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KINEMATIC, KINETIC AND ELECTROMYOGRAPHIC CHARACTERISTICS OF YOUNG ADULTS WALKING WITH AND WITHOUT PARTIAL BODY WEIGHT SUPPORT

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ABSTRACT

Objective: The aim of this study was to analyze the kinematic, kinetic and electromyographic characteristics of young adults walking on a fixed platform without a vest and with partial body weight support (PBWS) of 0, 10, 20 and 30%. Method: Eight young adults (mean age: 22.2 years) were videotaped walking on a walkway that contained a force plate embedded in its middle portion, to record the ground reaction force (GRF) components. Reflective markers were placed on the main anatomical points of the lower limbs in order to acquire kinematic data, and surface electrodes were attached to the tibialis anterior and gastrocnemius medialis muscles in order to record electromyographic muscle activity. Results: Significant differences among the five experimental conditions were observed with regard to spatial-temporal variables, the maximum and minimum angles for the thigh, knee, and ankle, and the amplitudes of the anteroposterior horizontal and vertical GRF components. Generally, the greatest changes occurred with PBWS of 30%. Conclusion: It is important to take into consideration the compensations to walking patterns that occur with PBWS, in planning therapeutic interventions. Moreover, to better define the use of suspended weight systems in rehabilitation programs, further investigations should be conducted in order to verify the walking patterns on fixed platforms among populations with movement disorders.

Key words: gait, biomechanics, ground reaction force.

INTRODUCTION

Weight support systems have been increasingly utilized for walking reeducation. The growth in implementation of this resource lies in the fact that weight suspension may decrease the biomechanical restrictions, thereby facilitating progressive weight support and improving the dynamic balance responses[1-3]. Thus, partial body weight support (PBWS) systems may be used in intervention cases in which the patient presents severely compromised locomotion[4], thus providing the possibility of walking reeducation. However, the alterations in walking patterns that result from the use of this system are in need of further examination. One of the first investigations aimed at investigating the effects of PBWS on walking was carried out by Finch, Barbeau & Arsenault[5], who examined healthy adults walking on a monitored treadmill with PBWS of 30, 50 and 70%. In general, these authors found that, as the percentage of PBWS increased, the support phase duration, total time with double support and angular displacement of the hip and knee decreased, and the activation amplitudes for the tibialis anterior, gastrocnemius and gluteus medius muscles were modified[5].

Other studies have examined the changes in walking patterns among healthy young adults on a treadmill with PBWS of 30%[6,7]. Again, the duration of the support phase and the activation amplitudes for the gastrocnemius and anterior tibial muscles decreased when there was PBWS of 30%. More recently, Threlkeld, Cooper, Monger, Craven & Haupt[8] analyzed the walking of healthy youths on a treadmill with a fixed speed of 1.25 m/s, with PBWS of 0, 10, 30, 50 and 70%. These authors found that the biggest changes occurred with PBWS of 50 and 70%, such that the cadence decreased, the length of the step increased and, at the beginning of the balance phase, the angular displacement of the hip and knee decreased and that of the ankle increased.

Although the studies mentioned above have contributed towards better understanding and consequently a better foundation for clinical interventions with PBWS, there is still a need to explain several matters relating to the effects of PBWS on walking patterns. One of the problems found in comparing the studies is the lack of uniformity in the methodological procedures, and therefore a more detailed and controlled comparison of walking patterns with PBWS is necessary. However, more importantly, whereas the effects of PBWS on walking patterns have been examined among adults and patients walking on treadmills, very little is known about the effects of PBWS on walking patterns on fixed floors. In this light, the present study had the aim of analyzing the kinematic, kinetic and electromyographic characteristics of walking among young adults without vests and with PBWS of 0, 10, 20 and 30%, on fixed floors.

METHODS

Participants

Eight university students (4 female and 4 male) without orthopedic or neurological impairments and with an average age of 22.2 years (±1.58) participated in this study. All the participants were informed about the experimental procedures and signed a consent statement that had been approved by the Research Ethics Committee of the Bioscience Institute, UNESP, Rio Claro Campus (Report 1735, dated March 17, 2004). All the participants completed the task they were asked to perform, with no sample loss.

Procedures

Firstly, the participants' body mass and height were obtained. Then, self-adhesive disposable surface electrodes of 2.5 cm in width, 3 cm in length and 3 cm of center-to-center spacing were attached to the belly of the tibialis anterior and gastrocnemius medialis muscles of both legs, in the direction of the muscle fibers, to record the electromyographic activity. The muscle region to which each electrode was attached was waxed and cleaned with alcohol before the electrode was attached. All the electrodes were connected by cables to a transmitting unit that pre-amplified and transmitted the signals generated by the activation of the muscles (MCS1000, EMG System do Brasil). Next, reflective markers were attached to the greater trochanter, femoral condyle, fibular malleolus and fifth metatarsus of both legs, in order to record the kinematic data.

A walkway of 7 x 1 x 0.03 m (length, width and height, respectively) was set up and a force platform (Kistler, 9286A) was embedded in the center of this walkway, under a rubber mat, to record data relating to the vertical and horizontal components of the ground reaction force (GRF). A video camera (Panasonic, M9000) was positioned perpendicularly to the walkway, at a distance of 6 m from its center, to enable filming of the participants in the sagittal plane. A metal structure of 5 x 3 m (length and height, respectively), with a track in the upper central region, was set up to enable weight suspension for the participants (Figure 1). A combination of an electric motor, pulley system and steel cable (PA 200 ELECTRIC HOST) was coupled to the track and to a vest (PETZ, Light C70) consisting of safety belts for the participants' trunk, pelvis and upper thighs. Furthermore, a load cell (Alfa Instrumentos, SV200) connected to an amplifier (MCS1000, EMG System do Brasil) was coupled to the steel cable close to the vest. Thus, when the motor was activated, it was possible to regulate the length of the steel cable, thereby lowering or raising the vest that the participant was wearing, and thus varying the weight suspension, which was recorded by the load cell.

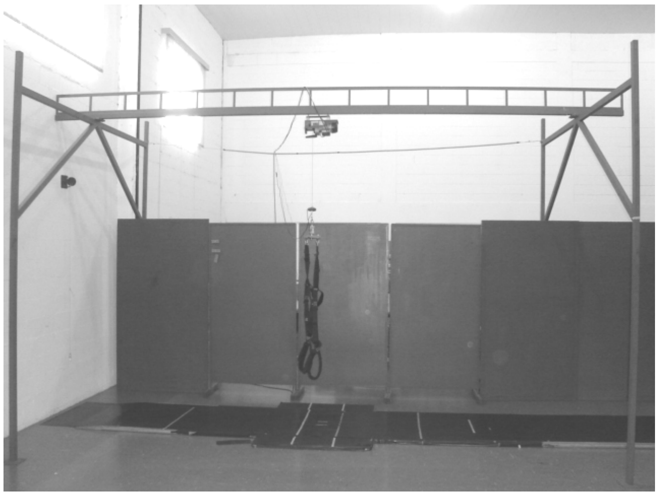

Figure 1. Partial view of experimental setting, showing the walkway, metal structure, motor and vest.

To define the percentage PBWS for the participants, the body mass (kg) of each of them was taken into consideration in order to calculate the amount of suspension. For this, the participants remained in a comfortable upright position while the height of the system was adjusted according to the predetermined percentage PBWS (0, 10, 20 or 30% of the total body mass), such that this percentage could be viewed on a computer screen.

The acquisition frequency for the signals coming from the force platform, electromyograph and load cell was 1000 Hz and the filming was done at 60 Hz. The signals from the force platform, electromyographic (EMG) activity and load cell were recorded using an analog data acquisition unit from the Optotrak system (Optotrak 3020, NDI) system. Synchronization of these data with the filmed data was achieved by using the instant at which the ipsilateral heel touched the contact surface, thus indicating the beginning of the walking cycle. This instant was identified on the filming and in the anteroposterior component of the GRF.

Before starting the task, the participants went through a familiarization period and, after this period, each of the conditions was repeated 10 times. The first 10 repetitions were done without the vest and the order of the other conditions was defined randomly. Under all conditions, the participants were asked to walk at a self-selected comfortable speed along the walkway.

Data analysis

Three repetitions for each experimental condition were selected, and one walking cycle from each repetition was analyzed. Each cycle corresponded to an interval between two consecutive touches of the ipsilateral heel on the contact surface. This selection took into consideration the best viewing of the markers, whether the walk was performed without interruption, image positioning in the central viewing area of the camera, foot contact with the force platform and the electromyographic data with the least noise.

The images from the selected cycles were captured and the markers positioned at the anatomical points were automatically digitized using the Ariel Performance Analysis System (APAS; Ariel Dynamics, Inc.) program. During the digitizing process, the occurrence of the main events of the walking cycle were determined: first ipsilateral heelstrike (IHS1), second ipsilateral heelstrike (IHS2), contralateral heelstrike (CHS), ipsilateral toe-off (ITO) and contralateral toe-off (CTO). Following this, the "x" and "y" coordinates of each marker were transformed into the metric system and then filtered using a second-order Butterworth digital filter with a cutoff frequency of 4 Hz. These data were used to calculate the descriptive, temporal and angular variables of walking.

The descriptive variables provided information about the length, duration, cadence and speed of the walking cycle, and the temporal variables provided information about the duration of the support phase and its subphases: first double support, simple support and second double support. In the case of angular variables, the maximum displacement (in flexion) and minimum displacement (in extension) of the thigh, knee and ankle were calculated. The coordination between the legs was analyzed by calculating the relative phase between the right and left legs, thus obtaining the temporal difference between the contacts of the ipsilateral and contralateral heels in relation to the total duration of the walking cycle. The force platform data were filtered with a fourth-order Butterworth digital filter, with a cutoff frequency of 6 Hz and normalized according to the respective participant's body weight. For the anteroposterior horizontal component of the GRF, the peaks corresponding to the deceleration and acceleration of the body were obtained. In the same way, for the vertical component, the first and second peaks and the trough were also obtained. Since these peaks and the trough of the vertical component were not distinct under the conditions of PBWS of 10, 20 and 30%, the GRF values under these conditions were defined based on temporal events of the walking cycle: the first peak obtained at the instant when the ITO occurred; the second peak at the instant when the CHS occurred; and the trough at the instant when the contralateral leg was halfway through the balance phase.

The EMG activity data were filtered with a fourth-order Butterworth digital filter, high-pass and cutoff frequency of 3 Hz, to obtain the linear envelope. Finally, because the stride duration was different between the cycles, all the data (kinematic, kinetic and electromyographic) were normalized in time from 0 to 100%, 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.

Statistic Analysis

To compare the five experimental conditions, six analyses of multivariance (MANOVAs) and two analyses of variance (ANOVAs) were performed, taking the five task conditions as the factors (without vest and with PBWS of 0, 10, 20 and 30%), and these factors were treated as repeated measurements. The dependent variables were the length, duration, cadence and speed of the stride for the first MANOVA; the duration of the first double support, simple support and second double support for the second MANOVA; angular displacements in flexion and extension for the thigh, knee and heel for the third and fourth MANOVAs, respectively; magnitudes of the peak acceleration and deceleration of the horizontal component of the GRF for the fifth MANOVA; amplitudes of the first peak, second peak and trough of the vertical component for the sixth MANOVA; support phase duration for the first ANOVA; and relative phase between legs for the second ANOVA. When the condition factor indicated a difference in the analyses, univariate tests and *post hoc* tests with Bonferroni adjustments were used. In all analyses, the significance level was maintained at 0.05. The analyses were performed using the SPSS software (SPSS for Windows – version 10.0).

RESULTS

Descriptive and temporal variables

Table 1 presents the mean values (± standard deviation) for the descriptive and temporal variables of the stride under the five experimental conditions. While some differences were observed for the variables of duration, cadence and speed among the experimental conditions, no difference in the temporal organization of the walking stride was observed.

Angular variables

Table 2 presents the mean values (± standard deviation) of the angular displacements of the thigh, knee and ankle and of the relative phase for coordination between the legs, under the different experimental conditions.

Table 1. Mean values (± standard deviation) for the descriptive walking variables of length, duration, cadence and speed, and the temporal variables of support phase, first double support, single support and second double support during the walking cycle, under the five experimental conditions (no vest and PBWS of 0, 10, 20 and 30%) (n= 8).

	Experimental Conditions				
	No vest	**0%**	**10%**	**20%**	**30%**
Descriptive					
Length (m)	1.39 ± 0.09	1.27 ± 0.19	1.19 ± 0.26	1.32 ± 0.14	1.28 ± 0.20
Duration (s)	1.37 ± 0.14[a,b]	1.53 ± 0.20	1.57 ± 0.23[a]	1.70 ± 0.32	1.72 ± 0.31[b]
Cadence (stride/s)	0.73 ± 0.07[a,b,c]	0.66 ± 0.09[d]	0.64 ± 0.09[a,e]	0.62 ± 0.10[b]	0.59 ± 0.10[c,d,e]
Speed (m/s)	1.03 ± 0.15[a,b,c]	0.85 ± 0.21	0.78 ± 0.24[a]	0.81 ± 0.21[b]	0.77 ± 0.22[c]
Temporal (%)					
Support phase	64.98 ± 2.34	66.94 ± 2.82	66.64 ± 3.53	64.54 ± 3.30	3.35 ± 2.75
First double support	13.20 ± 1.84	16.82 ± 2.61	14.67 ± 2.40	14.10 ± 3.65	13.00 ± 2.88
Single support	36.56 ± 1.55	34.49 ± 2.13	35.48 ± 2.09	35.42 ± 3.33	36.37 ± 3.67
Second double support	15.21 ± 1.27	15.64 ± 2.32	16.48 ± 3.34	15.02 ± 3.16	13.98 ± 3.55

Note: Same letters indicate significant differences ($p < 0.05$) between the respective experimental conditions.

Table 2. Mean values (± standard deviation) for the flexion and extension displacement of the thigh, knee and ankle, and relative phase during the walking cycle, under the five experimental conditions (no vest and PBWS of 0, 10, 20 and 30%) (n= 8).

	Experimental Conditions				
	No vest	**0%**	**10%**	**20%**	**30%**
Displacement in flexion (º)					
Thigh	16.01 ± 3.38[a]	19.32 ± 2.98	18.04 ± 3.82	15.65 ± 1.97[a]	14.54 ± 1.56
Knee	64.13 ± 6.11	61.65 ± 8.28	64.30 ± 3.56	61.68 ± 4.03	56.89 ± 4.20
Ankle	20.46 ± 2.47[a,b,c]	16.86 ± 2.45	14.14 ± 3.05[a]	14.94 ± 2.58[b]	13.95 ± 3.12[c]
Displacement in extension (º)					
Thigh	31.59 ± 1.45[a,b,c]	27.23 ± 3.43[a]	26.98 ± 2.76[b]	28.94 ± 4.15	26.31 ± 3.33[c]
Knee	4.16 ± 3.94	1.41 ± 1.43[a,b]	4.15 ± 3.93	3.89 ± 1.42[a]	5.14 ± 3.60[b]
Ankle	15.78 ± 2.23[a,b]	14.48 ± 3.49	13.20 ± 3.26	12.25 ± 2.23[a]	11.16 ± 1.29[b]
Relative phase (%)	49.76 ± 1.17	51.31 ± 1.07	50.15 ± 1.18	49.53 ± 1.59	49.36 ± 3.07

Note: Same letters indicate significant differences ($p < 0.05$) between the respective experimental conditions.

Ground reaction force

The mean trajectories of the anteroposterior horizontal and vertical components of the GRF are presented in Figure 2. While the anteroposterior horizontal component presented distinct deceleration and acceleration phases under the different experimental conditions (upper part of Figure 2), the two peaks and trough of the vertical component were only well-defined under the conditions of no vest and with PBWS of 0% (lower part of Figure 2).

The mean values (± standard-deviation) of the deceleration and acceleration peaks of the anteroposterior horizontal component, and the first and second peaks and trough of the vertical component of the GRF are presented in Table 3.

Electromyographic activity

The mean trajectories of the EMG activity of the tibialis anterior and gastrocnemius medialis muscles during the walking cycles, under the different experimental conditions, are presented in Figure 3. In general, the pattern was similar for the two muscles and, particularly for the gastrocnemius medialis, the amplitude seemed to have increased under the condition of 0% PBWS and decreased with PBWS of 10, 20 and 30%, in relation to the condition without the vest.

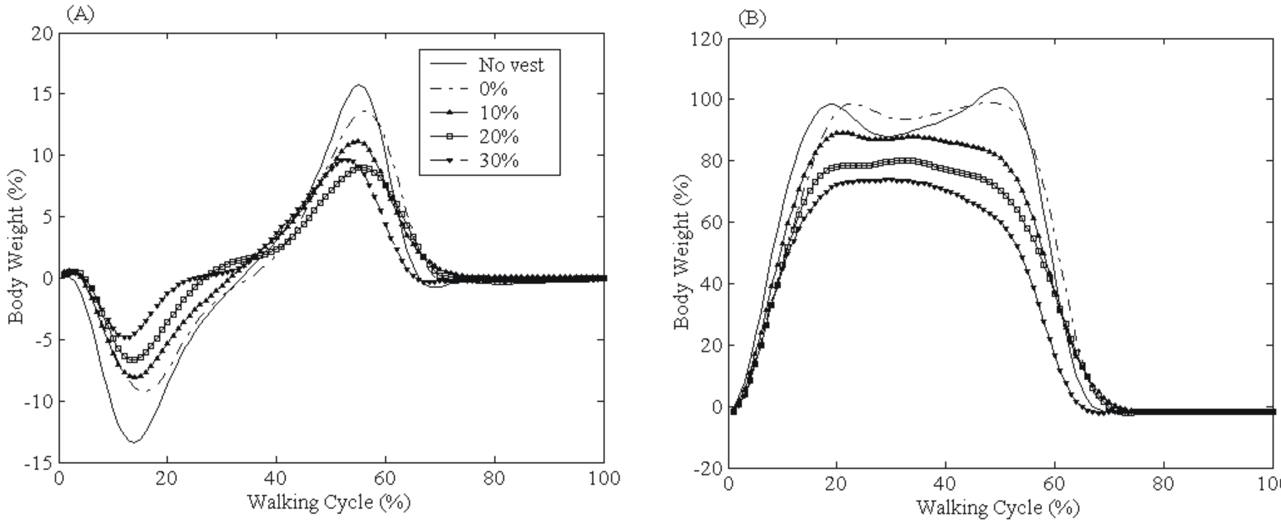

Figure 2. Mean trajectories of anteroposterior (A) and vertical (B) ground reaction force components during the walking cycle, under the five experimental conditions.

Table 3. Mean values (± standard deviation) for deceleration and acceleration peaks, from the anteroposterior GRF component and for the first peak, second peak and trough from the vertical GRF component during the support phase, under the five experimental conditions (no vest and PBWS of 0, 10, 20 and 30%) (n= 8).

	Experimental Conditions				
	No vest	**0%**	**10%**	**20%**	**30%**
Horizontal component (%)					
Deceleration peak	13.63±2.64[a,b,c,d]	9.82±2.60[a,e]	8.79±2.95[b,f]	6.71±2.11[c,f]	5.33±0.96[d,e]
Acceleration peak	16.22±3.05[a,b]	15.20±3.49	12.20±2.12[c]	10.35±1.96[a,c]	10.44±2.99[b]
Vertical component (%)					
First peak	97.86±2.44[a,b,c]	99.88±2.81[d,e,f]	89.26±4.78[a,d,g,h]	78.41±4.99[b,e,g,i]	71.33±3.06[c,f,h,i]
Second peak	103.15±4.98[a,b,c]	97.81±5.62[d,e,f]	80.56±4.44[a,d,g,h]	71.51±4.60[b,e,g,i]	61.58±4.55[c,f,h,i]
Trough	87.73±7.28[a,b,c]	93.82±3.68[a,d,e,f]	87.03±5.38[d,g,h]	79.03±5.68[b,c,g,i]	72.94±3.85[c,f,h,i]

Note: Same letters indicate significant differences ($p < 0.05$) between the respective experimental conditions.

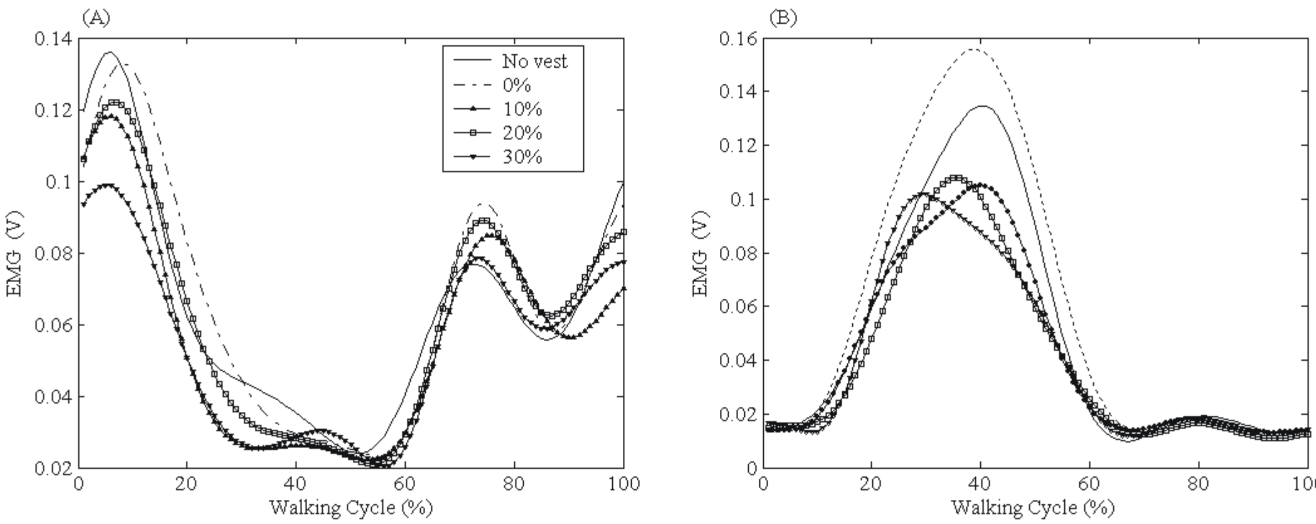

Figure 3. Mean trajectories of electromyographic (EMG) activity from tibialis anterior (A) and gastrocnemius medialis (B) muscles during the walking cycle, under the five experimental conditions.

DISCUSSION

This study had the objective of analyzing the kinematic, kinetic and electromyographic alterations to the walk of young adults with and without PBWS. The results from this study showed that there were changes in the stride patterns, angular displacement and GRF components. These changes reflected the need to adapt the stride while walking under different PBWS conditions.

In general, walking under the PBWS conditions of 10, 20 and 30% was slower, with a lower number of strides than under the conditions without a vest and with 0% PBWS. But even with these alterations, the temporal organization of the stride was not altered by PBWS. However, with PBWS of 10, 20 and 30%, changes in angular displacements were found. This pattern may the associated with adjustment of the strides and is in accordance with the results found for walking on a treadmill with PBWS of 30%[6-8]. Nevertheless, direct comparison between the results from the present study and from previous studies must be made cautiously, since the biomechanical and neurophysiological demands are different between walking on a fixed floor and walking on a treadmill[9].

In general, these young adults adjusted their walking pattern such that their walk became more conservative. This strategy may have been used a means of adapting to the demands imposed by the PBWS system and may also have been due to the fact that the participants had no previous experience of walking under these conditions. Moreover, there may have been neurophysiological changes to the rhythmic activation of the distention receptors of the hip flexors and ankle flexors in the initial support phase and, subsequently, of the hip extensors and ankle extensors in the final support phase[10]. It must be emphasized that activation of these receptors is important in generating the walking pattern[11] and providing the sensory feedback needed for adjusting the control parameters of the system for the next cycle[12].

Under the condition with the vest on the motored treadmill, trunk control can be facilitated and the balance maintenance demands are eased[13,14]. Furthermore, Chen, Patten, Kothari & Zajac[15] found a reduction in energy expenditure during walking with weight suspension, in populations with movement abnormalities. This result may have been due to the positioning of the trunk with the vest, which minimized the postural responses and facilitated the movement of the legs. In addition to these alterations, the changes under the condition with the vest in the present study may also be attributed to the movement of the system responsible for PBWS, which might explain the possible difference in the amplitude of EMG activity of the gastrocnemius medialis muscle, due to the greater effort made by the participants to displace the suspension system. However, with increased PBWS percentages, the EMG activity amplitudes of this muscle seem to have diminished, thus suggesting that there is a lower need for neural recruitment to respond to the new PBWS condition. These suggestions corroborate the results found for the GRF components, considering that, with PBWS, the participants generated a lower force during the initial contact of the foot with the ground and during propulsion. These matters are especially relevant and could be taken into consideration during the rehabilitation process among patients with neurological and/or skeletal muscle abnormalities, thereby facilitating the walking reeducation overall or refining the segmental responses[2,3]. Also, the therapeutic advantages of using fixed ground with PBWS would be used, thus enabling training for walking within a context that is close to reality and further facilitating the transference of what was learnt[4].

In general, the results from the present study demonstrated clearly that the walking pattern among young adults is partially altered with PBWS. These alterations involve characteristics of the walking pattern and the functional and biomechanical needs. Thus, it is evident that the pattern that emerges in situations in which the body weight of the participants is partially relieved is different from what is observed under normal conditions of locomotion. These observations indicate that the use of PBWS systems for intervention and rehabilitation processes among patients with walking abnormalities must be better examined, since patients produce patterns that differ from those that would be used in walking on fixed ground and without body weight support. Furthermore, considering the efficacy of this intervention situation, there is a need for better understanding of the reasons and mechanisms involved in the clinical situation.

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