Minimizing Postural Demands of Walking While Still Emphasizing Locomotor Force Generation for Nonimpaired Individuals

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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 the 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 the 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. In addition, 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 et al. [2] (1992) 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]–[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 [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]–[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 sidelying [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 3/8 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 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.

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).

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 ($F(1.4, 27.4) = 1712.6, p \leq 0.0001$). The support apparatus minimized vertical GRFs to 15.8% body weight, 95% CI [12.7 to 19.0] when walking without effort (Fig. 2A).

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 unloading 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 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 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 the treadmill (0.62 J/kg, 95% CI [0.59 to 0.66]).

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 eccentrical 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 eccentricaly 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.

REFERENCES


