

Biomechanical effects of body weight support with a novel robotic walker for over-ground gait rehabilitation

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Abstract Body weight support (BWS) promotes better functional outcomes for neurologically challenged patients. Despite the established effectiveness of BWS in gait rehabilitation, the findings on biomechanical effects of BWS training still remain contradictory. Therefore, the aim of this study is to comprehensively investigate the effects of BWS. Using a newly developed robotic walker which can facilitate pelvic motions with an active BWS unit, we compared gait parameters of ten healthy subjects during a 10-m walk with incremental levels of body weight unloading, ranging from 0 to 40 % at 10 % intervals. Significant changes in joint angles and gait temporospatial parameters were observed. In addition, the results of an EMG signal study showed that the intensity of muscle activation was significantly reduced with increasing BWS levels. The reduction was found at the ankle, knee, and hip joints in the sagittal plane as well as at the hip joint in the frontal plane. The results of this study provide an important indication of increased lateral body balance and greater stabilization in sagittal and frontal plane during gait. Our findings provide a better understanding of the biomechanical effects of BWS during gait, which will help guide the gait rehabilitation strategies.

Keywords Body weight support · Pelvic motion support · Lateral stability · Over-ground walking · Muscle activation

1 Introduction

Impairments in the musculoskeletal system after neurological disorders may hinder individuals from performing gait [12, 16]. Gait rehabilitation aims at restoring abnormal gait patterns and can be achieved through an appropriately designed regime that includes over-ground walking training, body weight support (BWS), and strength training [5, 14]. BWS training, especially, promotes better functional outcomes by reducing gravitational force acting on the body and by increasing stability with external assistance [22, 27, 33]. In a study investigating the effects of BWS treadmill training with 100 stroke survivors, Barbeau concluded that gait training with BWS significantly increased all clinical outcomes in terms of walking speed, endurance, balance, and motor recovery, proving that BWS is an effective training method for improving gait and postural abilities [3]. Similarly, Peurala et al. [29] reported that gait training outcomes with BWS were largely improved after intensive gait training. Furthermore, promising results in gait parameters after BWS training have been reported in patients with neurological disorders such as spinal cord injuries [17, 32], cerebral palsy [1, 7, 16, 21], and Parkinson's disease [30, 31].

Although an improvement in gait functionality has been clearly documented, the biomechanical effects of BWS still remain contradictory in both treadmill and over-ground-based rehabilitation devices. Threlkeld et al. reported decreased cadence, stance phase and double limb support (DLS) time with increased step length at 50 and 70 % levels of BWS as compared with 0 % BWS, while Lewek reported that use of BWS did not alter how stance time and step length were manipulated during treadmill walking [22, 34]. In addition, Van Hedel et al. [34] reported abnormal

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over-activated muscle profile changes at rectus femoris, lateral and medial hamstring, and vastus medialis during mid-stance with increased BWS level, while Burnfield et al. [6] found reduced EMG mean activation at the gluteus medius and vastus lateralis, and decreased duration at soleus in an BWS environment without differences in flexor muscles. An explanation for the inconsistencies in biomechanical effects of BWS is that there is a strong reliance on the devices used for the experiments, implying that the role of stance phase stability is shared with the mechanical structure of the devices. Another possible reason for the inconsistency observed may be the use of an overhead harness in the treadmill-based BWS device which causes a different gait pattern from an actual over-ground gait. It has been reported that walking on a treadmill can lead to greater cadence, forward tilted trunk motion, and an increased vertical acceleration [2], which potentially affects gait dynamics as a result of different sensorimotor feedback and proprioceptive input compared to over-ground walking [4, 26, 36]. Moreover, overhead harness BWS systems which are mounted on the treadmill-based devices can often cause restriction to pelvic motion in the horizontal plane, which can severely deviate the gait kinematics and temporospatial parameters from normative patterns [28, 35]. Celestino et al. [7] reported that BWS with over-ground walking can significantly improve the gait functional outcomes compared to treadmill walking, showing that walking over the ground promoted gait patterns of cerebral palsy patients more similar to their typically developing peers.

Therefore, the use of an over-ground BWS system allowing pelvic motion facilitation has two significant advantages: it provides a natural gait pattern which is more similar to typical controls and a better understanding of the biomechanical effects of BWS to guide gait rehabilitation. The effects of BWS during over-ground gait have been studied in both healthy elderly populations and individuals with chronic stroke, using a robotic gait device called KineAssist [5]. Self-selected gait speed decreased with increased levels of BWS for the healthy subjects, whereas gait speed was increased by 18 % for the post-stroke subjects with increased BWS level compared to the 0 % BWS condition. These findings imply that the BWS with over-ground walking might have better functional outcomes for clinical applications. However, both research projects investigated kinematic or temporospatial gait parameters without looking at the patterns of muscle activity [5, 7].

To fill the knowledge gaps in the previous studies, a novel robotic over-ground walking system (robotic walker) with pelvic motion support has been developed (Fig. 1). The details of the walker are described in the next section. Therefore, the aim of this study is to comprehensively investigate the biomechanical effects of BWS in terms of

gait kinematics, temporospatial parameters, and muscle activation in healthy subjects using the robotic walker.

One important criterion for effective and successful BWS training is that the load applied to the lower limb to support body weight and to move the limbs should be reduced [9, 19]. Another critical factor is that the patterns of muscle activity should not be altered or deviated from the normal patterns [11]. In this regard, we hypothesized that an increased level of BWS will (1) change gait temporal parameters including single limb support (SLS) and DLS time, (2) reduce muscle activation time and amplitude without altering normative patterns, and (3) increase the lateral stability which can be shown by reduced amplitudes of hip ab/adductor muscles [19].

2 Methods

2.1 The robotic walker with BWS system

A BWS module is implemented on top of a novel over-ground walking system with pelvic motion support called robotic walker (Fig. 1a) [24]. This walker is capable of supporting a patient's pelvic motion through six DoFs for natural gait patterns. The use of a BWS system which provides active unloading of the body mass of the subject to the desired percentage with unrestricted pelvic motion is proposed for effective BWS training (Fig. 1b). Therefore, the walker allows the pelvis and trunk to move vertically with pelvic antero-posterior (AP), medio-lateral (ML), and rotation (RT) movements, as well as pelvic tilt and obliquity. The BWS actuator provides all-in-one control through a proportional-integral-derivative controller, drive, and motor integrated into one compact component; the active body weight of the subject is maintained via the vertical axis of the force/torque (FT) sensor during dynamic walking.

2.2 Subjects and experimental protocol

Ten healthy young subjects (age: 25.1 ± 4.4 years old, weight: 62.1 ± 9.1 kg, height; 168 ± 5.0 cm) with no known history of gait disorders, lower extremity injuries, and neurological disease participated in this study. All subjects signed a consent form which was approved by the Institutional Review Board of the National University of Singapore.

Eight high-speed motion capture cameras (Vicon, Oxford, UK) and a sixteen-channel wireless bipolar electromyography (EMG) were used to obtain gait kinematic data, temporospatial parameters, and muscle activation data. Fifteen retroreflective optical markers were placed on the subject's pelvis and lower limb, and precise 3D positions of the markers were recorded with a sampling

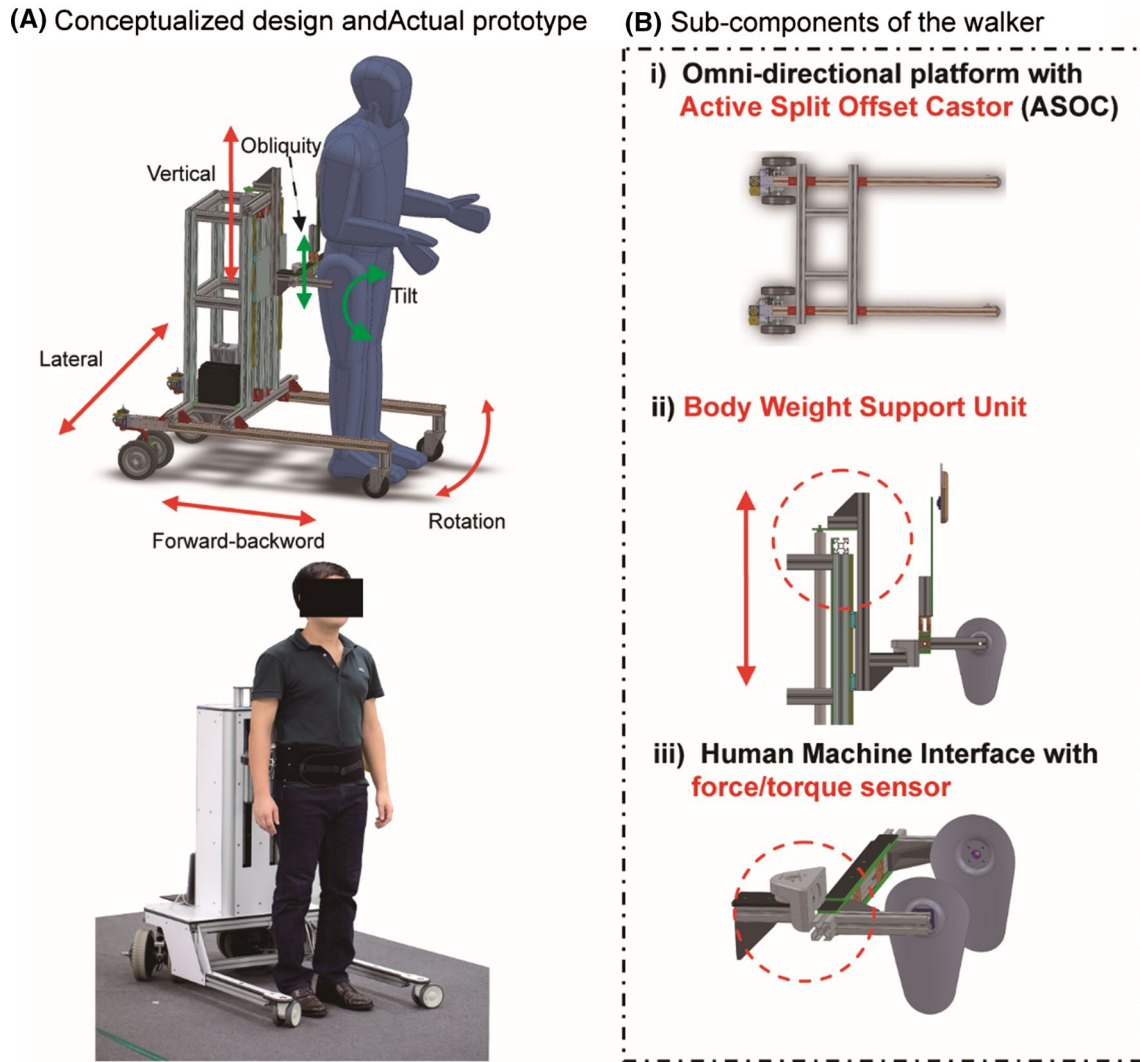


Fig. 1 **a** Conceptualized design and actual prototype of the novel robotic walker for pelvic motion support. **b** The system consists of omnidirectional mobile platform with ASOC, pelvic and trunk

frequency of 100 Hz (Nexus 1.8.3, Vicon, UK). Nine EMG (Delsys, USA) sensors were attached to the tibialis anterior (TA), gastrocnemius (GA), soleus (SOL), vastus medialis (VM), rectus femoris (RF), biceps femoris (BF), gluteus maximus (G_{Max}), gluteus medius (G_{Med}), and adductor longus (AL) according to the SENIAM protocol to quantify muscle activity [13], and electromyograms were recorded with sampling frequency at 1000 Hz. All of the experimental instruments were time synchronized. With all of the instrumentation in place, the subjects were first instructed to walk around with the walker for 5–10 min to become acclimatized. In the actual experiment, the subjects were instructed to walk on a 10-m-distance walkway with incremental amount of BWS forces from 0 to 40 % of body weight at 10 % intervals. Experimental conditions consisted of the 0, 10, 20, 30, and 40 % BWS level. All

motion support brace unit with active BWS actuator, human–machine interface with FT sensor

subjects were asked to walk through more than three successful trials with the different experimental conditions.

2.3 Data analysis and statistics

2.3.1 Kinematic parameters and temporospatial gait parameters

3D trajectories of 15 optical markers were low-pass-filtered through a zero-lag fourth-order Butterworth filter with a cutoff frequency of 6 Hz [18]. During each trial, heel strike (HS) and toe-off (TO) events were determined by using the vertical trajectory and velocity of the foot [25]; two strides in the middle of the walkway were used to calculate temporospatial gait parameters for each trial. Ankle, knee, and hip joint angles in the sagittal plane and their minimum,

maximum, and range of motions (RoMs) were calculated based on the positions of each marker, and the temporospatial gait parameters were further analyzed using MATLAB. The spatial parameters (stride length, step length, step width) were calculated based on the distance of markers on the left and right foot, and the temporal parameters (gait velocity, stride time, stance, swing, SLS, and DLS time, and percentage of stance phase) were determined based on HS and TO time. The stride and step length were normalized over the subjects' leg length.

2.3.2 EMG analysis

EMG signals were firstly band-pass-filtered with a zero-lag fourth-order Butterworth filter with a cutoff frequency between 2 and 200 Hz. The band-passed signals were rectified and then normalized by a maximum voluntary contraction value which was collected from overall experimental trials to obtain relative amplitude of the signal. The rectified and normalized EMG signals were then low-pass-filtered at 10 Hz for a linear envelope [23]. The enveloped signals were then used to calculate activation amplitude and duration for quantitative analysis. The mean EMG amplitude was determined in the overall gait cycle, stance, and swing phase. The temporal signals were considered activated if the relative amplitude exceeded 20 %; otherwise, the signals were defined as inactive. Finally, the EMG activation durations were calculated in the overall gait cycle, stance, and swing phase.

2.3.3 Statistics

One-way ANOVA was used to test significance among the conditions for the joint kinematics, temporospatial gait parameters, and EMG mean amplitude and duration with Tukey's post hoc analysis to contrast differences among the experimental conditions.

3 Results

3.1 Provision of BWS force with the robotic walker

Figure 2 shows the amount of vertical force acting on the FT sensor with increasing levels of BWS. As expected, with increasing BWS, subjects exerted a significantly greater unloading force for vertical movement ($p < 0.001$).

3.2 Gait kinematics

Table 1 shows the joint kinematics. No significant differences were found in ankle joint with increasing BWS

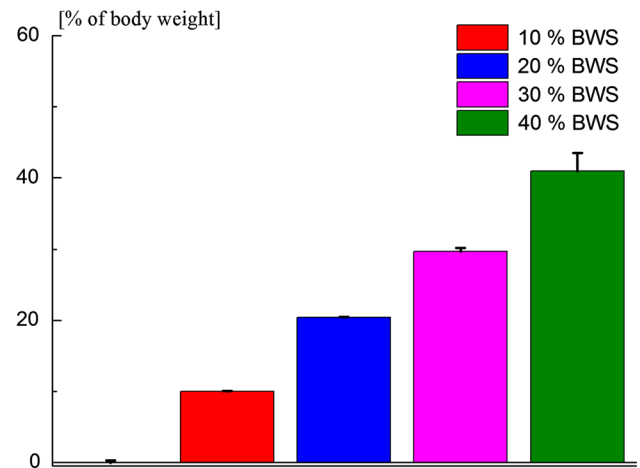


Fig. 2 Amount of vertical force exerted on FT sensor with increasing level of BWS

level. However, maximum knee flexion and RoM were significantly reduced at 40 % BWS compared to 0 % BWS ($p < 0.001$). Similarly, the maximum hip extensions were reduced at 20, 30, and 40 % BWS ($p < 0.001$) and hip RoMs were also diminished at 30 and 40 % BWS in comparison with 0 % BWS ($p < 0.001$) (Table 1).

3.3 Temporospatial gait parameters

Temporospatial gait parameters for the different conditions in accordance with BWS level are shown in Table 2. Gait spatial parameters such as normalized stride and step length, and step width were not influenced, while temporal parameters were significantly altered by the level of BWS. Specifically, as BWS level increased, a significant increase in gait velocity at 10 and 40 % BWS and shortened stride and stance time at 10, 20, 30, and 40 % BWS were observed compared to 0 % BWS ($p < 0.001$). In addition, swing time was longer at 40 % BWS; SLS and DLS time and percentage of stance phase at 10–40 % of BWS were significantly shortened compared to 0 % BWS ($p < 0.001$).

3.4 EMG parameters

Figure 3 depicts the enveloped EMG profiles in the lower extremity muscles, recorded at self-selected speed for designated BWS conditions. Clear indication of decreased muscle activation patterns can be observed in most of the major muscles, especially for ankle dorsi/plantar flexion and hip flexion/extension, with respect to 0 % BWS condition. For comprehensive analysis of the influence of varying amounts of BWS unloading on EMG activity, the

Table 1 Minimum, maximum, and range of motions of ankle, knee, and hip joint angles

Conditions	Ankle			Knee		Hip		
	Plantar flexion	Dorsiflexion	RoM	Flexion	RoM	Flexion	Extension	RoM
0 % of BWS	-11.41 ± 9.06	10.37 ± 4.95	21.78 ± 9.52	52.28 ± 8.33	52.38 ± 6.91	27.32 ± 8.16	-7.15 ± 7.37	34.47 ± 4.76
10 % of BWS	-15.13 ± 9.89	10.11 ± 5.17	25.24 ± 10.28	49.98 ± 9.06	49.62 ± 7.51	29.81 ± 9.03	-3.46 ± 7.35	33.27 ± 5.76
20 % of BWS	-13.71 ± 10.01	8.37 ± 4.67	22.08 ± 9.16	48.06 ± 8.69	48.90 ± 8.13	30.47 ± 7.87	0.19 ± 6.62**	30.28 ± 6.03
30 % of BWS	-15.02 ± 10.05	8.63 ± 4.73	23.66 ± 9.82	47.80 ± 8.86	48.32 ± 8.01	29.17 ± 6.65	1.19 ± 6.78**	27.98 ± 5.44**
40 % of BWS	-14.02 ± 10.04	8.62 ± 4.71	22.64 ± 8.98	46.33 ± 9.41**	47.27 ± 8.78**	29.41 ± 7.35	2.43 ± 8.31**	26.98 ± 6.90**

* Statistical difference from 0 % BWS, $p < 0.05$

** Statistical difference from 0 % BWS, $p < 0.001$

Table 2 Temporospatial parameters with increasing level of BWS

Conditions	0 % of BWS	10 % of BWS	20 % of BWS	30 % of BWS	40 % of BWS
	Mean ± SD	Mean ± SD	Mean ± SD	Mean ± SD	Mean ± SD
Normalized stride length	0.97 ± 0.14	0.98 ± 0.17	0.96 ± 0.15	0.98 ± 0.14	0.98 ± 0.14
Normalized step length	0.51 ± 0.06	0.51 ± 0.08	0.49 ± 0.08	0.50 ± 0.06	0.50 ± 0.07
Step width (m)	0.14 ± 0.03	0.14 ± 0.05	0.14 ± 0.04	0.15 ± 0.04	0.15 ± 0.05
Velocity (m/s)	0.366 ± 0.07	0.413 ± 0.14*	0.395 ± 0.06	0.406 ± 0.06	0.413 ± 0.06*
Stride time (s)	2.31 ± 0.24	2.06 ± 0.28**	2.04 ± 0.17**	2.05 ± 0.25**	2.00 ± 0.16**
Stance time (s)	1.61 ± 0.21	1.34 ± 0.31**	1.29 ± 0.19**	1.29 ± 0.21**	1.21 ± 0.15**
Swing time (s)	0.69 ± 0.12	0.72 ± 0.21	0.75 ± 0.12	0.77 ± 0.15	0.80 ± 0.16**
SLS time	1.15 ± 0.19	0.99 ± 0.31*	0.97 ± 0.26*	0.99 ± 0.16*	0.91 ± 0.43**
DLS time	0.47 ± 0.16	0.35 ± 0.15*	0.33 ± 0.21*	0.30 ± 0.17**	0.32 ± 0.32*
% of stance phase (%)	69.83 ± 4.46	64.82 ± 0.69**	63.19 ± 5.66**	62.61 ± 5.82**	60.30 ± 6.79**

* Statistical difference from 0 % BWS, $p < 0.05$

** Statistical difference from 0 % BWS, $p < 0.001$

changes in both amplitude and duration were computed in this study.

3.4.1 Mean amplitude of EMGs

For the quantitative measurements of amplitude domain, the mean EMG activation in overall gait cycle (Fig. 4), and stance and swing phase (Table 3) are indicated. In overall gait cycle, TA mean amplitude at 30 and 40 % BWS, and GA at 20, 30, and 40 % were significantly reduced compared to 0 % BWS ($p < 0.05$). In addition, the G_{Med} (30 and 40 % BWS) and AL (20, 30, and 40 % BWS) were significantly reduced. The mean EMG amplitude in the stance phase showed similar patterns to that in the overall gait cycle, but AL muscle amplitude in the swing phase was significantly reduced at 30 and 40 % ($p < 0.05$) compared to 0 % BWS. Despite significant reductions found in ankle dorsi/plantar flexor and hip flexor/extensor muscle activation, no significant differences were observed in Sol, VM, RF, BF, and G_{Max} muscles.

3.4.2 Activation duration of EMGs

For the quantitative measurements of the temporal domain, the activation duration of EMG in the stance and swing phase was calculated (Fig. 5; Table 4). During the stance phase, for ankle joint plantar flexor and dorsiflexor, duration of TA at 30 and 40 % of BWS ($p < 0.001$) and duration of GA at 20, 30, and 40 % of BWS ($p < 0.001$) were significantly shortened as compared to 0 % BWS. For knee extensor and hip flexor, the duration of VM at 40 % and duration of RF at 20 % were significantly shortened as compared to 0 % of BWS. Likewise, hip joint flexor/extensor and ad/abductor, G_{Max} in all conditions ($p < 0.001$), G_{Med} in 30 and 40 % ($p < 0.05$), and AL in all conditions ($p < 0.001$) were significantly reduced compared with 0 % of BWS. In addition, during the swing phase, slight but significant changes were found at GA at 20, 30, and 40 % BWS, Sol in 40 % BWS, BF in 30 % BWS, and G_{Med} in 30 % of BWS (Table 4).

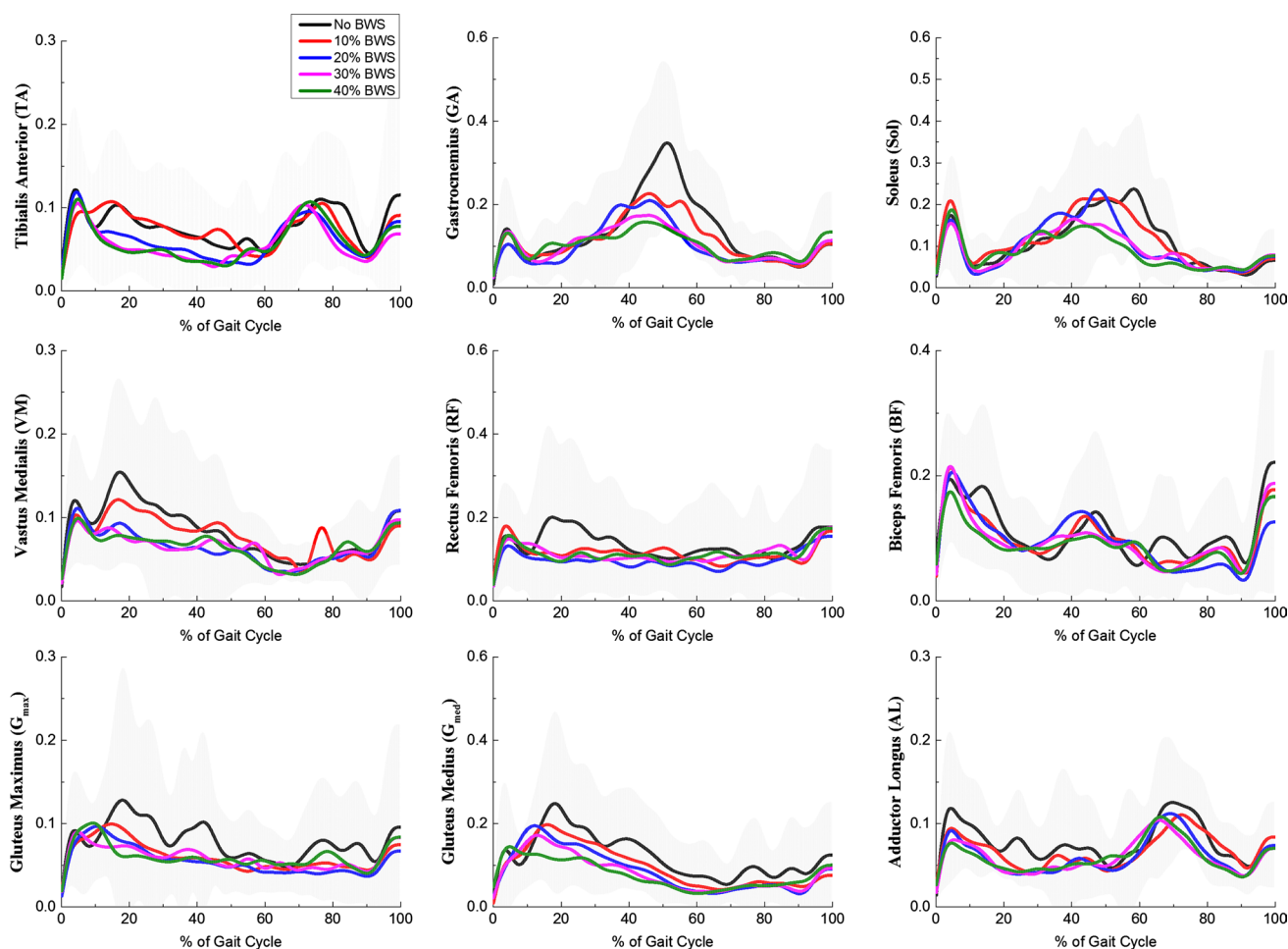


Fig. 3 Enveloped EMG profiles from nine major muscles during walking with BWS. The *black line* and *gray line* show averaged EMG profiles and its' SD in 0 % BWS. *Red, blue, pink, and green*

line shows the enveloped EMG profiles in 10, 20, 30, and 40 % BWS, respectively (color figure online)

4 Discussion

This study examined the unique effects of BWS with the robotic walker in terms of joint kinematics, temporospatial gait parameters, and muscle activations. The walker can achieve normal gait patterns without altering any joint kinematics and EMG activation through pelvic motion facilitation during walking [24]. It is important to note that the achievement of this natural gait is the key addition of this study.

4.1 Kinematics and gait temporospatial parameters

The mean peak angles and RoM showed a significant inverse relation with BWS levels. In particular, decreased peak flexion and RoM of the knee at 40 % BWS, and the maximum extension and RoM of the hip at 20, 30, and 40 % BWS level were observed. In contrast, the ankle joint kinematics were not significantly influenced by

increased BWS levels. These results support and contradict previous research. Fischer et al. [10] reported a significant reduction in maximum knee flexion and maximum hip flexion with increased BWS. These results are similar to ours showing a decrease in peak knee flexion at mid-swing. However, for hip joint kinematics the previous study showed a decrease in hip flexion while our results showed increased maximum hip extension with increased BWS level. It is important to note that the gait kinematic patterns are highly dependent on the devices used for the experiments [10], thus these kinematic differences can be caused by the pelvic motion support characteristics of the robotic walker.

With increasing amounts of bodyweight unloading, the product of spatial gait variables appears to be less significantly affected while temporal parameters were largely influenced with increased BWS levels. Contrary to previous assessment of over-ground walking with BWS using healthy individuals [5], the participants of our study walked

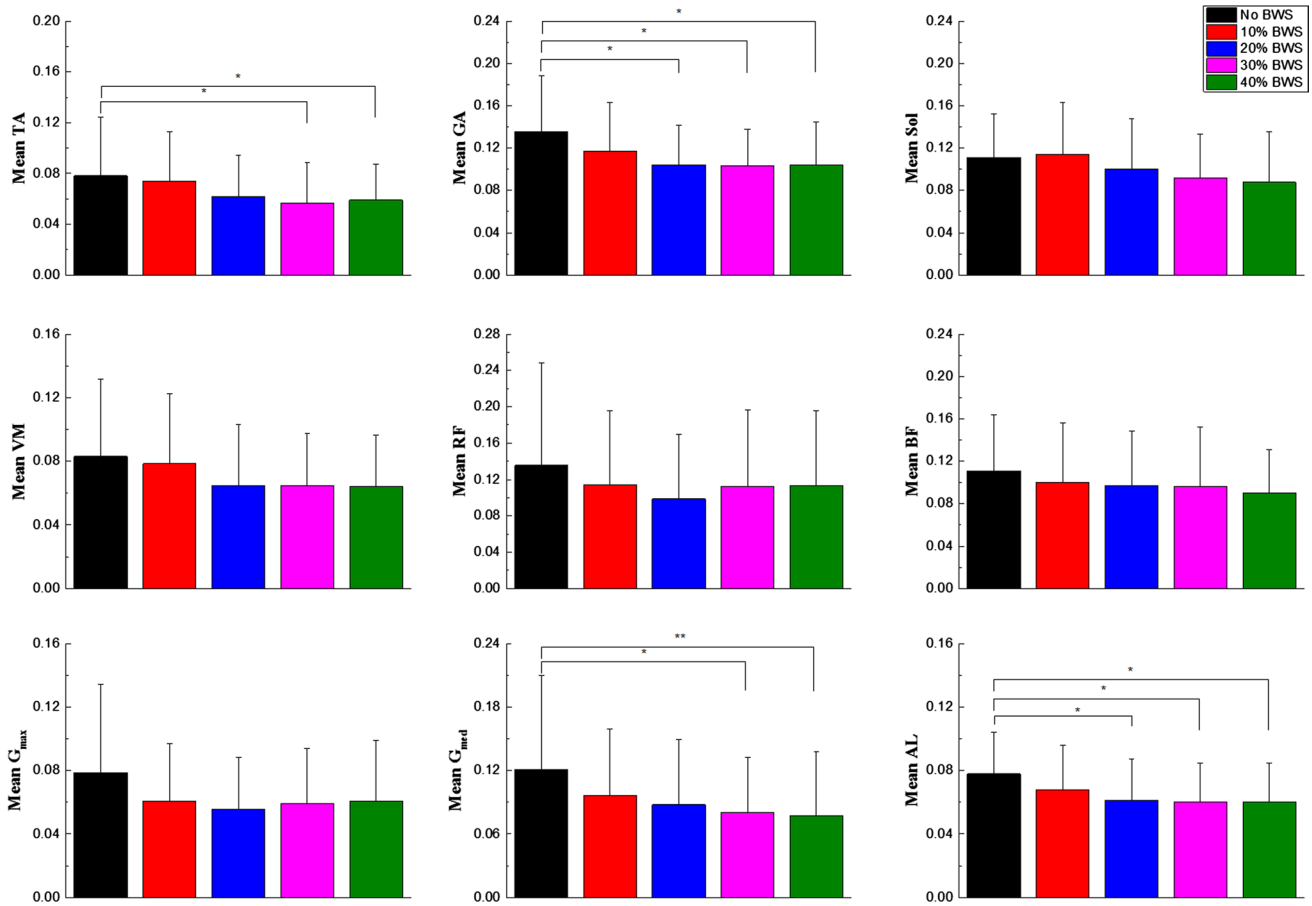


Fig. 4 Averaged EMG amplitude from nine major muscles during walking with BWS. The *black bar* shows averaged EMG amplitude in overall gait cycle at 0 % BWS. *Red, blue, pink, and green bars*

show the EMG amplitudes in 10, 20, 30, and 40 % BWS, respectively (color figure online)

faster in the higher BWS levels than in the 0 % bodyweight unloading condition. Increase in gait speed at these BWS conditions was accompanied by shorter stride and stance time. This contradiction may have arisen from the different mechanical structure and control strategies of the devices used for the experiments. The reduction in the absolute duration of the stride over the gait cycle observed in this study is mainly attributable to a decrease in the stance duration, as the duration of the swing phase does not change much with varying amounts of bodyweight unloading, except at 40 % BWS level. Such systematic decline in the proportion of the stance phase in the gait cycle at all BWS conditions indicates improved stability of the subjects, as patients generally remain with at least one foot in contact with the ground during unsteady walking. In addition, especially during DLS, additional effort should be exerted by pushing-off at the ankle or powering the hip to maintain a steady walking speed [20]. Thus, the significantly shortened stance, SLS, and DLS time may be explained by the reduced time required for leg transition and increased step

frequency rather than stride length in the BWS conditions. These results confirmed our first hypothesis that the BWS unloading will shorten the stance phase including SLS and DLS time.

4.2 EMG amplitude and duration

The EMG data suggest a specific mechanistic cause that underlies the observations in the kinematic and temporal gait changes in this study. In agreement with our second hypothesis, the results of our study showed that the intensity of muscle activation at all joints in the sagittal plane and that at hip joint in the frontal plane were significantly reduced with increasing BWS levels.

At the ankle joint, the use of BWS slightly but significantly reduced the ankle dorsiflexor in the overall gait cycle and stance phase. It can be explained that the use of 30 or 40 % of BWS may reduce ankle dorsiflexor load during mid-stance for weight acceptance rather than helping to elevate the foot during the swing phase (Fig. 4; Table 3).

Table 3 Averaged EMG amplitude of 9 major muscles in stance and swing phase

Conditions	0 % of BWS Mean ± SD	10 % of BWS Mean ± SD	20 % of BWS Mean ± SD	30 % of BWS Mean ± SD	40 % of BWS Mean ± SD
<i>Stance phase</i>					
TA	7.34 ± 4.72	7.33 ± 4.43	5.70 ± 3.69	5.22 ± 3.62	5.11 ± 3.38*
GA	16.41 ± 6.00	13.91 ± 6.09	12.15 ± 4.06**	12.01 ± 3.81**	11.60 ± 4.69**
Sol	13.75 ± 6.09	14.28 ± 6.52	12.44 ± 6.17	10.99 ± 5.38	10.65 ± 0.02
VM	9.54 ± 6.29	8.84 ± 5.21	6.85 ± 4.46	6.86 ± 3.84	6.81 ± 4.02
RF	13.86 ± 11.60	11.87 ± 8.60	9.48 ± 6.53	10.82 ± 8.10	10.90 ± 7.74
BF	11.49 ± 6.41	11.03 ± 6.58	11.67 ± 7.22	10.48 ± 6.92	9.81 ± 5.08
G _{Max}	8.38 ± 5.58	6.53 ± 3.54	6.17 ± 3.29	6.30 ± 3.30	6.26 ± 3.46
G _{Med}	14.13 ± 10.02	11.82 ± 7.62	10.80 ± 7.24	9.57 ± 6.29*	8.70 ± 6.61**
AL	7.40 ± 2.76	6.04 ± 2.68	5.56 ± 2.49*	5.80 ± 2.69*	5.74 ± 2.63*
<i>Swing phase</i>					
TA	8.72 ± 6.06	7.52 ± 4.65	7.09 ± 3.52	6.58 ± 3.35	7.38 ± 3.89
GA	8.31 ± 5.42	7.64 ± 4.53	7.19 ± 4.73	7.23 ± 4.03	8.14 ± 5.01
Sol	6.22 ± 3.46	6.15 ± 3.62	5.45 ± 2.80	5.79 ± 3.04	5.31 ± 3.23
VM	6.03 ± 3.06	5.98 ± 4.57	5.79 ± 4.56	5.77 ± 3.56	5.74 ± 3.61
RF	12.92 ± 11.49	10.69 ± 7.61	10.64 ± 8.73*	12.13 ± 9.74	12.14 ± 9.84
BF	10.37 ± 6.49	8.15 ± 6.07	5.99 ± 5.04	8.14 ± 6.29	7.55 ± 5.27
G _{Max}	6.93 ± 5.86	5.21 ± 3.99	4.46 ± 3.64	5.21 ± 4.26	5.69 ± 5.35
G _{Med}	8.41 ± 8.28	5.51 ± 5.64	4.99 ± 6.56	5.22 ± 4.04	5.86 ± 6.43
AL	8.51 ± 3.09	8.15 ± 4.28	7.16 ± 3.35	6.39 ± 2.76*	6.57 ± 3.21*

* Statistical difference from 0 % BWS, $p < 0.05$

** Statistical difference from 0 % BWS, $p < 0.001$

In contradiction to Lewek's [22] study reporting unchanged plantar flexor muscle activity with BWS, the amplitude and duration in the GA muscle were linearly and significantly reduced at the 20, 30, and 40 % BWS conditions without altering the normative EMG pattern. As a critical component for body propulsion, the reduced GA muscle activity with increased gait velocity and decreased DLS time in high BWS conditions can be explained as the body propulsion was achieved with relatively little active muscle powering compared with no BWS condition [19, 20].

Slightly reduced EMG amplitudes were found in the knee flexor and extensor such as VM, RF, and BF, but no significant differences were observed in higher BWS levels compared to 0 % BWS (Fig. 4; Table 3). Although the effects of BWS on the knee joint were minor, the activation duration of VM at 40 % and activation duration of RF at 20 % were significantly reduced as compared to 0 % BWS (Fig. 5).

For the hip joint flexor and extensor, there was a significant reduction in G_{Max} duration with increases in the BWS, due to the reduced maximum hip extension at high BWS levels. Furthermore, remarkable muscle activation changes were found in the hip ab/adductors. Both amplitude and duration of G_{Med} and AL were significantly reduced with increased BWS level. A strong hip adduction torque is required in the loading response period, followed by the

rapid transfer of BW onto the limb, and these demands continue throughout the stance period [28]. The G_{Med} muscle decelerates the rapid drop of the pelvis over the loading response period and maintains lateral stability during dynamic walking. Thus, the reduced G_{Med} muscle activation may be attributed to the reduced lateral momentum created in the pelvis and trunk induced by the unloading forces, following the smaller effort required for balancing the body in the horizontal plane. Furthermore, the hip flexion in pre-swing is initiated by both AL and RF muscles, while the AL muscle is further activated in pre-swing to restrain the abducting torque at the hip generated by BW falling toward the other limb [28]. The significantly reduced muscle load in AL implies an abductor torque that must be restrained to preserve weight-bearing balance with relatively little exertion. The reduced G_{Med} and AL support our third hypothesis that BWS systems should be able to increase lateral stability by reducing hip ab/adductor muscle amplitudes.

4.3 Clinical implications

Firstly, for effective and successful gait rehabilitation, the BWS levels selected must satisfy the normal sensorimotor input and extract the proportionately scaled motor responses required for normative gait [11]. However, the overhead harness BWS scheme often restricts pelvic lateral

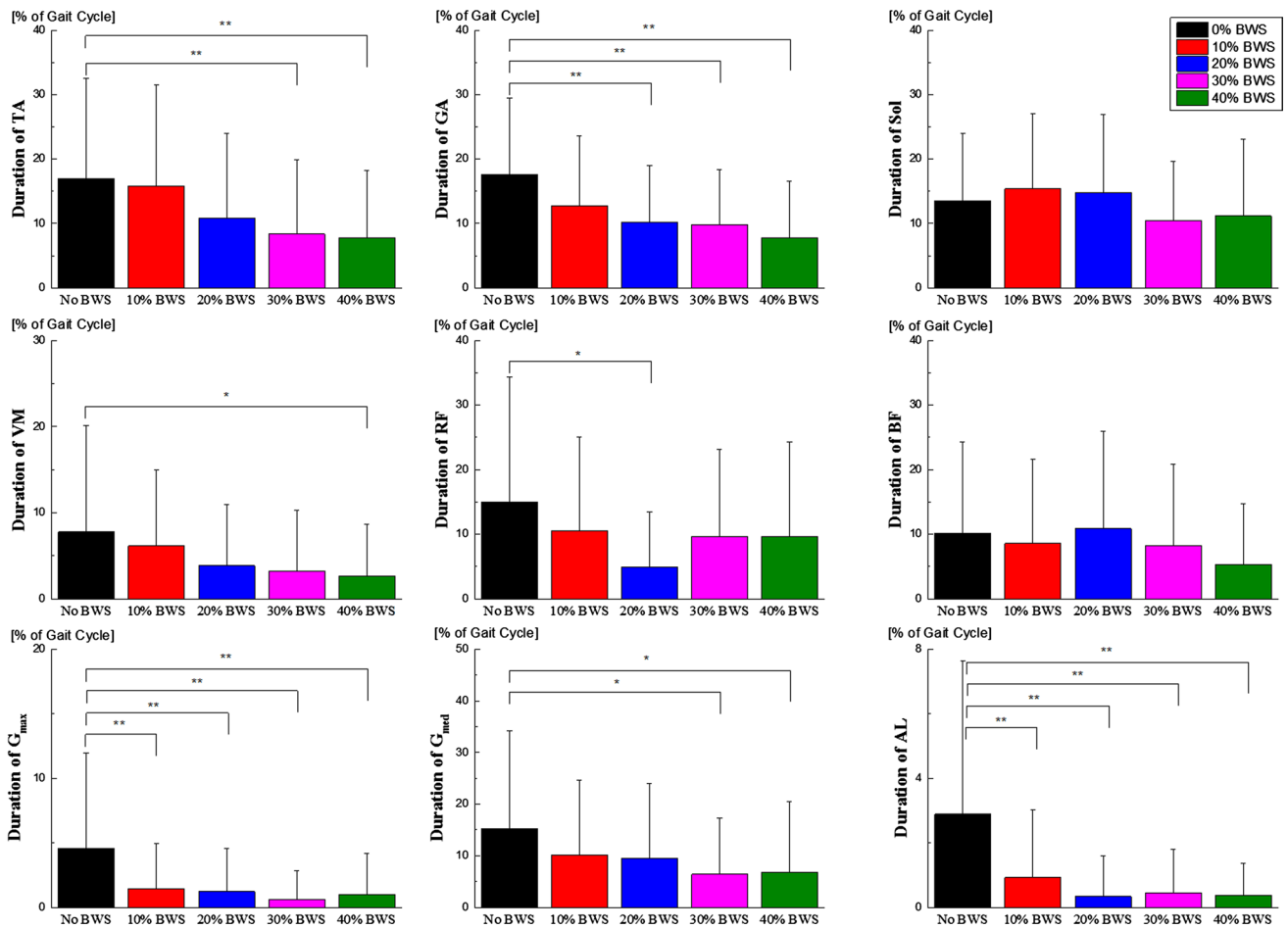


Fig. 5 Averaged EMG activation duration from nine major muscles during walking with BWS. The black bar shows averaged EMG activation duration in stance phase at 0 % BWS. Red, blue, pink, and green bars show the EMG amplitudes in 10, 20, 30, and 40 % BWS, respectively

and rotational movement, resulting in abnormal gait patterns [15, 35]. The restriction of the pelvis would finally affect gait functional outcomes after gait rehabilitation. The walker, which can facilitate 6 DoFs of pelvic motion combined with BWS ability, may provide neurologically challenged patients with afferent sensory feedback with linearly decreased muscle activation without altering normative EMG excursion during BWS training (Fig. 3).

Secondly, a previous study pinpointed that pelvic lateral displacement in patients with acute hemiparetic stroke was significantly increased to keep the body balanced from dysfunction of voluntary joint movements [8]. Such reduced G_{Med} and AL activation as the BWS level increased are of great importance in clinical trials for keeping patients' body laterally balanced, and this has a significant implication in increasing energy efficiency for maintaining lateral stability of neurological patients.

Last but not least, our findings show that the DLS time was shortened and muscle activation was decreased with

increasing BWS level. Particularly, despite the decreases in the EMG activity at the ankle joint, there was no notable change in kinematic variables at the ankle. This finding is important for the clinical application suggesting that with increased BWS levels, less ankle muscle effort is required to perform the same activity. Thus, the muscular effort at ankle joint can be reduced while providing the same amount of gait outcome, proving relatively high muscle efficiency at the ankle joint with increasing level of BWS.

Therefore, the BWS training with the robotic walker is expected to increase effectiveness of gait training by lowering muscular effort and increasing patients' mobility and stability.

5 Conclusion

The aim of this study has been to sufficiently investigate lower limb kinematics, temporospatial gait parameters,

Table 4 Averaged EMG activation duration of 9 major muscles in stance and swing phase

Conditions	0 % of BWS Mean ± SD	10 % of BWS Mean ± SD	20 % of BWS Mean ± SD	30 % of BWS Mean ± SD	40 % of BWS Mean ± SD
<i>Stance</i>					
TA	17.02 ± 15.51	15.86 ± 15.63	10.85 ± 13.14	8.34 ± 11.60**	7.78 ± 10.43**
GA	17.65 ± 11.89	12.73 ± 10.85	10.17 ± 8.78**	9.86 ± 8.53**	7.78 ± 8.82**
Sol	13.50 ± 10.53	15.39 ± 11.72	14.80 ± 12.17	10.51 ± 9.16	11.16 ± 11.92
VM	7.78 ± 12.33	6.16 ± 8.86	3.84 ± 7.15	3.20 ± 7.10	2.66 ± 5.96*
RF	15.02 ± 19.30	10.51 ± 14.54	4.96 ± 8.54*	9.69 ± 13.46	9.70 ± 14.56
BF	10.15 ± 14.15	8.60 ± 12.97	10.90 ± 15.03	8.28 ± 12.57	5.34 ± 9.44
G _{Max}	4.56 ± 7.39	1.47 ± 3.51**	1.24 ± 3.33**	0.64 ± 2.22**	0.99 ± 3.22**
G _{Med}	15.21 ± 19.05	10.21 ± 14.43	9.51 ± 14.49	6.41 ± 10.84*	6.81 ± 13.67*
AL	2.89 ± 4.76	0.93 ± 2.09**	0.34 ± 1.25**	0.46 ± 1.34**	0.36 ± 1.01**
<i>Swing</i>					
TA	11.37 ± 9.68	8.70 ± 7.95	8.61 ± 8.35	7.38 ± 7.57	10.46 ± 9.06
GA	5.14 ± 6.65	2.73 ± 4.25	1.52 ± 3.76**	1.49 ± 3.2**1	2.16 ± 4.49*
Sol	3.00 ± 4.20	2.28 ± 4.05	1.19 ± 2.41	1.40 ± 3.15	0.66 ± 1.70**
VM	0.91 ± 3.20	1.51 ± 3.88	1.25 ± 3.94	0.84 ± 2.82	0.95 ± 2.72
RF	8.58 ± 11.73	4.40 ± 7.67	4.40 ± 8.68	7.29 ± 11.64	8.46 ± 11.70
BF	5.14 ± 6.16	3.13 ± 5.31	1.80 ± 3.88**	3.04 ± 5.65	2.61 ± 4.27
G _{Max}	1.75 ± 4.20	0.37 ± 1.34	0.23 ± 1.13	0.70 ± 2.54	1.18 ± 3.57
G _{Med}	4.64 ± 9.26	2.08 ± 6.75	1.45 ± 4.94	0.45 ± 1.55**	1.89 ± 5.37
AL	1.73 ± 3.44	2.02 ± 3.66	1.51 ± 2.96	0.78 ± 1.47	1.35 ± 2.41

* Statistical difference from 0 % BWS, $p < 0.05$

** Statistical difference from 0 % BWS, $p < 0.001$

and EMG activity to address the question of how the gait variables adapt to reduced gravity over the gait cycle. Our unique robotic walker successfully reduced gravitational force and loading during gait. The findings of this study demonstrate the decreased intensity and duration of muscle activation without altering normal pattern, and high muscle efficiency for weight bearing and propulsion in the sagittal plane and for lateral balance. The findings of this study help guide the rehabilitation strategies and the future design of assistive robotic devices by highlighting the effectiveness of BWS gait training aimed at lowering muscle effort and increasing the stability of the patient. Although we only observed healthy individuals, our findings shed some light on determining the possible load-related sensory mechanisms that affect locomotor output.

People with neurological disorders would respond differently than non-impaired healthy subjects with increased amounts of BWS. Therefore, further studies with neurologically challenged patients suffering from gait impairments will be conducted in order to assess the effectiveness of over-ground BWS gait training without pelvic restriction. Thus, the use of a BWS system during over-ground walking will be recommended as a useful intervention strategy for gait rehabilitation.

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