

**DIFFERENCES BETWEEN TARGETED AND MEASURED BODY WEIGHT
SUPPORT WITH THE USAGE OF A BODY WEIGHT SUPPORT SYSTEM**

A Thesis

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By
Robin Cropper
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Abstract

The purpose of the study was to determine if the body weight support system (BWSS) maintains the targeted percentage of the subject's body weight support (BWS) during walking and to explore the relationship between unloading and ground reaction force. Sixteen healthy college students (mean age: 22 years old) were recruited as subjects. Tests were conducted using an instrumented treadmill with a BWSS. The BWSS harness was secured around the lower abdomen and pelvis of each subject for effective body weight support. Vertical ground reaction force was measured by force platforms embedded in the treadmill. A force sensor was attached to the top of the harness to measure the actual amount of support force being exerted by the BWSS while the subject walked on the treadmill. The subject's body weight was supported at targeted levels of 0%, 15%, 30%, and 45%. Subjects walked at speeds of 0.447, 0.671, 0.894, and 1.117 meters per second for each level of targeted body weight support (TBWS). As speed increased, the maximum support force (SF_{max}) increased while the minimum support force (SF_{min}) decreased. As the levels of TBWS increased, the SF_{max} as well as the SF_{min} increased. The maximum support force was affected by the different body weight support levels and fluctuated from 5.08% to 11.22% above the TBWS. The minimum support force fluctuated from 0.40% below to 13.05% above the targeted body weight support. The fluctuations were also affected by walking speed. SF_{max} deviation ranged from 6.12% to 10.03% above TBWS across speeds while the SF_{min} deviated from 7.06 to 8.58% above TBWS across speeds. As speed increased, maximum ground reaction force also increased. As the levels of BWS increased, the maximum ground reaction force decreased. The BWSS did not sustain the targeted percentage of BWS and actually

supported a lesser percentage of the subject's body weight throughout the trials. Vertical ground reaction forces were reduced with the use of the BWSS but were disproportional to the TBWS levels. These findings are important in providing both safe and efficient treatment for future patients in producing maximum benefits in therapy.

Introduction

Body weight support (BWS) is a task-specific therapy that enables subjects to practice complex gait cycles repeatedly (Werner, 2002). The body weight support system (BWSS) has been widely used in clinics and rehabilitation settings for therapy.

Improvements in temporal/distance gait variable such as stride length, cadence, gait speed, and swing and stance symmetry as well as electromyographic patterns of patients were found with the usage of body weight support training (Hesse et al, 1997).

By supporting a certain percentage of the patient's body weight during gait, the retraining may facilitate the expression of a more normal gait pattern (Suteerwattanon et al, 2002). If this particular therapy is being used to limit weight bearing, a determination should be made as to whether it actually reduces the vertical ground reaction force and maintains the correct percentage of body weight being supported throughout therapy to be an effective treatment.

People with severe injuries to the lower extremity often experience pain and weakness during weight bearing activities (Flynn et al, 1991). This leads therapists who are designing exercise programs to avoid having their patients perform weight bearing activities and turn to more non-weight bearing activities. Unfortunately, functional weight bearing activities during rehabilitation should be introduced to promote the goals involved with the progression towards the patient's everyday activities of living (Gardner et al, 1998). Other devices such as crutches and walkers can be used to help reduce the amount of loading on an injured limb, but the amount of load reduction being measured cannot possibly be precise due to individual accounts of usage.

Harness supported treadmill ambulation has been promoted to provide reduction in weight bearing on healing tissues, reduction of energy cost of treadmill ambulation, rehabilitation of patients with strokes and spinal cord injuries, and retraining people with amputations (Field-Fote & Tepavac, 2002). Harness supported treadmill ambulation refers to decreasing an individual's effective body weight by a predetermined amount using a supporting harness and counterbalance system that accommodates the rise and fall of the body during treadmill ambulation (Murray et al, 1993). The BWSS attempts to provide consistent BWS and postural support to promote the coordination of the lower extremities (Miller et al, 2002).

This BWS modality has been used in the management of stress fractures, tendon repair, osteoarthritis, and lumbar intervertebral disc pathology (Flynn et al, 1991). The decrease in weight supported is aimed to minimize the demands on the muscles allowing the patient to develop more useful and efficient movement strategies (Miller et al, 2002). The idea is to at first minimize the stress level on the muscles in order for the patient to walk in a somewhat normal manner. The repetition and consistent nature of walking on a moving treadmill helps the patient to repeatedly practice the movement under controlled conditions (Suteerawattananon et al, 2002).

The use of BWS and treadmill training has also recently been applied to a new type of treatment involving electrical stimulation to injured limbs. This method has been shown to increase walking speed in patients with paresis due to incomplete spinal cord injury and cerebrovascular accident (Field-Fote and Tepavac, 2002). People with spasticity from spinal cord injury have more normal gait patterns with the use of BWS,

showing an increase in walking speed, stride length, and single limb support time with a decrease in double limb support time (Visintin and Barbeau, 1989).

Neurological deficits can lead to lack of proper muscular activation leading to an inability to sustain a lower amount of weight and poor balance. Balance has been defined as maintaining the projected trajectory of the body's center of mass within the area defined by the base of support (Hanke and Rogers, 1992).

Deficits in weight acceptance, single limb stance, and limb progression result from impairments in strength and motor control associated with neurological damage (Gardner et al, 1998). The training geared toward these deficits mainly focuses on bearing weight, shifting of weight, and balance performed separately usually before they can be incorporated into locomotion (Gardner). Body weight support during treadmill walking provides simultaneous balance and locomotion training.

The BWSS was found to be beneficial in retraining gait for people with neurological deficits (Finch et al, 1991). When BWS is used in gait retraining for people who have suffered from a stroke, the therapy begins early after the brain lesion to take advantage of the inherent flexibility of the neural elements (Nilsson et al, 2001).

In therapy, maintaining the exact percentage of BWS used is not nearly as important as the idea of decreasing the amount of support as the patient recovers from injury. Effective weight bearing through the affected limb is part of criteria used to determine the patient's readiness for decreasing BWS throughout treatment (Miller, 2001). When BWS is reduced (i.e., a smaller percentage is supported), the patient is getting closer to full recovery and can place more weight on the injured limb. Therefore, reducibility of measurements is very important for the purpose of quantifying progression

(Wilson et al, 2000). Treatment protocols based on invalid or unreliable measurements may not adequately focus on underlying dysfunction. As a result, this may lessen therapeutic effectiveness (Hanke and Rogers, 1992).

Physical therapists customarily measure their patients' movement performance. The BWSS enables therapists to step back and evaluate the patients' gait without having to be at their side for support (Miller et al, 2002). The methods used to evaluate their patients and plan the correct treatment protocols must be justified with respect to reliability and validity in order to provide accurate information and minimize misleading interpretations.

The decision to decrease BWS during therapy is judged by observational analysis by the therapist and is based on the quality of gait demonstrated by the patient (Miller, 2001). Quality of gait refers to the amount of weight that can be placed on the affected limb along with consistent and symmetrical steps (Miller). Performance plateaus over several days also influence decisions about this decrease. The ideal goal is for the patient to support more weight on the injured limb and improve the quality of gait as therapy progresses.

Review of Literature

The recovery of locomotion following interactive training with graded weight support in the adult spinal cat has led to the idea that the removal of body weight could be a useful method in human gait retraining therapy (Finch et al, 1991). Animal studies suggest that normal gait can be recovered using a program that provides support to body weight with support decreasing as abilities improve. The training of walking on a

treadmill with BWS is intended to optimize locomotor-related sensory inputs which may enhance the timing and coordination of motor activity (Dobkin, 1999).

The initiation of gait from a motionless position requires a shift from being relatively stationary to moving. Gait initiation is a task that challenges the balance control system by forcing an individual from a state of stable balance to continuously unstable posture while walking (Halliday et al, 1998). BWS has been shown to be a critical factor of BWS treadmill training for improving balance (Mudge et al, 2003).

Kinetics, the study of forces responsible for joint movements, has been used often to examine human movement. Comparing center of mass and center of pressure offers insight to the movement. The center of mass is the point on the body that moves in the same way that a single particle would move if subjected to the same external force, or the point at which the weight of the body can be considered to act (Rodgers and Cavanagh, 1984). The center of pressure is the time varying signal recorded from a force platform in the plane of the floor (Winter et al, 1990).

Force acting at some distance from the axis of rotation from the body produces a moment or torque that can cause the body to rotate (Novacheck, 2001). When moments become unbalanced, movement occurs. As the body changes its position, the distance between the center of mass and the center of pressure increases. This makes the subject less stable and requires active postural control to return the center of mass to its stable position within the base of support (Martin et al, 2002). Vertical excursion is the peak to peak amplitude of the vertical body's center of mass displacement. This part of the body's center of mass was found to increase with walking speed (Gard et al, 2004).

Hass and Burden (2000) used a Kistler force plate as a standard to assess the validity of the measurements of a balance performance monitor. Differences in the measurements that were collected from the monitor and force plate resulted in concurrent validity with the assessment using correlation and regression. This same idea is used in this study using two types of force measurements. The measurements collected from the force sensor on the BWSS and the force plates within the treadmill will aid in the assessment of validity of the body weight support system to help determine if it maintains the correct percentage of weight being supported.

The vertical ground reaction force should give a more accurate reading to help determine what weight is being displaced from the BWSS. Due to force being measured, the changes in support force are not always parallel to the ground reaction force. Therefore, both measurements are taken from the force sensor of the BWSS and the force plates in the treadmill.

The establishment of the reliability regarding the measurements of the kinetic variables that underlie dynamic transitions from bipedal to single limb stance within a healthy adult population is needed in order to gather reference information and enhance the ability of clinicians to make a more valid inference regarding weight transfer function (Hanke & Rogers, 1992).

No kinetic data have been thoroughly ascertained to help establish the extent to which the actual amount of force produced is proven to be beneficial. Flynn et al (1991) used the Zuni exercise system. This particular system suggested that the vertical ground reaction force was reduced with BWS compared to full weight. This device included a spring that adjusted to the person's vertical oscillations experienced during walking.

The collection of valid force plate data from the treadmill is not always possible because insufficient step length may allow only one foot to have contact with a particular plate in the step cycle (Rose et al, 1991). The step cycle must be continuous requiring bilateral foot strikes for each cycle. Force fluctuations are essentially negligible at low BWS levels but become significant at high body weight levels (Ivanenko et al, 2002). Such items must be considered for proper data processing and interpretation.

The ground reaction force is defined as the force applied upon the foot by the ground during gait. It reflects the acceleration of the body's center of mass during locomotion. This force is three dimensional. It can be broken down into three parts known as vertical, fore-aft, and medial-lateral forces. Of the three parts of ground reaction force, the vertical part represents what is commonly referred to as the weight bearing function of the leg (Cook et al, 1997).

The magnitude of the vertical component includes the force required to continually oppose gravity and the force needed to move the body's center of gravity up and down with each step (Siegler et al, 1982). The vertical ground reaction force varies continually from the instant of initial contact until the foot leaves the supporting surface and has peak magnitudes in excess of a subject's body weight. Vertical ground reaction forces have been found to be reliable and repeatable measures of gait (Kadaba et al, 1989). The ground reaction force helps to measure the amount of force generated throughout the step cycle of gait.

The characteristics of the vertical ground reaction force can be affected by many factors. These factors include body mass and proportions, abilities of coordination and balance, and walking speed. Many studies have shown that the characteristics of ground

reaction force are affected especially by walking speed (Andriacchi et al, 1977 and Gard et al, 2004).

The gait cycle is the basic unit of locomotion. The gait cycle begins when one foot strikes the ground, which is known as initial contact. The cycle ends when this same foot strikes the ground again. The gait cycle is divided into swing phase and stance phase. The stance phase periods are as follows: initial contact, loading response, mid-stance, terminal-stance, and pre-swing. The swing phase periods are initial-swing, mid-swing, and terminal-swing.

Initial contact is defined as the moment when the foot touches the floor. The loading response is the reaction of the limb as it absorbs the impact. The period of single limb support during which the body progresses over a stationary foot is mid-stance. Terminal-stance is the period in the gait cycle in which the body moves ahead of the supporting foot and weight begins to fall on the contralateral limb. The final stance phase period of pre-swing is the transitional period of double support, during which the limb is rapidly unloaded in preparation for swing. Both feet are in contact with the ground and the limbs are alternately loaded and unloaded as body weight is transferred from one leg to the other (Novacheck, 2001).

Initial swing is the point in which the limb is lifted from the floor and initial advancement of the thigh to achieve toe clearance and forward propulsion is assumed. During mid-swing, the limb is advanced further in order to achieve a vertical tibial position. Continued tibial advancement toward full knee extension, deceleration of the thigh, and maintenance of the foot position are included in terminal swing. This completes the full cycle from initial contact to terminal swing.

The highest force produced during walking occurs in double stance phase while the lowest force is generated during single stance phase due to the body's higher center of mass (Herzog et al, 1989). During walking, the body extends over a somewhat stiff stance limb, and the center of mass reaches its highest point at the middle of stance phase. As a result, the gravitational potential energy of the center of mass is maximized while kinetic energy is minimized at the middle of stance phase (Farley and Ferris, 1998).

Between touchdown and mid-stance in gait, the forward velocity of the center of mass decreases as the trunk arcs upward over the stance foot (Lee and Farley, 1998). The kinetic energy of the center of mass reaches its maximum during the second half of stance as it loses the gravitational potential energy. At this point the center of mass moves downwards as the forward velocity of it increases (Lee and Farley, 1998).

Significant decreases in total double limb support time have been observed across BWS trials with the percentage of total double limb support time for the full weight bearing being longer than any values attained with using the BWS (Finch et al, 1991). An increased total double limb support time perhaps results from uneven loading and a greater balance demand (Eke-Okoro and Larsson, 1984). Thus with BWS, the subjects require less balance control to walk supported on a treadmill, as reflected in the reduced total double limb support time (Finch et al, 1991).

The percentage of single limb support time was unaffected by walking speed in the full weight bearing trials but increased slightly across BWS trials in the study performed by Finch (1991). The percentage of stance decreased significantly and was mainly affected by BWS across the different levels (Finch). The speed changes in gait are

produced mainly by changes in stance duration time, which decreases with an increase in speed (Finch).

The decrease observed in percentage of stance and total double limb support time across BWS trials may have resulted from a pure BWS effect or interaction between BWS levels and the set walking speed (Finch et al, 1991). It appeared that the BWS effects on the overall percentage of stance were greater for the percentage of total double limb support time by slightly increasing the percentage of single limb support (Finch et al, 1991).

The results of the combined intervention of increasing BWS and decreasing walking speed suggest that these body weight effects may have been underestimated (Finch et al, 1991). Because the subjects were assigned specific walking speeds per body weight level, the walking speeds may not have corresponded with the amount of BWS being removed for each subject, resulting in a higher than normal variability.

The purpose of this particular study was to determine if the BWSS maintains the correct percentages of body weight being supported and if the vertical ground reaction force was reduced in the process. This system is widely used, thus knowing whether it sustains the precise level of support throughout trials is important for therapists to consider for it to be used most effectively. Factors such as variations in speed levels and percentages of BWS were reported in order to determine their influences with this particular modality.

Methods

Subjects

Sixteen subjects were chosen from Louisiana State University (mean age: 22 and mean weight: 138 lbs) to participate in this study. The study took place in the biomechanics laboratory located within the kinesiology department. The subjects were screened in order to find out certain health information (Appendix D). This included the subjects not having history of surgery or exhibiting any pain within the last six months in their shoulders, back, hips, thighs, knees, ankles, or feet. Informed consent was obtained from the volunteers prior to experimentation. All subjects wore loose fitting clothes such as shorts, t-shirt, and tennis shoes.

Experimental Protocol

All procedures were explained to the subjects prior to experimentation. Each subject walked on a motor driven treadmill (Kistler Gaitway, Amherst, NY) at speeds of 0.447, 0.671, 0.894, and 1.117 meters per second. Body weight support levels of 0%, 15%, 30%, and 45% of subjects' body weight were supported by a modified overhead harness of the body weight support system. There were sixteen trials for each subject. One trial was set for one body weight support level at a particular speed. The orders of BWS levels were randomly chosen and the walking speeds were randomly assigned to each particular body weight support level for each subject.

There were sixteen 20-second trials performed by each subject. This specific amount of time chosen was to allow a sufficient amount of force data to be collected from the force platform as well as the force sensor while the subject walked at the lower speeds on the treadmill. There were approximately ten foot strikes throughout the gait cycles

measured for each trial. The subject walked only for a few seconds on the treadmill at each targeted level in order to get accustomed to the device. Subjects were not allowed to use the treadmill handrails for balance because the use might affect the amount of ground reaction force generated and present unwanted oscillations. Equipment information is presented in the appendices (Appendix A, B, and C).

Body Weight Support System (BWSS)

Each subject was supported in a modified harness over the treadmill (Appendix A). The harness supported the subject primarily about the pelvis and lower abdomen to avoid interfering with lower limb movement. The chest strap of the harness was placed around the subject's torso at the level of the xiphoid process, while leg straps were placed around each upper thigh for additional support. The harness was fastened comfortably enough for the subject in order to maintain a comfort level even when the maximum BWS was being supported. When the subject walked without BWS, he or she still remained secured in the harness.

The BWSS (Vigor, MI, Appendix A) consisted of an overhead frame with a pulley attached. The harness was fastened to the cable of the pulley suspended above the subject. The frame and cable system of the BWSS provided displacement of a targeted amount of weight from the subject using a compressor (Jun-Air, Denmark, Appendix C) associated with the Vigor system. A dial located on the system was used to attain the correct level of support for the support.

The BWSS was adjusted to allow enough vertical movement of the subject's center of gravity so that the subject would maintain a normal gait but not to the point where the subject would lose posture (Wilson et al, 2000). The BWSS data were

collected using a force sensor. The force sensor was attached to the top of the harness to measure the actual amount of force being exerted downward by the body weight support system while the subject walked on the treadmill. This sensor measured the maximum support force (SF_{max}) and the minimum support forces (SF_{min}). The SF_{max} measures the maximum force in each step cycle while the SF_{min} measures the minimum force in each cycle.

The force sensor was calibrated by attaching weights to the overhead support and attaining accurate measurements of force generated downward. After calibrating the body weight system, subjects were then supported at 100% of their body weight (total suspension). A dial located on the BWSS reads the amount of weight being supported. The targeted level of BWS provided during walking was then calculated from this total weight measurement. The dial was set to allow displacement of the targeted level of the subject's body weight at the levels of 0%, 15%, 30%, and 45%. The levels of TBWS were adjusted while the subject was in a static position.

The unloading effects of the BWSS were expressed by maximum vertical ground reaction force data measured by the force platforms within the treadmill. The amount of force was compared using the maximum and minimum forces detected from the force sensor. This was used to determine how the measured maximum and minimum forces from the sensor differed between the measured weight being supported and the targeted levels of body weight support in comparison to the vertical ground reaction force measured from the treadmill.

Kistler Gaitway Instrumented Treadmill

The Kistler Gaitway instrumented treadmill (Type 9810S1x, Appendix B) contains a piezoelectric ground reaction force measurement system found within the device. The force plates, located one in front of the other embedded within the treadmill collected vertical ground reaction force data during walking. Readings of the subjects' weight were also measured with the use of the force plates. The Kistler force platform was connected to a charge amplifier which was linked by means of a digital converter to a personal computer loaded with the Gaitway data collection and analysis software (Version 2.04).

The Gaitway software determined peak vertical forces and the force-time integral for each of the footsteps produced by the subject. It also included a patent algorithm which distinguished left and right foot strikes. The average ground reaction force was measured throughout the gait cycles. This mean vertical component of ground reaction force was used to help validate the targeted level of weight being distributed between the BWSS and force plates. The force platform sensors were statically and dynamically calibrated using weights in order to determine if the same amount of weight was being displaced throughout the plates properly. The same amount of weight should be equal on both plates. Some trials were repeated because of non-continuous force data collection.

Peak force was measured as the highest ground reaction force which occurred anytime during the stance phase and was expressed in percent body weight. No attempt was made to determine whether the peak force occurred in early or late stance since the study was mainly focused on the fluctuations or deviations in the forces. The mean and standard deviation of force data in each trial were calculated.

Although placement of the foot on the force plate was critical for measurement, subjects are often not instructed not to look at the force plate. If the subject was told to place the foot in a certain area on the force plate during the trial, it was known as targeting. Targeting on the force plate was found not to be a significant effect on the variability of the maximum vertical ground reaction force (Grabiner et al, 1995).

To avoid alterations in the gait pattern of the subjects, they were not specifically instructed to step on the force plate. The subject was instructed to keep his or her head up while walking to minimize any tendency to target the force plate. An acceptable trial was found when the entire foot struck the force plate properly, no part of the other foot struck the force plate, and no focus was put on foot placement on the force plate by the subject.

Data Analysis

Resultant forces (SF_{\max} and SF_{\min}) collected from the force sensors of the body weight support system were determined by converting the measures using the formula obtained through the force sensor calibration. The percentage of error from the TBWS was then calculated by dividing the measure by each subject's body weight and then multiplying by 100. This measure was then subtracted from the corresponding TBWS level percentage. The maximum ground reaction force was measured in newtons from the force plates within the treadmill and was then converted into pounds. The percentage of weight was then calculated the same way as the maximum and minimum forces from the sensor as well as the percent error from TBWS.

Statistical analysis of the data was performed using two-factor repeated measures of ANOVA to determine the amount of variance displaying how the different factors involved produce fluctuations in the amount of all forces measured by the subjects. This

included the maximum and minimum support forces from the sensor used with the body weight support system as well as the ground reaction force measured from the force plates within the treadmill. It was used to ascertain whether the amount of forces produced by the subjects were affected by the various levels of speed, the levels of body weight being supported, or both factors.

Results

This results of this experiment established that all of the forces measured were affected by locomotion speed and the targeted percentages of body weight supported (TBWS). The maximum support force, SF_{max} , (Figure 1) increased while the minimum support force, SF_{min} , (Figure 2) decreased as the treadmill speed increased. The SF_{max} was influenced by the variations in speed showing fluctuations ranging from 6.12% to 10.03% while the SF_{min} ranged from 7.06 to 8.58% above the amount of TBWS. As speed increased, the maximum ground reaction force also increased.

As the TBWS increased, the SF_{max} as well as the SF_{min} increased. The SF_{max} was affected by the different TBWS levels and fluctuated from 7.03% to 11.22% above the TBWS while the SF_{min} fluctuated from 0.4% below to 13.05% above the TBWS. As the level of TBWS increased, the maximum ground reaction force decreased. This force deviated from 0.35% to 13.65% above the TBWS.

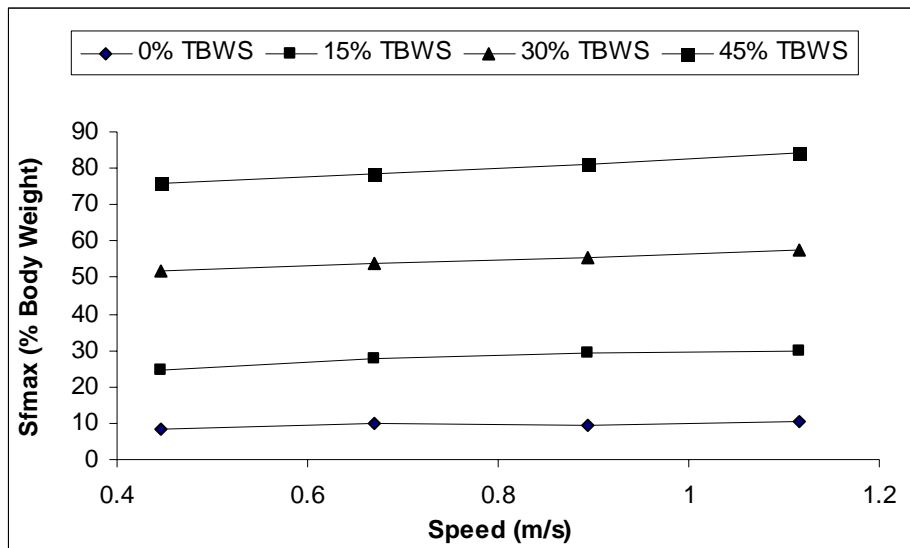


Figure 1. Maximum support force measured from the BWSS. The x-axis represents the speed that the subject walked on the treadmill measured in meters per second. The y-axis represents the force expressed in the percent of subjects' body weight. The symbols represent the mean force at the different targeted BWS levels.

Although there were certain levels of BWS targeted at different speed conditions, the forces exhibited a great deal of fluctuation. These fluctuations can be demonstrated by the change of the SF_{\max} and SF_{\min} . These changes can either be presented as the absolute value of the support forces or the relative value of these forces to its targeted level.

The absolute SF_{\max} and SF_{\min} changed as a result of both speed and the level of TBWS. The maximum support forces (Figure 1) were affected by TBWS levels as well as the walking speed of the subject ($p = 0.0261$). As the subjects' walking speed increased, the SF_{\max} increased. For example, at 15% TBWS, the maximum support force measured 24.33% at 0.447 meters per second and increased to 29.86% body weight support at 1.117 meters per second..

If the TBWS of the subject increased, then the SF_{\max} increased. At the speed level of 1.117 meters per second at 0% TBWS, the maximum support force measured 10.48% BWS and increased to 84.47% BWS at the 45% TBWS. As the level of TBWS increased, the effect of the locomotion speed was more profound. For example, when the speed increased from 0.447 to 1.117 m/s, the SF_{\max} increased 3.91% for the 0% TBWS and 4.19% for 45% TBWS.

The minimum support forces (Figure 2) were also affected by TBWS levels as well as the walking speed of the subject ($p = 0.0132$). As the subjects' walking speed increased, the SF_{\min} resulted with a decrease in force being measured. At 30% TBWS, the maximum support force measured 28.82% at 0.447 meters per second and decreased to 26.52% body weight support at 1.117 meters per second.

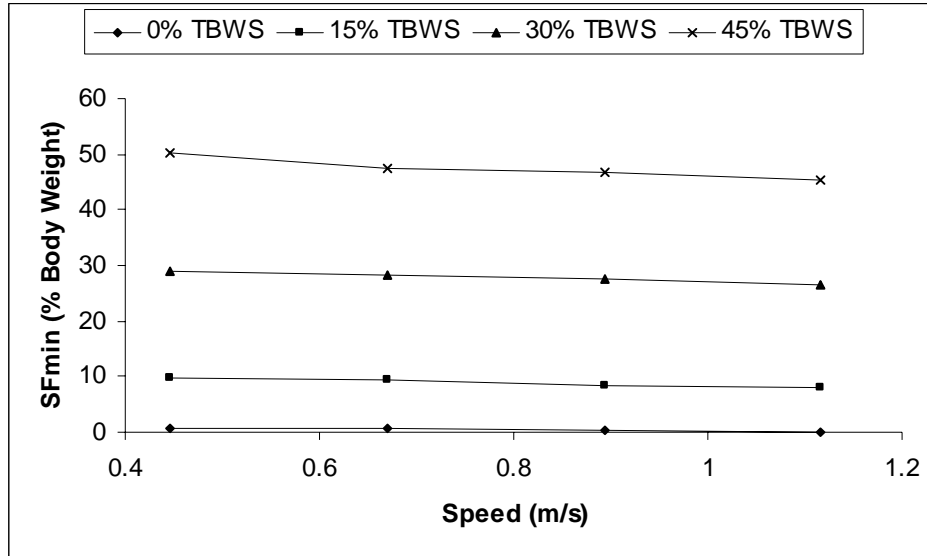


Figure 2. Minimum support force measured from the BWSS. The x-axis represents the speed that the subject walked on the treadmill measured in meters per second. The y-axis represents the force expressed as percent of subjects' body weight. The symbols represent the mean force at the different targeted BWS levels.

If the percentage of TBWS of the subject increased, then the SF_{\min} produced also increased. At the speed level of 0.894 meters per second at 0% TBWS, the maximum support force measured 0.47% BWS and decreased to 46.82% BWS at the 45% TBWS. As the level of TBWS increased, the effects of the various speed levels were more evident. For example, when the levels of speed increased from 0.447 to 1.117 m/s, the minimum support force decreased 0.67% for the 0% TBWS and 4.726% for the 45% TBWS.

The absolute SF_{\max} and SF_{\min} deviated from the TBWS significantly due to the variation in the speed levels ($p < 0.0001$). Deviations are the fluctuations between the targeted body weight support and the measured body weight support from the forces measured.

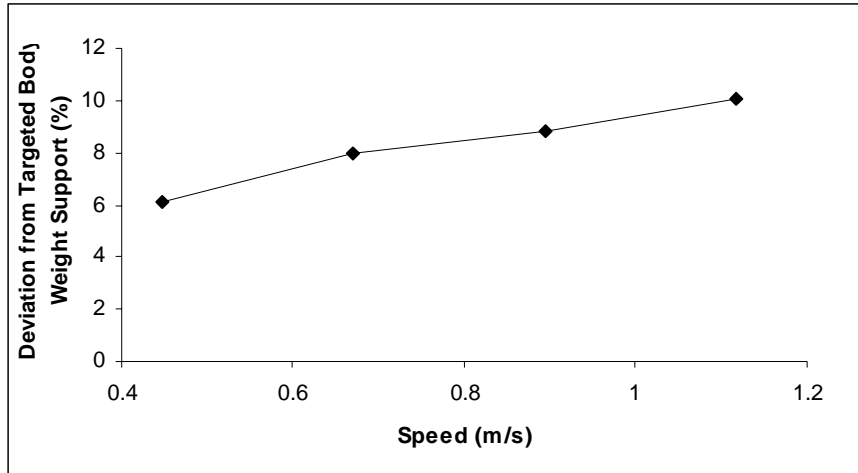


Figure 3. Maximum support force deviation from targeted BWS levels due to speed levels measured in percentages. The x-axis describes the speed of the subject walking on the treadmill measured in meters per second. The y-axis describes the deviations from the targeted body weight measured in percentages. The symbols represent the amount of deviation from BWS levels targeted.

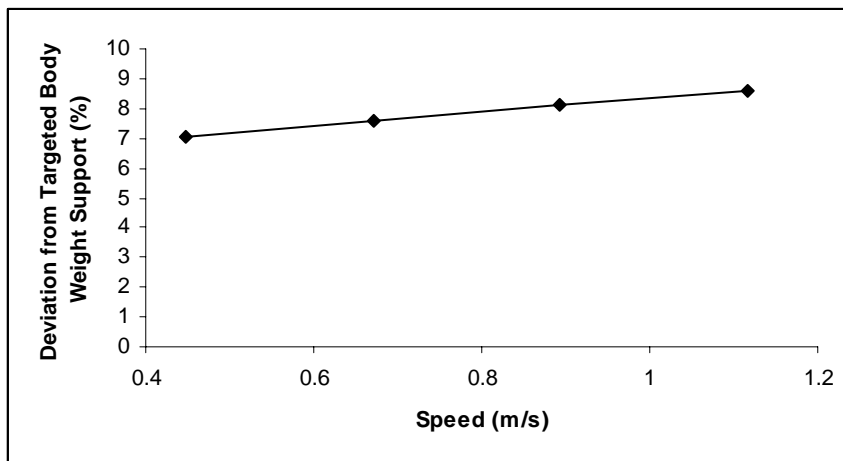


Figure 4. Minimum support force deviation from targeted BWS due to speed levels. The x-axis describes the speed of the subject walking on the treadmill measured in miles per hour. The y-axis describes the deviations from the targeted body weight measured in percentages. The symbols represent the amount of deviation from the targeted body weight support levels.

The maximum support force (Figure 3) exhibited deviations ranging from 6.12% to 10.03% above the TBWS levels while the minimum support force (Figure 4) ranged

from 7.06% to 8.5% above the TBWS levels as a result of the various walking speed levels. As the levels of walking speed increased, the amount of deviation from TBWS for both SF_{\max} and SF_{\min} increased as well.

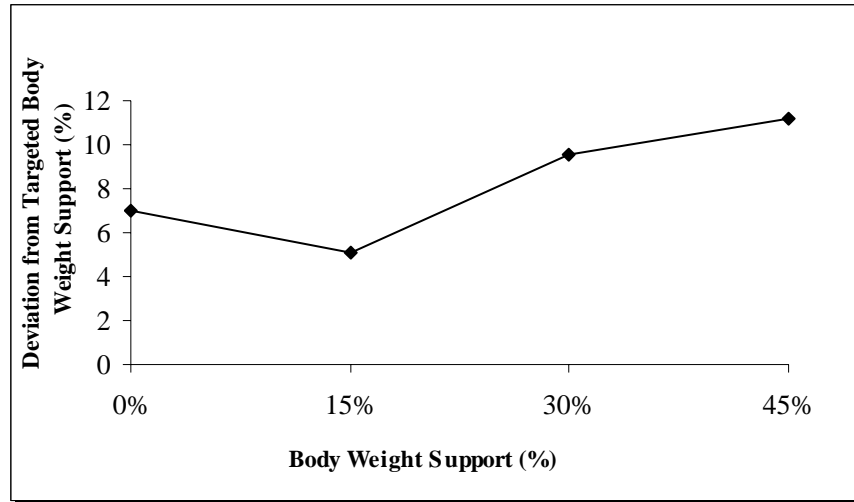


Figure 5. Maximum support force deviation from targeted BWS due to BWS levels measured in percentages. The x-axis describes the different targeted BWS levels in percentages. The y-axis describes the deviations from the targeted body weight support measured in percentages. The symbols represent the amount of deviation from targeted body weight support levels.

The absolute SF_{\max} and SF_{\min} established significant deviations from the TBWS levels (Figures 5 and 6). The maximum support force presented deviations fluctuating from 5.08% to 11.22% above the TBWS in correlation to the relative BWS. The minimum support force deviated from a range of 0.40% below to 13.05% above the TBWS levels. As the levels of BWS increased, the result was an increase in the amount of deviation from these targeted percentages.

Maximum ground reaction force generated by the subjects was proven to be significantly affected by the targeted body weight support levels TBWS. As the subjects' walking speed increased, the maximum ground reaction force increased. The mean force

at 0.447 meters per second measured 498.98 newtons and increased to 525.7 newtons at 1.117 meters per second throughout the trials.

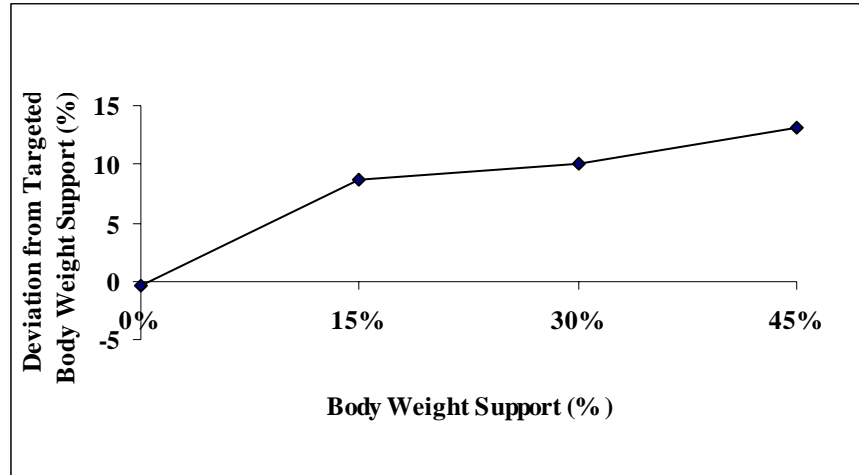


Figure 6. Minimum support force deviation from targeted BWS. The x-axis describes the different TBWS levels in percentages. The y-axis describes the deviations from the targeted body weight measured in percentages. The symbols represent the amount of deviation from the targeted body weight support level.

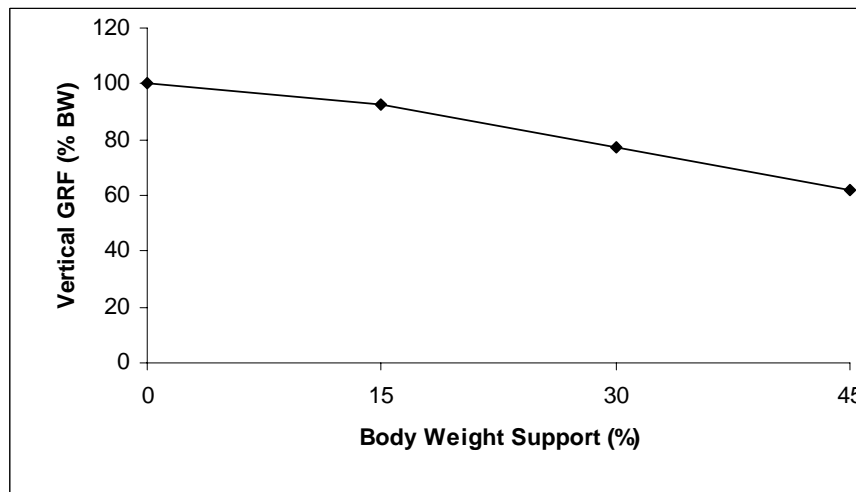


Figure 7. Maximum ground reaction force affected by targeted body weight support. The x-axis describes targeted body weight support levels. The y-axis describes the vertical ground reaction force measured in percent body weight. The symbols represent the mean force measured.

The maximum ground reaction force was significantly affected by the targeted body weight support levels ($p < 0.001$). The maximum ground reaction force (Figure 7) decreased with increased BWS. The average force measured corresponding to the effects of body weight support began at 622.43 newtons at 0% targeted body weight support and decreased to 381.14 newtons at the 45% targeted body weight support.

The maximum ground reaction force (Figure 8) presented deviations fluctuating from 0.35% to 13.65% above the TBWS in correlation to the relative BWS. If the body weight support system is maintaining 45% of the subjects' weight, then the resultant force measured by the force plates in the treadmill should result as accepting 55% of the body weight. The measurement was actually 61.77% which would lead to a difference of 13.65% above the weight.

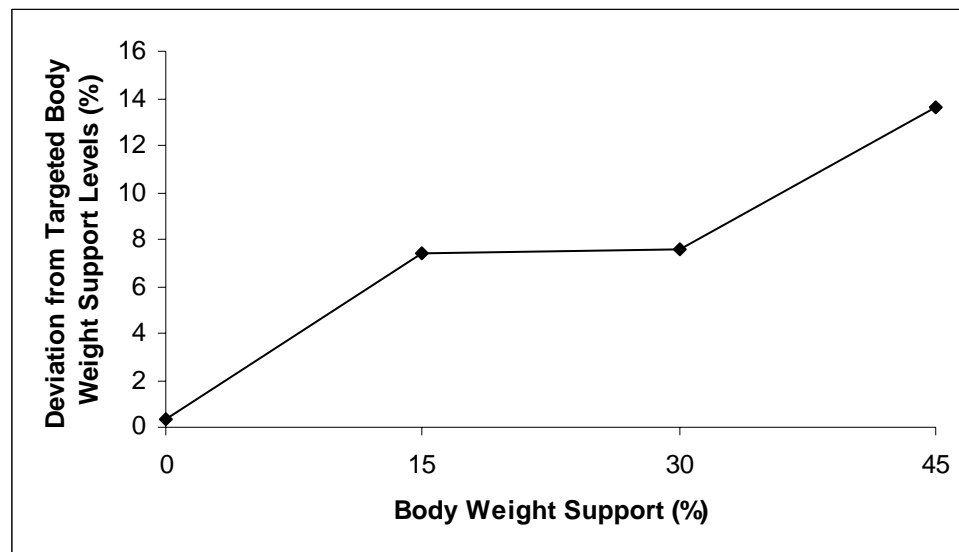


Figure 8. Maximum ground reaction force affected by BWS levels. The x-axis describes the targeted levels of body weight support. The y-axis describes deviations from the targeted body weight measured in percentages. The symbols represent the mean force measured.

Discussion

The purpose of this study was to report on the accuracy of precise measurements of the body weight support system. The results of this study support the idea that the body weight system does not maintain the targeted body weight support (TBWS). All of the forces measured were affected by levels of BWS and speed. Both the maximum as well as the minimum support forces conferred that the correct percentage was not maintained with values measuring above the targeted level of BWS.

The effects of the levels of speed varied between the maximum and minimum support forces. As the amount of speed increased, the SF_{\max} increased while the SF_{\min} decreased. More force is measured in the maximum support force due to the greater force produced in double stance phase. Less force is measured in the single stance phase. The effect of BWS levels on the percentage of weight being supported varied from 5.08% to 11.22% above TBWS for maximum support force while minimum support force ranged from 0.40% below to 13.05% above TBWS. The maximum ground reaction force ranged from 0.35 to 13.65% above the TBWS.

The body support system did reduce the vertical ground reaction force. These findings are in agreement with the results of Flynn (1991). In the current study, an increase in BWS resulted in a decrease in ground reaction force. Less force was measured due to an increase in the amount of BWS. When only a small percentage of weight was being supported, more force was exerted upon the ground by the subject.

The increase in speed levels produced a higher measurement of ground reaction force which explains why speed was one of the most important determinants of ground reaction force as suggested by Andriacchi (1977). This was probably due to the vertical

excursion of the body's center of mass that is found to increase with walking speed (Gard et al, 2004). The center of mass and gravity were changing throughout the trials due to the speed and levels of body weight support. This particular point is not accounted for in the measurement of targeted levels of support since the subject is measured statically and the center of mass does not move. Once the subjects begin to walk this factor becomes relevant to fluctuations in force measured. Further kinematic data collection could be used to aid in this assessment. This kinematic data would obtain information involving the center of mass and center of pressure factors affecting the forces collected.

The results suggest that the BWSS and Kistler force platform display validity in their measurements. The amount of vertical ground reaction force measured was similar to the amount of subject's body weight force that was not supported by the body weight system. The amount of force measured from the plates within the treadmill should reflect the amount of weight not being supported by the BWSS. The data collected were seemingly close but not exact. The amount of reduction was disproportional to the targeted body weight support levels. At the 15% targeted body weight support level, the mean force collected from the treadmill plates was actually 92.39%. It should have measured only 85% of the subjects' weight but this shows that the BWSS was under the targeted body weight support level by 7.39%.

The fact that the BWSS does not maintain the targeted percentage of body weight of the subject throughout all of the trials at the different levels of BWS as well as speed leads one to wonder why this modality is so widely used effectively. The protocols used by therapists based on unreliable or not exact measurements may not focus on the proper dysfunction of the patient and lessen therapeutic effectiveness (Hanke and Rogers, 1992).

The lack of precise BWS levels is most important during the single limb support of stance phase. The single support phase is the important focus in therapy. This phase is important because it involves the patient's need for the most support. The amount of force placed on the injured limb is greatest at this point (Herzog 1989). A greater percentage of BWS is needed in order for a patient's healing tissues to have a better chance of withstanding the amount of weight being placed upon them. Any additional weight placed on injured muscles or tissues will impede on the correct healing process for that patient as well as the progress of the treatment designed by the therapist. If the correct percentage is maintained throughout this fragile state, progress may increase more quickly and effectively. As the patient increases his or her abilities, the amount of variance in the percentages may not affect the patient as much as being a hindrance in treatment.

In the study by Murray (1993), harness supported treadmill ambulation was suggested to decrease an individual's body weight by a predetermined amount using a harness and counterbalance system to accommodate the rise and fall of the body during treadmill ambulation. The idea of the BWSS was to attempt to provide consistent BWS in the study by Miller (2002). Contrary to these two ideas about the BWSS, the results of this study suggest otherwise. All forces measured exhibited deviations from the targeted body weight support levels due to speed and BWS levels. Fluctuations in the measured forces indicate that this system does not accommodate for the rise and fall of the body and the percentage of BWS are not consistent during walking on the treadmill.

Conclusion

This study was designed to further investigate to what extent of support is physically done by this particular body weight support system. The study mainly focused on the capability of this system to constantly maintain the proper targeted body weight support level and to understand the idea of unloading and vertical ground reaction force. The BWSS appears to be a valid instrument for predictably reducing the vertical ground reaction force during walking.

Results provide evidence to support the idea that this system does not maintain the precise targeted body weight support level. As speed increased, the maximum support force (SF_{max}) increased while the minimum support force (SF_{min}) decreased. As the percent of TBWS increased, the SF_{max} as well as the SF_{min} increased. The measured support forces (SF_{md}) affected by the different body weight support levels fluctuated from 5.08% to 11.22% below the TBWS. The SF_{md} fluctuated from 6.11% to 10.03% below the targeted body weight support. The fluctuation was affected by walking speed. As speed increased, maximum ground reaction force also increased. As the percentage of BWS increased, the maximum ground reaction force decreased.

Maintaining the precise percentage of BWS is an important idea that needs to be addressed. A device needs to be manufactured in order to control or provide the correct amount of weight displacement of the targeted amount of BWS throughout all BWS and speed levels. Because the subjects' percentages of body weight are measured statically, there is no account for the change in the body's center of mass as the subject is walking from the BWSS. More research is necessary to further our understanding of the use of BWS systems in the rehabilitative process of patients.

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Appendix A

Vigor BWSS



www.vigorequipment.com

Vigor Neuro I

- Designed for low maintenance operations.
- Free standing closed metal frame (may be mounted to the floor).
- Pneumatic assist up to 255 lbs.
- Easy to adjust fall guards.
- Units can be customized to any ceiling height.
- Floor space minimum requirement: 4'Wx8'L (can be adapted to wider treadmills if necessary).

Appendix B

Kistler Gaitway
Instrumented Treadmill



Appendix C

JUN-AIR Compressor



JUN-AIR compressors are built as complete compressed air units with thermal cut-out device, adjustable pressure switch, safety valve, gauge and manual drain cock on the air receiver - and are easily installed.

- exceptionally quiet
- effective and trouble free
- original
- functional and aesthetic
- handy and easily portable
- compact and rugged
- easy to operate/minimum maintenance
- anti corrosion
- treated low in electrical consumption
- 2 years' warranty

Specifications

Power supply	100/120 volts
Frequency	60 Hz
Motor HP	0.45
Motor kW	0.40
Max current	6.2 amps
Displacement	2.12 CFM
Max pressure	120 psi
Tank size	6.6 gallon
Machine height	21 5/8"
Machine width	15"
Machine depth	15"
Weight	64 lbs
Noise level	45 dB(A)1m
Pumping time	270 sec 0-120 psi

Appendix D

Consent Form

Study Title: Differences between Targeted and Measured Body Weight Support with the Usage of a Body Weight Support System

Performance Site: The Motor Behavior Lab of the Louisiana State University
Kinesiology Department

Investigator: The following investigators are available for questions about this study, M-F 8 a.m.-5 p.m.

Robin Cropper, B.S. (337) 351-1584
Li Li, Ph.D (225) 578-9146

Purpose of the Study: The purpose of this research study is to determine how unloading the body using a harness system effects how we walk.

Subject Inclusion: Subjects will be normal, healthy students between the ages of 18 and 40 who have no history of surgery or pain within the last six months to their shoulders, back, hips, thighs, knees, ankles or feet.

Number of Subjects: 16

Study Procedures: Testing will be completed in one session lasting approximately one hour. Prior to beginning testing each subject will receive training instructions. You will be asked to walk on a treadmill at various speeds wearing a walking harness that will support your body at certain percentages of your body weight. Force plate data will be taken while you are walking.

Benefits: The study may yield valuable information regarding the effect of BWS on walking. It is possible that this information may be useful in treating individuals who have difficulty walking.

Risks: There is a very slight possibility that you may lose your balance while walking on the treadmill.

Measures Taken to Reduce Risk:

You will be wearing a harness that is attached to very sturdy apparatus. The harness and the apparatus system are designed for individuals to walk on a treadmill. If you happened to stumble, the harness would support you and you would not fall. The investigator will ensure that the harness is attached correctly to each subject and is very aware of maintaining the safety of individuals while they are walking under their supervision.

Right to Refuse:

Participation in this study is voluntary. You may choose not to participate without penalty or loss of any benefit to which you might otherwise be entitled.

Privacy:

This study is confidential. Results of the study may be published, but no names or identifying information will be included in the publication. Subject identity will remain confidential unless law requires disclosure.

Financial Information: There is no financial compensation for participating in this study.

Withdrawal:

You may withdraw from the study at any time without penalty or loss of any benefit to which you might otherwise be entitled.

Removal:

The investigator may remove you from the study in the unlikely event of any unforeseen safety issues.

Signatures:

This study has been discussed with me and all my questions have been answered. I may direct additional questions regarding study specifics to the investigators. If I have questions about subjects' rights or other concerns, I can contact Robert C. Mathews, Chairman, LSU Institutional Review Board, (225) 578-8692. I agree to participate in the study described above and acknowledge the researchers' obligation to provide me with a copy of this consent form if signed by me.

Subject Signature _____

Date _____

Vita

Robin Cropper is currently enrolled as a master's student at Louisiana State University. She graduated from Port Barre High School located in Port Barre, Louisiana, in 1998. Cropper then attended college at Louisiana State University at Eunice located in Eunice, Louisiana, where she majored in pre-physical therapy. After two years, she then attended Louisiana State University. Cropper majored in kinesiology and graduated with a Bachelor of Science degree from the university in May of 2002. After receiving her Master of Science degree, she plans to attend physical therapy school.