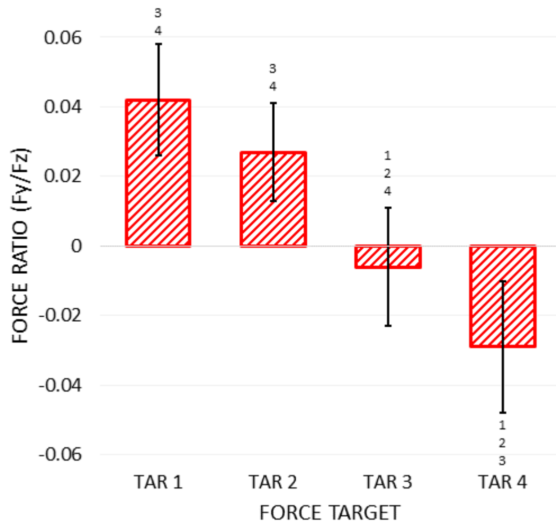

Figure 7. Paretic limb force ratios across targets (TAR) during externally stabilized walking for participants who hit all 4 targets. Superscript numbers reflect significant post hoc comparisons at $p < 0.05$.

Results of our primary hypothesis regarding total work during BWS vs. externally stabilized walking

The paretic limb did not generate significantly different total positive work between BWS and externally stabilized walking but did reduce total negative work (Table 4). Total positive work was significantly correlated with the force target achieved ($\rho = 0.58$, $p = 0.02$), which we expected since larger force generation was necessary to hit higher vertical force targets. Total negative work was also significantly correlated with force target achieved ($\rho = -0.49$, $p = 0.05$) indicating that participants who achieved higher targets also relied on negative work to do so. The breakdown of work across joints looked very similar between conditions for power generation, but different for power absorption. The small reduction in total negative work during externally stabilized walking appeared to be driven by the ankle, while the knee actually increased negative work during externally stabilized compared to BWS walking.

Table 4. Paretic limb positive and negative work for BWS vs. externally stabilized walking. Mean (SD) except for total positive work which is median [range].

	BWS	Externally Stabilized	p value
Total Positive Work (J/kg)	0.37 [0.07 to 0.58]	0.31 [0.11 to 0.91]	0.91
Ankle Positive Work (J/kg)	0.11 (0.06)	0.11 (0.08)	
Knee Positive Work (J/kg)	0.08 (0.06)	0.07 (0.04)	
Hip Positive Work (J/kg)	0.14 (0.09)	0.12 (0.14)	
Total Negative Work (J/kg)	-0.37 (0.10)	-0.30 (0.10)	≤ 0.01
Ankle Negative work (J/kg)	-0.21 (0.09)	-0.09 (0.05)	
Knee Negative Work (J/kg)	-0.13 (0.08)	-0.19 (0.10)	
Hip Negative Work (J/kg)	-0.04 (0.03)	-0.03 (0.02)	

Limb angles between walking conditions for the paretic limb

We observed significantly reduced leading limb angles during externally stabilized walking of -8.5 degrees (SD = 3.6) versus BWS walking of -12.7 degrees (SD = 2.8) ($p \leq 0.001$), and a significant increase in trailing limb angle during externally stabilized walking of 20.4 degrees (SD = 4.2) versus 16.6 degrees (SD = 6.0) ($p \leq 0.01$) during BWS walking. Neither limb angle was significantly correlated with force target achieved ($p > 0.05$).

Section III. BWS vs. externally stabilized walking for the nonparetic limb

Temporal characteristics and vertical GRF comparisons demonstrated that the nonparetic limb behavior was well controlled between BWS and externally stabilized walking conditions.

Stride and stance time between conditions for the nonparetic limb

Participants maintained cadence between BWS and externally stabilized walking trials with their nonparetic limb, evidenced by no condition x level interaction ($p = 0.44$), main effect of level ($p = 0.39$), or main effect of condition ($p = 0.35$) for stride duration. Average stride duration was 1.63 sec 95% CI [1.33 to 1.93] during BWS walking versus 1.70 sec 95% CI [1.42 to 1.98] during externally stabilized walking.

We also observed no differences in stance time between BWS and externally stabilized walking for the nonparetic limb ($p = 0.14$). We did not detect a main effect of level ($p = 0.39$) or condition x level interaction ($p = 0.37$). Average stance duration was 1.24 sec 95% CI [0.97 to 1.52] for BWS walking and 1.15 sec 95% CI [0.90 to 1.40] for externally stabilized walking.

Average GRFs between conditions for the nonparetic limb

We observed a bimodal vertical and alternating braking and propulsive GRF during BWS walking versus a unimodal vertical GRF with little to no braking component for externally stabilized walking (Figure 8). Average vertical forces decreased across force levels ($F(1.8, 28.8) = 373.1$, $p \leq 0.0001$). We detected a significant condition x level interaction ($F(2.0, 32.6) = 5.5$, $p \leq 0.01$) reflecting differences in the way the

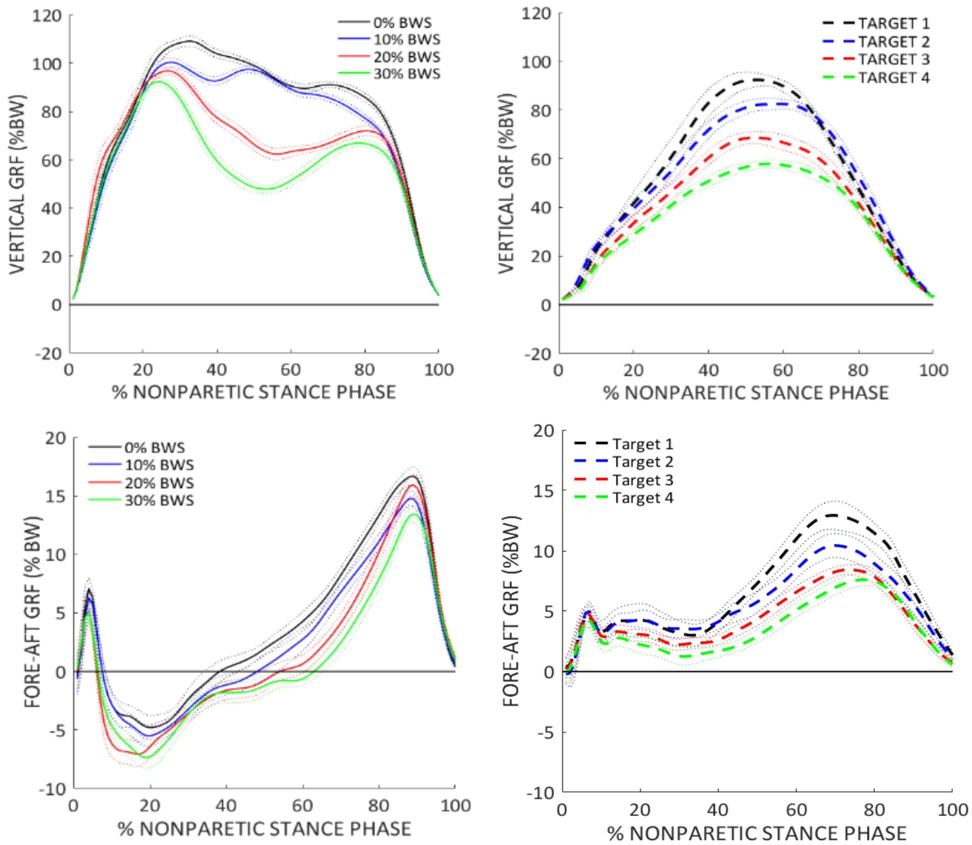

Figure 8. Average vertical (top) and fore-aft (bottom) GRF profiles of a representative individual poststroke across steps performing BWS walking (left column; solid lines) vs. externally stabilized walking (right column, dashed lines).

nonparetic limb decreased in vertical force between conditions. We also detected a main effect of condition ($F(1, 16) = 249.8$, $p \leq 0.0001$) indicating that average vertical GRFs were higher during BWS walking than externally stabilized walking (Table 5). Post hoc comparisons for externally stabilized walking showed significant differences ($p \leq 0.0001$) between all vertical force targets, indicating that we achieved distinct force categories with our selection procedure to retain steps falling within ± 5% of the average vertical force for a given target.

Table 5. Average vertical GRFs for BWS vs. externally stabilized walking for the nonparetic limb. Mean [95% CI].

	0% / TAR 1	10% / TAR 2	20% / TAR 3	30% / TAR 4
BWS walking (% BW)	75.5 [72.4 to 78.6]	68.5 [65.8 to 71.1]	61.1 [58.6 to 63.7]	54.0 [51.9 to 56.1]
Externally stabilized walking (% BW)	55.0 [52.5 to 57.6]	48.1 [45.7 to 50.4]	42.9 [41.1 to 44.6]	36.4 [34.7 to 38.1]

*All post hoc comparisons between externally stabilized targets (TAR = target) significantly different at $p \leq 0.0001$

Force ratios between conditions for the nonparetic limb

Nonparetic limb force ratios were not significantly different between conditions ($p = 0.56$) ($Fy/Fz = 0.07$, SE (0.01) for BWS walking vs. $Fy/Fz = 0.08$, SE (0.01) for externally stabilized walking). We did not observe a condition x level interaction ($p = 0.13$) or a main effect of level ($p = 0.15$).

Positive and negative work between conditions for the nonparetic limb

Total positive work decreased across force levels and was larger for the nonparetic limb during BWS compared to externally stabilized walking, evidenced by main effects of level ($F(1.6, 25.1) = 40.0$, $p \leq 0.0001$) and condition ($F(1,16) = 31.4$, $p \leq 0.0001$) (Figure 9). We did not detect a significant condition x level interaction ($p = 0.66$).

Total negative work was lower during externally stabilized compared to BWS walking ($F(1, 16) = 41.5$, $p \leq 0.0001$). We also observed a condition x level interaction ($F(1.5, 24.1) = 7.4$, $p = 0.006$) and main effect of level ($F(1.6. 25.5) = 71.3$, $p \leq 0.0001$). A follow-up repeated measures ANOVA for externally stabilized walking revealed a

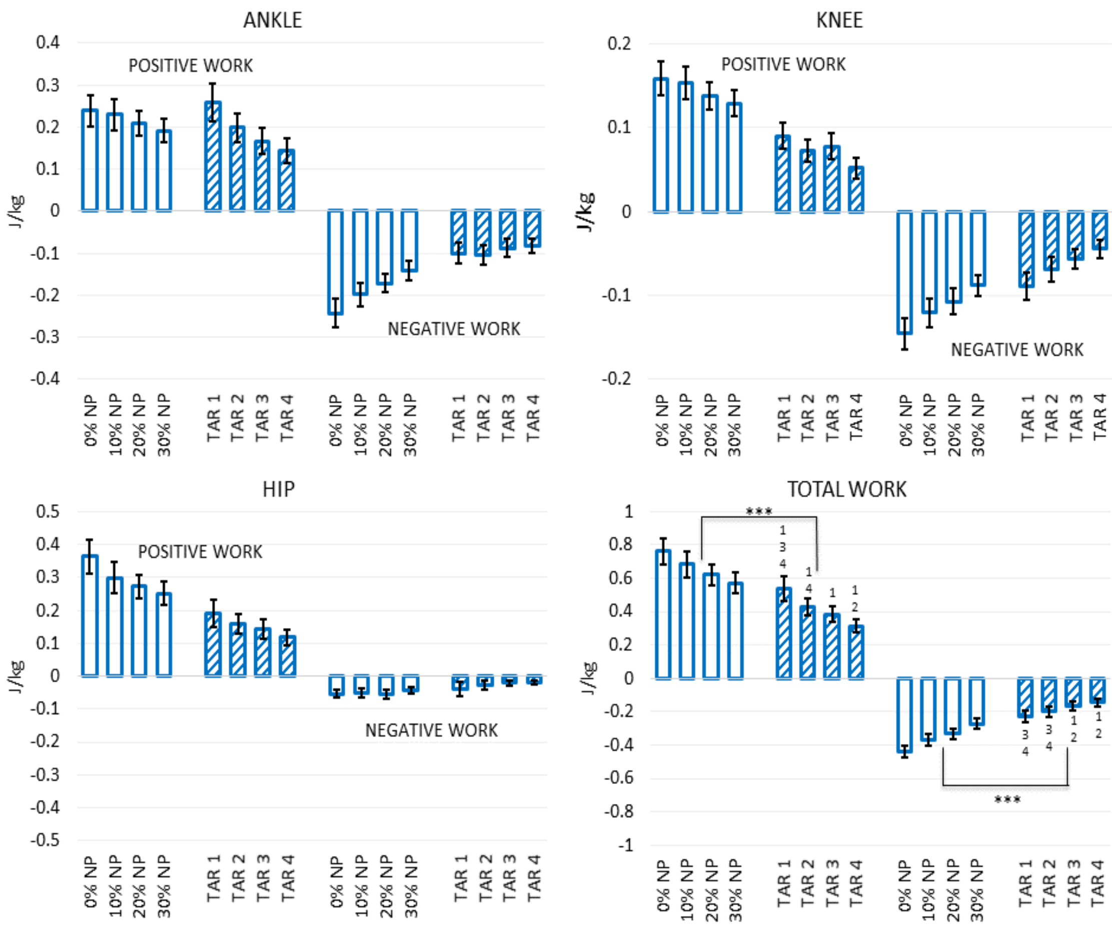

Figure 9. Positive and negative work across the lower-limb joints (ANKLE, KNEE, HIP) and sum of joints (TOTAL) for the nonparetic limb during BWS (open bars) vs externally stabilized (dashed bars) walking. Values are mean (SE). Significant pairwise comparisons reflected as numbers above or below each set of bar graphs for total work. All significant at Bonferroni corrected p≤0.01. *** indicates significant main effect of condition p≤0.0001.

main effect of level (F (1.6, 26.0) = 16.3, $p \leq 0.0001$), indicating that negative work decreased with decreasing vertical force target. Post hoc comparisons did not detect differences between targets 1 versus 2 ($p = 0.19$) and 3 versus 4 ($p = 0.12$) but all other comparisons were significant ($p \leq 0.01$) (Figure 9).

Leading and trailing limb angles between conditions for the nonparetic limb

Participants placed their nonparetic limb closer to their body during externally stabilized compared to BWS walking ($F(1, 16) = 19.1$, $p \leq 0.0001$). We did not detect a condition x level interaction ($p = 0.42$) or main effect of level ($p = 0.20$). Average leading limb angles during externally stabilized walking ranged from -11.8 degrees 95% CI [-14.7 to -8.8] at the highest force target 1 to -10.8 degrees 95% CI [-13.3 to -8.2] at the lowest force target 4.

Participants did not exhibit differences in trailing limb angle between walking conditions ($p = 0.72$). We did not detect a condition x level interaction ($p = 0.10$) or main effect of level ($p = 0.20$). Average trailing limb angle during externally stabilized walking ranged from 19.1 degrees 95% CI [15.5 to 22.7] at the highest force target 1 to 17.0 degrees 95% CI [14.0 to 20.1] at the lowest force target 4.

DISCUSSION

Our primary goal was to determine whether fully minimizing postural demands of walking via an apparatus that provided external control of upright orientation, full upper body and trunk mass vertical weight support, and minimal accelerations acting on the body COM would result in more appropriate Fy/Fz ratios of the paretic limb during the second half of the stance phase. We further expected increased positive but reduced negative work performed by the paretic limb compared to walking under various levels of BWS where posture and locomotion still had to be coordinated. Our primary hypothesis was not supported in that we did not observe more appropriate Fy/Fz ratios during externally stabilized walking or increases in positive work generated by the paretic limb.

Participants had varying ability to achieve vertical force targets during externally stabilized walking

Nine of seventeen participants did not achieve all four target levels of force during externally stabilized walking. This may have occurred for a variety of reasons. Weakness of the paretic limb during functional activities is well characterized for individuals poststroke (51–54), and strength deficits are shown to be related to gait performance at both comfortable and fast walking speeds (55). It is possible that the inability to achieve higher force targets was also related to weakness in this study. Alternatively, impaired accuracy or coordination deficits may instead explain the difficulty in achieving higher force targets. Stroke survivors exhibited impaired ability to sense dynamic load during walking in a study by Chu et al. (2014) (56), which could have affected their accuracy in attaining force targets despite visual feedback. Abnormal muscle activation patterns have also been shown to reduce muscle force generation poststroke (57).

Limb loading characteristics are altered with decreased magnitude of paretic-limb loading, prolonged rate of loading, and quicker offloading (58). Participants in this study may have had difficulty in generating force rapidly enough to achieve the higher force targets and still maintain walking cadence. Finally, the ability to achieve higher vertical force targets was positively related to both total positive and negative work, suggesting that individuals who attained all four targets did so through a combined strategy of power generation and absorption. This is in stark contrast to our previous observations of a force-generation-driven strategy nonimpaired individuals employed to achieve force targets (46) and also different than the force-generation strategy used by the nonparetic limb in this study.

The paretic limb did not exhibit more appropriate force ratios during externally stabilized walking

Contrary to our expectations, the paretic limb did not alter the Fy/Fz ratio during externally stabilized walking. We only observed a modest relationship between Fy/Fz ratios and force target attained, so some individuals who generated higher forces did not necessarily better direct these forces during the second half of the stance phase. Boehm and Gruben (2015) (43) suggested that individuals poststroke have a neural preference for directing GRFs anterior to their COM. Their observations in pedaling suggested that despite removing the need to control upright orientation during a pedaling task, stroke survivors maintained this anteriorly directed force bias (59). These findings are in contrast to those of Liang and Brown (2013) (14) who observed more appropriate ratios of shear-to normal crank force against the pedal during a seated pedaling task with minimal postural demand. While we minimized postural demands of walking in the present study, the body and limb configurations of participants in our support apparatus were more similar to the nonseated pedaling task from Liang and Brown (2013) (14) where they observed inappropriate force ratios and excessive extensor muscle activity. Body position may have a role in inappropriate muscle activity underlying Fy/Fz ratios. For example, Lewek et al (2006) (60) observed that individuals poststroke had greater quadriceps activity in a neutral compared to flexed hip when controlling for vestibular input. They speculated that this finding could be due to group II spindle afferents of the hip flexors exerting strong influence over vasti muscles, eliciting inappropriate muscle coupling.

Unaltered Fy/Fz ratios were accompanied by unchanging total positive work during externally stabilized compared to BWS walking. These findings are somewhat surprising given the considerable improvement that we observed in paretic trailing limb angle. Participants increased their trailing limb position by an average of 3.6° compared to BWS walking, much larger than the proposed minimal detectable change of 1.0° (61). Trailing limb angle is known to be a strong predictor of propulsive force generation (50,62); thus, we would speculate that improvements in the degree of trailing limb angle should have been accompanied by increased Fy/Fz ratio and power generation during late stance.

Techniques like functional electrical stimulation and passive assistance from an exoskeleton are shown to collectively improve peak propulsion, trailing limb angle, and swing phase knee flexion of participants with hemiparesis (63–66), and also promote longer limb loading in single-limb support (67). However, locomotor improvements accomplished through artificially stimulating muscles like the plantarflexors and dorsiflexors to perform locomotor functions do not necessarily reflect capability to use these muscles appropriately without assistance. The externally stabilized walking apparatus used in this study enabled participants to actively attain greater trailing limb angles because they were stabilized and prevented from losing their balance. The lack of increase in total positive work and invariant Fy/Fz ratios suggests that increasing trailing limb angle without assistance from stimulation or an exoskeleton does not necessarily enhance locomotor function of stroke survivors.

While we did not conduct statistical comparisons for each individual joint, knee joint negative work was increased during externally stabilized versus BWS walking. The timing of this negative work warrants further investigation since power absorption at the knee is functionally related to controlling knee flexion during pre-swing (68). Differing gait compensations poststroke can reflect either a knee that collapses into flexion during stance with greater eccentric load placed on the quadriceps, or a hyperextended knee that instead relies on passive knee stability (69). While the externally stabilized environment required no more than support of limb weight, the task of hitting the vertical force targets may have elicited these patterns.

Additional investigation of muscle activity and joint angles will be beneficial to determine whether requirements for active force generation during externally stabilized walking elicited excessive activity from muscles like the vasti, inappropriate muscle timing in relation to gait phase, or inappropriate muscle coactivations (57). For example, reduced paretic propulsion has been related to lack of independent modulation of the distal and proximal extensor muscles of the lower limb (30,70). Antagonist muscles to the plantarflexors (e.g., tibialis anterior) have also been negatively related to paretic propulsion during preswing (5). We are interested in whether inappropriate activity patterns like these were present in our participants poststroke, and if so, whether they persisted during externally stabilized walking.

Minimizing postural demands of walking does not engender improvements in locomotor control poststroke

The findings of the present study support weakened locomotor control poststroke, as suggested by our adapted-postural locomotor control interaction model presented in the introduction (Figure 10, dotted red lines). In addition, based on our findings we extend this model to suggest brainstem-derived control over locomotor function of the paretic limb as opposed to excessive postural influence preventing locomotor pathways from being appropriately expressed (Figure 10, thick red lines). The paretic limb exhibited diminished locomotor function, evidenced by small Fy/Fz ratios and low power generation even when the externally stabilized walking environment mechanically decoupled these locomotor force generation behaviors from postural influence. This finding supports the proposed brainstem-mediated control over the paretic limb following stroke (17,25,26,52). Minimizing the need for postural behaviors associated with reactive and anticipatory muscle activity did not allow muscles involved in generating forward propulsion to more appropriately perform this task. It is therefore possible that brainstem-mediated control over locomotor function of the paretic limb is an innate neuromotor bias of the nervous system poststroke as others have also proposed (Figure 10, thick red lines) (16,43,52).

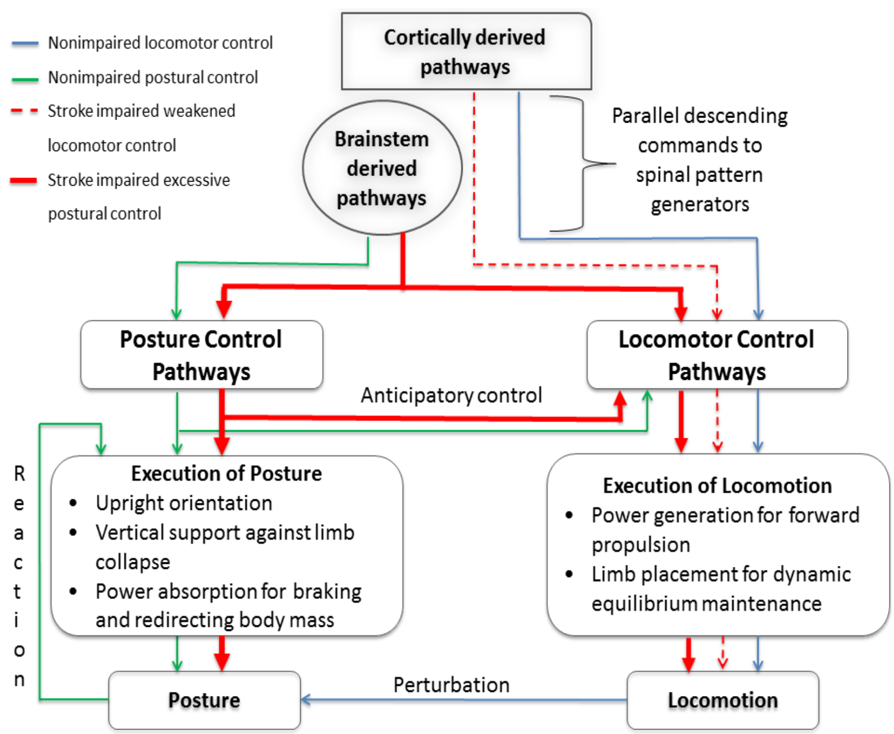

Figure 10. Updated postural-locomotor interaction model to reflect brainstem-derived control of locomotor function poststroke.

While reducing reactive and anticipatory control over postural orientation, vertical weight support, and COM accelerations may have reduced the need for reticulospinal involvement due to its role in regulating postural muscle tone (71) there still may have been excessive influence from other brainstem-derived pathways like the vestibulospinal pathway, which also plays an important role in the maintenance of postural equilibrium (72). Lewek et al. (2006) (60) observed considerable increases in vasti activity of stroke survivors elicited by vertical orientation of the head. Externally stabilizing individuals in an upright posture may not necessarily decrease input from gravireceptors to these postural networks, which could potentially allow excessive influence to persist. Finally, our efforts to match vertical GRFs between walking conditions may have invoked activity in these pathways similar to those that occur in response to vertical weight support due to similar load-related feedback (52).

Provision of BWS resulted in invariant Fy/Fz ratios for both limbs

BWS provides vertical offloading of the limbs during walking enabling individuals who have difficulty supporting their full body weight to undergo walking rehabilitation (73,74). The BWS mechanism used in this study successfully reduced vertical weight bearing on the limbs, which was accompanied by proportional scaling of propulsive forces, resulting in invariant Fy/Fz ratios across BWS levels. The Fy/Fz ratio reflects the proportion of GRF that is directed to propel the body forward (numerator) and support body weight (denominator) and both requirements were still present under conditions of BWS. The scaling of forces was similar for both limbs; thus, the asymmetry in force ratio between limbs persisted with the provision of BWS. Such scaling of GRF components is a typical behavior of nonimpaired individuals walking in reduced gravity environments (19) (Graham et al. 2018 unpublished) (39,41,73,75,76). Proportional scaling of Fy and Fz GRF components with increased BWS is a normal, albeit scaled, response of the paretic limb to the provision of BWS, similar to that of the nonparetic limb and nonimpaired individuals under BWS walking conditions (39,41,75,76).

Both limbs decreased total positive and negative work with increasing BWS

Interestingly, while positive work was larger for the nonparetic limb, negative work was not different between limbs. This finding suggests that the paretic limb played a typical postural, but diminished locomotor role during BWS walking (77,78). Smaller paretic-limb positive work appeared to be due to smaller values across all lower-limb joints compared to the nonparetic limb. These findings are consistent with those of Combs et al. (2012) (35) who did not observe improvements in the percentages of positive joint work performed between limbs following an eight-week BWS training protocol. Taken together it seems that BWS does not engender acute improvements in kinetic symmetry between limbs and this asymmetric relationship between limbs persists with training.

Limb angles were larger for the nonparetic compared to paretic limb during BWS walking

The paretic limb was placed closer to the body at foot strike compared to the nonparetic limb across BWS levels. Neither limb altered its leading limb angle as BWS increased, which is in contrast to our previous findings in nonimpaired individuals who reduced leading limb angles with increasing BWS when walking at both normal (1.0 m/s) and slow (0.5 m/s) speeds (19) (Graham et al. unpublished). Leading limb placement is critical to restore stability of the moving body, and BWS should reduce this requirement since it minimizes accelerations of the trunk (79–82). However, individuals poststroke are shown to exhibit excessive and asymmetrical trunk movements during walking (83,84) and since BWS does not provide full postural support, participants in this study may have required a similar leading limb angle at all levels of BWS to maintain stability.

While not significantly different than those of the nonparetic limb, trailing limb angles were smaller for the paretic limb across BWS levels. Considered along with the smaller leading limb angles exhibited by the paretic compared to nonparetic limb, this finding suggests that on average participants took shorter steps with their paretic limb. We did not conduct a sub-analysis to determine if there were directional differences in the group with some individuals taking longer paretic than nonparetic limb steps and vice versa, which is a well described finding of previous studies (85,86). However, paretic leg ankle kinetics have been shown to be reduced in all participants regardless of their paretic step ratio (86), suggesting that separating individuals into sub groups would not likely alter our Fy/Fz ratio or work results.

The nonparetic limb exhibited similar force ratios during externally stabilized compared to BWS walking

The nonparetic limb responded as we expected during externally stabilized walking. Participants were able to achieve all vertical force targets and exhibited Fy/Fz ratios comparable to BWS walking. Importantly, GRF profiles and Fy/Fz ratios during externally stabilized walking for the nonparetic limb were similar to our previous findings in nonimpaired individuals (19) (Graham et al. 2018 unpublished) demonstrating that participants were able to understand and accomplish this novel walking task.

Total negative work decreased during externally stabilized compared to BWS walking and was similar in magnitude to that of our previous investigation of nonimpaired individuals (46). Power absorption functions to decelerate the body's forward momentum and provide vertical support (87). Since externally stabilized walking did not require these postural functions we expected this decrease in total negative work. Unexpectedly, we also observed a decrease in total positive work for the nonparetic limb during externally stabilized walking. This finding is in contrast to that of nonimpaired individuals walking in this environment who significantly increased total positive work compared to walking without the external support apparatus (46). There are several possible explanations for this behavior. Locomotor force generation in this externally stabilized walking condition does not enable the use of inverted pendulum mechanisms of energy transfer (77,88). It is likely that this group of participants poststroke were relatively more deconditioned (55,89) compared to our previous investigations of healthy young participants. These individuals may have been less able to replicate power generation through purely active mechanisms. The observed reduction in total power appeared to occur because of reductions at the knee and hip but not the ankle. The nonparetic hip, in particular, has also demonstrated weakness in other constrained locomotor-like force production tasks (52).

Limitations

One primary limitation to the present study was the bilateral versus unilateral nature of the walking environments. BWS walking allowed normal interaction between limbs, while this presentation of the externally stabilized environment emphasized control over only one limb at a time making it difficult to investigate compensatory strategies that involved coordination between limbs. It would be interesting to observe participants poststroke walking in this externally stabilized environment while targeting forces with both limbs simultaneously. However, the difficulty level of such a task would likely require practice and multiple training sessions would be necessary before collecting data. The constraints of the externally stabilized walking environment also only enabled us to observe sagittal plane walking mechanics. Important mechanical work related to postural control is also performed in the frontal plane by the hip abductors (90). Compensatory walking patterns involving frontal plane mechanics like hip hiking are also commonly observed poststroke (91). Finally, half of our participants were unable to attain all vertical force targets in this study, limiting the complexity of our statistical analyses.

Conclusions and future directions

Stroke survivors with chronic hemiparesis exhibited greatly reduced Fy/Fz ratios on the paretic versus nonparetic limb during BWS walking and did not exhibit more appropriate ratios during externally stabilized walking where postural influence was minimal relative to locomotor requirements. While we anticipated that minimizing postural requirements for anticipatory and reactive responses during walking would help individuals to better express locomotor capacity, this was not the case. Instead, our results indicated that individuals may rely on postural control mechanisms to compensate for loss of voluntary locomotor control poststroke. Despite increases in trailing limb angle, power generation also did not increase during externally stabilized compared to BWS walking. While the nonparetic limb considerably reduced power absorption associated with postural control during externally stabilized compared to BWS walking, decreases in power absorption for the paretic limb were very small between walking conditions. Taken together, the paretic limb appeared to still exhibit postural functions in an environment where they were not required for successful performance of the task.

Future investigations of muscle activity underlying Fy/Fz ratios and associated power generation and absorption are needed to confirm whether the paretic limb used the same neuromotor control strategy for both BWS and externally stabilized walking conditions. If so, this finding would suggest that limited motor flexibility in the control of the paretic limb may be responsible for impairment characteristics of hemiplegic walking poststroke.

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SUMMARY

My primary aims throughout the investigations conducted for this dissertation were to discover basic neural control mechanisms underlying interactions between postural and locomotor control in nonimpaired and poststroke walking. I investigated the separate versus coordinated effects of postural and locomotor control mechanisms on appropriate foot-force generation during walking where postural and locomotor control mechanisms had to be coordinated versus where postural control was mechanically decoupled from locomotor force generation behaviors. I was particularly interested in control over the paretic limb during the propulsive period of the stance phase.

In chapter 1, I introduced the background and significance of appropriately directed foot forces in nonimpaired walking and the consequences of misdirected foot forces on walking function poststroke (23–25,75). I described a theoretical model of appropriate interactions between feedforward and feedback postural responses with locomotor control mechanisms. I proposed that inappropriate postural influence from operational postural networks was preventing weakened cortically derived locomotor control pathways from being appropriately expressed resulting in misdirected foot forces during walking poststroke (58,63,69,70,76). I suggested that if postural and locomotor control pathways were separate yet coordinated control systems (59,60) they could be experimentally manipulated to mechanically decouple postural influence from locomotor force generation behaviors. I expected that minimizing postural demands of walking

could more clearly reveal undisturbed locomotor control behaviors and be a potential way to facilitate more appropriate foot-force direction by the paretic limb as characterized by a ratio of Fy/Fz GRF. To test these ideas, I developed a novel experimental device and approach to address my hypotheses.

Role of body-weight-support treadmill training for improving walking ability poststroke

One common approach to manipulating postural demands of walking is through BWSTT (10). In chapter 2, I presented comparisons from a 6-week randomized clinical trial of BWSTT with and without challenging walking skills on clinical walking and balance outcomes of stroke survivors. We investigated the efficacy of challenging skill practice for $N = 29$ participants ≥ 5 months poststroke who exhibited slow CWS at baseline by incorporating challenging mobility skills into one of two comparable BWSTT protocols ($n = 15$ participants performing BWSTT with walking skills; $n = 14$ participants performing only BWSTT), and further isolated the effects of skill practice through eliminating use of handrails or assistance from both training protocols. We expected both groups to improve CWS pre to post training but expected that performing challenging mobility skills would improve overground CWS for the group performing mobility skills to a greater extent than the group only performing BWSTT. Our hypotheses were partially supported as both groups improved; however, there were no differences in CWS between groups.

While participants collectively improved CWS, average gains did not reach the minimal clinically important difference of 0.16 m/s (77). However, we observed both “responders”, meaning individuals who improved CWS $\geq$ 0.16 m/s and “nonresponders” who improved ≤ 0.16 m/s. Our observed speed improvements also changed the community ambulation status (78,79) of many participants pre to post training. We saw more limited community (pre: n = 3 vs. post: n=7 between 0.4 to 0.8 m/s) compared to household (pre: n=16 vs. post: n=10 < 0.4 m/s) ambulators following training. While we ensured that each participant was challenged during training, we did not adapt skills to meet each participant's unique requirements; thus, an individualized approach might have elicited greater gains in walking speed. Based on the theoretical underpinnings of this dissertation, another reason why speed improvements may have been minimal was because BWS only provided partial postural support during training and still required participants to coordinate posture with locomotor control.

Development of a novel experimental apparatus for reducing postural demands during walking

In Chapter 3, I introduced a novel experimental approach to minimize reactive and anticipatory postural demands of walking. I described the fabrication of a novel postural support apparatus that served to create an externally stabilized walking environment through (1) externally stabilizing the trunk in an upright position, 2) fully offloading the trunk and upper body mass from the lower limbs, but still allowing participants to make contact with the treadmill surface with their feet to generate forces, and 3) minimizing need to control body COM accelerations by holding participants in place. We expected that this apparatus would allow N = 20 nonimpaired participants to minimize behaviors associated with postural control including trunk range of motion, vertical weight bearing on the limbs, and power absorption to brake and redirect body mass (19,29,80–84). I demonstrated that this apparatus indeed minimized postural functions during walking of nonimpaired individuals, evidenced by reduced trunk motion in flexion/extension, lateral flexion, and transverse rotation, minimized peak vertical GRFs to limb weight, and reduced total positive and negative work compared to walking with typical postural demands.

In addition, using visual feedback, participants successfully matched vertical forces during supported walking to those of walking with typical postural demands. Even under these matched vertical force demands nonimpaired participants minimized total lower-limb power absorption related to postural control. However, they exhibited greater power generation than during typical walking, suggesting that this externally stabilized walking environment did not allow for normal inverted pendulum behavior of the body during the stance phase (28). There was no longer a forward "fall" of the body COM in response to gravity, which we aimed to achieve. However, since participants could not take advantage of inverted pendulum mechanisms of energy transfer, forces generated by the body were likely replaced through purely active, as opposed to a mix of active and passive, mechanisms.

Nonimpaired individuals reduced behaviors associated with postural control during externally stabilized walking

This novel demonstration provided evidence that I had fabricated a walking paradigm that could allow investigation of locomotor force generation behaviors with minimal reactive and anticipatory postural influence. However, this novel walking environment required further investigation of the neuromechanical strategy underlying postural and locomotor functions before using it to investigate my primary research question regarding individuals poststroke. Thus, in Chapter 4 I presented findings from an additional investigation of nonimpaired participants generating locomotor force with and without requirements for postural coordination. In this investigation we manipulated locomotor force generation requirements of walking for N = 16 nonimpaired individuals walking at two speeds (1.0 m/s and 0.5 m/s) under two experimental conditions: 1) under increasing amounts of BWS, which partially reduced postural demands, but did not fully minimize them relative to locomotor demands (73,85) and 2) externally stabilized walking within the novel support apparatus that minimized postural demands relative to locomotor requirements while allowing participants to generate matched vertical forces to those of BWS walking using visual force feedback.

We introduced our primary outcome variable for this dissertation research, the Fy/Fz ratio, which reflected the proportion of GRF that was directed to propel the body forward (Fy) and to support body weight (Fz) in late stance. This ratio characterized the overall direction of GRFs in late stance and allowed us to observe appropriate GRF direction of nonimpaired individuals when walking in these experimental conditions (20,86). We expected that the Fy/Fz ratio would remain unchanged, reflecting postural and locomotor coordination, during BWS walking but that the direction of this ratio would not be constrained by postural influence during externally stabilized walking. In support of our expectations, the Fy/Fz ratio remained invariant and lower-limb joint moments scaled across BWS levels reflecting a similar neuromechanical strategy when walking with requirements to coordinate posture and locomotion. Average muscle activity of muscles directly involved in vertical weight support like the plantarflexors decreased with increasing BWS; however, other lower-limb muscles did not change their magnitude of activity.

In contrast, we demonstrated that when the postural support apparatus minimized postural requirements during externally stabilized walking, the motor strategy underlying locomotor force generation was dependent instead on the magnitude of vertical force generated in the absence of anticipatory and reactive postural influence (Figure 1). The

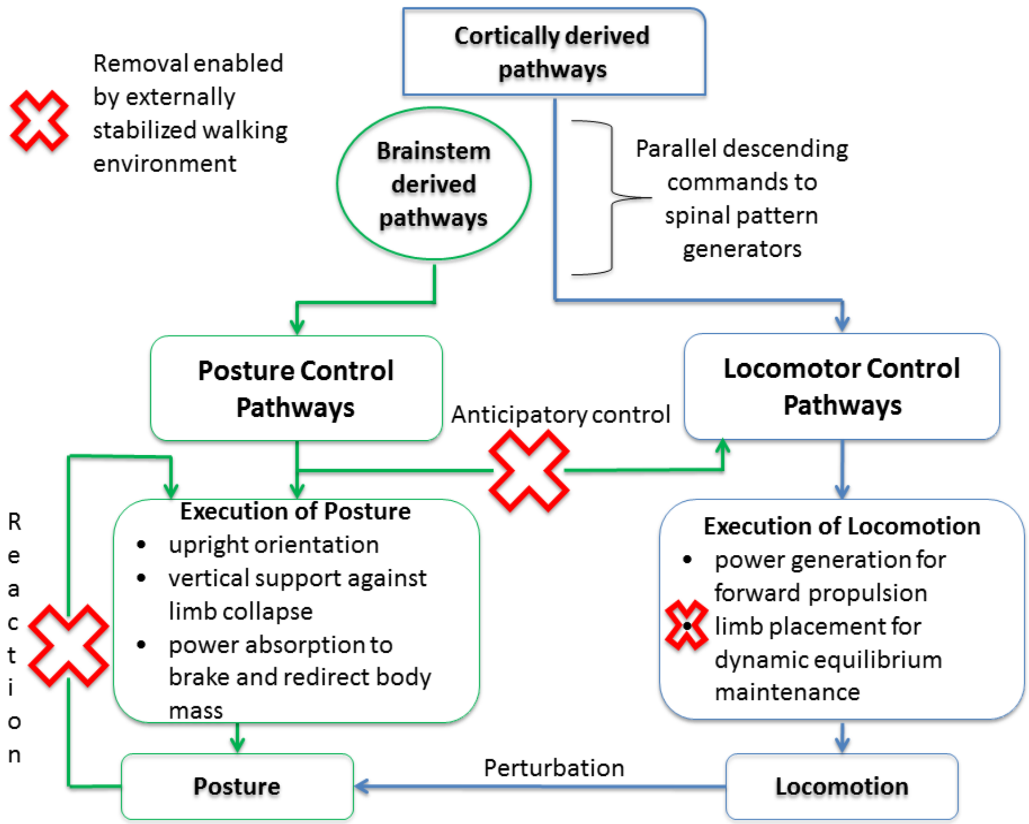

Figure 1. Revised postural-locomotor interaction model to reflect decoupling of postural influence from locomotor force generation behaviors during externally stabilized walking. Red Xs reflect decoupling of postural behaviors from locomotor output.

Fy/Fz ratio decreased demonstrating a relative decoupling of the Fy and Fz GRF components as force demands decreased. We further demonstrated that the Fy/Fz ratio was not speed dependent when we externally stabilized participants with no requirements to maintain dynamic equilibrium. Based on these findings we suggested that BWS reduces vertical loading requirements placed on the limbs, but still imposes postural demands through remaining requirements support body weight, maintain upright orientation, and control body accelerations, which was reflected by the similar but scaled neuromechanical strategy we observed across BWS levels. When the postural support apparatus mechanically decoupled posture from locomotion, locomotor force generation instead reflected muscular effort without activity associated with anticipatory or reactive postural influence while retaining similar joint moment and muscle activity profiles associated with locomotor behavior. We concluded that mechanically decoupling posture and locomotor control functions during walking offered a unique opportunity to probe locomotor control of populations that have difficulty coordinating posture with locomotion like individuals with chronic poststroke hemiparesis.

Individuals poststroke use postural mechanisms to compensate for loss of voluntary locomotor control

Equipped with an experimental environment that would allow me to investigate locomotor force generation of individuals poststroke with minimal need to coordinate postural control, in Chapter 5 I investigated locomotor force generation behaviors of N=17 individuals ≥ 6 months poststroke with residual hemiparesis walking at their comfortable speed in two environments: 1) under the provision of 0 to 30% BWS, which partially reduced postural demands of walking but did not fully minimize them relative to locomotor requirements and 2) under matched vertical forces but with full postural support provided through external stabilization from the novel postural support apparatus that mechanically decoupled locomotor force generation behaviors from postural influence by removing the need for reactive and anticipatory postural responses. We expected that participants poststroke would exhibit smaller Fy/Fz ratios compared to the nonparetic limb during BWS walking, and that the relationship of Fy/Fz ratios between limbs would not change with the provision of BWS.

In contrast, we expected more appropriate Fy/Fz ratios during externally stabilized walking when the support apparatus decoupled postural influence from locomotor force generation. Our primary hypothesis regarding Fy/Fz ratios of the paretic limb was not supported. The paretic-limb Fy/Fz ratio was not more appropriately directed during externally stabilized walking. However, we supported aspects of our theoretical model of excessive postural influence predominating over weakened locomotor control pathways. We extended this model through the results of this study to suggest brainstem-derived control over weakened locomotor behaviors of the paretic limb, instead of excessive postural influence preventing locomotor behaviors from being fully expressed (Figure 2). The paretic limb exhibited much smaller Fy/Fz ratios compared to the nonparetic limb during BWS walking that were almost entirely directed toward support with little to no contribution to forward progression. This behavior is consistent with the paretic limb acting like a support strut during walking with little locomotor function. The small Fy/Fz ratio did not change with minimal postural support provided via BWS, suggesting that alleviating vertical loading requirements placed on the paretic limb did not alter the postural-behavior-dominant control strategy during walking.

Externally stabilized walking also did not alter this postural-behavior-dominant control strategy over locomotor behaviors for the paretic limb. Power generation characterized by positive work performed by the paretic limb did not increase during externally stabilized walking compared to with BWS. In fact, the paretic limb exhibited

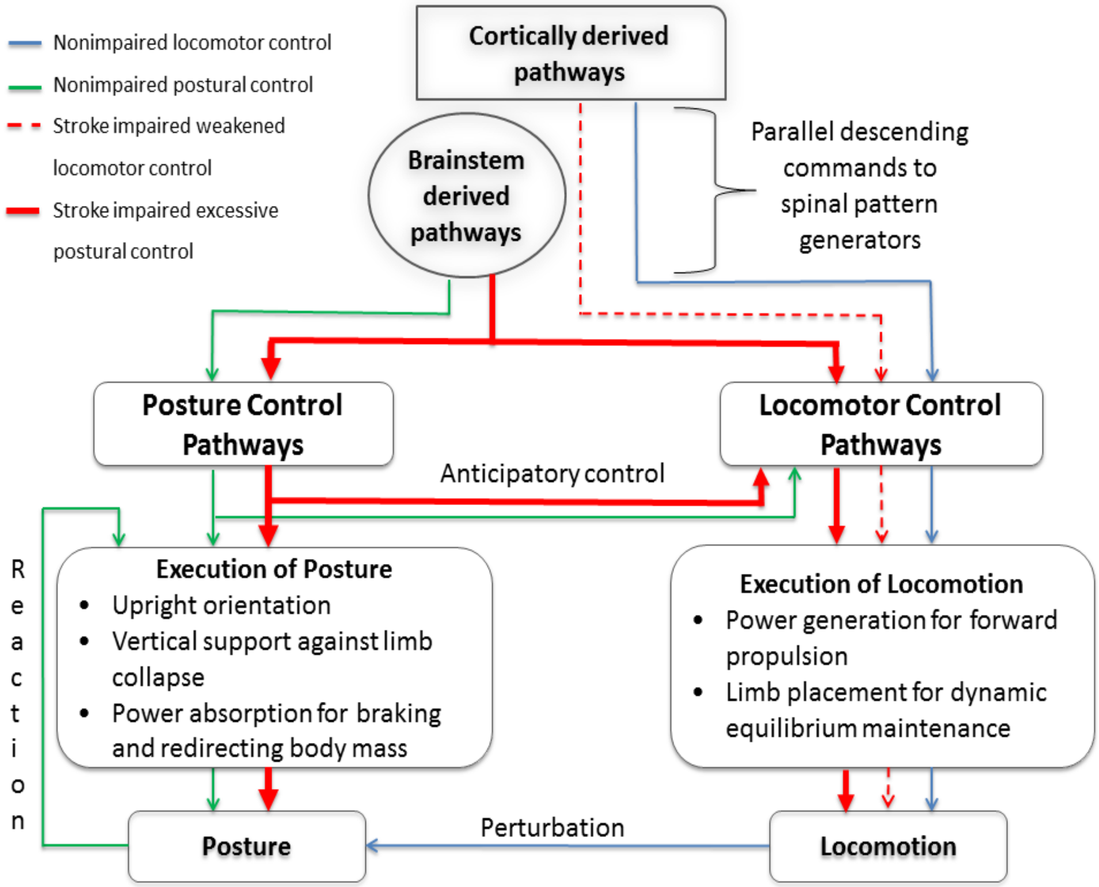

Figure 2. Revised postural-locomotor interaction model to reflect brainstem-derived control over weakened locomotor pathways.

much the same behavior during walking in this novel supported environment, with invariant positive work and Fy/Fz ratios. Despite the lack of more appropriate locomotor force generation, the paretic limb increased trailing limb angle by 3.6° in this novel walking environment, which is considerably greater than the proposed minimal clinically important difference of 1.0° (87). This surprising finding indicated that when participants achieved greater trailing limb angles without passive assistance for propulsion generation from interventions like FES or an exoskeleton (50–52) they could not actively take advantage of this ideal limb position for appropriate locomotor force generation (35). The paretic limb appeared to rely on postural functions in an environment where they were not required for successful performance of the task. These findings suggest that individuals poststroke may rely on postural control mechanisms to compensate for loss of voluntary locomotor control poststroke.

Future investigations and clinical implications

This important work requires future investigations to more fully characterize locomotor force generation behaviors of stroke survivors with and without requirements to coordinate postural control. We need additional investigations of muscle activity underlying Fy/Fz ratios and associated power generation and absorption to further confirm whether the paretic limb used the same neuromotor control strategy for both BWS and externally stabilized walking conditions. For example, co-contraction indices between antagonist muscles like dorsiflexors versus plantarflexors and quadriceps versus hamstrings (24,88,89) will provide important information regarding whether stroke survivors were capable of exhibiting more appropriate muscle coordination and phasing during externally stabilized versus BWS walking that we were unable to detect through Fy/Fz ratios and mechanical work. Alternatively, inappropriate muscle couplings could have worsened or remained unchanged when we decoupled postural from locomotor functions during walking. If so, this finding would suggest that limited motor flexibility in the control of the paretic limb may be responsible for impairment characteristics of hemiplegic walking poststroke, or, that reducing postural requirements of walking negatively impacts the preferred postural-dominant control strategy of the paretic limb.

Clinical implications associated with this dissertation work include the finding that BWS engenders modest speed improvements for individuals with poststroke hemiparesis ≥ 5 months post ictus, but that BWSTT protocols need further optimization and individualization in order to elicit CWS gains above the minimal clinically important difference for stroke survivors. Acute provision of BWS did not promote kinetic symmetry between the nonparetic and paretic limbs during walking in this study. While BWS may offer clinicians a method of vertically offloading the limbs during walking, enabling individuals who have difficulty supporting their full body weight to undergo walking rehabilitation (74,90), it is important to recognize that training in such an environment may promote strengthening of compensatory strategies instead of promoting interlimb symmetry.

Control pathways responsible for generating locomotor patterns are weakened poststroke (66,70,71) and decoupling postural influence from locomotor force generation did not promote more appropriate expression of locomotor behaviors. In light of these findings, it is possible that some individuals poststroke are more appropriate for rehabilitation efforts focused on strengthening appropriate locomotor patterns while others may be better suited toward exercise programs to improve overall conditioning. The relatively small sample size of stroke survivors in Chapter 5 of this dissertation did not allow for dividing participants into those who had evidence for remaining locomotor capacity versus those who did not. Future investigations are necessary to determine the efficacy of methods directed at promoting remaining appropriate locomotor function of the paretic limb, even for individuals who rely on postural mechanisms for locomotor control. Alternatively, rehabilitation approaches designed to capitalize on the constrained use of this limb may be essential to promote continued physical activity of stroke survivors and minimize the occurrence of secondary health conditions (12,91).

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APPENDIX A

IRB APPROVAL

THE UNIVERSITY OF ALABAMA AT BIRMINGHAM
Office of the Institutional Review Board for Human Use

470 Administration Building
701 20th Street South
Birmingham, AL 35294-0104
205.934.3789 | Fax 205.934.1301 | irb@uab.edu

APPROVAL LETTER

TO: Graham, Sarah Anne

FROM: University of Alabama at Birmingham Institutional Review Board
Federalwide Assurance # FWA00005960
IORG Registration # IRB00000196 (IRB 01)
IORG Registration # IRB00000726 (IRB 02)

DATE: 08-Jun-2018

RE: IRB-150422006
Interactions Between Posture and Locomotion in Nonimpaired and Poststroke Walking

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The IRB reviewed and approved the Continuing Review submitted on 22-May-2018 for the above referenced project. The review was conducted in accordance with UAB's Assurance of Compliance approved by the Department of Health and Human Services.

Type of Review: Full (Institutional Review Board 01 (UAB))
Determination: Approved
Approval Date: 08-Jun-2018
Approval Period: One Year
Expiration Date: 05-Jun-2019

The following apply to this project related to informed consent and/or assent:

- Waiver (Partial) of HIPAA

Please note the following:

- The IRB reviewed the conflict of interest related to the research.
- At your next revision or renewal the following changes should be made to your consent form:

  1. Revise the protocol number from "*F*" to "*IRB-*" in the title section of page 1 and the HIPAA page.
  2. Revise the Confidentiality section to align with the new wording in the Sample Consent from our website.
  3. Revise page 5 so that the header, Voluntary Participation and Withdrawal, and the text-block are both on the same page.
  4. Revise Payment for Participation in Research to include the sentence "*ask study staff about the*

method of payment that will be used for this study (e.g., check, cash, gift card, direct deposit)" at the end of the section.

5. Revise the Questions section to remove the sentence *"You may also call this number in the event the research staff cannot be reached or you wish to talk to someone else."*

Documents Included in Review:

- consent.180521
- ipr.180521

Investigator's Progress Report

Form version July 30, 2015

☒Continuing Review (Complete Items 1-12) —OR— ☐Final Report—all protocol-related activities are complete, including data analysis (Complete Items 1-11, and Item 13)	—FOR—	☐Expedited Review —OR— ☒Convened (Full) Review

1. Dates		
Today's Date	5/20/18	To help avoid delay, respond to all required items in the format provided, and include requested materials.
Starting Date of Project	6/24/15	If previous approval expires before approval is officially re-issued by the Office of the IRB, all work on the protocol must cease.
Current IRB Expiration Date	6/20/18	The IRB recommends applying for continuing review *4-6 weeks* before expiration of current approval. (See schedule.)

2. Principal Investigator (PI)			
Name (with degree)	Sarah Anne Graham, MA	Blazer ID	sarahg83
Department	Physical Therapy	Division	Rehabilitation Science PhD
Office Address	Building 516 20th Street South	Office Phone	205-996-3024
E-mail	sarahg83@uab.edu		
PI Contact who should receive copies of IRB correspondence (Optional)			
Name		E-mail	
Phone			

3. UAB IRB Protocol Identification	
Protocol Number	F150422006
Protocol Title	Interactions between Posture and Locomotion in Nonimpaired and Poststroke Walking
Study Sponsor(s)	American Heart Association
OSP Assigned Number (9 digits)	000512112
Note. *If the source or amount of funding for this project has changed or a new OSP # has been assigned to the protocol, include the new or revised funding application and/or provide the new OSP Assigned Number:*	

4. Purpose

In two or three sentences, briefly summarize the purpose of this protocol, and related studies if applicable. Please use non-technical language, and write for adults with general knowledge rather than for specialists.

▶ This study investigates differences in walking behavior (movements, forces, and muscle activity) between walking conditions in a robotic device called the KineAssist where an individual is fully supporting their body weight and controlling their upright position (i.e., typical walking), versus conditions where an external support apparatus provides full body weight and upright position support (i.e., walking with minimal postural demand). Participants walk over a moving treadmill belt while their movements are captured via motion capture cameras, their forces are measured via force plates embedded under the treadmill belt, and their muscle activity is captured via surface (i.e., skin attached) electromyography sensors. The goal of this study is to determine how the nervous system regulates walking behavior when postural demands of supporting body weight and controlling upright orientation are minimized. We compare these characteristics between nonimpaired individuals and individuals with poststroke hemiparesis (i.e., one-sided weakness).

5. Screened, entered, or otherwise accessed by the UAB Investigator(s). Include numbers for individuals, specimens, data records, charts, etc., as applicable to the protocol.

5.a. Number screened for study entry since the start of the project?	52 total 36 nonimpaired 16 poststroke
5.b. Number entered in study since the start of the project? (See Total in 5.e.)	52
5.c. Number entered in study since the last IRB review?	22 total 10 nonimpaired 12 poststroke
5.d. What is the age range for all participants entered in the study since the start of the project (e.g., 18-65)?	22 to 71 years

5.e. Complete the table below for cumulative enrollment for each racial and ethnic category. Copy/paste the entire table for additional groups (e.g., controls, sub-studies) if needed.										
	Ethnic Categories									
	Not Hispanic or Latino			Hispanic or Latino			Unknown/Not Reported Ethnicity			
Racial Categories	Female	Male	Unknown/ Not Reported	Female	Male	Unknown/ Not Reported	Female	Male	Unknown/ Not Reported	Total
American Indian/ Alaska Native										0
Asian	1	1								2
Native Hawaiian or Other Pacific Islander										
Black or African American	1	2								3
White	17	12								29
More Than One Race					2					2
Unknown or Not Reported										0
Total										36

☐ Check the box at the left if demographic information is not available (e.g., not collected for screening; collecting only specimens or data records and did not have access to the information).

Participants Poststroke

5.e. Complete the table below for cumulative enrollment for each racial and ethnic category. Copy/paste the entire table for additional groups (e.g., controls, sub-studies) if needed.										
	Ethnic Categories									
	Not Hispanic or Latino			Hispanic or Latino			Unknown/Not Reported Ethnicity			
Racial Categories	Female	Male	Unknown/ Not Reported	Female	Male	Unknown/ Not Reported	Female	Male	Unknown/ Not Reported	Total
American Indian/ Alaska Native										0
Asian										0
Native Hawaiian or Other Pacific Islander										0
Black or African American	3	4								7
White	3	5			1					9
More Than One Race										0

Unknown or Not Reported										0
Total										16

☐ Check the box at the left if demographic information is not available (e.g., not collected for screening; collecting only specimens or data records and did not have access to the information).

6. Conflict of Interest Review Board (CIRB)

Does the Principal Investigator, the institution, or any other person listed on this protocol have a financial conflict of interest, as defined by the UAB CIRB, related to this research? **If No, continue with Item 7.** **If Yes, in the space below, provide the names of the individuals who have a conflict and indicate whether or not a management plan is in place for each person listed.**	☒Yes ☐No

▶ David A Brown, one of the principle investigators, has a financial interest in the treadmill-KineAssist walking and balance robot being used in this study. David Brown has disclosed this information in the informed consent and also filed conflict of interest information.

7. Information Since the Date of Last IRB Review

- Mark at least one checkbox to indicate the type(s) of information received since the Date of Last IRB Review.
- Please summarize each type of information, and provide details and copies as requested.

7.a. You received multi-center trial reports that you have not previously forwarded to the IRB. Attach a copy and, in the space below, provide the date and source of the report, and summarize the findings and any recommendations:	☐Yes ☒No Multi-Center Trial Report
▶	
7.b. You received data and safety or other monitoring reports (e.g., DSMB, sponsor site visit). Even if you have already forwarded a copy to the IRB, attach a copy and, in the space below, provide the date and source of the report, and summarize the findings and any recommendations:	☐Yes ☒No Data Safety or Other Monitoring Report
▶	
7.c. You learned of literature published about this research. Attach the publication or provide its web address, and summarize the published findings here:	☐Yes ☒No Published Literature
▶	
7.d. You learned of other relevant information regarding this research, especially about risks associated with the research. Attach a copy of the source and/or summarize below, and check "Other Information" at right. Also check "Affects Willingness" if this information might affect a participant's willingness to continue in the research, and describe the effects on participants here:	☐Yes ☒No Other Information ☐Yes ☒No Affects Willingness
▶	
7.e. You have received another type of information. Summarize the information, including details relevant to participants here:	☐Yes ☒No Other Type of Information
▶	

8. Reportable and Non-reportable Problems

8.a. Have there been any "reportable events" since the IRB's last continuing review of the project? "Reportable events" are those that may constitute unanticipated problems involving risks to participants or others. If yes, attach the UAB Problem Report (even if already reported to the IRB); also attach the UAB Problem Summary Sheet completing Table A; Provide brief narrative summary (2-3 sentences) of any trends or increases in frequency or severity noted for all events over the life of the project, or enter "None noted" here:	☐Yes ☒No Reportable Events since last continuing review (Table A)
▶	
8.b. Have participants experienced harms (expected or unexpected, serious or not serious) that do not meet the UAB IRB criteria for "reportable events" since the IRB's last continuing review of the project? Attach UAB Problem Summary Sheet completing Table B; provide brief narrative summary (2-3 sentences) of any trends or increases in frequency or severity noted for all events over the life of the project, or enter "None noted" here:	☐Yes ☒No Other Events since last continuing review (Table B)
▶	
8.c. Have there been any reportable or non-reportable events over the life of the project? Attach UAB Problem Summary Sheet completing Table A and/or B as appropriate. Note the UAB Problem Summary Sheet is a cumulative report for all events over the life of the project. Provide brief narrative summary (2-3 sentences) of any trends or increases in frequency or severity noted, or enter "None noted" here:	☐Yes ☒No Any reportable or non-reportable events over the life of the project
▶	

9. Events Since the Date of Last IRB Review Mark at least one checkbox to show event(s) that have occurred since the Date of Last IRB Review. Please summarize all events, and provide specific details and/or copies as requested.	
9.a. You have had one or more problems obtaining informed consent. Briefly describe the problem(s) here:	☐Yes ☒No Consent Problems
▶	
9.b. You have received complaints about the research. Briefly describe the number and nature of the complaints:	☐Yes ☒No Complaints
▶	
9.c. One or more participants withdrew, or were withdrawn from, the research. Indicate here the number of withdrawals and the reason for each:	☐Yes ☒No Withdrawals
▶	
9.d. Participants have experienced research-related benefits. For example, "60% of participants in the treatment group appear to have reduced symptoms or reduced severity of symptoms, compared with 10% in the placebo group." Briefly describe the benefits here:	☐Yes ☒No Benefits
▶	
9.e. The risks, potential benefits, or both of this research have changed. Briefly describe the changes here:	☐Yes ☒No Change in Risk or Benefit
▶	

9.f. Does the research involve minors (<18 years of age)? If the study is still open to accrual or the participants are still receiving protocol driven intervention, the PI must either (a) confirm the previously assigned Children's Risk Level (CRL) number or (b) reassign a new CRL and give the reason it has changed in the space provided below:	☐Yes ☒No
►	
9.g. Events have occurred that relate to participant safety but do not fit into the categories listed above. Briefly describe the events here:	☐Yes ☒No Other Events
►	

10. Protocol and/or Informed Consent Modifications **Check the applicable boxes to indicate modifications made since Date of Last IRB Review (Yes to 9.a.) or requested with this renewal (Yes to 9.b. or 9.c.). Please provide the details and materials requested.**	
10.a. Previous Modifications	
Since the last IRB review, have you made modifications to the protocol, consent process, consent document or change in personnel?	☐Yes ☒No
If Yes, have the modifications been approved by the IRB?	
☐Yes—Provide a copy of each amendment form stamped "Approved" by the IRB during this approval period. ☐No—In the space below, justify making the modification without prior IRB approval:	
►	
10.b. Modifications To Protocol Requested With This Renewal	
Are you requesting IRB review of changes to the protocol (e.g., procedures, personnel, recruitment)? If so, check "Yes" and describe them in the space below. If adding personnel, (1) provide full name and UAB department/division, (2) indicate role in research, (3) and address whether the personnel has a financial conflict of interest. If removing personnel, please provide name(s) of personnel being removed. Indicate beside the name whether you are adding or removing each individual.	☐Yes ☒No Protocol Changes
►	
10.c. Modifications To Consent Requested With This Renewal	
Are you requesting IRB review of changes to the consent process and/or form(s)? If so, check the applicable "Yes" box and, in the space below, describe the changes.	☐Yes ☒No Consent Process Changes ☐Yes ☒No Consent Document Changes
If the changes affect the consent form(s), indicate the number of consent and/or assent forms used for this protocol, and describe the changes to each form: **(a) describe all changes to IRB-approved forms and the reasons for them;** **(b) describe the reasons for the addition of any materials (e.g., addendum consent); and** **(c) indicate either (1) how and when you will reconsent enrolled participants or (2) why reconsenting is not necessary.** **Also, indicate the number of forms changed or added. For new forms, provide 1 copy. For revised documents, provide 3 copies:** **• a copy of the currently approved document (showing the IRB approval stamp, if applicable),** **• a revised copy highlighting all proposed changes with "tracked" changes, and** **• a revised copy for the IRB approval stamp.**	
►	

11. Gene Therapy, Gene Transfer, Recombinant DNA

If this study involves	☐Gene therapy	☐Gene transfer	☐Recombinant DNA	☒N/A – go to item 12.
	Complete this item, and include memorandum with original signatures of Gene Therapy Review Panel addressing the risk-benefit ratio, any recommendations, and the CRL if applicable.			

11.a. Has the Panel's assessment of the risk-benefit ratio of this project changed? If yes, please explain below.	☐Yes ☐No Risk-Benefit Change
▶	
11.b. Does the Panel have any recommendations regarding the protocol or the consent form? If yes, please explain below.	☐Yes ☐No Panel Recommendations
▶	

12. Continuing Review—Complete only if you want to renew IRB approval so that protocol-related activities can continue.

12.a. Accrual Status—Indicate whether the study is "NOT YET OPEN," "OPEN," or "CLOSED" (described below) and provide the details requested for that accrual status.

NOT YET OPEN: No individuals have been screened or entered.	☐Not Yet Open
OPEN: The study could still enroll more individuals, add more specimens, review more records, etc. • Attach a copy of the most recently approved consent form(s) OR note in the space below that the IRB has waived informed consent and/or use of a consent form (waiver of documentation of informed consent). • Describe plans for future accrual, enrollment, or recruitment here:	☒Open
▶	
CLOSED: No more individuals will be enrolled, no more specimens or records will be added.	☐Closed
If the study is closed, is a consent form being submitted for review? If "Yes," explain why in the space below.	☐Yes ☐No Closed & Consent Form
▶	
• Indicate the date closed to accrual:	Date Closed:
• Choose one status to describe accrued participants, specimens, records:	Check ONE Status Below:
One or more is/are still receiving procedures as defined in the protocol (therapy, intervention, follow-up visits, etc.)	☒On protocol procedure
All are off protocol-driven procedures, in long-term follow-up only	☐In long-term follow-up
All are off protocol-driven procedures, in data analysis only	☐In data analysis

12.b. Describe any interim findings from this research. Please note that the IRB expects to receive findings on any protocol approved for 5 years.

▶We published a manuscript on our first subset of 20 nonimpaired individuals. These findings, including movement kinematics, kinetics, and muscle activity, suggest that nonimpaired individuals minimize behaviors related to postural control during walking when postural demands of walking are minimized. This paper can be viewed at https://ieeexplore.ieee.org/stamp/stamp.jsp?arnumber=8340168

13. Final Report—Complete only if you want to end IRB approval after all protocol-related data analyses are complete and no further work on the protocol will be done.

13.a. On what date were the final data analyses completed?	Final Date:
13.b. Summarize the final findings from this protocol and provide copies of any publications:	

▶
13.c. Who will be responsible for managing and storing the data records, including any and all research-related electronic files and paper documents?

Name	
UAB Dept/Div, or Employer	
Work Address	
Daytime Telephone	

13.d. Describe the storage plan. How will data records be stored—on paper, computers, or both? How will they be protected from damage, unauthorized release, loss, and theft? How long will the data be stored? Where will the records be stored?	
▶	
13.e. At the end of the storage period, will the data records be destroyed, archived, or transferred? Describe the plan in detail.	☐Destroy ☐Archive ☐Transfer
▶	

Note. Specimens may be stored only if/as described in the IRB-approved protocol. Data records must be stored as described in the sponsor's protocol or contract if applicable, and/or in the UAB Health System Record Retention Policy. Anyone wishing to use these data or specimens for secondary research purposes or for purposes preparatory to secondary research must obtain prior IRB review and approval.

Signature of Principal Investigator: Sarah A. Graham **Date:** 5/20/18

FOR IRB USE ONLY – Expedited Review

Change to Expedited Category Y / N

No change to IRB's previous determination of approval criteria at 45 CFR 46.111 or 21 CFR 56.111 ☐

Signature (Chair, Vice-Chair, Designee) Date