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Assessment of the effects of body weight unloading on overground gait biomechanical parameters

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ABSTRACT

Background: Gait rehabilitation with body weight unloading is a common method of gait rehabilitation for clinical subjects with neurological and musculoskeletal impairments. However, the efficiency of this method was hard to assess given the confounding variables walking modality (treadmill vs. overground) and subjects' inability to maintain a comfortable speed when pulling a body weight unloading system by which they were suspended. By controlling the gait modality (overground) and devising a mechanical device that pulled the system at a constant speed, this study could examine the unique effects of body weight unloading on the biomechanical parameters of healthy subjects walking overground at comfortable speed.

Methods: Ten healthy subjects were instructed to walk overground under a control (no suspension vest) and three (0%, 15%, 30%) body weight unloading experimental conditions. Hip, knee and ankle spatiotemporal, kinematic, and kinetic measures were recorded for all conditions (six trials per condition).

Findings: ANOVA showed no changes in cadence, speed and stride length, a reduction in double limb support and increase in single limb support. Pairwise comparisons of gait parameters under 0%,15% and 30% body weight unloading conditions indicated significant reductions in lower joint kinematics and kinetics with increased body weight unloading. Additionally, despite changes in the peak values of kinematic and kinetic measures, the curvature patterns remained unchanged.

Interpretation: This study shows that overground gait with up to 30% body weight unloading reduces joint loads without modifying gait curvature patterns or the plantarflexion angle. Several clinical applications for gait reeducation conducted in situ with unloading are enumerated.

1. Introduction

Body weight unloading (BWU) conducted on treadmills has become a common method of gait rehabilitation for patients with neurological and musculoskeletal impairments (Barbeau et al., 2004; Hesse et al., 1994; Lee and Hidler, 2008; Mangione et al., 1996; Patiño et al., 2007; Threlkeld et al., 2003). The assumption behind this method is that the partial support of clinical subjects' body weight with a BWU suspension vest during treadmill walking will alleviate the load applied on the lower joints and the related pain experienced when patients start walking, thereby allowing them to generate the locomotor patterns (Sousa et al., 2009) and sensorimotor input essential for successful gait correction (Dietz, 2009). Gait rehabilitation on treadmills with a BWU suspension vest was, therefore, recommended early after injury or surgery to induce sensory stimulation (Threlkeld et al., 2003), and improve gait speed, balance and locomotion (Dickstein, 2008; Lamontagne and Fung, 2004; Perry and Davids, 1992; Perry et al., 1995; Schmid et al., 2007; Sousa et al., 2009; Van Hedel et al., 2006).

One of the main assumptions of gait research and rehabilitation on treadmills was that treadmill and overground gait patterns were similar enough so that gait corrections on treadmills could easily transfer to daily overground walking. However, research comparing treadmill and overground gait of healthy subjects refutes this assumption by showing that albeit small, the modifications of gait biomechanical parameters observed on a moving belt — the treadmill, were nevertheless significant (Murray et al., 1985; Riley et al., 2007; Strathy et al., 1983). In comparison to overground gait, healthy subjects' gait on treadmills became more conservative as exhibited by a significant reduction in speed and peak hip and knee flexion and extension which resulted in reduction in range of motion (Murray et al., 1985; Riley et al., 2007; Strathy et al., 1983). Similarly, significant modifications in healthy subjects lower joints' ground reaction forces (GRF) were observed during treadmill gait (Kram et al., 1998; Riley et al., 2007). In comparison to overground gait, healthy subjects' treadmill gait exhibited a significant decrease in ankle dorsiflexion and knee extension moments, and a significant increase in hip extension moments (Lee and Hidler, 2008; Riley et al., 2007).

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Research comparing overground vs. treadmill gait with BWU indicated significant deviations in gait motor patterns and kinematics during treadmill walking (Alton et al., 1998; Lee and Hidler, 2008; Threlkeld et al., 2003; Van Hedel et al., 2006). Healthy subject's foot on a treadmill was observed to be pulled posteriorly and to remain for a longer period of time on the moving belt which required proximal knee and hip kinematic adjustments to permit foot clearance (Threlkeld et al., 2003). The observed differences in gait patterns led to the conclusion that treadmill gait does not replicate overground gait. Consequently, the unique effects of BWU on gait parameters could not be assessed as long as research was conducted on treadmills (Carollo and Matthews, 2002; Harris and Smith, 1996; Hesse et al., 1997; Ivanenko et al., 2004, 2006; Lee and Hidler, 2008; Murray et al., 1985).

A limitation to be addressed prior to conducting overground gait research with BWU was related to healthy subjects' inability to maintain a comfortable overground walking speed when having to pull the BWU system to which they were attached. Consequently subjects' gait speed variability, which was easily controlled on treadmills with BWU, constituted a potentially confounding variable that had to be controlled prior to conducting this study. To meet this challenge a mechanical device was designed to pull the BWU system at a constant speed. No longer having to pull the BWU system, healthy subjects' gait speed variability overground was controlled thus allowing us to pursue the objectives of this study which were to examine the unique effects of BWU on gait spatiotemporal, kinematic and kinetic parameters under conditions that replicate daily walking.

We expected that partial reduction of healthy subjects' body weight, performed with a BWU system during overground walking, would reduce lower joint loads (kinetics), and result in modifications and changes of gait spatiotemporal and kinematic parameters to allow for gait curvature patterns to remain unchanged.

The hypotheses of this study were as follows: A reduction of 0%, 15% and 30% body weight load during overground walking will result in modifications and changes of healthy subjects' gait (1) spatiotemporal parameters, (2) kinematic parameters, and (3) kinetic parameters without modifying the gait kinematic and kinetic overall curvature patterns over the gait cycle.

2. Methods

2.1. Subjects

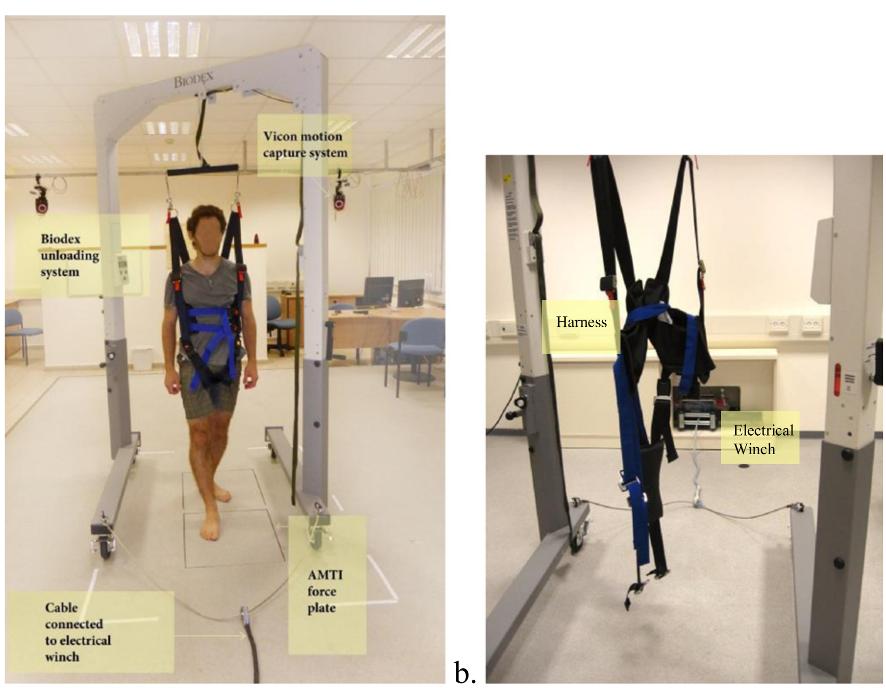
Ten male subjects were recruited for this study. The sample variability was controlled by including only healthy subjects with no previous history of lower extremity joint injuries or gait impairments. Additionally, all the subjects recruited were approximately the same age, height and weight: mean (SD) age in years was 23.8 (3), mean (SD) height was 1.72 (0.06) m and mean weight was 67.7 (5.7) kg. The study was approved by the Internal Review board, and informed consent was obtained from all subjects before data collection.

2.2. Instrumentation

The Biodex Unweighing System (Biodex Co., Shirley, NY, USA) (Fig. 1a) was used to manipulate subjects' body weight and accommodate for the vertical displacement of the center of gravity during overground gait under controlled BWU levels. This system includes a suspension vest with shoulder straps, a pelvic belt and a groin piece attached to the belt. Once suspended in the BWU device in an upright position, a pulley system lifted the subjects until the predetermined BWU level was reached, as a function of their body weight (in kg), and indicated on the screen of the Biodex system.

Previous gait research on treadmills with BWU has shown that a reduction of more than 30% BWU resulted in significant distortions of healthy subjects' biomechanical parameters. Subjects' gait speed and step length were greatly reduced (Lewek, 2011; Threlkeld et al., 2003; Van Hedel et al., 2006). Also observed were significant reductions in hip and knee kinematics (Finch et al., 1991; Threlkeld et al., 2003), in knee and ankle flexion moments, and in hip extension moments (Goldberg and Stanhope, 2013), and a significant increase in the ankle plantarflexion angle (Threlkeld et al., 2003; Van Hedel et al., 2006). Based on these findings, researchers concluded that a reduction of more than 30% body weight during treadmill walking resulted in a pronounced distortion in hip movement and impaired the ability of the ankle plantarflexors to produce propulsive forces. Learning from previous research (Lewek, 2011; Meinders et al., 1998; Sousa et al.,

a.



b.

Fig. 1. a. The BWU Biodex system, b. The BWU system connected to the electrical winch.

2009), the body weight reduction applied in this research was up to 30%.

As mentioned before, to circumvent the problem of healthy subjects' inability to maintain a comfortable overground gait speed, when having to pull the BWU system to which they were suspended, an especially tailored mechanical device was designed. This device is composed of an electric winch, installed in the wall facing the Biodex system, and a cable that connects the winch to the Biodex system, thus allowing one to pull the system across the floor at a controlled speed (Fig. 1b) without the subject's assistance.

Gait research on treadmills with BWU, which easily controlled for healthy subjects' gait speed by manipulating the moving belt, showed that setting subjects' gait speed below 0.7 m/s resulted in significant distortions of gait phases, duration and lower joint kinematic trajectories (Van Hedel et al., 2006). Additionally healthy subjects' overground walking speed reported in previous research ranged between 1.1 and 1.2 m/s (Bohannon, 1997; Öberg et al., 1993). Consistent with these findings the speed at which the winch pulled the BWU system was set at 1.1 m/s thus enabling to maintain a comfortable overground walking speed while suspended to the BWU system.

2.3. Procedure

The walking task included a control (no vest) and three experimental conditions, 0%, 15% and 30% BWU, manipulated by having subjects wear the Biodex suspension vest. Subjects were instructed to walk overground along a 10 m walkway while stepping on two AMTI OR6-7-1000 force plates, placed in tandem along a walkway under the control and experimental conditions. Each one of the control and BWU experimental conditions included six trials. During an accommodation period, a metronome helped subjects maintain a comfortable gait speed that matched the speed at which the winch pulled BWU system.

2.4. Measurements

Using the Vicon Plug-In-Gait model (Vicon®, 2002), reflective markers were placed on the subjects' anterior superior illiac spine (ASIS), posterior superior illiac spine (PSIS), lateral thigh, lateral shank, lateral femoral epicondyle, lateral ankle malleolus, second metatarsal head, and heels of both legs. The Vicon Knee Alignment Device (Vicon Motion Systems Ltd, Oxford Metrics Group, UK) was used to define the three-dimensional alignment of the knee flexion/extension axis of the subjects prior to the walking trials.

An eight-camera infrared Vicon motion analysis system with two force plates tracked and recorded the position of the reflective markers at 100 Hz. The Plug-In-Gait model (Vicon®, 2002) was used to calculate hip, knee and ankle joint kinematics and kinetics based on the reflective marker positions, subjects' anthropometric measurements and force plate data. Spatiotemporal, kinematic and kinetic data from the force plates and marker positions were exported for analysis to Vicon Polygon 3.5.1 software and to a custom-written Matlab program.

2.5. Spatiotemporal, kinematic and kinetic parameters

The spatiotemporal measures included the variables speed, cadence, and stride length over a gait cycle and durations of the stance phase: single limb support (SLS) and double limb support (DLS). Kinematic measures of the hip, knee, and ankle joints included the global maximum and minimum angular displacement in the sagittal plane. These included the hip peak flexion at loading response (hip angle 1-HA1), hip peak extension at terminal stance (hip angle 2-HA2), knee peak extension at terminal stance (knee ankle 1-KA1), knee peak flexion at initial swing (knee angle 2-KA2), ankle peak dorsiflexion angle during stance (ankle angle 1-AA1) and peak plantarflexion angle during swing (ankle angle 2-AA2), the range of motion (ROM) in the sagittal plane, and kinematic joint trajectories.

While previous studies operationally defined gait kinetic measures by the GRF, this study expanded this definition by adding additional kinetic measures namely the peak extension moment, peak flexion moment, and sagittal impulse of the hip, knee and ankle, and joints' kinetic trajectories. The specific lower joint peak moment values examined were for the hip peak extension at loading response (hip moment 1-HM1) and peak flexion at terminal stance (hip moment 2-HM2), for the knee peak extension at loading response (knee moment 1-KM1) and peak flexion at midstance (knee moment 2-KM2) and for the ankle peak dorsiflexion at loading response (ankle moment 1-AM1) and peak plantarflexion at terminal stance (ankle moment 2-AM2). The sagittal plane impulses that were examined included the flexion impulse of the hip during stance, the flexion impulse of the knee during midstance and the plantarflexion impulse during stance of the ankle.

2.6. Data analysis

Data from the force plates were analyzed and used to determine the initial heel strike and toe off, and divided into gait cycles accordingly. Each gait cycle corresponded to one interval between two consecutive contacts of the ipsilateral heel with the force plate. The spatiotemporal, kinematic and kinetic measures were averaged across six trials to obtain one mean value per variable for each subject for the control and each one of the three 0%, 15% and 30% BWU experimental conditions. Hip, knee and ankle kinematic and kinetic data were normalized to a full gait cycle, such that 0% corresponded to the touch of the ipsilateral heel on the contact surface and 100% to the subsequent touch by the same heel.

A repeated measures Analysis of variance (ANOVA) was conducted with all the dependent measures under three 0%, 15% and 30% BWU experimental conditions as independent measures. The control condition (no suspension vest) during which subjects walked at a selected comfortable speed was excluded from our ANOVA. Given that the BWU system was not programmed to reproduce the individuals speed of each subject but set to a standard walking speed of approximately 1.1 m/s and given that many gait parameters are dependent on subjects' walking speed, it was assumed that no valid conclusion could be drawn when comparing the dependent biomechanical measures under the control condition with a self-selected gait speed and the three experimental conditions with an imposed comfortable gait speed, therefore the exclusion. Any significant main effect of conditions indicated by ANOVA on the dependent measures was further analyzed with post hoc t-tests.

3. Results

3.1. Spatiotemporal measures

Table 1 presents the mean values (SD) of the spatiotemporal variables over the gait cycle under the control and three experimental BWU conditions.

Repeated measures ANOVA (2df) conducted with the spatiotemporal dependent measures under three 0%, 15%, and 30% BWU experimental conditions indicated a significant ($P < 0.001$) main effect of BWU conditions for the SLS and DLS. These findings suggest that increased BWU levels did not have an impact on subjects' speed, cadence or stride length.

To further verify this assumption, post-hoc *t*-tests were performed to compare pairwise spatiotemporal measures under the 0% vs. 15%, 15% vs. 30% and 0% vs. 30% BWU conditions (Table 1). Post-hoc *t*-test comparisons of spatiotemporal measures under 0% vs. 15% BWU, and 0% vs. 30% indicated significant differences: a significant increase in SLS and a decrease in DLS. Comparison of these measures under 15% vs. 30% BWU conditions indicated a significant decrease only in DLS. No significant changes in subjects' speed, cadence or stride length were indicated when 0%, 15% and 30% BWU levels were compared pairwise.

Table 1
Spatiotemporal variables by conditions — mean (standard deviation).

N = 10	Control — no harness	Experimental BWU Conditions		
		0%	15%	30%
	$\overline{X}$(SD)	$\overline{X}$(SD)	$\overline{X}$(SD)	$\overline{X}$(SD)
Measures				
Speed (m/s)	1.29 (0.1)	1.09 (0.1)	1.12 (0.03)	1.12 (0.1)
Cadence (steps/min)	113.3 (8.4)	106.9 (7.5)	108.7 (7.4)	110.2 (8.6)
Stride length (m)	1.37 (0.1)	1.21 (0.1)	1.23 (0.1)	1.23 (0.1)
DLS (s)	0.2 (0.02)	0.22 (0.03)[a,b]	0.17 (0.03)[c]	0.12 (0.02)
SLS (s)	0.43 (0.03)	0.45 (0.03)[a,b]	0.48 (0.04)	0.48 (0.04)

T-tests showing significant differences in spatiotemporal measures for paired comparisons:
[a] Significant difference ($P < 0.05$) between 0% and 15% BWU.
[b] Significant difference ($P < 0.05$) between 0% and 30% BWU.
[c] Significant difference ($P < 0.05$) between 15% and 30% BWU.

3.2. Kinematic measures

Table 2 presents the mean values (SD) of lower joint kinematics under the control and three experimental BWU conditions.

A repeated measures ANOVA (2df) was performed on all kinematic dependent measures with three 0%, 15% and 30% BWU levels as independent measures. A significant main effect ($P < 0.001$) of BWU levels was indicated for the peak flexion angle of the hip, knee, and ankle, and for knee peak extension angle. The peak extension angle of hip and ankle were not affected by changes in BWU conditions. Additionally, whereas a main effect ($P < 0.001$) of BWU level was observed for the RoM of the hip and knee, no such effect was indicated for the ankle RoM.

Post-hoc *t*-tests allowed to examine differences in hip, knee and ankle kinematic angular measures under pairwise comparisons (0% vs. 15%, 15% vs. 30% and 0% vs. 30% BWU) of experimental conditions (Table 2). Comparison of 0% vs. 15% BWU conditions showed a significant ($P < 0.001$) decrease in the peak flexion and RoM of the hip. In contrast, no significant changes were indicated on the parameters peak extension of the hip, peak flexion, peak extension and RoM of the knee and ankle. Comparisons of parameters in the 15% vs. 30% BWU conditions showed a significant ($P < 0.01$) decrease in peak flexion and RoM of the hip, knee and ankle. Additionally, while a significant decrease in knee peak extension angle was indicated under 30% as opposed to 15% BWU levels, the peak extension angle of the hip and the ankle (plantarflexion) were not affected by the increase in BWU level. Comparisons of the 0% vs. 30% BWU conditions showed a significant ($P < 0.01$) decrease in the peak flexion and RoM of the hip, and knee and in the peak extension of the knee. No significant changes were observed on any of the ankle kinematic parameters dorsiflexion and plantarflexion angles and RoM were not significantly modified.

3.3. Sagittal kinematic trajectories

Fig. 2 shows the averaged sagittal kinematic trajectories of the (a) hip, (b) knee and (c) ankle over a gait cycle for all subjects under the control and three 0%, 15% and 30% BWU experimental conditions.

As indicated in Fig. 2 the curvatures of the kinematic trajectories under control, 0%, 15% and 30% BWU showed mild changes in peak angular values that were examined (Table 1), yet the curvature pattern remained similar under the control and experimental conditions. The significant reductions in the peak kinematic measures included the hip peak flexion (HA1), knee peak extension (KA1), knee peak flexion (KA2) and ankle peak dorsiflexion angle (AA1).

3.4. Kinetic measures

Table 3 presents the mean values (SD) of lower joint kinetics in the sagittal plane under the control and three experimental BWU conditions.

A repeated measures ANOVA (2df) was conducted on the lower joint kinetic dependent measures in the sagittal plane with three 0%, 15% and 30% BWU experimental levels as independent measures. As shown in Table 3, a significant ($P < 0.001$) main effect of BWU level was indicated for the peak hip flexion moment, peak knee extension moment, and for flexion impulses of the hip and knee. While a significant main effect of BWU level was also shown for the ankle peak plantarflexion moment and ankle sagittal impulse, no such effect was indicated for the ankle peak dorsiflexion moment.

Post-hoc *t*-tests were performed to find differences in kinetic measures moments and impulses, under paired comparisons 0% vs. 15%, 15% vs. 30% and 0% vs. 30% of BWU levels. Comparisons of kinetic

Table 2
Kinematic parameters of the hip, knee, and ankle by conditions — mean (standard deviation).

N = 10		Experimental conditions			
Measures [deg]		Control — no harness	0%	15%	30%
		$\overline{X}$(SD)	$\overline{X}$(SD)	$\overline{X}$(SD)	$\overline{X}$(SD)
Hip	Peak extension (HA1)	−8.95 (6.4)	−5.53 (6.1)	−4.33 (7.1)	−5.38 (7.1)
	Peak flexion (HA2)	34.8 (7.7)	30.8 (6.5)[a,b]	27 (6.7)[c]	25.3 (7)
	Range of motion	43.7 (4.3)	36.4 (3.8)[a,b]	32.7 (3.5)[c]	27 (3.2)
Knee	Peak extension (KA1)	3.3 (5.3)	4.87 (3.4)[b]	4.98 (3. 4)[c]	7.11 (3.2)
	Peak flexion (KA2)	62.2 (4.5)	59.2 (4.5)[b]	58.8 (5.2)[c]	56.8 (6.4)
	Range of motion	58.9 (4.9)	54.3 (4.8)[b]	53.9 (5.9)[c]	49.7 (6.5)
Ankle	Peak plantarflexion (AA1) (extension)	−15.3 (3.5)	−14.1 (4.4)	−15.5 (6.4)	−14.6 (5.6)
	Peak dorsiflexion (AA2) (flexion)	13.7 (2.8)	13.9 (2.8)	13.9 (2.6)[c]	12.2 (3.3)
	Range of motion	29 (5.4)	28 (4.7)	29.4 (6.5)[c]	26.8 (6.4)

T-tests showing significant differences in kinematic measures for paired comparisons:
[a] Significant difference ($P < 0.05$) between 0% and 15%.
[b] Significant difference ($P < 0.05$) between 0% and 30%.
[c] Significant difference ($P < 0.05$) between 15% and 30%.

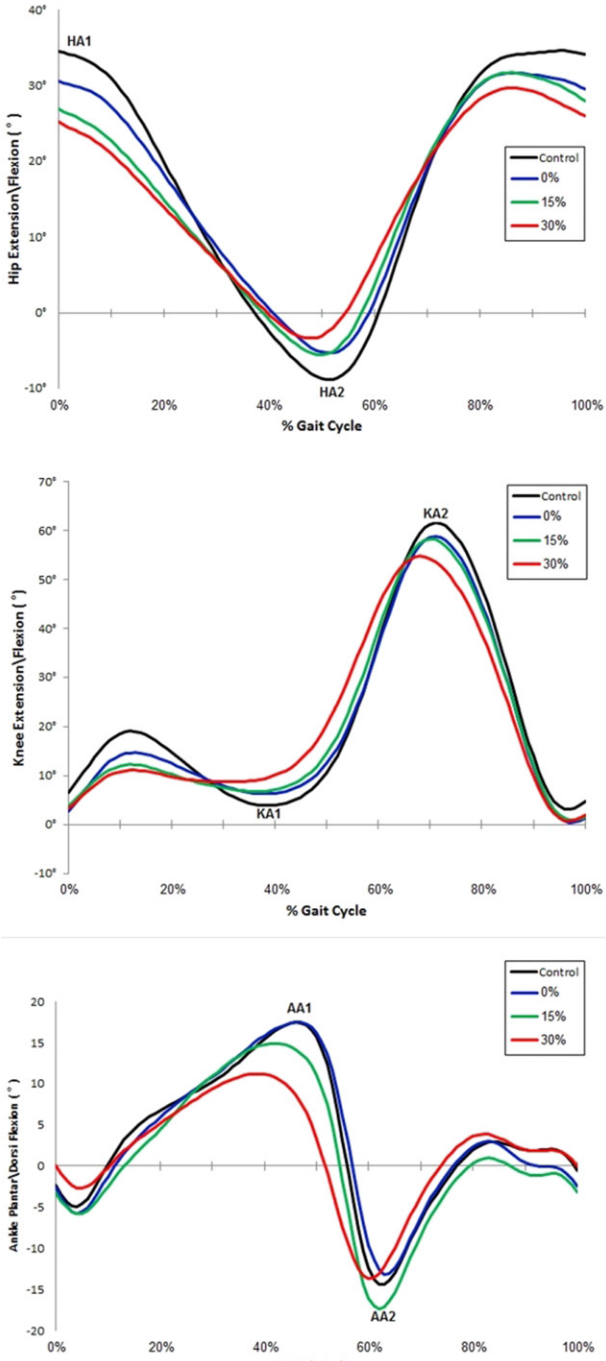


Fig. 2. Mean kinematic angular trajectories with extension and flexion peaks of the (a) hip, (b) knee, and (c) ankle during a gait cycle under three BWU experimental conditions. (HA1 — hip angle 1; HA2 — hip angle 2; KA1 — knee angle 1; KA2 — knee angle 2; AA1 — ankle angle 1; AA2 — ankle angle 2).

parameters in the sagittal plane under 0% vs. 15% BWU conditions (Table 3), indicated a significant ($P < 0.001$) decrease in the peak flexion moment and impulse of the hip, and in the peak extension moment of the knee. Similarly, the ankle plantarflexion moment and impulse were significantly reduced. In contrast the ankle dorsiflexion moment was not affected by the changes in BWU levels.

Post-hoc *t*-test pairwise comparisons of lower joints moments in the sagittal plane under the 15% vs. 30% BWU and 0% vs. 30% BWU conditions (Table 3), indicated a significant ($P < .001$) decrease in the peaks of the hip flexion moment, the knee extension moment and ankle plantarflexion moment. Additionally, hip, knee and ankle flexion impulse were significantly reduced. No significant changes in ankle dorsiflexion were observed when BWU conditions were compared pairwise.

3.5. Sagittal kinetic trajectories

Fig. 3 shows the averaged sagittal kinetic trajectories of the (a) hip, (b) knee and (c) ankle over a gait cycle for all subjects under the control and three experimental BWU conditions.

Fig. 3 shows that despite the reduction in moment magnitude during mid and late stance with increased BWU, the curvatures of kinetic trajectories were quite similar. The significant reductions included the hip peak flexion at terminal stance (hip moment 2-HM2), the knee the peak extension at loading response (knee moment 1-KM1) and the ankle peak plantarflexion at terminal stance (ankle moment 2-AM2).

4. Discussion

This study examined the unique effects of 0%, 15% and 30% BWU on the spatiotemporal, kinematic and kinetic overground gait parameters of healthy subjects walking overground at a comfortable speed, once a specially tailored mechanical device pulled the BWU suspension system at a constant speed. Previous research has shown that gait analysis conducted on a moving belt (treadmill) does not replicate overground walking (Riley et al., 2007; Strathy et al., 1983; Van Hedel et al., 2006). Consequently, the unique effects of BWU on gait parameters had to be assessed in situ, during overground walking (Lewek, 2011; Threlkeld et al., 2003; Van Hedel et al., 2006).

The first hypothesis assumed that a body weight reduction of 0%, 15% and 30% performed with a BWU system would result in small albeit significant changes in healthy subjects' spatiotemporal gait parameters during overground walking. This hypothesis was supported in this research. In line with previous research (Patiño et al., 2007; Van Hedel et al., 2006), once suspended by a BWU vest, prior to applying any body weight reduction, (control vs. 0% BWU condition) resulted in subjects reducing their speed. However, given that this effect could have been an artifact of the winch pulling the BWU system at a constant speed, comparisons of the spatiotemporal parameters under the control vs. 0% BWU level were left out of our analysis. Comparisons of gait spatiotemporal parameters under 0%, 15%, and 30% BWU levels showed that while healthy subjects' gait temporal organization — speed, cadence and stride length were not changed, the increased BWU levels was significantly related to small but nevertheless significant changes in SLS and DLS. Increasing the BWU level from 0% to 30% resulted in an increase in SLS and a decrease of DLS. Given the close connection between SLS and kinematics, an increase in SLS may suggest an increase in RoM of clinical subjects undergoing overgound gait training with BWU. Sousa et al. (2009) reported similar findings with clinical patients undergoing overground gait rehabilitation with BWU. These patients were observed to achieve greater balance and stability especially during SLS.

Hypothesis 2 assumed significant changes in healthy subjects' gait kinematic parameters under 0%, 15% and 30% BWU. This hypothesis was supported by our findings. A significant inverse relationship was indicated between an increase in BWU levels and a decrease in the peak flexion and RoM of the hip and knee in the sagittal plane. In contrast, the ankle kinematic parameters examined — dorsiflexion angle, plantarflexion angle and RoM were not significantly modified. These findings suggest that the modifications and changes observed in hip and knee angular parameters over the gait cycle under 0% to 30% BWU levels allowed the ankle RoM and plantarflexion angle to remain unchanged. This finding is discrepant with those of previous research (Van Hedel et al., 2006), reporting significant distortions in ankle kinematics. This discrepancy may be explained by the difference in the

Table 3
Kinetic parameters — mean (standard deviation) values for the peak extension and flexion moments and impulses of ankle, knee and hip under control and three experimental BWU conditions.

N = 10		Control — no harness	Experimental conditions 0%	15%	30%
		$\overline{X}$(SD)	$\overline{X}$(SD)	$\overline{X}$(SD)	$\overline{X}$(SD)
	Measures				
Hip	Peak extension moment (HM1)	6.2 (1.4)	4.76 (1.1)	4.71 (1.1)	4.72 (1.3)
	Peak flexion moment (HM2)	−5.87 (0.8)	−5.14 (0.9)[a,b]	−4.05 (0.7)[c]	−2.93 (0.5)
	Flexion impulse	−131.5 (20.2)	−135.7 (30.1)[a,b]	−120.5 (24.9)[c]	−88.5 (15)
Knee	Peak extension moment (KM1)	2.5 (1.1)	2.36 (0.9)[a,b]	2.1 (0.8)[c]	1.51 (0.5)
	Peak flexion moment (KM2)	−1.79 (0.6)	−0.83 (0.8)	−0.69 (0.5)	−0.8 (0.5)
	Flexion impulse	35.9 (13.8)	36.4 (17.3)[b]	32.8 (17.7)[c]	17.07 (7.9)
Ankle	Peak plantarflexion moment (AM1)	8.59 (0.9)	7.38 (0.8)[a,b]	5.71 (0.5)[c]	4.62 (0.5)
	Peak dorsiflexion moment (AM2)	−1.1 (0.4)	−1.2 (0.3)	−1.21 (0.2)	−1.11 (0.3)
	Plantarflexion impulse	224.5 (16)	195.2 (40.5)[a,b]	141.5 (22.6)[c]	123.5 (22.5)

T-tests showing significant differences in kinetic measures for paired comparisons:
[Moment units] = (% BW · Ht); [impulse units] = (% BW · Ht · % GC).
[a] Significant difference ($P < 0.05$) between 0% and 15%.
[b] Significant difference ($P < 0.05$) between 0% and 30%.
[c] Significant difference ($P < 0.05$) between 15% and 30%.

walking (overground vs. treadmill) modality. In contrast to the present study, previous gait research was conducted on a treadmill, which may have acted as a confounding variable interacting with the effects BWU on gait kinematics. As we assumed, small changes in peak kinematic measures under 0%, 15% and 30% BWU conditions were observed. However, despite these changes the curvature patterns remained very similar under the three BWU levels. This finding led to the conclusion that a body weight reduction of up to 30% does not distort kinematic curvature patterns over the gait cycle.

The third hypothesis assumed significant changes of gait kinetic parameters — lower joint moments and impulses under 0%, 15% and 30% BWU. This hypothesis was also supported by our findings. A significant inverse relationship was indicated between increased BWU levels and a decrease in the peak moments of the hip flexion, knee extension, ankle plantarflexion moments and sagittal impulses of these joints over the stance phase of the gait cycle. No changes were indicated in the hip extension moment, the knee flexion and the ankle dorsiflexion moments. Previous studies on overground gait with BWU operationally defined gait kinetics by the ground reaction force (GRF) and reported a reduction in healthy subjects' ability to generate force at initial contact of the foot with the ground and during propulsion (Patiño et al., 2007). Similar findings were reported with clinical subjects who exhibited a limitation in hip movement and difficulty to produce the force required for forward propulsion during overground rehabilitation with BWU (Sousa et al., 2009). The present study reported similar findings concerning significant reduction of the plantarflexion moment of the ankle with increased levels of BWU. No such modification was indicated for the ankle dorsiflexion moment and the ankle plantarflexion angle. Additionally, while the changes in the peak kinetic measures were observed, the kinetic curvature patterns remained similar under the three experimental conditions suggesting a reduction in peak kinetic measures rather than distortion of kinetic patterns over the gait cycle.

The importance of this study cannot be stressed enough since it shows that healthy subjects, during overground gait with BWU, exhibit very specific kinetic and kinematic patterns which the treadmill system may obscure. Conducting overground gait studies with BWU allowed us to examine such patterns in such a way that slight modifications, which are nevertheless important, could be shown.

The findings of this study have broad implications in the field of gait rehabilitation. Once comfortable walking speed is maintained, a body weight reduction of up to 30% may be safely applied to correct gait deficits under conditions that replicate daily walking, i.e., overground. Sousa et al. (2009) who tested clinical subjects during overground rehabilitation with BWU reported that being suspended by a BWU vest allowed for subjects' vertical alignment and stability of the trunk throughout gait. Achieving vertical alignment and stability of the trunk on their own (due to the BWU vest) will give clinical subjects the confidence to undergo rehabilitation early after surgery or trauma to regain balance and locomotion without fearing they will fall if unsupported by the physical therapist. An additional benefit of the BWU vest is that it will enable the physical therapist to focus on clinical subjects' locomotor patterns and quality of overground gait rather than assisting the patient for support (Sousa et al., 2009). Future studies with clinical subjects will verify these assumptions.

The kinetic results of the present study indicated an inverse relationship between increased BWU levels and a decreased magnitude of the internal joint moments suggesting that the reduction in body weight decreased the ground reaction force measured by the force plates. The reduced moments observed at initial contact of the foot and propulsion (stance) during overground gait with BWU suggest facilitation of locomotor patterns for clinical subjects who cannot apply the same force as healthy subjects. These findings have wide-ranging implications in the field of gait rehabilitation of patients with neurological and skeletal muscle abnormalities.

Another application of overground rehabilitation with BWU may target patients with Osteoarthritis (OA), a degenerative disease affecting the knee of over 25% of the senior population. These patients usually suffer of knee OA characterized by a pronounced knee adduction moment (KAM), which results in pain during any load-bearing activity such as walking (Andriacchi et al., 2009; Block and Shakoor, 2009; Brandt et al., 2006; Felson et al., 1992; Kito et al., 2010). So far, reduction of the KAM through rehabilitation interventions has not been successful to improve gait. The reduction of loads on OA patients' lower joints and not simply on the KAM during overground rehabilitation with BWU may reduce the level of pain OA patients experience while improving kinematic and kinetic trajectories under conditions that replicate daily walking. An additional potential benefit of partial body weight reduction is to encourage obese patients with gait abnormalities to undergo gait rehabilitation.

In conclusion by relieving loads on joints while maintaining kinematic and kinetic gait patterns, overground gait with up to 30% BWU stands out as an efficient method of gait re-education allowing clinical subjects to perform the task of walking, in situ, on a surface that is used in daily walking activities. As gait biomechanical parameters improve the BWU levels may be progressively reduced until daily locomotion without any body weight reduction is achieved, thereby facilitating the transfer of what has been learned in rehabilitation to everyday living conditions. By charting the modifications and changes in healthy subjects' lower joint kinematics and kinetics at various stages of the gait cycle under 0%, 15% and 30% BWU levels, these changes could

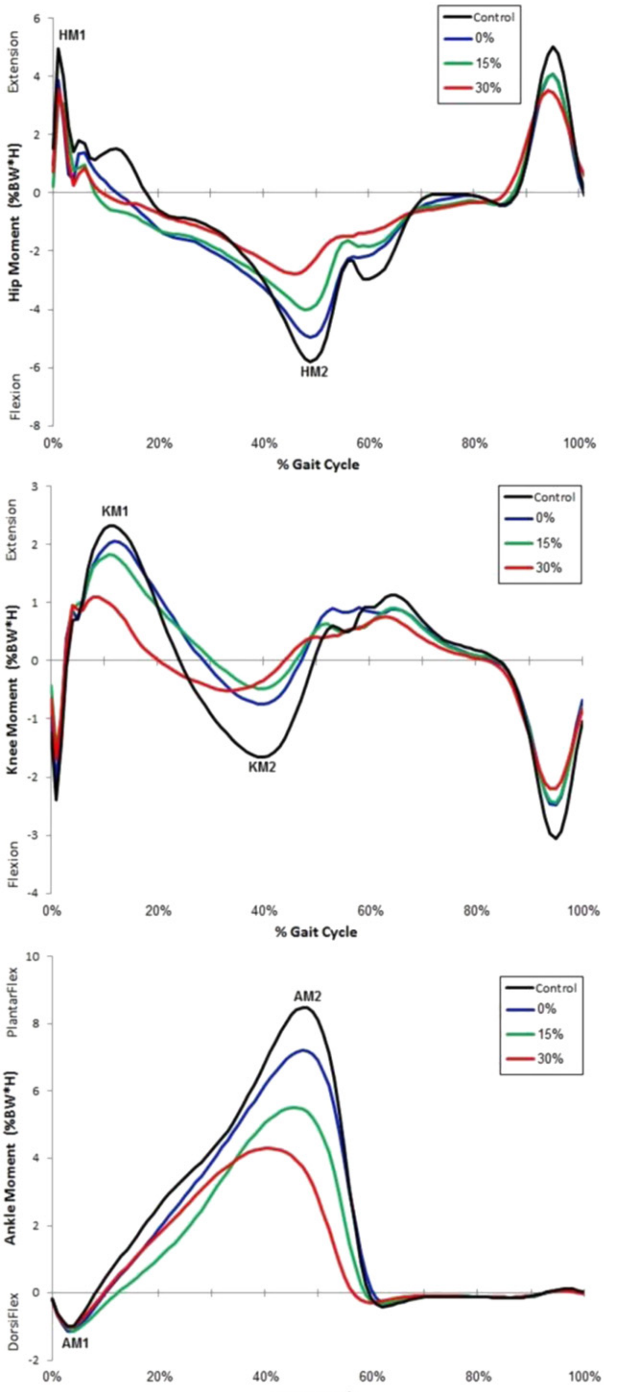


Fig. 3. Mean kinetic trajectories in the sagittal plane of the (a) hip, (b) knee, and (c) ankle during a gait cycle under three BWU experimental conditions. (HM1 — hip moment 1; HM2 — hip moment 2; KM1 — knee moment 1; KM2 — knee moment 2; AM1 — ankle moment 1; AM2 — ankle moment 2).

be subsequently used to determine deviations of clinical subjects' biomechanical parameters and consequent gait corrections to be achieved under various levels of BWU.

4.1. Limitations of this study

This study only measured young healthy subjects. Future studies will examine the gait spatiotemporal, kinetic and kinematic parameters of healthy elderly subjects and thereafter with clinical elderly subjects suffering from OA or other gait impairments. Testing clinical (young or elderly) subjects undergoing gait rehabilitation with BWU in situ, under normal daily walking conditions will increase the reliability of the findings of the present study.

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