REHAB Tech

Monash Rehabilitation Technology Research Unit

Effects of harness supported walking on able-bodied subjects and lower limb amputees.

Honours Project

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CHAPTER ONE

INTRODUCTION

Rehabilitation of the lower limb amputee (LLA) must focus on restoring the amputee to an overall functional level that is as full as possible considering their environment and physical condition (Vitali, et al., 1986). It is widely accepted that early mobilisation and weight bearing will improve post-operative results (Siddiqi, et al., 1994). Advantages include reduced oedema and pain, fewer problems with joint contractures and muscle condition, and profound psychological benefits. Studies indicate that early gait re-education will also reduce hospitalisation and disability time (Gerhardt, et al., 1982).

However, for early mobilisation and gait training to be successful, it is critical to control the amount of weight bearing and to support the patient in an upright position (Pillar, et al., 1991). Factors that often limit or delay this process are pain, vulnerable stump tissues, residual skin tolerance, and the patient's confidence in transferring weight onto the prosthetic limb (Hunter, et al., 1995). So as to avoid a delayed rehabilitation and thus to achieve the benefits of early mobilisation, partial weight relief has traditionally been provided through the use of parallel bars and assistive ambulation devices.

The use of parallel bars and assistive devices, however, is associated with a number of disadvantages (Dickstein, et al., 1992; Pillar, et al., 1991; Hunter, et al., 1995; Visintin & Barbeau, 1994; Harburn, et al., 1993). Firstly, ambulation requires additional energy (McBeath, et al., 1974; Pagliarulo, et al., 1979). Secondly, the amount of weight relief provided cannot be adequately controlled or measured. Thirdly, the use of such aids necessitates a modestly high level of upper body strength and control (and at least one intact upper extremity). Fourthly, an asymmetric gait pattern tends to result, the patient strongly favouring the uninvolved lower limb. Finally, although patient safety within these devices is greater than unsupported walking, there remains a real risk of falling. Consequently, the clinician must spend a considerable amount of time simply ensuring patient safety.

A possible solution to the above mentioned problems encountered in traditional gait re-education may be an external system that relieves a portion of the patient's body weight. Such a system has been designed and constructed at Rehab Tech, Melbourne, with the aim of assisting patients during gait rehabilitation. The system comprises an overhead track mounted to the ceiling, within which a motorised carriage runs (Searle, 1996). Connected to this carriage is a pneumatic actuator, which is in turn attached to a body jacket harness via a rope. Through wearing the harness, a patient is provided with an adjustable, controlled and measurable amount of weight relief. The amount of weight relief is kept constant throughout the gait cycle, despite the vertical position of the body's centre of gravity, due to the action of the pneumatic actuator. Carriage movement, controlled by feedback from the angle of the rope, responds to the periods of walking. The system allows unrestricted walking along a 15 metre walkway while providing controlled support and weight relief.

Using a device such as this may allow correct and entire walking movements at an early stage by stabilising the trunk and enabling controlled weight transfer and loading of the affected limb. Among the potential benefits are:

a. safety -- the user should be absolutely confident that they cannot fall, allowing them to concentrate on re-learning to walk and increasing confidence;
b. the patient is able to learn to walk relatively unencumbered, free from the constraints of parallel bars and assistive devices;
c. a percentage of the patient's body weight can be relieved during early gait rehabilitation when pain and tissue breakdown may otherwise prevent mobilisation -- as pain and tissue tolerance increase, the amount of weight bearing can be progressively increased;
d. the clinician and patient may direct more attention to walking technique rather than being concerned about safety and balance;

e. the harness support system encourages a normal standing posture with correct alignment of the shoulders over the hips, whereas assistive devices encourage a slight forward lean that may contribute to compensatory movements at other joints; and,
f. training gait under such conditions may lead to a more straightforward transition to unassisted overground walking.

It may be postulated that these advantages would lead to a more expedient rehabilitation process, bringing with this the associated physical, social and economical benefits.

CHAPTER TWO

LITERATURE REVIEW

The concept of relieving a portion of body weight during locomotor rehabilitation is not a recent discovery. Hydrotherapy and assistive ambulation devices have been used for many years to unload the affected lower extremity(s) and provide support and stability (Harrison, et al., 1992). However, only recently have overhead harness support systems been utilised to accomplish these goals. The use of overhead harness support systems in rehabilitating the gait of spinal cord injured and stroke (hemiplegic) patient's has been relatively well reported.

2.1 SPINAL CORD INJURED PATIENTS

Finch and Barbeau (1985) proposed that the relief of body weight may "facilitate the expression of gait patterns and therefore could be considered a therapeutic tool in gait retraining of patients who have neurological impairments." From this proposal a number of studies have eventuated, investigating the effects of harness support systems on the gait of neurological patients.

Wainberg and Barbeau (1985) studied spastic paretic patients walking on a treadmill at self selected velocities during full weight bearing (FWB) and with 20% and 40% body weight support (BWS). Their results indicate that with increasing BWS, the patients were able to attain higher comfortable walking speeds. Associated with the increase in speed were a decrease in cycle duration and stance duration ($p<0.01$), and an increase in stride length. The authors also observed a decreased duration of the proximal extensor muscle activity of the lower limb. Similarly, Barbeau, et al. (1987), reported increases in speed in seven spinal cord injured (SCI) patients with increased BWS. Walking velocity increased 15-38% at 20% BWS, and 10-77% at 40% BWS. Two patients, previously wheelchair bound, were able to walk with BWS provided by the harness system. The authors commented that the advantages of using a harness support system to retrain gait were, the ability to train simultaneously the three components of locomotion: posture, balance and stepping.

Visintin and Barbeau (1989) investigated the effects of supporting a portion of body weight on the gait pattern of seven spastic paretic patients during treadmill ambulation. An overhead harness system provided the patients with 0% and 40% BWS at their maximum comfortable walking speed. The global effects seen with 40% BWS were a general decrease in electromyographic (EMG) mean burst amplitude for the lower limb muscles, with instances of more appropriate EMG timing. These results were supported by Dietz, et al., (1994) who studied EMG activity of the leg muscles in nine paraplegics using BWS and found similar decreases in EMG activity. Joint angular displacement patterns were also measured by Visintin and Barbeau (1989), BWS producing a straighter trunk and knee alignment during the weight bearing phase. An increase in single limb support time and a decrease in percentage total double support time were evident at 40% BWS. Increases in stride length and velocity were also seen, confirming the findings of previous authors. The patients in this study reported easier and less fatiguing walking as evidenced by the fact that their endurance increased by up to twice as long. Barbeau, et al., (1993) found very similar improvements in a single case study of one chronic spastic paretic subject. Danakas, et al., (1991) demonstrated that the improvements found in kinematic and EMG patterns at 40% BWS, by Visintin and Barbeau (1989), could be retained and carried over to full weight-bearing conditions.

The effects of a combined drug therapy and intensive harness supported treadmill training program on two wheelchair bound SCI patients were reported by Fung, et al., (1990). Both patients were provided with at least 50% BWS initially, which was reduced to 0% BWS over the training period, after which both patients were able to ambulate overground with the assistance of crutches. Norman, et al., (1995) describe a similar finding: two

SCI patients previously unable to walk were able to ambulate effectively using the BWS system, their speed of locomotion increasing as the treatment progressed.

Barbeau and Blunt (1991) and Visintin and Barbeau (1994) studied eight spastic paretic subjects at 0% and 40% BWS during treadmill locomotion. These authors were interested in the effects of BWS and parallel bars on gait. With 40% BWS both studies reported improvements in EMG and kinematic patterns that were consistent with other researchers' results. However, when the parallel bars were removed at 0% and 40% BWS, further improvements were seen. Even though the gait was laborious and difficult without parallel bars at 0% BWS, some improvements were noted. Visintin and Barbeau (1994) reported that removing the parallel bars at 40% BWS with asymmetrically involved SCI patients produced a more normal swing phase. The hip, knee and ankle flexion that were present when walking without the parallel bars were not evident under conditions where the subject was able to compensate by using the parallel bars. Barbeau and Blunt (1991) continued on to study two spastic paretic patients, progressively reducing the amount of BWS provided. Improvements in the kinematic pattern were observed, producing a smoother more symmetrical gait pattern, and allowing the patients to ambulate overground following the training program.

Progressive BWS rehabilitation programs using eight spastic paretic patients have been shown to improve functional muscle ratings and to produce phasic muscle activity and stepping movements in paralysed limbs (Wernig and Muller, 1991 & 1992; Dobkin, et al., 1992). Significant increases in speed and duration of treadmill locomotion, and improvements in EMG and kinematic patterns were also observed. Using 153 SCI patients, conventional gait training was compared to treadmill training under a harness support system. The treadmill training program produced superior results in locomotor capability upon completion of the treatment (Wernig, et al., 1995).

2.2 STROKE PATIENTS

Similar results to those reported above have been found in hemiplegic populations. Visintin, et al., (1988) reported results of a single case study on a chronic hemiplegic patient trained with BWS reduced from 60% to 20% as kinematic and EMG patterns improved. Increases in gait velocity and near normal EMG patterns were evident toward the end of the program. Hesse, et al., (1994 & 1995) reported significant ($p<0.01$) increases in velocity, cadence and stride length, plus 24% improved stance symmetry and 123% greater swing symmetry. The authors observed that vertical force diagrams indicated better symmetry of loading and less variability with BWS. However, these diagrams were not published in the article. Improvements in functional ambulation category were also reported. Increases in velocity and duration of treadmill walking were also confirmed by Malouin, et al., (1992) using an intensive treatment program that incorporated harness supported treadmill training.

Pillar, et al., (1991) investigated the effects of a harness support system on hemiplegic and able-bodied (AB) gait. Velocity, stance and swing duration's were recorded, along with subjective evaluations by the subjects and the investigators. Subjects were tested during free ambulation on a walkway and then harnessed to the system with no weight relief and with 20% weight relief. Unlike all of the previous studies discussed, the overhead harness support system in this case was located above a walkway, not over a treadmill. A motorised carriage followed the subject while they traversed the walkway. Pillar, et al., (1991) observed increases in walking velocity in the patient group in both harnessed conditions, larger effects seen at 20% relief. The ratio between duration of stance on the unaffected lower limb to the affected lower limb decreased, while the corresponding ratio for swing increased. These changes occurred even when the patient was harnessed with no weight relief, and became more pronounced with weight relief. The result of these changes was that the patient was spending relatively more time weight-bearing on the affected lower extremity; stance and swing symmetry more closely approximated normal values. Both the patients and the investigators felt that the system facilitated gait.

It appears that harness supported rehabilitation produces significant improvements in the gait patterns of neurological patients. It is difficult to isolate the effects of the treadmill, BWS (harness) and the manual assistance given to stepping in many of these studies. Presumably it is a combination of these influences that produced improvements in kinematics, EMG activity and temporo-spatial parameters. Although lower limb amputation is very different from a neurological impairment, some of the same disabilities arise, including the inability to cope with weight bearing and reduced balance. It may therefore be hypothesised that some of the advantages seen in this group of patients will be valid for the lower limb amputee.

2.3 ALTERNATIVE SYSTEMS

A variety of alternative weight relieving systems and their effects on various patient groups have been discussed in the literature. The REHABOT and AID-1 systems consist of a harness suspended from a rotating arm that is attached to a central shaft. These systems provide controlled and prescribed weight relief while the patient walks 360° around the central shaft, optionally holding onto the handrail. Such devices have been used extensively in Japan for rehabilitating gait in orthopaedic and central nervous system disorders. Qualitative evidence lends support for their value in rehabilitating gait of two lower limb fracture patients (Siddiqi, et al., 1994) and patients with various other locomotor disabilities (Kawamura, et al., 1993; Ide, et al., 1993). Dickstein, et al., (1992) describe a novel walking aid, the weight relieving walker, which partially relieves the body weight of an ambulating patient. This device incorporates an ischial bar that provides the weight relief and a forward propulsion mechanism. Data from twenty six patients of varying pathologies indicate that patients were able to walk 6-7 times faster and much farther with the aid of the system. The ZUNI™ (SOMA, Inc., Austin, TX), another example of a weight relieving system, mechanically offsets the patient's weight in 1 kg increments to provide weight relief. This device, incorporating a harness to support the patient, must be used over a treadmill for gait training. Kelsey and Tyson (1994) provide case study evidence for its use in rehabilitating two professional athletes following a lower limb fracture and tendon rupture, respectively.

Kline (1994) recognised the problems with conventional exercise therapy for patients who suffered painful osteoarthritis (OA) of the knee. She proposed an alternative approach using the ZUNI™ to unload the lower limbs during treadmill training. The purpose of her investigation was to determine the effects on heart rate (HR), oxygen consumption (VO_2) and pain perception. Twenty seven OA patients were tested at 0%, 20% and 40% BWS. The results of the study confirmed the hypotheses that increased BWS results in decreased HR and VO_2. Although non-significant, pain responses at 0% BWS were found to increase as the level of exercise increased, this effect was not as large, nor did it occur as soon with 20% and 40% BWS. Therefore, one may interpret these findings to conclude that unloading does reduce perceived pain.

2.4 ABLE-BODIED SUBJECTS

Several authors have investigated the effects of harness support systems on AB gait. Pillar, et al., (1991) noted the restriction that the harness system has on the acceleration of the body's centre of gravity, resulting in reduced velocity while wearing the harness. This effect was more pronounced with weight relief. Stance and swing duration's remained unchanged in this study. With 40% BWS, Barbeau and Blunt (1991) observed decreased flexion of the trunk, hip and knee and decreased maximum extension of the hip. Other effects noted were increased dorsiflexion at terminal swing and at heel contact. Decreased EMG amplitude of the lower limb muscles was also recorded with 40% BWS.

Finch and Barbeau (1985) reported decreases in speed with increasing BWS (up to 70%). At 50% and 70% BWS kinematic data from four AB subjects indicated that gait cycle time increased by at least 10%; percentage stance duration decreased by 5% while percentage swing duration increased by 5%. These results were supported by Finch, et al., (1991). Additional results reported by Finch were a decrease in total double limb support time with increasing BWS, plus an increase in percentage single limb support time. Other effects observed with increasing BWS were a reduction in mean burst amplitude of the muscles required for weight acceptance and for push off, plus an increase in the mean burst amplitude of tibialis anterior (active during the swing phase).

Hewes, et al., (1967) replicated lunar gravity and studied the kinematics of AB walking and running. This may be viewed as an extreme case of BWS. The outcome was that subjects walked and ran 60% slower than normal, and exhibited decreased hip, knee and ankle angular movements.

From the results reported here, it would appear that harness supported walking with BWS has an adverse effect on AB gait. This may be due to the physical restriction placed on the body by the harness or due to the reduced potential for kinetic energy generation at the push-off phase of the gait cycle. The decreased total double limb support time and increased single limb support time could have implications for balance training. During the BWS conditions, subjects were forced to support their body weight (albeit less body weight) on a single limb for longer periods of time.

Overhead harness systems have also been designed simply for the purposes of assessing standing balance. Examples of this are reported by Harburn, et al., (1993) and Hill, et al., (1994).

2.5 LOWER LIMB AMPUTEES

Very little research has been conducted using LLAs and harness support systems. Adler, et al., (1987) described the results of a treadmill training program for a bilateral transtibial amputee. The amputee wore an upper body harness for "physical and psychologic support" while training on the treadmill. However, it was not the purpose of the study to investigate the effects of the harness. Similarly, Palma and Hoyle (1992) described a harness support system that was suspended over a treadmill in order to aid rehabilitation of LLAs. They commented that using the harness system allows a physical therapist to work on weight shifting, balance, cadence and speed with reduced risk or fear of falling.

Hunter, et al., (1995a & b) investigated the energy expenditure of transtibial amputees during harness supported treadmill ambulation. Seven healthy transtibial amputees and ten healthy matched AB volunteers participated. Subjects were tested on a treadmill at two speeds ($0.67ms^{-1}$ and $1.34ms^{-1}$) at FWB, 20% BWS and 40% BWS. The harness support system used was the ZUNI™ Incremental Weight bearing System. During the last minute of each trial, rate of perceived exertion, HR, and VO_2 measures were collected. Caloric expenditure (kl/min) was calculated using metabolic conversion equations. As expected, both groups demonstrated significantly lower HR's,

VO_2's and kl/min at 1.34 ms^{-1} when under 40% BWS. Although many other results were not found to be significant, the trends appeared to be reliable indicators; increasing BWS resulted in decreased HR, VO_2 and kl/min at both speeds in both groups.

The results of the above study indicate that harness supported treadmill ambulation can allow a patient who requires a partial weight bearing status to exercise weight bearing muscles, bone and connective tissue at a lower energy cost. This minimises the negative physiological effects of disuse and immobilisation, and it allows rehabilitation to commence earlier. The authors of the study state reasons for using this system in amputee rehabilitation:

a. to reduce weight bearing on new residual limbs;
b. for safety assurance of the patient and therapist;
c. for balance assistance during gait training;
d. to eliminate the need for assistive devices; and,
e. for pain relief.

Although the theory behind the use of harness support devices for rehabilitation of LLAs appears to be sound, few studies have been published. The potential benefits should be enough to provoke further research in this area. It is obvious that harness support systems have proved very valuable in other patient populations, with beneficial results being universally reported. There is some preliminary evidence that amputees will benefit from the use of harness support during gait rehabilitation. It is the purpose of the present investigation to explore more thoroughly the precise effects of a harness support system on certain LLA and AB gait parameters.

CHAPTER THREE

DETAILS OF THE HARNESS SYSTEM

3.1 TECHNICAL DETAILS

An overhead harness system was designed and constructed at Rehab Tech, Melbourne, to assist gait rehabilitation of lower limb amputees and other patients with locomotor disabilities (Appendix A). The overhead harness is used in conjunction with an established walkway, unlike most previous systems that utilise a treadmill. Influencing the decision to construct a harness system over a walkway rather than a treadmill was the fact that walkway ambulation more closely simulates normal ambulation. A number of authors have reported differences between treadmill and overground walking (Arsenault, et al., 1986; Charteris & Taves, 1978; Hwang, et al., 1994; Murray, et al., 1985; Strathy, et al., 1983; Wall & Charteris, 1980). Another influencing factor was that using a walkway allowed subjects to pass over an existing force platform.

The construction of the harness system can be considered under the following areas (Searle, 1996).

3.11 Rail

After considering the available rail options, a hollow profile rail design was chosen. This form of rail can withstand loads up to 650 kg and it is compatible with commercially available carriages.

3.12 Motorisation

A DC permanent magnet electric motor was chosen for its ability to respond rapidly to changes in velocity. The motor chosen allows a maximum velocity of 1.8 ms^{-1} (deemed fast enough for rehabilitation purposes).

3.13 Weight Relief

Supporting the user's weight is a critical factor in the design of such a weight relieving system. The method of weight relief must be comfortable, unrestrictive, and sufficiently strong. A harness was constructed to meet these criteria. The final design consists of an adjustable canvas body jacket that encircles the thorax. Weight bearing is distributed over the total contact area of the jacket, from the pelvis to the shoulders. The two primary weight bearing areas are provided by:

a. an adjustable, padded waist belt that encircles the pelvis between the greater trochanters and the iliac crests; and
b. padded supports in the axilla region.

The harness is connected to a crossbar above the user's head by D-rings on both shoulders, this crossbar being in turn connected to the carriage via a rope. The harness is easily and quickly donned and doffed. Two adjustable harnesses cover a broad spectrum of body sizes.

The decision to employ a body jacket harness was made in order to maximise weight bearing area and hence comfort, while minimising restriction of the lower limbs, pelvis and upper limbs as much as possible. Ratliff, et al., (1993) suggest a harness that distributes the lifting force over a large body surface area to reduce cardiopulmonary compromise and to lessen discomfort. Parachute or mountaineering harnesses, or variants of these, have been utilised in many of the previous studies. However, it was felt that the crotch loops in these harnesses would restrict lower limb movement and provide a possible source of discomfort. It has also been suggested that these harness types encourage a flexion and abduction position of the hips and forward tilt of the trunk (Norman, et al., 1995; Barbeau, et al., 1987). If the current harness is likened to a lumbosacral orthosis it is reasonable to expect that it may restrict the extremes of hip flexion and extension, and trunk motion (Lantz and Schultz, 1986). However, because the subjects for whom it is intended are generally not capable of walking at fast speeds with long strides, restriction of the extremes of hip and trunk excursion is usually of no practical significance (Norman, et al., 1995).

Ralston (1965) investigated the effects of torso immobilisation on energy cost when using a body jacket cast. Increased energy expenditure by about 10% was seen over a wide range of speeds. Presumably, the restriction provided by the harness used in the present study would be far less than the restriction of a body cast and hence any increases in energy expenditure should be less than 10%.

Weight relief of the user could be provided a number of ways. Most previous systems have simply employed static bearing. This method of suspension, although simple and cost effective, does not allow for vertical movement of the body's centre of gravity during the gait cycle. Spring suspension, another available method, allows vertical movement of the body. However, a constant upward force cannot be maintained due to the changing force provided by the spring at different positions of its stretch. Motorised suspension could be used, the motor adjusting the amount of upward force instantaneously according to the vertical position of the user. Such a system, however, would be costly and complicated in design. The method of weight relief chosen for this system was provided by pneumatic actuation (Figure 1). The weight of the user is relieved by closing a pressure regulator that causes a desired amount of air to be compressed into the piston and a proportional shortening of the rope. Tension in the rope will not change as a result of the vertical walking motions because the action of the piston compensates for such movements.

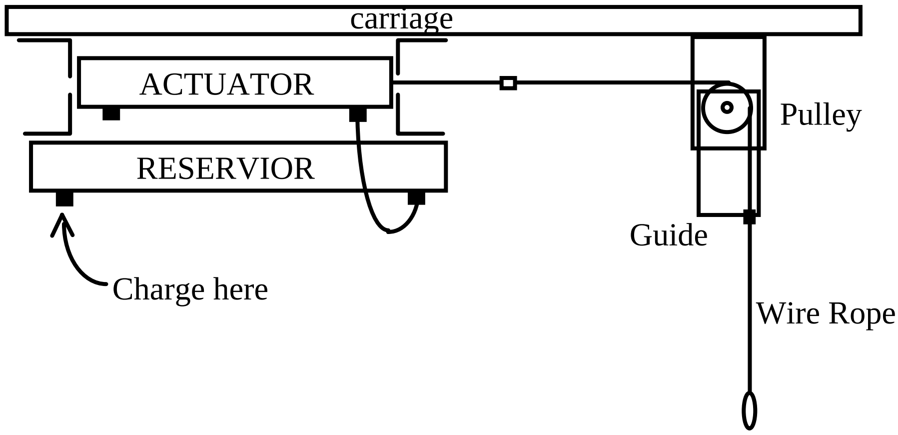

Figure 1. Configuration of the pneumatic weight relief system

3.14 Control

In order to free the subject from the need to drag the carriage behind them, a control system was designed. The control mechanism needed to be dependent upon and synchronised to the periods of walking. Maximum sensitivity and responsiveness were desired. To effect this, feedback from the carriage is provided through a potentiometer mounted at the axis of the rope guide. The potentiometer returns a signal proportional to the angle between vertical and the position of the user (Fig. 2). Thus, a velocity request signal is sent to the motor controller directly from the potentiometer.

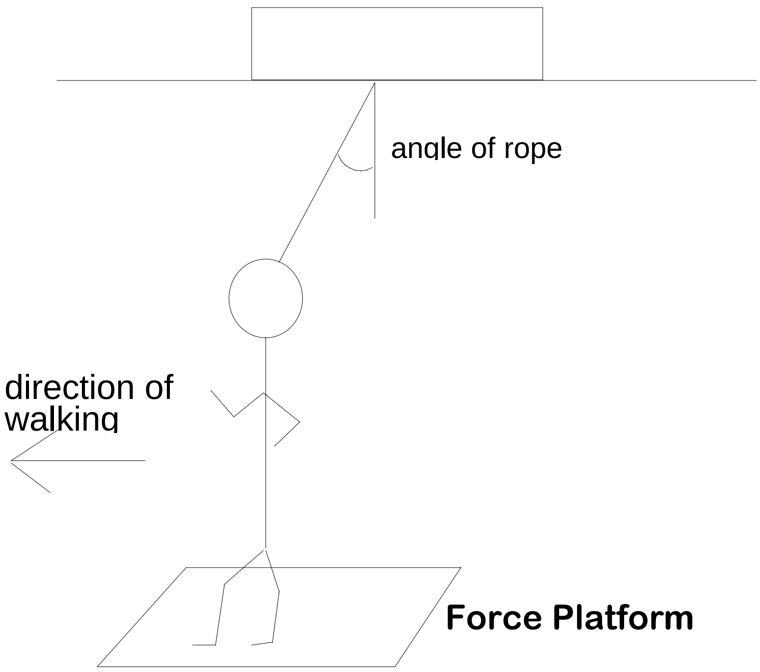

Figure 2. Feedback from the potentiometer returns a signal proportional to the angle between vertical and the position of the user

CHAPTER FOUR

DISCUSSION OF POTENTIAL OUTCOMES AND HYPOTHESES

4.1 POTENTIAL ADVANTAGES AND DISADVANTAGES

There are several advantages of using a system such as this when rehabilitating LLAs. Positively, it offers:

a) early mobilisation with controlled partial weight bearing in a completely safe environment;
b) removal of the restrictions imposed by conventional parallel bar training so that, the clinician and patient will be able to focus more attention on gait technique rather than safety;
c) reduced risk of work-place injury to the health professional;
d) physical and psychological benefits resulting from early mobilisation;
e) controlled reduction in weight relief as the amputee gains confidence and tissue tolerance; and
f) a sense of physical security that might be psychologically valuable (even with full weight-bearing status).

Along with the proposed advantages come a number of foreseeable disadvantages of using this system. Potentially these include:

a) the system is relatively noisy and the environment within which it is used may be threatening;
b) the harness may restrict the extremes of body motion, and compromise cardiopulmonary efficiency (as discussed previously); and
c) as suggested by Pillar, et al. (1991), the system may hinder the forward acceleration of the head, arms and trunk, thus restricting AB gait.

The overhead harness system at Rehab Tech features improvements over previous devices reported in the literature. The most significant of these include:

a. the control method, freeing the subject from the need to drag the carriage;
b. the relatively unrestrictive harness (i.e., minimal restriction of lower and upper limbs);
c. the pneumatic method of weight relief; and
d. the fact that this system can be used over a distance walkway, as well as over a treadmill.

It is felt that the negative effects of harness support systems reported previously in the literature may be reduced or even negated with the current system due to these factors.

4.2 HYPOTHESES

We now elucidate a number of main hypotheses that are to be tested. These are listed below and the method of testing is dealt with in Chapter 5.

Due to reasons discussed previously it is expected that the system will have an inhibiting effect on AB gait. Expected results include:

a. reductions in velocity, cadence and stride length, and an increase in gait cycle duration, these effects being more pronounced with increasing BWS;
b. temporal and loading symmetry of the lower limbs will remain unchanged;
c. vertical ground reaction forces will decrease proportionately with increasing the BWS; and
d. it is also hypothesised that the antero-posterior ground reaction forces will decrease with increasing BWS.

Due to the central support, inherent safety, and partial weight relief, it is expected that the system will have a positive effect on LLA gait. Among the expected results with increasing BWS are:

a. increased velocity, stride length, and cadence (toward their optimum post-rehabilitation);
b. decreased gait cycle duration;
c. greater stance and swing symmetry between the sound and prosthetic limbs;
d. greater symmetry of loading between the sound and prosthetic limbs, indicated in the vertical ground reaction force impulse;
e. decreased vertical and antero-posterior ground reaction forces;
f. decreased pain (if present); and
g. increased confidence reported subjectively.

CHAPTER FIVE

METHOD

5.1 SUBJECTS

Three groups of subjects were selected for participation in this investigation. These groups comprised; 1) young AB (YAB) subjects, 2) elderly AB (EAB) subjects, and 3) LLAs. Subjects within each group were chosen according to the following criteria:

- YAB subjects - age range 18 to 35 years; and
  - free from neuro-musculoskeletal disorders.

- EAB subjects - age range 45 to 80 years; and
  - free from neuro-musculoskeletal disorders.

- LLA subjects - either transtibial or transfemoral amputees;
  - recent amputation but able to tolerate up to one hour of intermittent walking; and
  - free from other compromising medical conditions.

Nine subjects were recruited for each group, yielding a total of 27 subjects. The nine YAB subjects (Table 1) were a convenience sample of friends and family of employees at Rehab Tech. These subjects ranged in age from 18 to 32 years (mean 24.8). Their mean body mass was 68.6 kg [standard deviation (SD), 17.6 kg] and mean height was 169.2 cm (SD, 6.2 cm). Similarly, the EAB subjects (Table 1) were a convenience sample with ages ranging from 45 to 80 years (mean 58.2). The mean body mass and height of these subjects were 69.8 kg (SD, 10 kg) and 168.7 cm (SD, 10.1 cm) respectively.

TABLE1. SUBJECT TABLE -- YOUNG AND ELDERLY ABLE-BODIED SUBJECTS

Young Able-bodied Subjects						Elderly Able-bodied Subjects				
Subject No.	**Sex**	**Age (yrs)**	**Weight (kg)**	**Height (cm)**		**Subject No.**	**Sex**	**Age (yrs)**	**Weight (kg)**	**Height (cm)**
YAB1	M	27	95.7	172		EAB1	F	64	50.1	157
YAB2	M	30	76.4	172		EAB2	M	51	70.9	181
YAB3	M	32	97.8	181		EAB3	M	61	81.0	182
YAB4	M	28	65.6	174		EAB4	F	50	61.2	168
YAB5	F	21	57.1	166		EAB5	M	64	67.7	175
YAB6	F	23	60.3	167		EAB6	F	59	76.9	167
YAB7	F	20	57.9	167		EAB7	F	80	68.9	156
YAB8	M	24	58.3	164		EAB8	M	50	82.5	174
YAB9	F	18	48.6	160		EAB9	F	45	69.1	158
mean		24.8	68.6	169.2		mean		58.2	69.8	168.7
std. dev.		4.8	17.6	6.2		std. dev.		10.7	10.0	10.1

TABLE 2. SUBJECT TABLE -- LOWER LIMB AMPUTEES

Lower Limb Amputee Subjects									
Subject No.	**Sex**	**Age (yrs)**	**Weight (kg)**	**Height (cm)**	**Amp. Level**	**Cause**	**Prosthesis**	**Time Post-amp (mths)**	**Time Post-prosthesis fit (mths)**
LLA1	M	63	73.5	167	TTA	PVD	PTB, cuff, SACH foot	4.2	3.0
LLA2	M	59	76.1	183	TTA	fracture, non-union	PTB, cuff, SACH foot	3.9	3.0
LLA3	M	22	86.8	183	TTA	trauma	PTB, cuff, Blatchford ankle, Seattle Lite foot	4.6	3.9
LLA4	M	55	104.8	183	TTA	charcot	PTB, cuff, SACH foot	16.7	14.6
LLA5	M	75	108.5	183	TTA	PVD	PTB, cuff, SACH foot	12.8	11.6
LLA6	M	39	95.6	185	TTA	trauma	PTB, cuff, Blatchford ankle, Seattle Lite foot	8.6	7.4
LLA7	F	22	110.1	175	TTA	trauma	PTS, multi-axial foot	5.8	0.5
LLA8	M	58	89.3	173	TTA	PVD	PTB, cuff, SACH foot	2.3	1.4
LLA9	M	61	100.6	183	TTA	tumour	PTB, cuff, Seattle Lite foot	33.5	32.0
mean		50.4	93.9	179.4				10.3	8.6
std. dev.		18.6	13.4	6.2				9.9	10.0

It was apparent after testing AB subjects that the physical requirements were higher than initially expected. The selection criteria for the LLAs was thus modified accordingly. A decision was made to include only transtibial amputees. Initially it was proposed that amputees would be tested prior to inpatient discharge (usually 6 to 8 weeks post-amputation), but, after testing two amputees at this stage it was evident that more “mature” amputees were required -- it was felt that most amputee subjects at such an early stage could not tolerate the testing procedures. Unfortunately, testing more experienced amputees may affect the results in an adverse manner (i.e., the expected benefits may not be as evident for this group). However, due to the required multiple comparisons, testing a group of ‘new’ amputees was not possible. The amputees who participated in the study (Table 2) ranged in age from 22 to 75 years (mean 50.4), had a mean weight of 93.9 kg (SD, 13.4 kg) and a mean height of 179.4 cm (SD, 6.2 cm). The mean time since amputation (or revision in one case, LLA8) was 10.3 months (range 2.3 to 33.5 months) and the mean time since initial prosthetic fitting was 8.6 months (range 0.5 to 32 months).

5.2 APPARATUS

5.21 Temporo-spatial Data

Temporo-spatial data was recorded using a Footswitch Stride Analyser (B & L Engineering, Santa Fe Springs, California). The Footswitch Stride Analyser is a microcomputer system designed to record foot-floor contact data. The footswitches are worn as insoles in the subject’s shoes. Contacts are provided in the heel, 5th metatarsal, 1st metatarsal and great toe areas. Each insole is connected to a small, lightweight recorder that is worn around the subject’s waist. A light-sensitive switch worn by the subject is activated by triggering lights set at a known distance apart along the walkway. This switch controls the recorder, providing signals that determine the elapsed time of the test. Powered by a nine volt battery, the device records the elapsed time and the footswitch patterns.

Following the completion of each trial, the recorder is connected to an IBM personal computer (PC) by means of an umbilical cable and the data are transferred through the recorder output port. Temporo-spatial gait parameters are calculated from the transferred data.

5.22 Ground Reaction Force Data

Ground reaction force data was recorded while the subject ambulated along the 15 metre walkway. An IBM PC with Bioware 2.0 software in conjunction with a Kistler force platform (9281B) and an 8 channel Kistler charge amplifier (9865B) allowed recording, storing and evaluation of ground reaction force data. A sampling rate of 250 hertz for a duration of between 2 to 4 seconds was used, depending upon the gait velocity of the subject. The Kistler force platform is mounted flush with the walkway and both are coated with the same non-slip surface.

5.3 PROCEDURES

Prior to testing, subjects were informed of the test procedures and a written consent form was completed (Appendix D). A brief explanation of the overhead harness system was given followed by recording of personal details. Appropriate size footswitch insoles were chosen for each subject and placed in the subject's shoes. These insoles were connected to a recorder that was worn around the subject's waist. The light-sensitive switch was secured to the upper arm of each subject with adhesive tape and connected to the recorder. The switch was positioned so as to avoid triggering through arm or leg movements (Herzog, et al., 1989).

Temporo-spatial Data: During each trial, subjects were required to walk at a comfortable self-selected velocity. Temporo-spatial data was recorded using the Footswitch Stride Analyser while the subject walked along the fifteen metre walkway. Triggering lights were placed between six and ten metres apart depending upon the subject's endurance and walking velocity, allowing a minimum of three complete strides to be recorded for each subject.

Ground Reaction Force Data: Ground reaction force data was recorded using the Kistler Force Platform while the subject walked along the same walkway. Subjects were required to complete a minimum of three successful passes over the force platform with each lower limb. Where 'targeting' of the force platform was recognised, the trial in question was rejected and repeated.

Each testing session began with a free-walking trial (i.e., the subject walked along the walkway without the harness system). Following the free-walking trial, subjects were harnessed to the overhead harness system. Subjects were then required to step onto the force platform while their body weight was recorded. Each subject was then given approximately 20% BWS, confirmed by static vertical ground reaction force collection. Subjects were required to familiarise themselves to 20% BWS conditions for a minimum of five minutes (the duration of familiarisation required was previously determined by pilot study tests of two YAB subjects; see Appendix B). For the duration of this familiarisation period, subjects were encouraged to ambulate along the walkway. It was hypothesised that the maximum amount of weight relief (20%) would require the longest familiarisation time, hence familiarisation to 20% BWS was given prior to the remaining test procedures. Following the familiarisation period, subjects were given an appropriate rest period, if required, prior to the next phase of testing.

Using a Latin square design (Table 3), subjects were randomly tested while harnessed to the system under the following conditions:

1) with no weight relief;
2) with 10% BWS; and
3) with 20% BWS.

The amount of BWS was confirmed by static vertical GRF collection using the force platform. A period of one to three minutes re-acclimatisation was given prior to testing under each of the above conditions. Appropriate rest periods were given.

TABLE 3. ORDER OF PRESENTATION OF WEIGHT RELIEF CONDITIONS

Subject Number	Order of Presentation		
Subjects 1,4,7	0% BWS	10% BWS	20% BWS
Subjects 2,5,8	10% BWS	20% BWS	0% BWS
Subjects 3,6,9	20% BWS	0% BWS	10% BWS

Twenty percent BWS was chosen as the maximum BWS level following pilot study tests of several subjects (Appendix B). It was concluded that greater than 20% BWS resulted in occasional loss of heel contact and made forward progression very difficult. 10% BWS was selected arbitrarily as a half-way point. Subjects were tested while wearing the harness system with no weight relief in order to distinguish the effects of weight relief from those of the harness system. The initial free-walking trial was not randomised amongst the harness trials; this choice minimised the inconvenience of donning and doffing the harness jacket. It was felt that this compromise would not affect the results significantly, since appropriate rest periods were given.

Following the testing sessions of the AB subjects a second free-walking trial was recorded. This allowed correlation with the first free-walking trial to determine any fatigue or learning effects present. In order to minimise the inconvenience and energy requirements of these subjects, a decision was made not to conduct a second free-walking trial with the LLAs. Since appropriate rest periods were given and the trials were randomised, the issues of learning and fatigue were not expected to influence the results significantly. It was felt that an idea of the effects on the AB subjects would be satisfactory to conclude the extent of their influence.

Following the completion of the session each subject completed a questionnaire regarding the effects of the harness system (Appendix C).

5.4 ANALYSIS

A Pearson's Correlation Test was performed on each variable to compare the 1st and 2nd free-walking trials, using Microsoft Excel software (Version 5.0).

To determine the statistical significance of the differences observed in the temporo-spatial data, 2 two-way repeated measures analysis of variances (ANOVAs) (group-by-condition) were performed on velocity and stance phase symmetry using Statistics Package for the Social Sciences (SPSS) (Minium, et al., 1993; Gravetter & Wallnau, 1992). Tukey's honestly significant difference (HSD) post-hoc tests, for repeated measures ANOVA, were used to locate the statistically significant differences (Gravetter & Wallnau, 1992; Kirk, 1995).

To determine the significance of the results observed for the GRFs, 5 three-way repeated measures ANOVAs (group-by-condition-by-side) were performed, using SPSS, on 1st vertical maximum, mid-stance minimum, 2nd vertical maximum, 1st horizontal maximum and 2nd horizontal maximum. Velocity was used as a covariate in the above ANOVAs, since the GRFs are velocity dependent. Using velocity as a covariate attempts to negate the effects of velocity when determining the statistical significance of the differences observed. The significance of the vertical loading symmetry was analysed using a two-way repeated measures ANOVA (group-by-condition). Tukey's HSD post-hoc testing, for repeated measures ANOVA, was used to determine the location of the statistically significant differences (Kirk, 1995).

CHAPTER SIX

RESULTS

6.1 CORRELATION BETWEEN 1st and 2nd FREE-WALKING TRIALS

Although the BWS trials were randomised to minimise the effects of learning and fatigue, free-walking trials for the YAB and EAB groups were repeated at the end of each testing session to determine the extent of these detrimental effects. The results of the multiple Pearson's correlations are presented in Table 4 below.

TABLE 4. RESULTS OF PEARSON'S CORRELATIONS BETWEEN 1ST AND 2ND FREE-WALKING TRIALS FOR YAB AND EAB GROUPS

CORRELATION BETWEEN 1st AND 2nd FREE-WALKING TRIALS				
Gait Variable	**YAB Right**	**YAB Left**	**EAB Right**	**EAB Left**
Gait Velocity	0.578	N/A	0.717	N/A
Stride Time	0.517	N/A	0.918	N/A
Stance Time	0.730	0.762	0.619	0.940
Vertical GRF 1st Max.	0.871	0.896	0.946	0.878
Vertical GRF Mid-stance Min.	0.757	0.891	0.948	0.886
Vertical GRF 2nd Max.	0.899	0.856	0.758	0.545
Vertical GRF Impulse	0.860	0.922	0.817	0.835
AP GRF Braking Max.	0.756	0.840	0.867	0.865
AP GRF Propulsion Max.	0.817	0.742	0.647	0.949

The results of the multiple Pearson's correlations presented above show a moderate to high positive correlation for all variables measured.

6.2 TEMPORO-SPATIAL DATA

6.21 Velocity

Figures 3-5 graphically display the effect of the harness condition on walking velocity for the YAB, EAB and LLA subject groups respectively. Walking velocity of YAB subjects is higher than that of the EAB group which was higher again than the amputee group [$F(2,24)=6.16$, $p=0.007$]. Tukey's HSD post-hoc testing (HSD=0.256) revealed that the group difference was between the YAB and LLA groups; the EAB group was not significantly different from either of these groups.

Inspection of these graphs show that as the level of BWS increases, the velocity consistently decreases in all three groups. Statistical significance of this condition effect was found [$F(3,72)=110.40$, $p<0.001$]. Post-hoc testing (HSD=0.074) revealed that all harness conditions were significantly different from each other.

A significant interaction (group-by-condition) effect was also present [$F(6,72)=2.64$, $p=0.023$]. Figures 4-6 indicate that the interaction effect may be caused by the fact that induced velocity variations for the YAB subjects differed from the other two groups; the difference between free-walking and 0% BWS in this group was larger than the minimal differences observed in the other two groups.

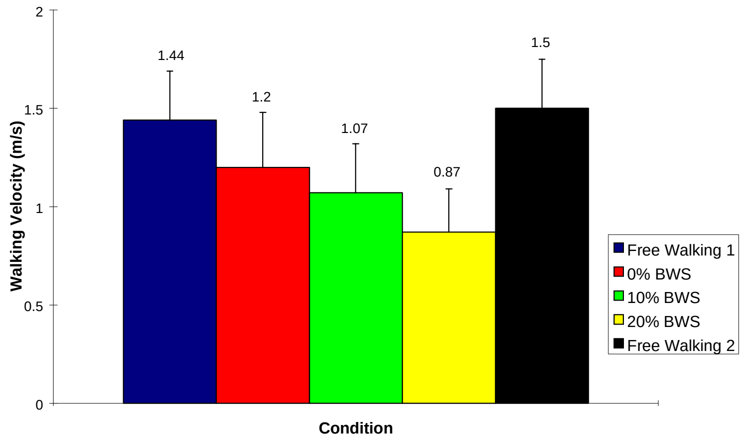

Figure 3. Walking velocity of YAB subjects with varying BWS

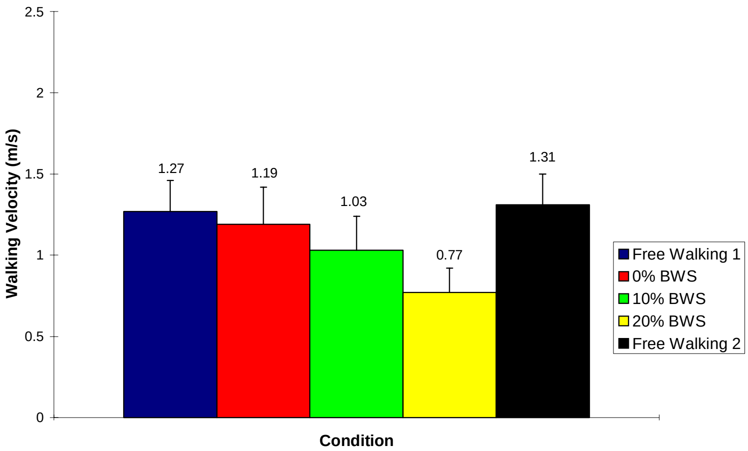

Figure 4. Walking velocity of EAB subjects with varying BWS

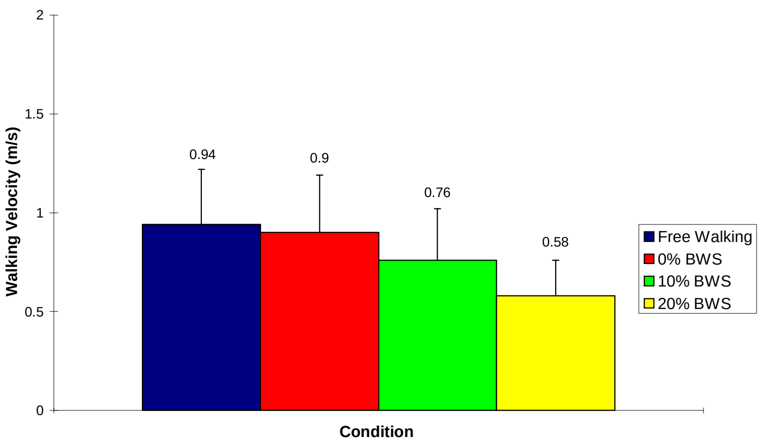

Figure 5. Walking velocity of LLA subjects with varying BWS

Note: Values above the error bars represent the mean of each condition.

6.22 Cadence

Figures 6-8 show the effect of BWS on cadence for the three subject groups. Since cadence is linearly related to velocity, the effects are similar: as BWS increases, cadence consistently decreases in all three groups. The largest differences are present between 0% and 10% BWS and between 10% and 20% BWS. Little difference is seen between free-walking and 0% BWS.

Cadence of the YAB and EAB groups were similar, while the cadence of both of these groups was greater than that of the amputee subjects.

6.23 Stride Length

Since stride length is also linearly related to velocity, one would expect stride length to decrease as velocity decreases (with increasing BWS). This expected result was evident in all three groups, and is displayed in figures 9-11. Differences between the three groups were similar to the differences observed for velocity.

6.24 Gait Cycle Duration

Figures 12-14 display graphically the effect of harness condition on gait cycle duration. As the level of weight relief is increased, gait cycle duration also increases. This effect is closely related to a decreased gait velocity (since gait cycle duration is inversely proportional to gait velocity). The average gait cycle duration of the LLA group was greater than that of the YAB and EAB groups, which were relatively similar.

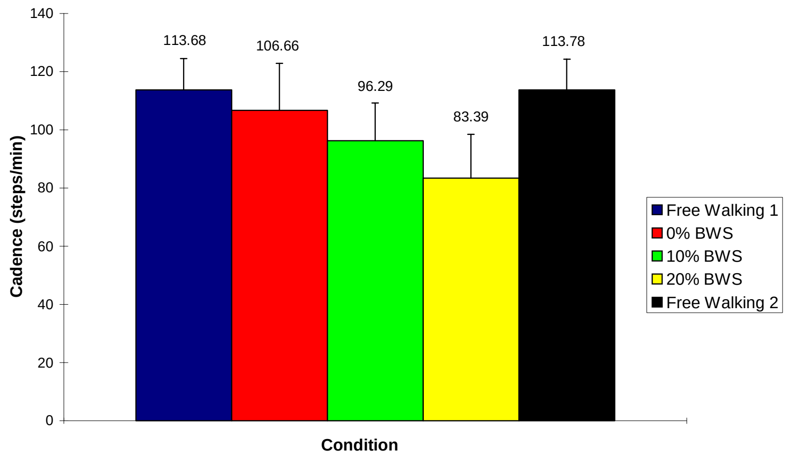

Figure 6. Cadence of YAB subjects with varying BWS

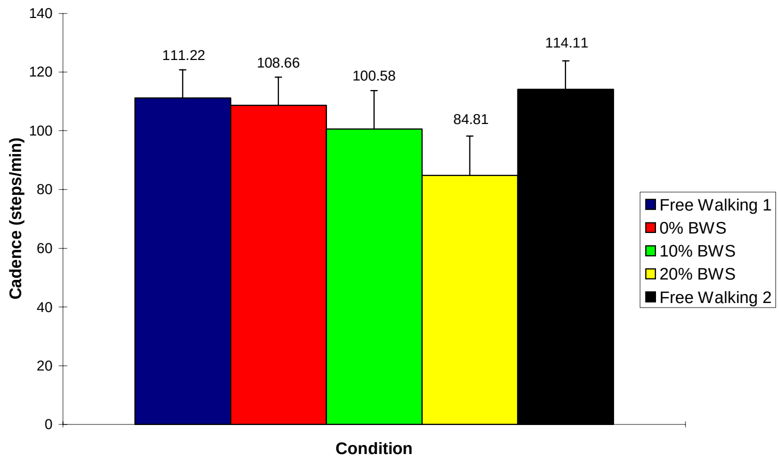

Figure 7. Cadence of EAB subjects with varying BWS

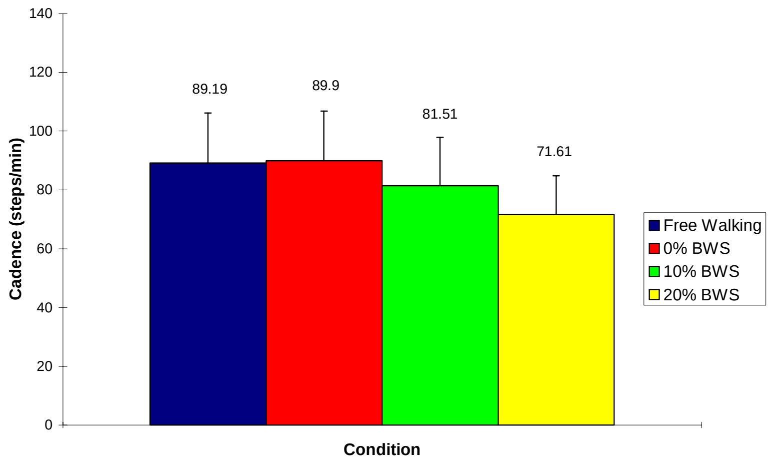

Figure 8. Cadence of LLA subjects with varying BWS

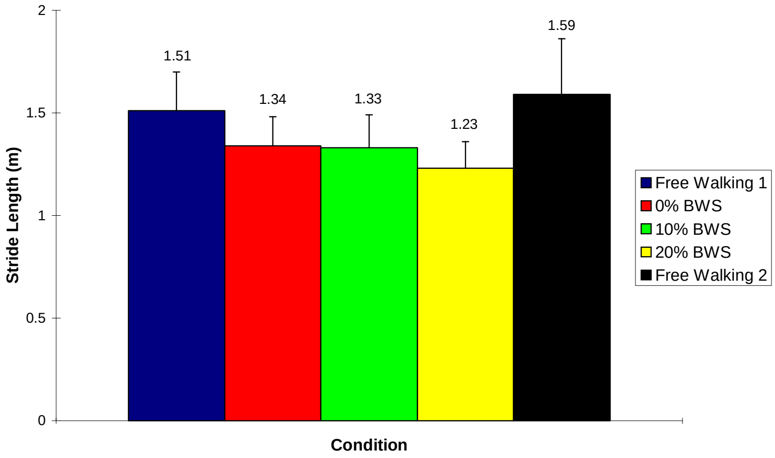

Figure 9. Stride length of YAB subjects with varying BWS

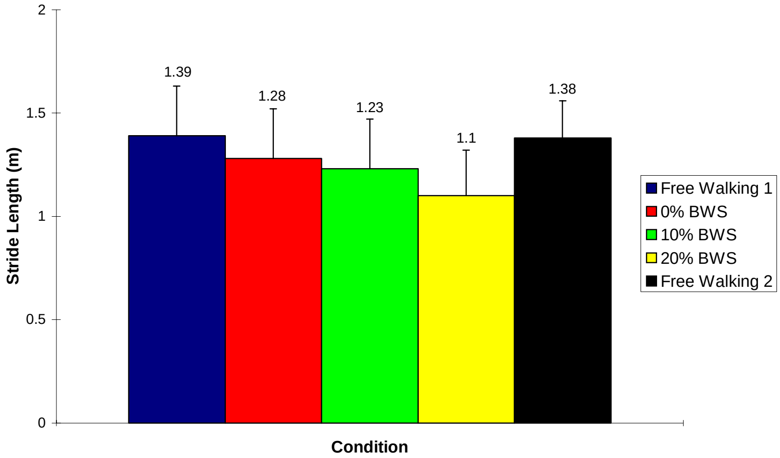

Figure 10. Stride length of EAB subjects with varying BWS

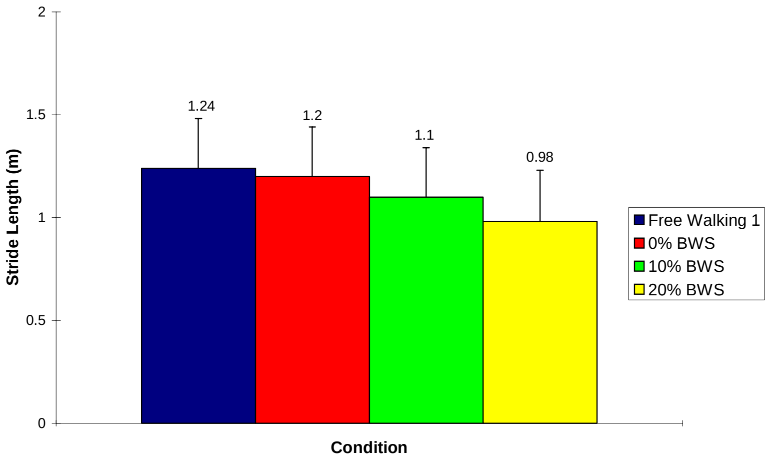

Figure 11. Stride length of LLA subjects with varying BWS

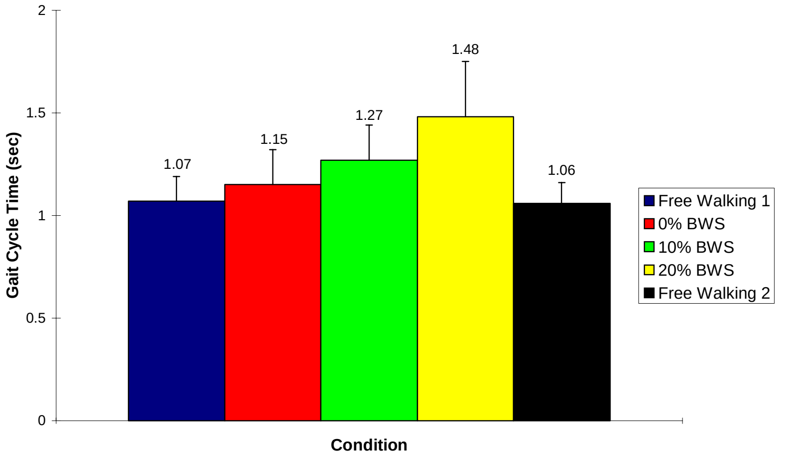

Figure 12. Gait cycle duration of YAB subjects with varying BWS

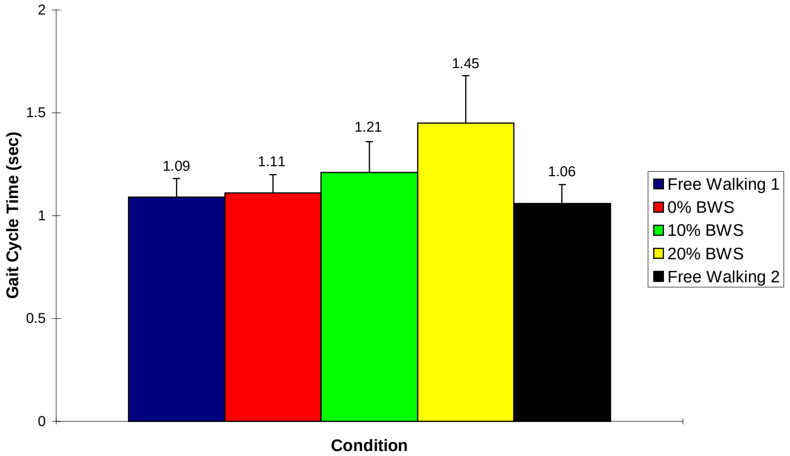

Figure 13. Gait cycle duration of EAB subjects with varying BWS

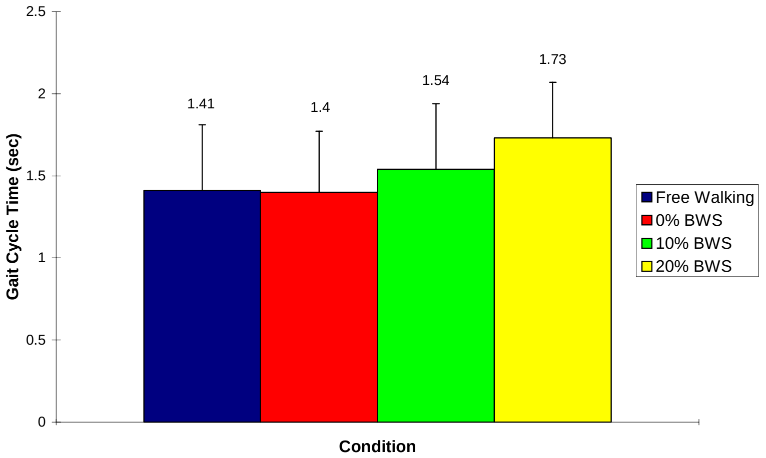

Figure 14. Gait cycle duration of LLA subjects with varying BWS

6.25 Stance Phase Symmetry

To obtain an index of stance phase symmetry, stance time on the left limb was divided by stance time on the right limb in the AB groups. In the LLA group, prosthetic stance time was divided by sound stance time to yield an equivalent symmetry index. Symmetry between the right and left limbs was found to be close to 100% for the AB groups and unaffected by the degree of weight relief (Figures 15 & 16).

The results of this study revealed that the stance symmetry of the LLA subjects ranged between 89.7% to 94% depending upon the degree of weight relief (Figure 17). A significant group effect [$F(2,24)=8.41$, $p=0.002$] was confirmed, Tukey's HSD post-hoc testing (HSD=4.82) revealing that the LLA group was significantly different from both AB groups (but the AB groups were not significantly different from each other).

Although there is a trend toward decreased stance symmetry with increasing BWS in the LLA subjects, this trend was not statistically significant [$F(3,72)=1.57$, $p=0.204$]. A significant group-by-condition interaction effect [$F(6,72)=3.33$, $p=0.006$] was present, caused by the fact that the stance symmetry of the amputee subjects was reduced at 20% BWS while the stance symmetry of the AB subjects remained relatively constant.

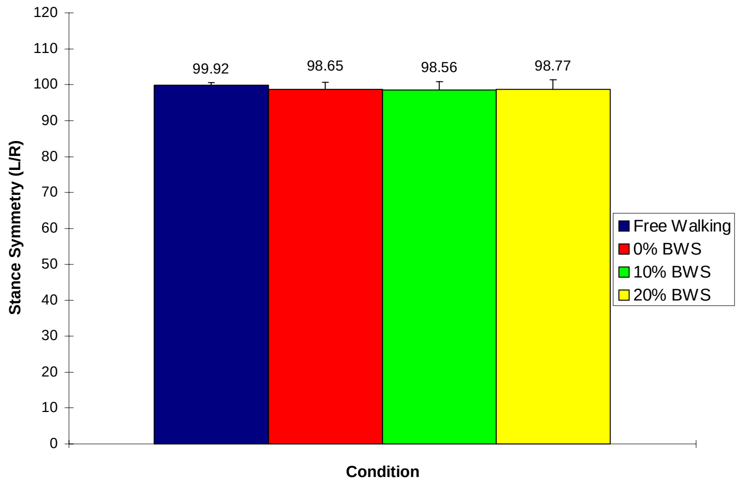

Figure 15. Stance symmetry of YAB subjects with varying BWS

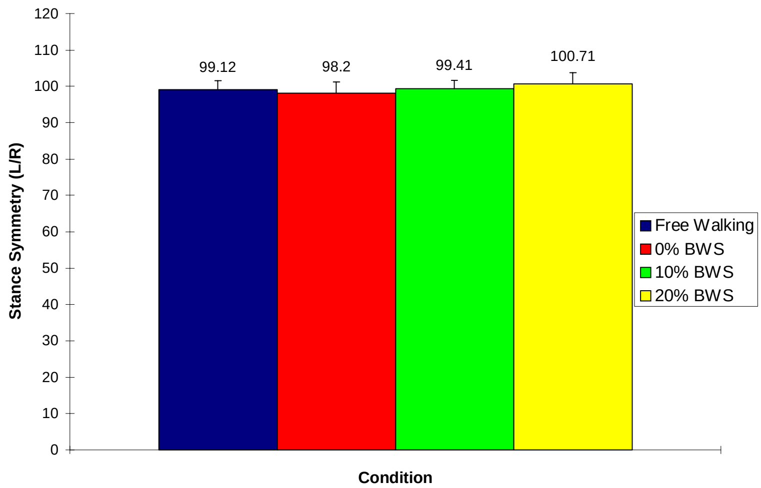

Figure 16. Stance symmetry of EAB subjects with varying BWS

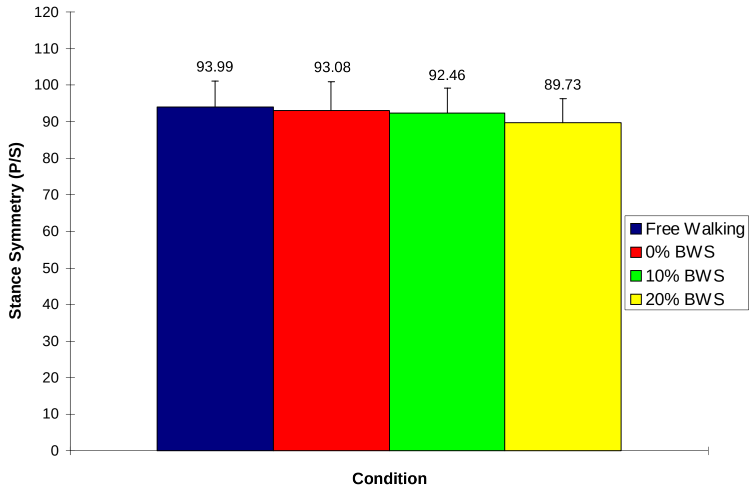

Figure 17. Stance symmetry of LLA subjects with varying BWS

6.3 GROUND REACTION FORCE DATA

6.31 Vertical Ground Reaction Force 1st Maximum

Figures 18-20 graphically display the means and standard deviations of this variable as a function of the test conditions for the three subject groups. These figures show that as the level of BWS increases the vertical GRF 1st maximum decreases accordingly. When statistical analysis was performed using velocity as a covariate, a significant condition effect was noted [$F(3,71)=15.14$, $p<0.001$]. Tukey's HSD post-hoc testing (HSD=3.49) revealed that the vertical 1st maximum at free-walking velocity was significantly greater than all harness conditions, but, the harness conditions were not significantly different from each other. Similar reductions are seen in all subject groups, confirmed by a non-significant interaction (group-by-condition) effect [$F(6,71)=1.23$, $p=0.300$]. No significant differences in magnitudes were observed between the YAB, EAB and LLA groups [$F(2,23)=0.15$, $p=0.862$]. However, the vertical 1st maximum of the prosthetic limb of the LLA subjects was less than the equivalent value on the sound limb, producing a significant 'side' effect [$F(1,24)=6.2$, $p=0.02$] (i.e., there was a significant difference between the two limbs). The significant group-by-side interaction [$F(2,24)=640.16$, $p=0.018$] confirms the asymmetry of the amputees compared to the AB groups.

6.32 Vertical Ground Reaction Force Mid-stance Minimum

Figures 21-23 display the effects of BWS on the vertical GRF mid-stance minimum. There appears to be random variation of this variable with the condition changes. However, when velocity was used as a covariate in the statistical testing a significant condition effect was observed [$F(3,71)=69.56$, $p<0.001$], with post-hoc tests (HSD=2.26) indicating that the mid-stance minimum at free-walking was significantly higher than all harness conditions and 0% BWS was significantly higher than 10% and 20% BWS.

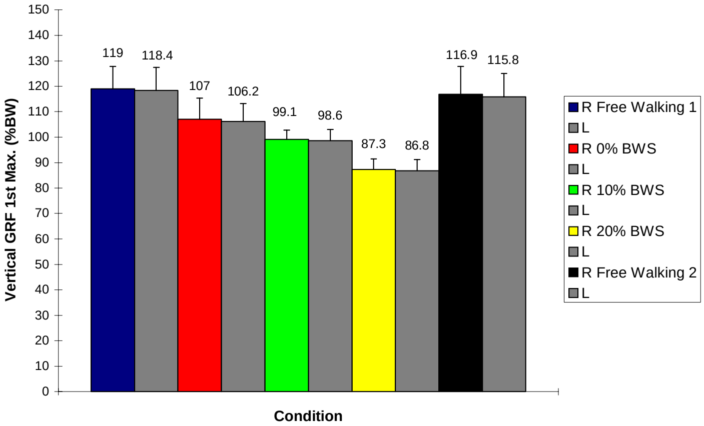

Figure 18. Vertical GRF 1st maximum of YAB subjects with varying BWS

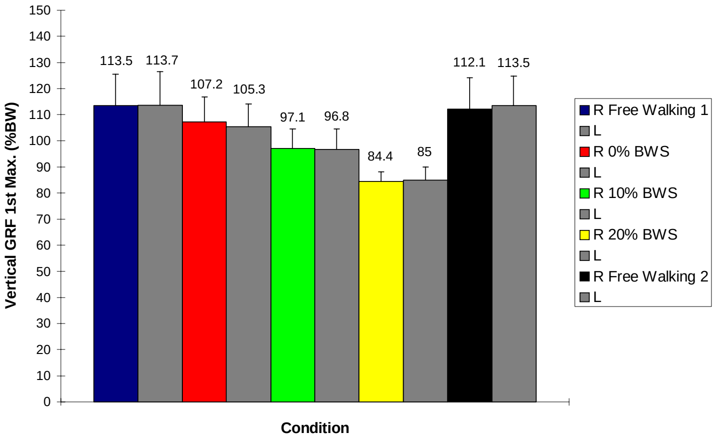

Figure 19. Vertical GRF 1st maximum of EAB subjects with varying BWS

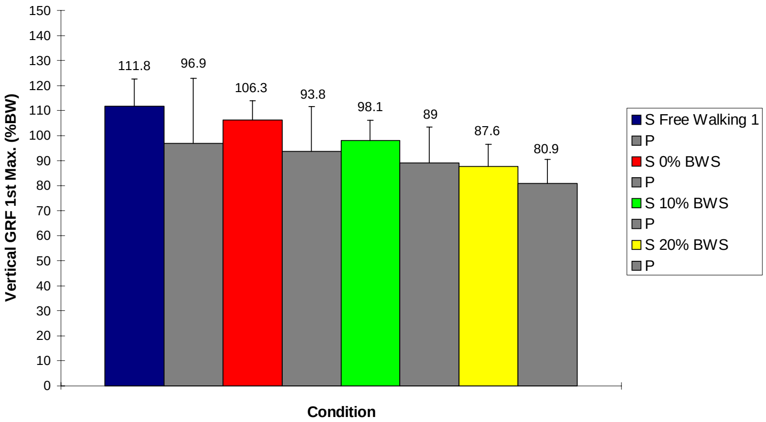

Figure 20. Vertical GRF 1st maximum of LLA subjects with varying BWS

Note: R=right, L=left, S=sound, P=prosthetic.

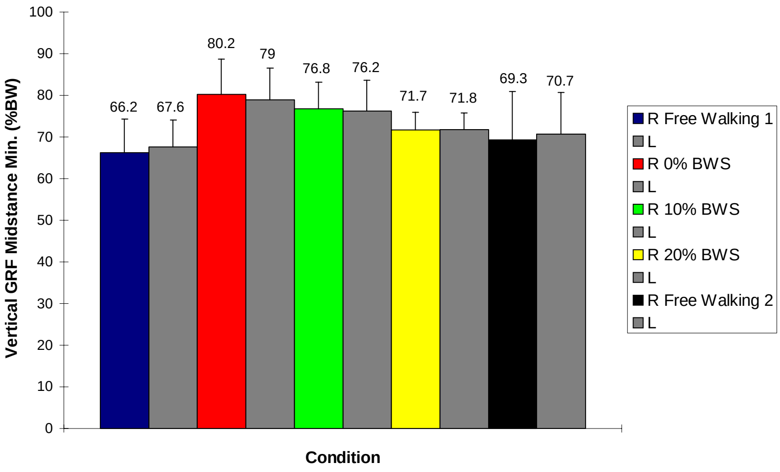

Figure 21. Vertical GRF mid-stance minimum of YAB subjects with varying BWS

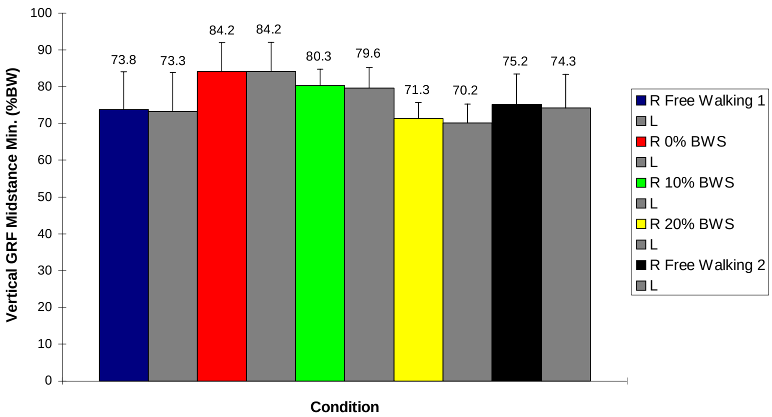

Figure 22. Vertical GRF mid-stance minimum of EAB subjects with varying BWS

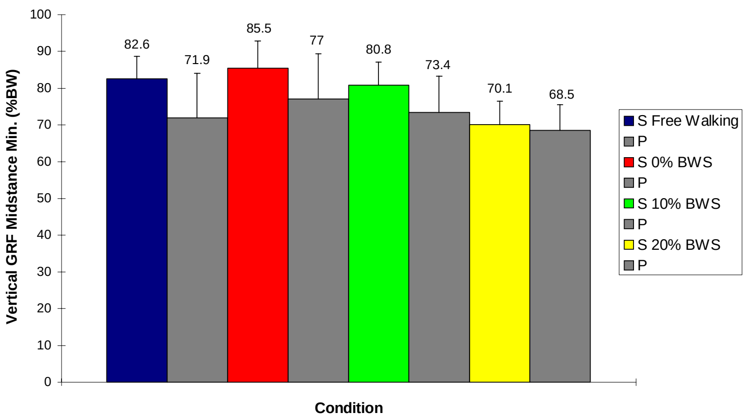

Figure 23. Vertical GRF mid-stance minimum of LLA subjects with varying BWS

The amputee subjects appeared to vary in different manner to the AB subjects, with less pronounced decreases in the mid-stance minimum GRF with BWS. This observation was confirmed by a significant group-by-condition interaction effect [$F(6,71)=5.19$, $p<0.001$].

Left-to-right symmetry of the YAB and EAB groups remained very close to 100%. Mid-stance minimum force of the sound limb of the LLA subjects is generally slightly greater than that of the YAB and EAB, groups although no statistical difference between groups was observed [$F(2,23)=1.62$, $p=0.219$]. A significant 'side' effect was detected [$F(1,24)=4.36$, $p=0.48$], values on the right(sound) limb being higher than those on the left(prosthetic). The non-significant group-by-side interaction [$F(2,24)=3.30$, $p=0.054$] indicates that all groups were similarly asymmetrical. A significant condition-by-side interaction [$F(3,72)=17.82$, $p=0.041$] may have been caused by the reduced asymmetry in the amputee group with 20% BWS.

6.33 Vertical Ground Reaction Force 2nd Maximum

The vertical GRF 2nd maximum is affected by the harness condition in a similar way to the 1st maximum. Figures 24-26 show that as BWS increases, the vertical GRF 2nd maximum decreases. Statistical analysis using velocity as a covariate revealed that the magnitude of this variable at free-walking velocity was significantly higher than for all harness conditions [$F(3,71)=37.04$, $p<0.001$] (HSD=2.53). The analysis also showed that the magnitude of the 2nd maximum at 10% BWS was higher than 0% BWS and significantly higher than 20% BWS. No significant differences between groups were observed [$F(2,23)=2.85$, $p=0.078$].

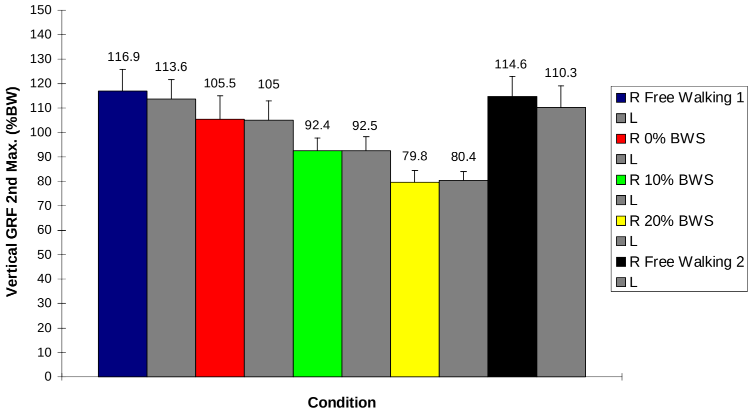

Figure 24. Vertical GRF 2nd maximum of YAB subjects with varying BWS

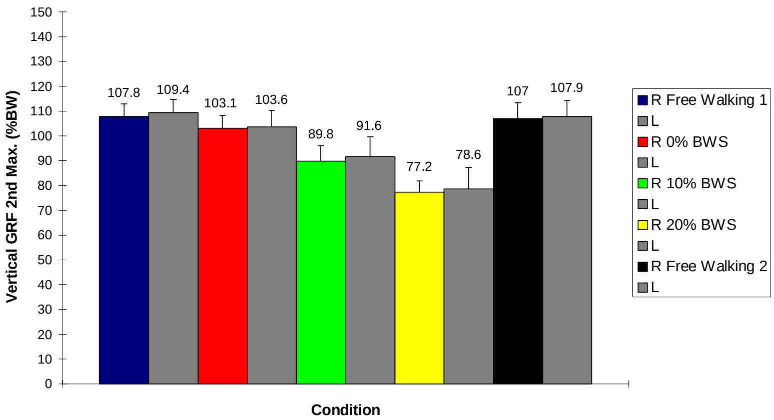

Figure 25. Vertical GRF 2nd maximum of EAB subjects with varying BWS

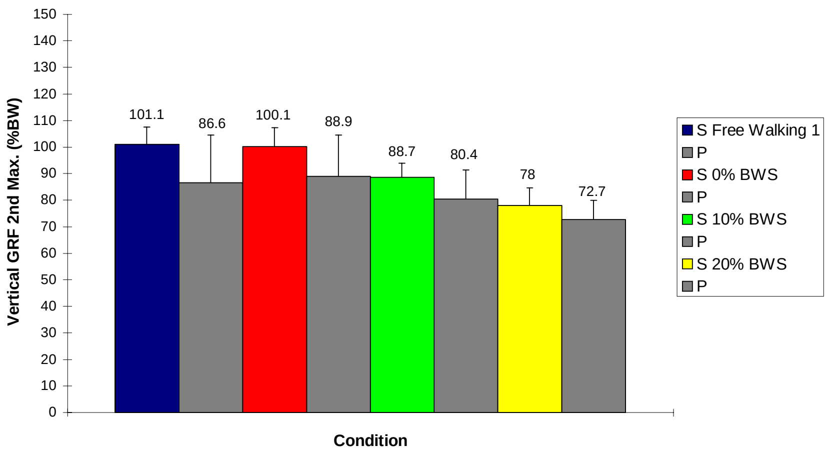

Figure 26. Vertical GRF 2nd maximum of LLA subjects with varying BWS

In the LLA group, the magnitude of the vertical GRF 2nd maximum is higher on the sound limb than on the prosthetic limb, although a significant 'side' effect was not recorded [$F(1,24)=3.82$, $p=0.062$]. However, a significant group-by-side interaction [$F(2,24)=632.07$, $p=0.019$] probably confirms the asymmetry of the amputees compared to the AB groups. Prosthetic/sound symmetry increases as the level of BWS increases (from 86% at free-walking to 93% at 20% BWS), producing a significant condition-by-side interaction effect [$F(3,72)=48.95$, $p=0.035$] .

6.34 Vertical Ground Reaction Force Loading Symmetry

To gain an index of lower limb loading symmetry the average force through the left (or prosthetic) limb was divided by the average force through the right (or sound) limb. Figures 27-29 display the effects of BWS on these loading ratios.

A significant group effect was observed [$F(2,24)=8.93$, $p=0.001$], Tukey's HSD post-hoc testing (HSD=10.89) revealing that the group difference was between the LLA group and the AB groups. That is, the amputees were significantly less symmetrical than both of the AB groups. Left/right symmetry of the AB groups ranged from between 97% to 101%, while prosthetic/sound symmetry ranged from 80.4% during free-walking up to 86.9% while harnessed with 20% BWS. Loading symmetry in the amputee group improved with increasing BWS, but the condition effect was not significant [$F(3,72)=0.15$, $p=0.932$]. The reason for the non-significance of this effect probably relates to the large variation between subjects. The non-significant group-by-condition interaction effect [$F(6,72)=1.26$, $p=0.288$] confirms that the amputees are similarly asymmetrical under all conditions.

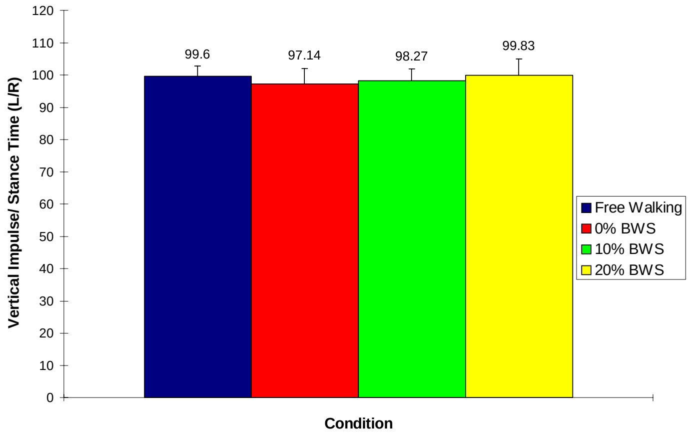

Figure 27. Vertical impulse symmetry of YAB subjects with varying BWS

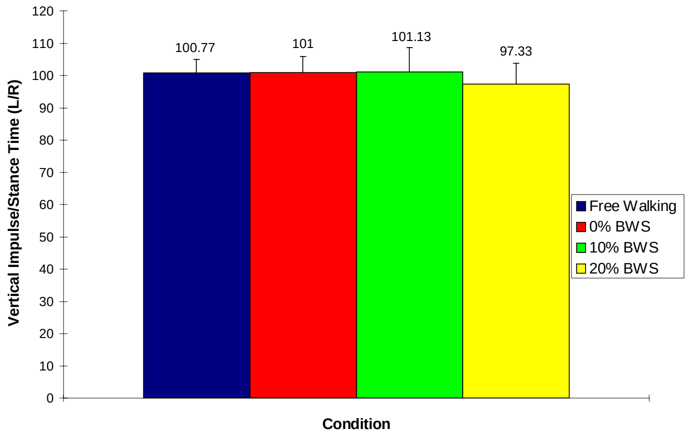

Figure 28. Vertical impulse symmetry of EAB subjects with varying BWS

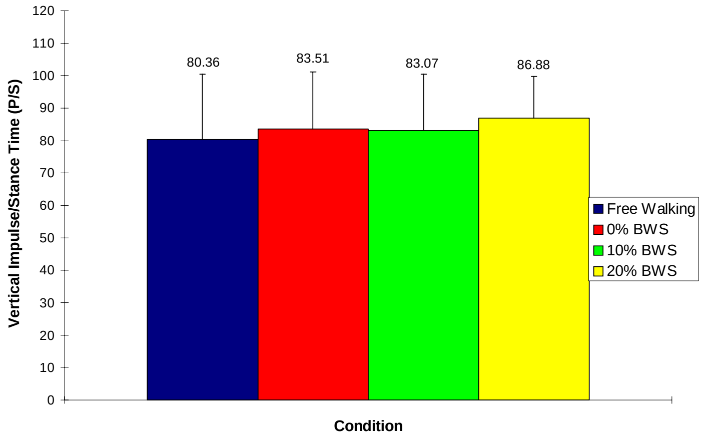

Figure 29. Vertical impulse symmetry of LLA subjects with varying BWS

6.35 Antero-posterior Maximum Braking Force

As BWS increases, the maximum braking force decreases accordingly in all three subject groups (Figures 30-32). However, when velocity is used as a covariate, the AP braking maximum for free-walking velocity is significantly higher than all harness conditions; the magnitude of this variable for 10% BWS is significantly greater than 0% BWS [$F(3,71)=11.71$, $p<0.001$] (HSD=1.74). No significant differences between groups were noted [$F(2,23)=1.41$, $p=0.264$].

The maximum braking force on the prosthetic limb of the amputees was between 55-70% of the value observed on the sound limb, resulting in a significant ‘side’ effect [$F(1,24)=8.73$, $p=0.007$]. The relative asymmetry of the amputee subjects was confirmed by a significant group-by-side interaction effect [$F(2,24)=56.03$, $p=0.034$].

6.36 Antero-posterior Maximum Propulsion Force

Reductions in the maximum propulsion force, similar to those observed for the braking force, are seen with increasing BWS (Figures 33-35). However, the reductions in the maximum propulsion force do not appear as large as those for the maximum braking force, although they still reach statistical significance [$F(3,71)=10.89$, $p<0.001$]. When the values are adjusted for the differences in velocity, the magnitudes of the AP propulsion GRF during free-walking was greater than that recorded for all harness conditions (HSD=0.96). The maximum propulsion force with 10% BWS was also greater than the values observed for 0% and 20% BWS. Similar asymmetry between the prosthetic and sound limbs of the amputees was also observed, although a significant ‘side’ effect was not detected [$F(1,24)=4.14$, $p=0.053$]. The significant group-by-side interaction effect [$F(2,24)=104.97$, $p=0.050$] is most likely explained by the relative asymmetry of the amputee subjects compared to the AB groups.

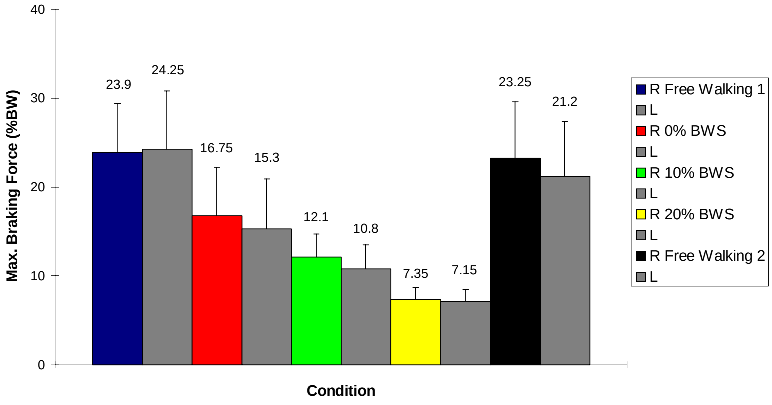

Figure 30. Maximum AP braking force of YAB subjects with varying BWS

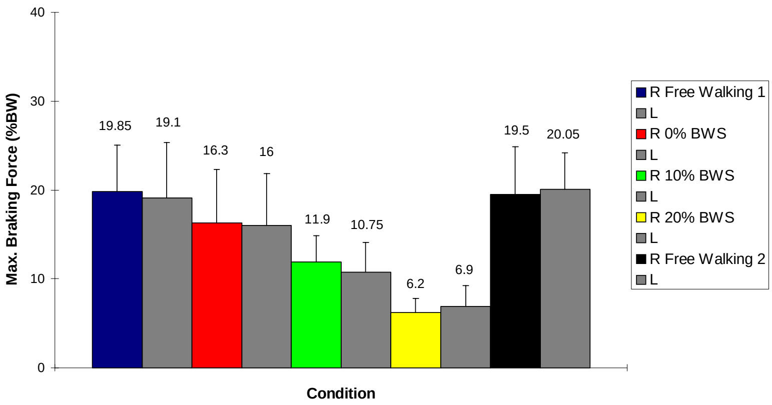

Figure 31. Maximum AP braking force of EAB subjects with varying BWS

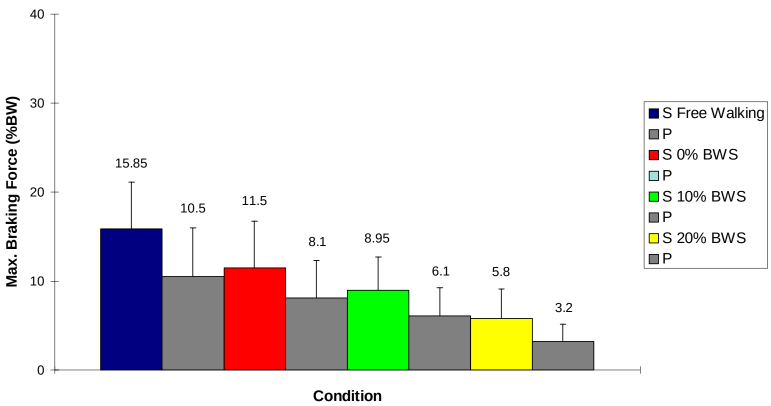

Figure 32. Maximum AP braking force of LLA subjects with varying BWS

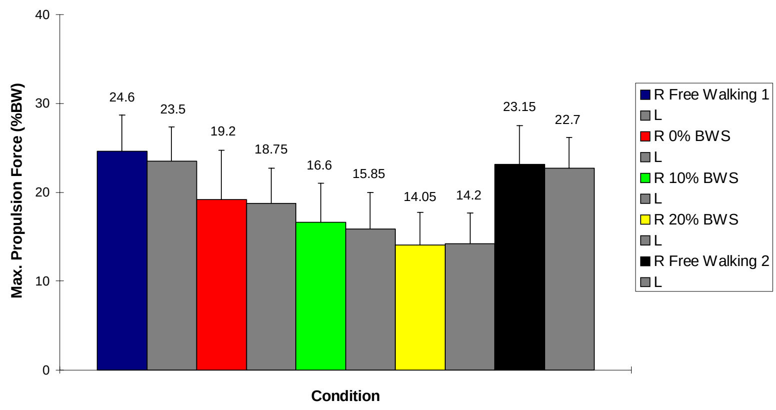

Figure 33. Maximum AP propulsion force of YAB subjects with varying BWS

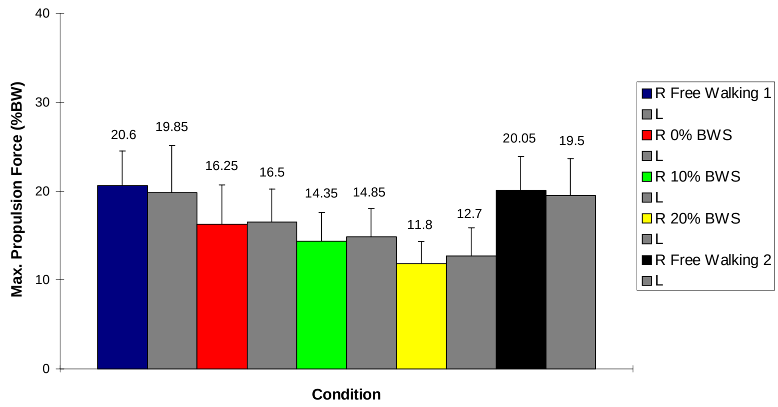

Figure 34. Maximum AP propulsion force of EAB subjects with varying BWS

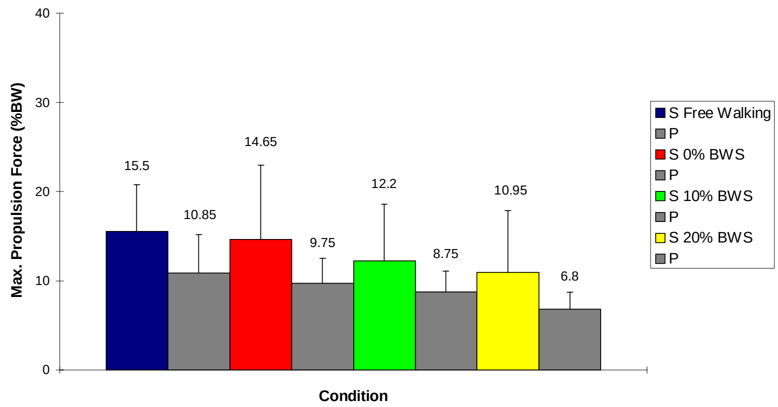

Figure 35. Maximum AP propulsion force of LLA subjects with varying BWS

6.4 Subjective Assessment

Results of the subject questionnaire are tabulated in Appendix C. Some of the most interesting and relevant findings are reported below. Of the four amputees who had initial pain, two felt that the harness system made no difference and two felt that it relieved the pain (no AB subjects reported any pain to begin with). Four of the nine amputees felt that the harness system increased their balance by either a little or a lot, while four EAB subjects reported small increases in balance. However, most subjects in all groups felt that the system had no effect on their balance. The majority of AB subjects felt that their safety was not altered by the harness system, while eight of the amputees felt very safe when using the system.

The vast majority of subjects in all three groups felt that the harness system would help in rehabilitating amputees, but, they were not as sure whether it had any advantages over parallel bars. Two thirds of the amputee group felt that the system had advantages over parallel bar training. Among the most frequent reasons given were:

1) increased security and safety;
2) increased confidence;
3) reduced pain;
4) more natural gait pattern; and
5) less weight required to be supported by upper limbs.

Among the most frequently reported disadvantages were:

1) the uncomfortable harness;
2) extra effort required to walk;
3) loss of traction experienced;
4) the noise of the system; and
5) although a few amputees experienced reduced distal stump pain, some of these reported increased AP distal stump pain.

CHAPTER SEVEN

DISCUSSION

7.1 TEMPORO-SPATIAL DATA

The free-walking temporo-spatial data of the AB subjects who participated in this study closely resembled the values reported by previous authors (Gonzalez, et al., 1974; Murray, et al., 1966; Murray, 1967). This lends support to the testing procedures and apparatus used for the purpose of this investigation, and confirms that the AB subjects were not atypical. Similarly, the free-walking temporo-spatial data for the LLAs in this study fell within the ranges reported by previous authors (Robinson, et al., 1977; Gonzalez, et al., 1974; Waters, et al., 1976; Huang, et al., 1979; Barth, et al., 1992; Hubbard & McElroy, 1994).

The results of the Pearson's correlations performed to compare the first and second free-walking trials of the YAB and EAB groups indicated a moderate to high positive correlation for all temporo-spatial variables included. These moderately strong positive correlations between the first and second free-walking trials confirms that learning and fatigue played little significant part in the results. This, coupled with the randomisation of the harness trials, increases the validity of the results.

The first experimental hypotheses relating to the temporo-spatial data were that increasing BWS would result in a decrease in velocity, cadence, and stride length and an increase in gait cycle duration for the AB subjects. The results of this investigation support these hypotheses. A significant condition effect was observed for velocity, and post-hoc testing indicated that all harness conditions were significantly different from each other. That is, as BWS increased, velocity significantly decreased with each successive change in harness condition. Associated with the decrease in velocity were a decrease in cadence and stride length and an increase in gait cycle duration.

Since these variables are proportional to velocity such effects would be expected. However, the relationship between velocity and these other temporo-spatial variables had not been previously tested under these conditions and was therefore worth examining. The reductions in stride length and cadence with the reductions in velocity are similar to those observed by Andriacchi, et al. (1977) for normal walking at varying velocities. Therefore, it does not appear that harness support significantly alters the basic mechanics of walking.

The above findings confirm the results of previous authors who have studied the effects of BWS on AB subjects. Pillar, et al. (1991) recorded reductions in velocity in six AB subjects with 0% and 20% BWS, using a similar harness system. Their results showed that subjects walked at an average of 90% and 65% of their free-walking velocity with 0% and 20% BWS respectively. Similar percentage reductions were observed in both of the AB groups in this study. YAB subjects walked at 83%, 74% and 60% of their free-walking velocity with 0%, 10% and 20% BWS, while at the same levels of BWS the EAB subjects walked at 94%, 81% and 61% of their free-walking velocity. Barbeau and Blunt (1991), Finch and Barbeau (1985) and Finch, et al. (1991) also noted reduced velocity with increasing BWS for AB subjects. Finch and Barbeau (1985) also reported increased gait cycle durations with 50% and 70% BWS. It is interesting that there was a significant reduction in velocity with 0% BWS. This observation would suggest a restriction placed on the subject by the harness system, not due to the weight relief. Some possible sources of this restriction are discussed later.

The results of this investigation did, however, contradict the hypotheses that increasing BWS would result in an increase in velocity, cadence and stride length and a decrease in gait cycle duration in LLA subjects. In contrast, the opposite was observed. Increasing BWS resulted in decreased velocity, cadence and stride length and increased gait cycle duration. The LLA subjects walked at 96%, 81% and 62% of their free-walking velocity while harnessed to the system with 0%, 10% and 20% BWS respectively. The magnitudes of the reductions observed were similar to those observed in the AB subject groups, a result that was not expected.

There may be a number of reasons why the above results were obtained. Firstly, the amputees who were selected to participate in this study were already competent walkers and had completed their in-patient gait rehabilitation programme. It may have been that these amputees had already reached velocities close to their optimal. For this reason, the expected beneficial effects of weight relief and security might have had little positive influence. The negative effects imposed by the harness system obviously outweighed the positive effects for this group of amputees (at least when considering the temporo-spatial variables).

Secondly, a number of design factors may have resulted in the negative effects observed for both the AB and LLA subjects. The harness jacket that was worn on the users upper body appeared to have caused a restriction on upper body and pelvic movement. Although it is felt that this restriction was minimal, it may have influenced the gait pattern in an adverse manner. This may have contributed to the reduction in velocity with 0% BWS.

The harness system was specifically designed to free the subject from the need to drag the apparatus behind them, but this effect (or at least the perception of it) apparently remained present. This may have been related to the rate of fall of the body's centre of gravity. The pneumatic weight relief system maintained a constant upward lift despite the position of the body's centre of gravity. However, when compared to free walking, the rate of fall of the body's centre of gravity during harness support would be much reduced. The effect of this phenomenon was that the subject was still 'falling' while trying to walk forward, giving the perception that they were dragging the system. This may explain the decreases in velocity observed with weight relief.

Finally, increasing BWS resulted in a reduced capacity to achieve push-off, evidenced in the reduced vertical GRF 2nd maxima and the antero-posterior GRF propulsion maxima (to be discussed). These reduced GRFs also contributed to the overall reduction in velocity. It is most likely that a combination of all of the above effects led to reduced velocity, cadence and stride length and the concomitant increase in gait cycle duration. Some of these detrimental influences may be minimised by design modifications, but, some are simply a result of the reduced weight. Possibly, with further design refinements and a group of ‘less experienced’ amputees, the hypothesised positive effects of the harness system may yet be realised.

Since velocity is important and reliable as a general gait descriptor, the influence that BWS has on velocity is of clinical significance. Generally, in gait rehabilitation, increased velocity indicates a functional improvement and is often one of the outcomes of effective rehabilitation. For this reason, it would appear that the harness system has no beneficial effects when considering the above temporo-spatial variables.

The second set of hypotheses were related to the stance phase symmetry of AB and LLA subjects. It had been postulated that the stance symmetry of AB subjects would be unaffected by the harness condition. The results of this study confirm the hypothesis. The stance symmetry of AB subjects was close to 100% for all conditions, and unaffected by the change in condition. It has been previously documented that transtibial amputees are temporally asymmetrical, that is, their prosthetic/sound stance time ratio is not 100%. Breakey (1976) and Robinson, et al., (1977) reported prosthetic/sound stance symmetry of transtibial amputees to be approximately 93-94%. The amputees in this investigation displayed a prosthetic/sound stance symmetry of around 90%. However, it was hypothesised that increasing BWS might result in an increase in the stance symmetry of these subjects. It was reasoned that weight relief, combined with central support plus increased safety and security, would allow the amputee to spend more time on the prosthetic limb and hence achieve increased temporal symmetry. The results of this study, however, did not confirm the above hypothesis. In fact, a non-significant trend toward increased temporal asymmetry was observed with increasing BWS.

There may be a few reasons why temporal symmetry did not improve as expected, among them:

1) the amputees may have already reached their optimal stance symmetry; and
2) the increased difficulty in achieving push-off may have caused the increased asymmetry. Amputees may spend a greater proportion of the gait cycle on their sound limb in an attempt to achieve adequate push-off, while reducing the proportion of time spent on the prosthetic limb to reduce the stump pressures during this more difficult push-off.

The clinical implications of this finding are that as BWS increases, temporal asymmetry also increases, whereas the opposite is desired.

7.2 GROUND REACTION FORCE DATA

The free-walking vertical and antero-posterior GRF data of the AB subjects who participated in this investigation were similar to previously reported results (Martin & Marsh, 1992; Menard, et al., 1992; Nilsson & Thorstensson, 1989; Andriacchi, et al., 1977; Jansen & Jansen, 1978). It may therefore be reasoned that the AB subjects who participated in this study were a reasonable representation of the AB population.

Similarly, the free-walking vertical and antero-posterior GRF data of the amputee subjects in this study also correlated well with values reported for transtibial amputees by previous authors (Menard, et al., 1992; Barr, et al., 1992; Suzuki, 1972). Again, this lends support to the validity of the test design and the apparatus used. Prior to the present investigation, no previous literature existed investigating the effects of BWS on GRFs. Hence, the new findings may serve as baseline data with which future studies may be compared.

The first experimental hypothesis relating to the GRFs was that vertical loading symmetry of AB subjects would be unaffected by the harness condition. The results are confirmatory of the hypothesis. Vertical loading symmetry of AB subjects remained close to 100%, despite the changes in BWS. Hamill, et al. (1984) investigated GRF symmetry during walking in AB subjects, also finding a high degree of loading symmetry in vertical GRFs between the right and left limbs.

The hypothesis for LLA subjects was that vertical loading symmetry would increase as BWS increased. The weight relief, central support plus enhanced safety and security may allow the loading of the lower limbs to become more symmetrical; diminished load-bearing pain and increased confidence would contribute to such an expectation. It has been well documented that transtibial amputees load their sound limb more than their prosthetic limb (Menard, et al., 1992; Suzuki, 1992 cited in Hurley, et al., 1990; Barth, et al., 1992). The transtibial amputees in this study displayed a degree of loading asymmetry, the sound limb being preferentially loaded. However, no significant improvement in vertical loading symmetry was observed. Loading symmetry in the amputee group increased from 80% at free-walking to 87% with 20% BWS, but this effect was not statistically significant. The implications of this trend are that transtibial amputees may weight-bear more symmetrically with increased BWS. This trend toward increased loading symmetry with increased BWS in the LLA subjects confirms the observations made by Hesse, et al. (1994 & 1995). Hesse, et al. (1994 & 1995) reported greater symmetry of loading of stroke patients with BWS (although no data were reported to substantiate their claims). However, one must be careful in inferring that BWS results in increased loading symmetry since statistical significance was not reached.

The remaining hypotheses concerning the GRFs were that both the vertical and AP GRFs would be reduced as BWS increased in all three subject groups. In an attempt to control for the effects of the reduced velocity with increased BWS and the differences in velocity between the groups, velocity was used as a covariate in the ANOVA calculations. This allowed the investigator to isolate the effects of BWS alone on these GRF variables. The results were a reduction in both the vertical GRF 1st and 2nd maxima as BWS increased. As BWS progressively increased, the 1st and 2nd maxima decreased accordingly. Reductions of approximately 6-10%, 15-17% and 26-27% were seen with 0%, 10% and 20% BWS respectively in the AB groups. The reductions seen in the amputee group were 3-5%, 8-12% and 17-22% with the increasing levels of BWS. It appears that the reductions in peak forces are close to, or greater than, the respective level of weight relief. However, when velocity was used as a covariate in the statistical analyses, few significant differences were observed. For both the 1st and 2nd vertical maxima, all harness conditions resulted in a reduction in these peak forces compared with free-walking. However, contrary to expectation, as the level of BWS increased these peak forces did not decrease, apart from the contribution ascribed simply to decreased velocities. It appears, therefore, that BWS reduces the peak vertical GRFs, but, the reduction is almost independent of the level of BWS. One would expect a reduction in vertical GRFs with increased BWS since the subject bears less weight on the ground. It may be that the variability within and between groups has overshadowed this effect.

Prosthetic limb values for these two variables were generally lower than those for the sound limb. Increased symmetry between limbs in the amputee group was observed, with 20% BWS for the vertical 2nd maximum. This finding may be of clinical significance since the 2nd maximum is related to the push-off phase of the gait cycle, and improvements in the push-off symmetry should result in a more symmetrical gait.

Significant differences in the vertical GRF mid-stance minimum were observed with the changes in BWS. When the effects of velocity are uncompensated, the influence of BWS on this variable appears random. However, when velocity is considered, increasing BWS results in a decreased in vertical mid-stance minimum. Again, the values of this variable during harness conditions are significantly less than during free-walking, with 10% and 20% BWS yielding decreased values compared with 0% BWS. Values of this variable on the prosthetic limb were significantly less than values on the sound limb.

If the effects of velocity are uncompensated, decreases in the 1st and 2nd maxima together with a relatively random variation of the mid-stance minimum has a flattening effect on the characteristic two-peak vertical GRF pattern. With harness support, there are reduced peak vertical forces on the sound and prosthetic limbs of the amputees, which may be of clinical significance when weight-bearing pain is an issue. Harness supported walking with BWS allows the amputee to ambulate (albeit at a slower speed) while reducing the vertical forces applied to both lower limbs, including the residual limb. This may be valuable when reduced, controlled weight-bearing is desired during the early stages of amputee rehabilitation.

When considering forces in the horizontal direction, the maximum braking and propulsion GRFs decreased with increasing BWS, in all three groups. However, when velocity was used as a covariate in the statistical analysis, other conclusions could be drawn. Peak braking and propulsion forces during free-walking were significantly greater than those during all harness conditions. Other differences were observed, but they were minimal and random in nature. Differences between the sound and prosthetic limbs of the amputees were, however, apparent, with values for the sound limb higher than those observed on the prosthetic limb. Harness support results in a decrease in the horizontal GRFs. However, the reduction does not seem to be consistent with the level of BWS in all cases.

It appears that BWS has a more pronounced impact on the maximum braking GRFs than on the propulsion GRFs. One possible explanation is that the reduced rate of falling of the body's centre of gravity would lessen the braking forces required in the horizontal direction; however, since forward velocity is more difficult to achieve with increased BWS, the subject would consciously attempt to increase the propulsive shear forces. This could be responsible for the subjective feedback from the amputee subjects that pain resulting from AP movement of the residuum within the socket was increased with BWS. Since additional force is required to achieve a 'normal' push-off, this additional force is transmitted to the anterior aspect of tibia within the socket, possibly causing an increase in residuum pain. It is also possible that the harness system does not apply a pure vertical unloading but, because of the inclination of the suspending rope, has a component of posteriorly directed shear. If this is the case, in order to maintain constant velocity, the braking impulse would have to be reduced and the accelerating impulse increased to offset the constant braking impulse of the harness system.

To summarise, it appears that increasing BWS results in instances of increased symmetry in the LLA group, together with reduced vertical and AP GRFs. These effects may be of clinical significance and would seem desirable during preliminary stages of amputee gait rehabilitation. Increased loading symmetry is one of the goals of rehabilitation, and may mitigate against any degeneration in the sound limb as well as in the prosthetic limb.

Lower GRFs that occur with increased BWS may also be of clinical significance during amputee gait rehabilitation. They may allow the amputee who can only tolerate a partial weight-bearing status to begin gait training in a completely safe and controlled environment with reduced risk of tissue breakdown.

CHAPTER EIGHT

CONCLUSION AND FURTHER INVESTIGATION

8.1 CONCLUSION

Recapitulating, the overhead harness system at Rehab Tech allows a controlled and quantifiable amount of weight relief to be provided to users, while allowing them to ambulate along a walkway. Increasing the amount of weight relief yielded a significant reduction in velocity and a reduction in cadence and stride length, together with increased gait cycle duration in the AB subjects. Contrary to expectation, the above effects were also observed for the transtibial amputees. The positive effects of safety, security and weight relief did not generate an expected increase in velocity in the amputee subjects. There may have been a number of reasons for this discrepancy between observed and expected results. Among the reasons are:

1) group demographics (specifically already trained subjects);
2) harness system design factors; and
3) the effects of the reduced GRFs.

It had been postulated that stance symmetry of the AB subjects would be unaffected by the changes in BWS, while the stance symmetry of the amputee subjects would increase. However, although the hypothesis for the AB subjects was confirmed, the stance symmetry of the amputee subjects decreased with increasing BWS. An increased difficulty in achieving push-off may be the cause of this.

Increased BWS produced lower vertical and AP GRFs for all three subject groups. This result may be of clinical importance during early gait training when a partial weight-bearing status is required or desired. Loading symmetry of the AB subjects remained unchanged while the loading symmetry of the amputee subjects increased with increasing BWS.

The harness system has both positive and negative effects on amputee gait and must, therefore, be viewed cautiously in relation to amputee rehabilitation. Although most subjects gave favourable feedback after using the system, further improvements and investigation are required before its adoption as a clinical tool in amputee rehabilitation. It is the opinion of the author that, when compared to parallel bar training, the overhead harness system does not offer enough substantial improvements to warrant its future utilisation in the rehabilitation of LLAs. Among the reasons that form the foundation of this statement are:

1) the cost of the overhead harness system would be substantially greater than the cost of parallel bars;
2) the simplicity and reliability, and therefore the ease of use, of parallel bars make them an attractive tool compared to the harness system; and
3) the harness system, as tested, failed to produce clear cut improvements in gait pattern.

As an alternative, the merits of the harness system are not strikingly great, but its disadvantages would not seem to disqualify it either.

Nevertheless, such a system could be used in the case of 'difficult' patients. Some examples of the types of patients that may benefit from the use of such a system are:

1) the obese;
2) the non-compliant;
3) those with high tissue breakdown susceptibility;
4) those with abnormally poor balance; and
5) those with upper body or upper limb weakness (ill-suited to parallel bars).

8.2 FURTHER INVESTIGATION:

Design modifications (which are currently underway) and further investigation should take place in the form of clinical trials to compare parallel bars and the weight relief system. A long-term study comparing the rehabilitation outcomes of a large group of pre-ambulatory amputees using an improved overhead harness system compared with a similar group of amputees rehabilitated in the conventional manner would be appropriate. Only following such a study can one fully gauge the effect of the overhead harness system on lower limb amputee rehabilitation. It would be interesting and valuable to conduct a fixed velocity analysis of the harness system to precisely distinguish the effects of velocity and BWS.

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APPENDIX A

OVERHEAD HARNESS SYSTEM AND WALKWAY

Figure 36. Overhead harness system and walkway

APPENDIX B

PILOT STUDY

During the initial stages of this investigation, while the testing protocol was being established, several AB subjects were asked to spend some time walking while harnessed to the system. Varying degrees of weight relief were provided and the subjects walked along the 15 metre walkway. A number of observations were made:

1) magnitudes of weight relief much greater than 20% resulted in an occasional loss of heel contact;

2) walking while harnessed to the system with greater than 20% BWS placed large physical requirements on the subject. It appeared that much additional energy was required to maintain forward velocity at a reasonable rate; and,

3) observation and personal experience seemed to indicate that a period of acclimatisation was necessary before a consistent gait pattern could be established.

Considering the first two observations, it was decided that 20% BWS would be selected as the maximum level of weight relief given. The period of acclimatisation required when weight relief was given was unknown. It was generally felt among those with experience using the harness system that accommodation periods of between 2 and 10 minutes were necessary before "one felt comfortable and natural" walking with weight relief. A literature review failed to reveal any specific time periods required to familiarise to such conditions. A pilot study was therefore conducted in an attempt to determine the time required for a number of variables to stabilise. Two previously inexperienced YAB subjects participated in the pilot study. These subjects were harnessed to the system with 20% BWS. It was reasoned that 20% BWS, being the maximum level to be administered, would require the longest familiarisation period.

Once the two subjects were harnessed to the system and 20% BWS was given, they were asked to ambulate along the 15 metre walkway while successive, repeated measurements were taken at 4 minute intervals (approximately) during a 20 minute period. A Footswitch Stride Analyser (B & L Engineering, Santa Fe Springs, California) was utilised to record temporo-spatial data while a Kistler Force platform recorded vertical and antero-posterior GRFs. The use of these apparatus allowed recording of: 1) gait velocity, 2) cadence, 3) stride length, 4) gait cycle duration, 5) right and left stance and swing times, 6) maximum vertical ground reaction force, 7) vertical GRF impulse, and 8) antero-posterior maximum braking and propulsion GRFs.

The results of the pilot study indicated very little change over time or random change over time of all temporo-spatial variables recorded (Figures 36-39). Vertical and antero-posterior GRFs changed inconsistently over the twenty minute period. Slight trends appeared with the maximum vertical (Figure 40) and antero-posterior braking (Figure 42) and propulsion (Figure 43) GRFs increasing throughout the period, while a trend toward a decrease in the vertical impulse (Figure 41) was observed. However, fluctuations over the twenty minute period made it difficult to conclude that a definite trend was apparent. It can be concluded from this pilot study that zero or very little acclimatisation is required in order for most of these variables to stabilise while walking with 20% BWS. Those variables that did not stabilise over the twenty minutes may require longer periods of acclimatisation.

Considering the results of this pilot investigation, it was felt that a period of at least five minutes acclimatisation was sufficient to allow most variables to stabilise with 20% BWS. Randomisation of the weight relief trials will also be employed to minimise the effects of learning and fatigue.

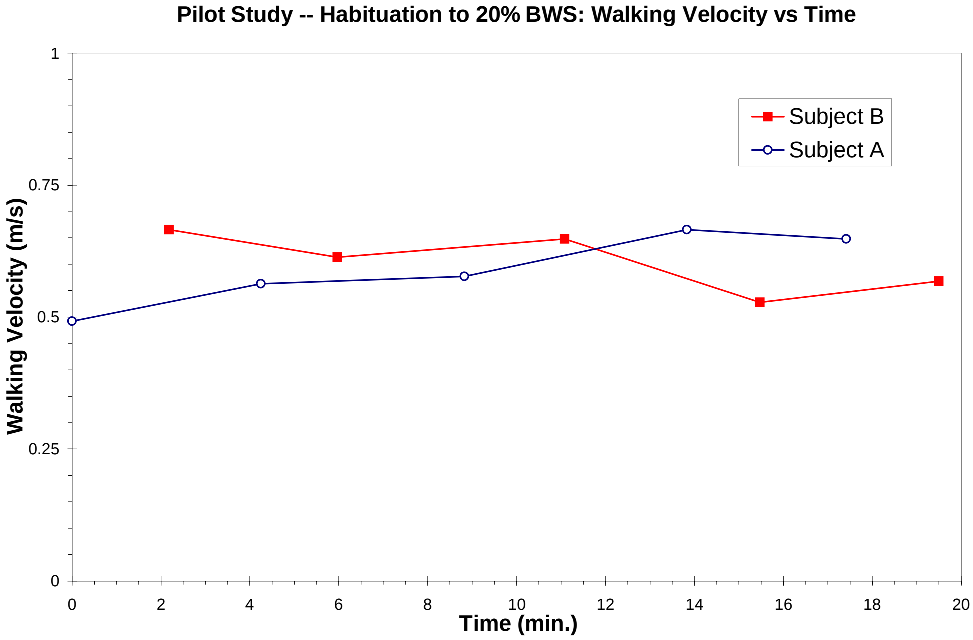


Figure 37. Variation of gait velocity over a twenty minute period.

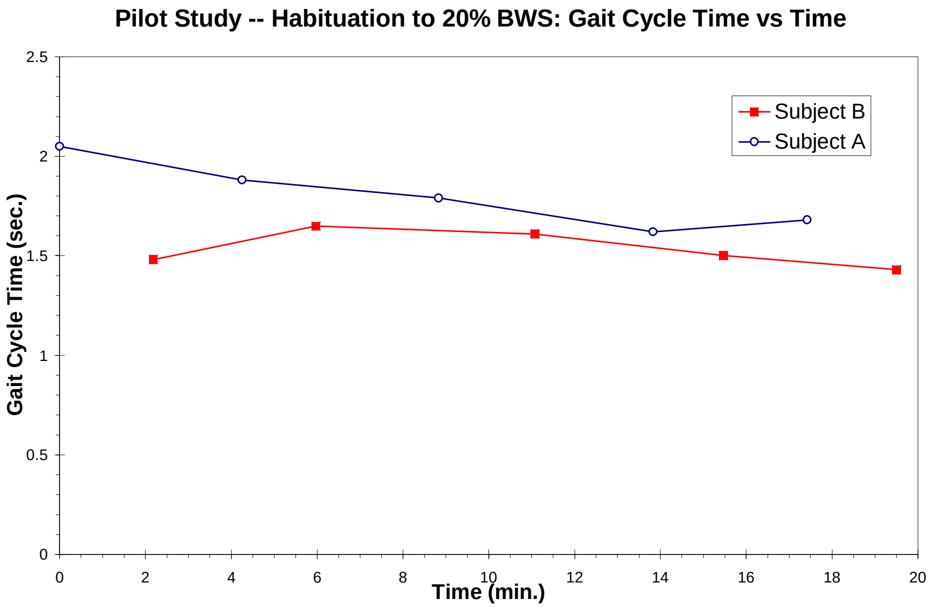
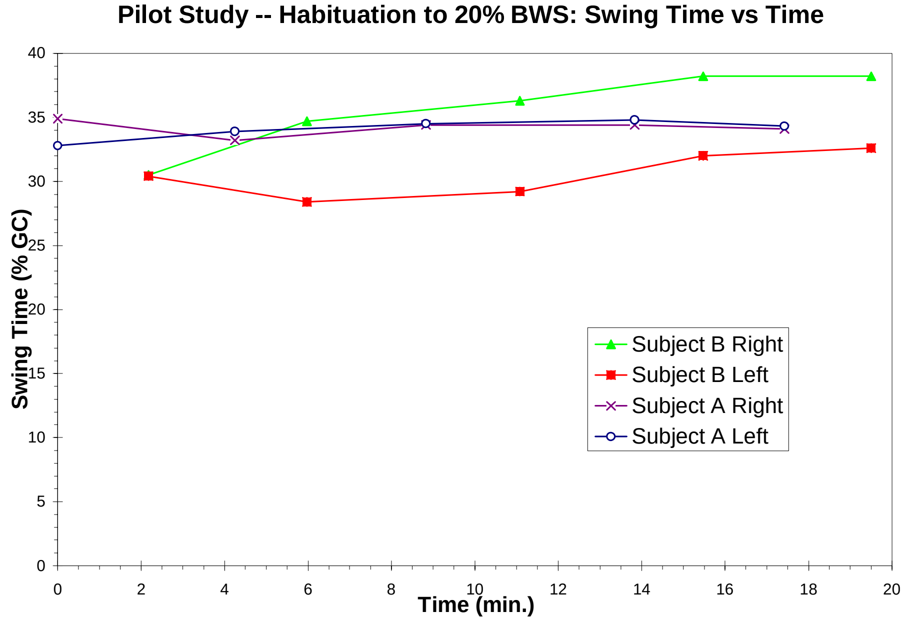
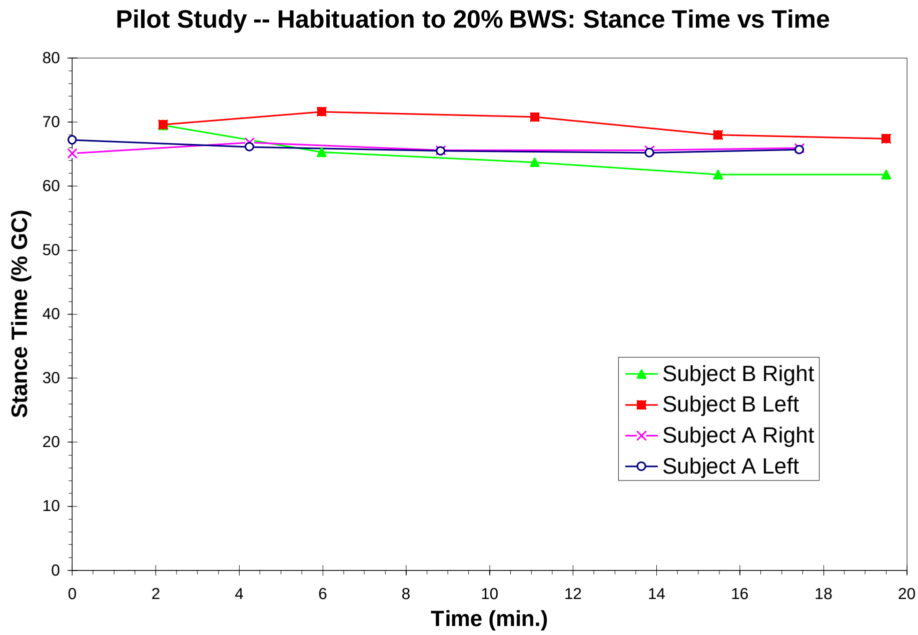


Figure 38. Variation of gait cycle duration over a twenty minute period.



<u>Figure 39.</u> Variation in swing time over a twenty minute period.



<u>Figure 40.</u> Variation in stance time over a twenty minute period.

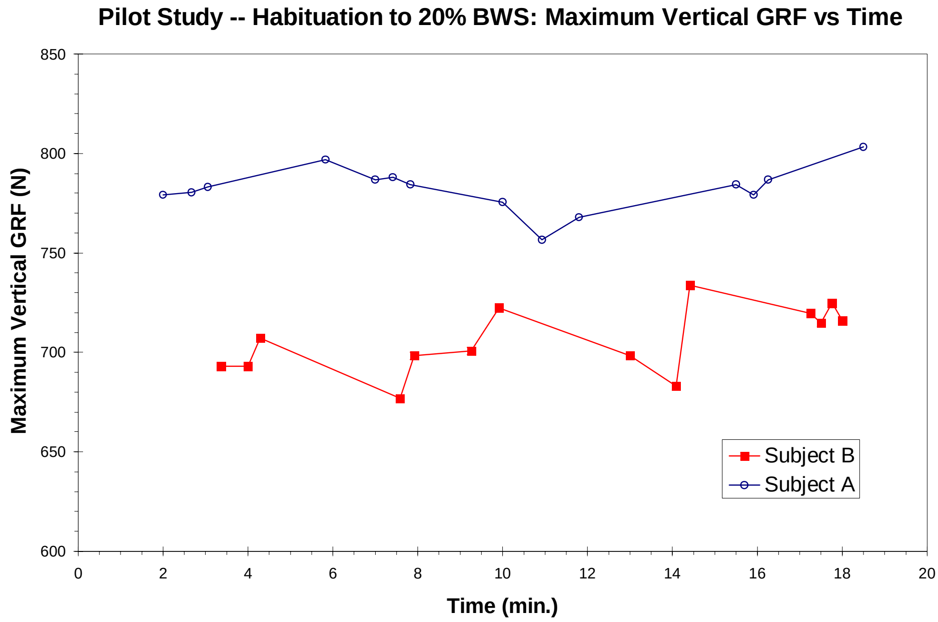


Figure 41. Variation of maximum vertical ground reaction force over a twenty minute period.

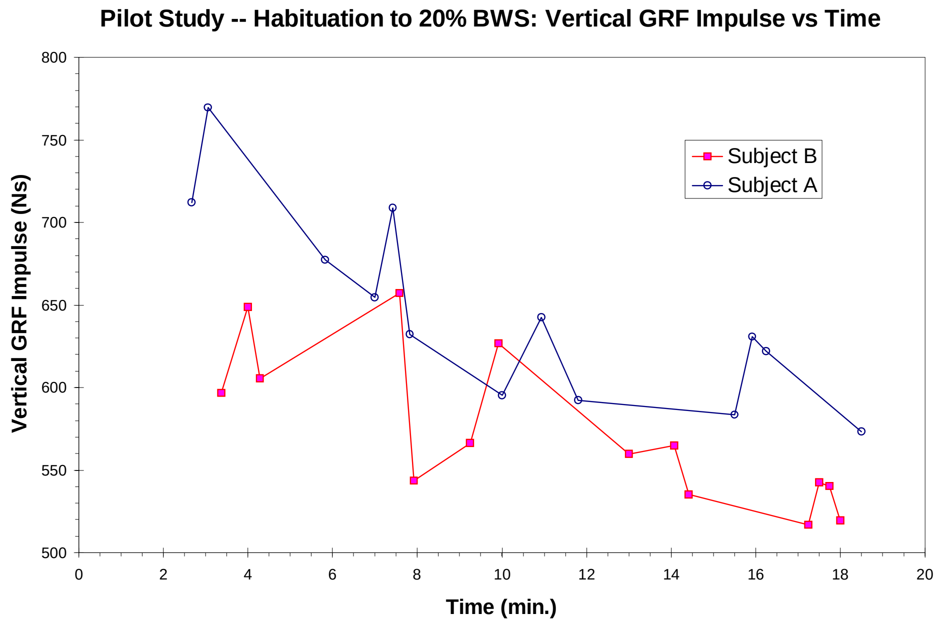


Figure 42. Variation of vertical ground reaction force impulse over a twenty minute period.

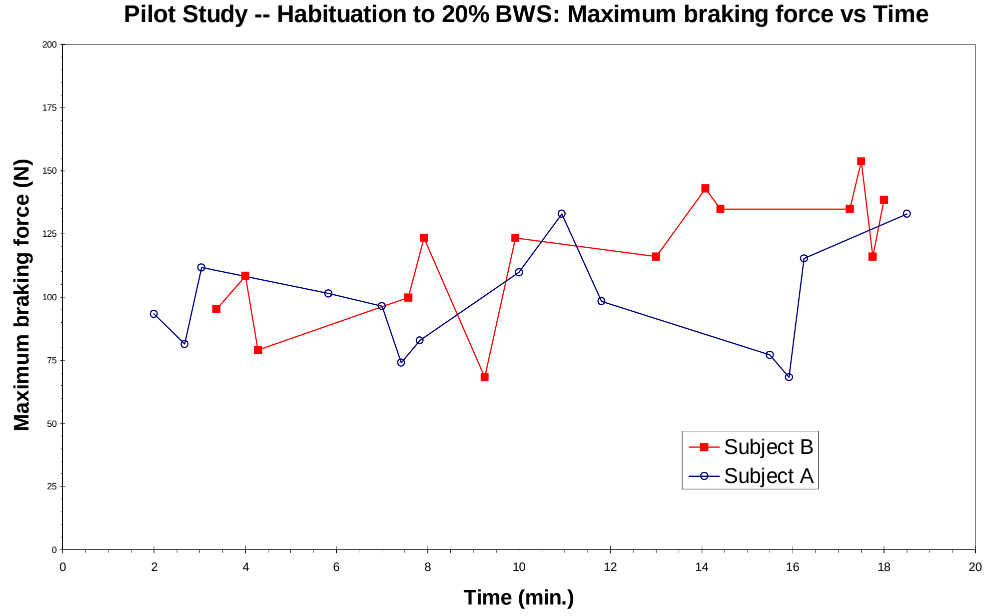


Figure 43. Variation of antero-posterior maximum braking force over a twenty minute period.

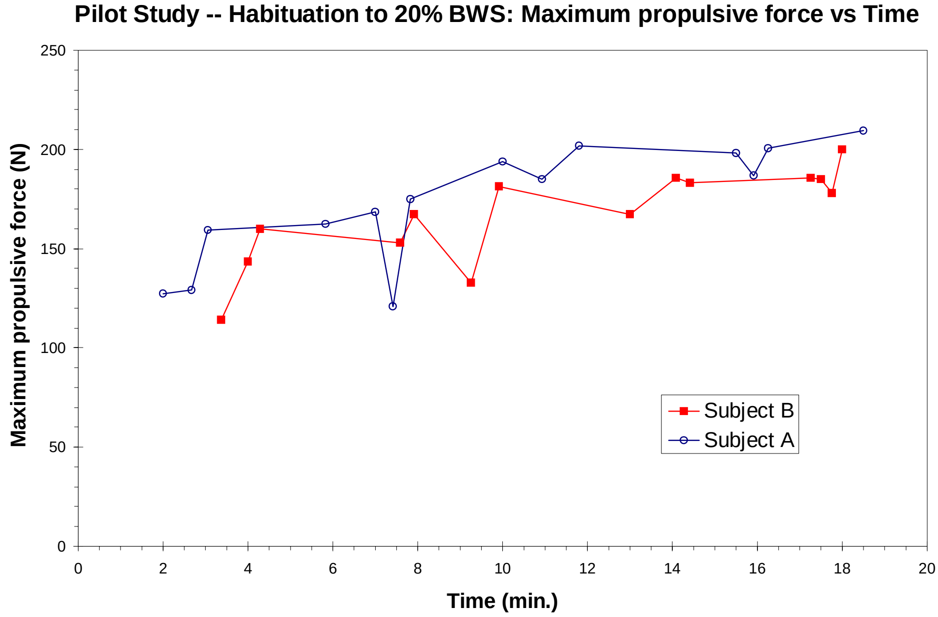


Figure 44. Variation of antero-posterior maximum propulsion force over a twenty minute period.

APPENDIX C

SUBJECT QUESTIONNAIRE

EFFECTS OF HARNESS SUPPORTED WALKING ON ABLE-BODIED SUBJECTS AND LOWER LIMB AMPUTEES

Investigator: Tim Jarrott

Q1. How would you rate the comfort of the harness?

	Responses		
	YAB	EAB	LLA
Very uncomfortable			
Uncomfortable	3	3	2
Not Noticed	2	2	2
Comfortable	4	4	4
Very Comfortable			1

Q2. How would you rate the comfort of the harness system when walking?

	YAB	EAB	LLA
Very uncomfortable			
Uncomfortable	6	5	2
Not noticed		3	2
Comfortable	3	1	4
Very comfortable			1

Q3. When harnessed to the system with no weight relief, do you feel that the harness system:

	YAB	EAB	LLA
Hindered your walking pattern a lot		1	
Hindered your walking pattern a little	6	5	6
Didn’t change your walking pattern	3	3	2
Helped your walking pattern a little			1
Helped your walking pattern a lot			

Q4. When harnessed to the system with 10% relief, do you feel that the harness system:

	YAB	EAB	LLA
Hindered your walking pattern a lot	1		
Hindered your walking pattern a little	8	8	5
Didn't change your walking pattern			1
Helped your walking pattern a little		1	2
Helped your walking pattern a lot			1

Q5. When harnessed to the system with 20% relief, do you feel that the harness system:

Hindered your walking pattern a lot	7	4	3
Hindered your walking pattern a little	1	5	4
Didn't change your walking pattern			1
Helped your walking pattern a little	1		1
Helped your walking pattern a lot			

Q6. With no weight relief, do you feel that you walked:

Much slower		1	
Slightly slower	4	2	3
At the same speed	4	5	5
Slightly faster	1	1	1
Much faster			

Q7. With 10% weight relief, do you feel that you walked:

	YAB	EAB	LLA
Much slower	2		
Slightly slower	5	7	6
At the same speed	2	1	3
Slightly faster		1	
Much faster			

Q8. With 20% weight relief, do you feel that you walked:

	YAB	EAB	LLA
Much slower	7	5	3
Slightly slower	1	3	5
At the same speed	1		1
Slightly faster			
Much faster		1	

Q9. How else do you feel that the harness system affected your walking?

YAB

- Imbalanced - extra weight
- Lost smooth forward motion (jerky)
- Lost forward acceleration - push off
- More effort required to ambulate
- Similar to hovering
- Horizontal force required to move forward increased with weight relief
- Felt like walking in water
- Felt as if dragging something with 20% BWS
- Noise distracting, off balance and restricted
- Reduced pressure on feet

EAB

- Upper body restriction
- Axilla pressure
- 20% BWS created shorter steps
- Resistance apparent
- Walked slower
- Didn't feel in control

LLA

- Loss of traction
- Dragging the system at 0% & 20% BWS
- Less pain/pressure on stump with BWS
- Backward drag at gait initiation
- Increased confidence
- Felt like walking into wind
- Forced upright posture
- Felt as though could walk up to 3-4 times further
- Increased security

10. Do you feel that the harness system:

	YAB	EAB	LLA
Relieved any pain present			2
Made no difference to any pain present			2
Increased any pain present			
No pain present to begin with	9	9	5

Q11. Do you feel that the harness system:

	YAB	EAB	LLA
Increased your balance a lot			2
Increased your balance a little	1	4	2
Made no difference to your balance	4	5	4
Decreased your balance a little	4		1
Decreased your balance a lot			

Q12. When compared to normal walking, do you feel that the harness system made you feel:

Very unsafe			
Unsafe	1		
No different	4	5	
Safe	3	2	8
Very safe	1	2	1

Q13. What effect do you feel the harness system had on loading of your legs (able-bodied -- left versus right; amputee -- sound versus prosthetic):

Load more evenly	2		4
Made no difference	6	9	5
Load more unevenly	1		

Q14. Do you feel that the harness system would help in rehabilitating amputees?

Yes	6	6	7
No			
Unsure	3	3	2

Q15. Do you feel that the harness system has any benefits compared to parallel bar training for amputees?

	YAB	EAB	LLA
Yes -- if so please list a them below.	3	5	6
No	1		1
Unsure	5	4	2

YAB

- Gradual increasing weight
- Psychological security
- Safety
- Upper body and arm strength not required
- Less weight on vulnerable stump

EAB

- Safety and balance
- Removes 4 point style of walking
- Weight relief from arms
- More normal, natural gait pattern

LLA

- More natural gait pattern
- Security and safety
- Confidence
- Less weight through arms
- Reduced pain early walking

Q16. Do you think there are any disadvantages using this system?

YAB

- Restrictive on upper body
- Axilla pressure
- Increased effort to walk
- Reduced ability to achieve push off
- Noise
- Reduced psychological security since there is nothing to hold onto

EAB

- Noise
- Increased muscle work
- Uncomfortable harness

LLA

- Loss of traction especially for first few steps
- Psychological reliance on system
- Can only move in a straight path
- Lower back pain experienced when weight relief removed
- Harness uncomfortable
- Restricted breathing
- More effort required to walk
- Experienced AP distal stump discomfort

APPENDIX D

SUBJECT INFORMED CONSENT FORM

EFFECTS OF HARNESS SUPPORTED WALKING ON ABLE-BODIED SUBJECTS AND LOWER LIMB AMPUTEES

SENIOR INVESTIGATOR: Timothy M. Bach, PhD -- Head, National Centre for Prosthetics and Orthotics.

INVESTIGATOR: Tim Jarrott -- Prosthetics and Orthotics Honours Student (La Trobe University).

The purpose of this investigation is to examine the effects of an overhead harness system on able-bodied subjects and lower limb amputees. The system being tested is capable of providing weight relief (upward lift to reduce the weight of the subject on the ground), safety and support to the user during walking. The harness that is worn on the upper body is a conventional harness (similar to a life vest) that provides support under the armpits and around the waist. Previous research using similar systems has indicated that various benefits are available.

Subjects will be required for one session for approximately one to two hours duration. In this session, subjects will be familiarised with the harness system and the testing equipment prior to testing. Following a period of accommodation, subjects will be required to walk along a 15 metre walkway under the following conditions:

1. normal walking;
2. walking while harnessed to the system (no weight relief);
3. walking while harnessed to the system with 10% weight relief; and,
4. walking while harnessed to the system with 20% weight relief.

Weight relief is provided by a constant upward lift on the jacket that the subject wears. While walking on the walkway under each of the above conditions, subjects will be required to wear inner soles in their shoes. These inner soles are attached to a small pack that is strapped around the waist. This will enable the investigator to determine the pattern of walking (speed etc.). Subjects will also be required to walk over a force platform (a metal plate that measures the weight/force in three directions) that is inset into the walkway. Appropriate rest periods will be given at regular intervals during the testing session. Following the test period, a questionnaire regarding the effects of the system will be administered. Video footage may be obtained for analysis purposes only, this data will not be published or presented unless further written consent is gained.

The investigator and an experienced and qualified clinician will be present at all times during the testing session. It is not expected that the subject will experience any pain or discomfort as a result of the testing apparatus or the procedures. This assumption is based on previous experience using the harness system. Subjects may choose to stop at any time during the tests if they feel discomfort or fatigue.

The results of this study will help investigators understand the effects of an overhead harness system on able-bodied subjects and lower limb amputees. This understanding could lead to regular use of a harness support system during amputee rehabilitation. The potential benefits may eventually lead to faster, safer and easier rehabilitation of lower limb amputees.

Any questions regarding the project titled EFFECTS OF HARNESS SUPPORTED WALKING ON ABLE-BODIED SUBJECTS AND LOWER LIMB AMPUTEES may be directed to the investigator, Tim Jarrott (c/- Rehab Tech -- Ph. 9528 1960), the senior investigator and project supervisor, Dr. Tim Bach (Head, National Centre for Prosthetics and Orthotics, La Trobe University -- Ph. 9285 5311) or the co-supervisor, Mr. Bill Contoyannis (Manager, Rehab Tech).

If you have any complaints about this project or have any questions which the investigator or the project supervisors have not been able to answer satisfactorily, you should write to Mr. David Williams, Secretary, FHEC, Faculty Office, Faculty of Health Sciences, La Trobe University, Bundoora, Victoria, 3083. You are free to withdraw your consent and discontinue participation at any time.

I, ..., have read and understood the information above. Any questions I have had, have been answered to my satisfaction. I agree to participate in this study, realising that I may withdraw at any time. I agree that research data collected during the study may be published or provided to other researchers, on condition that my name is not used.

Name of participant:...

Signature: ....   Date: ....../....../......

Investigator: ....

Signature: ....   Date: ....../....../......

Senior Investigator: ....

Signature: .... Date:..../....../........

APPENDIX E

Young Able-bodied Temporo-spatial Data								
Ambulation Type	Velocity (m/s)	Cadence (steps/ min)	Stride Length (m)	Gait Cycle (sec)	Right Stance (%GC)	Left Stance (%GC)	Right Stance (sec)	Left Stance (sec)
Free-walking 1								
YAB1	0.82	89.10	1.11	1.35	66.80	66.90	0.902	0.903
YAB2	1.69	113.00	1.80	1.07	63.80	63.50	0.683	0.679
YAB3	1.47	111.00	1.59	1.09	65.00	65.30	0.709	0.712
YAB4	1.49	111.00	1.61	1.08	65.60	65.70	0.708	0.710
YAB5	1.43	116.00	1.49	1.04	61.60	61.50	0.641	0.640
YAB6	1.41	119.00	1.43	1.01	62.40	62.20	0.630	0.628
YAB7	1.48	114.00	1.57	1.06	63.40	62.50	0.672	0.663
YAB8	1.62	121.00	1.60	0.99	60.70	61.00	0.601	0.604
YAB9	1.52	129.00	1.41	0.93	62.30	62.50	0.579	0.581
mean	1.44	113.68	1.51	1.07	63.51	63.46	0.681	0.680
std dev	0.25	10.85	0.19	0.12	1.99	2.05	0.094	0.095
Harness 0% BWS								
YAB1	0.80	88.70	1.09	1.35	66.60	66.90	0.899	0.903
YAB2	1.04	87.90	1.42	1.37	64.60	63.90	0.885	0.875
YAB3	1.02	93.70	1.31	1.28	68.60	66.00	0.878	0.845
YAB4	1.48	117.00	1.52	1.02	67.00	66.90	0.683	0.682
YAB5	0.94	95.60	1.17	1.26	63.60	61.20	0.801	0.771
YAB6	1.29	113.00	1.38	1.07	61.10	61.40	0.654	0.657
YAB7	1.15	109.00	1.26	1.10	65.60	62.50	0.722	0.688
YAB8	1.48	120.00	1.48	1.00	61.50	61.50	0.615	0.615
YAB9	1.60	135.00	1.43	0.89	60.70	61.1	0.540	0.544
mean	1.20	106.66	1.34	1.15	64.37	63.49	0.742	0.731
std dev	0.28	16.20	0.14	0.17	2.84	2.50	0.130	0.124
Harness 10% BWS								
YAB1	0.76	85.40	1.07	1.40	65.20	64.20	0.913	0.899
YAB2	0.93	81.10	1.38	1.48	64.50	64.20	0.955	0.950
YAB3	0.93	90.00	1.24	1.33	67.90	68.30	0.903	0.908
YAB4	1.21	101.00	1.45	1.19	66.20	62.10	0.788	0.739
YAB5	0.75	81.00	1.11	1.48	62.60	61.20	0.926	0.906
YAB6	1.18	104.00	1.37	1.16	60.40	59.90	0.701	0.695
YAB7	1.10	97.10	1.36	1.24	61.30	62.10	0.760	0.770
YAB8	1.38	110.00	1.51	1.09	61.90	59.90	0.675	0.653
YAB9	1.42	117.00	1.46	1.03	60.20	60.00	0.620	0.618
mean	1.07	96.29	1.33	1.27	63.36	62.43	0.804	0.793
std dev	0.25	12.85	0.16	0.17	2.72	2.76	0.124	0.125

Harness 20% BWS								
YAB1	0.54	67.40	0.97	1.78	66.50	66.00	1.184	1.175
YAB2	0.78	73.90	1.27	1.62	62.80	62.20	1.017	1.008
YAB3	0.63	64.70	1.16	1.86	66.70	67.40	1.241	1.254
YAB4	0.96	91.00	1.26	1.32	65.80	61.10	0.869	0.807
YAB5	0.66	70.00	1.12	1.71	62.40	60.40	1.067	1.033
YAB6	0.95	86.50	1.32	1.39	59.20	59.60	0.823	0.828
YAB7	1.03	94.00	1.32	1.28	62.00	60.80	0.794	0.778
YAB8	1.09	94.00	1.39	1.28	61.50	62.40	0.787	0.799
YAB9	1.16	109.00	1.27	1.10	61.90	61.80	0.681	0.680
mean	0.87	83.39	1.23	1.48	63.20	62.41	0.940	0.929
std dev	0.22	15.10	0.13	0.27	2.57	2.61	0.194	0.197
Free-walking 2								
YAB1	1.17	108.00	1.29	1.11	66.50	66.00	0.738	0.733
YAB2	1.87	100.00	2.24	1.20	64.10	64.30	0.769	0.772
YAB3	1.25	102.00	1.48	1.18	67.00	62.70	0.791	0.740
YAB4	1.61	117.00	1.65	1.02	64.30	64.90	0.656	0.662
YAB5	1.30	111.00	1.41	1.08	62.70	60.70	0.677	0.656
YAB6	1.55	124.00	1.50	0.97	62.40	61.50	0.605	0.597
YAB7	1.33	109.00	1.47	1.10	63.50	62.00	0.699	0.682
YAB8	1.60	121.00	1.59	0.99	61.10	61.40	0.605	0.608
YAB9	1.80	132.00	1.64	0.91	60.70	61.90	0.552	0.563
mean	1.50	113.78	1.59	1.06	63.59	62.82	0.677	0.668
std dev	0.25	10.56	0.27	0.10	2.17	1.82	0.081	0.071

Elderly Able-bodied Temporo-spatial Data								
Ambulation Type	**Velocity (m/s)**	**Cadence (steps/ min)**	**Stride Length (m)**	**Gait Cycle (sec)**	**Right Stance (%GC)**	**Left Stance (%GC)**	**Right Stance Time (sec)**	**Left Stance Time (sec)**
Free-walking 1								
EAB1	1.48	130.00	1.37	0.92	61.90	62.40	0.57	0.57
EAB2	1.40	101.00	1.66	1.19	62.70	62.50	0.75	0.74
EAB3	1.39	104.00	1.60	1.15	61.60	62.00	0.71	0.71
EAB4	1.25	109.00	1.38	1.10	63.50	60.20	0.70	0.66
EAB5	1.21	100.00	1.55	1.20	62.80	63.50	0.75	0.76
EAB6	1.09	111.00	1.19	1.09	66.10	66.20	0.72	0.72
EAB7	0.90	116.00	0.93	1.04	68.00	64.60	0.71	0.67
EAB8	1.47	111.00	1.59	1.08	65.20	65.40	0.70	0.71
EAB9	1.22	119.00	1.23	1.01	64.80	64.40	0.65	0.65
mean	1.27	111.22	1.39	1.09	64.07	63.47	0.70	0.69
std dev	0.19	9.51	0.24	0.09	2.12	1.88	0.06	0.06
Harness 0% BWS								
EAB1	1.42	128.00	1.33	0.94	62.80	62.50	0.59	0.59
EAB2	1.34	102.00	1.57	1.17	63.70	63.00	0.75	0.74
EAB3	1.43	108.00	1.59	1.11	61.80	62.00	0.69	0.69
EAB4	1.30	114.00	1.37	1.05	63.60	58.20	0.67	0.61
EAB5	1.05	92.90	1.36	1.29	62.80	63.30	0.81	0.82
EAB6	1.08	113.00	1.15	1.07	65.60	66.00	0.70	0.71
EAB7	0.75	109.00	0.83	1.10	68.40	65.30	0.75	0.72
EAB8	1.31	106.00	1.48	1.13	65.10	64.50	0.74	0.73
EAB9	1.00	105.00	1.14	1.15	65.90	64.3	0.76	0.74
mean	1.19	108.66	1.31	1.11	64.41	63.23	0.72	0.70
std dev	0.23	9.59	0.24	0.09	2.04	2.29	0.06	0.07
Harness 10% BWS								
EAB1	1.30	124.00	1.25	0.97	61.40	61.70	0.60	0.60
EAB2	1.20	91.20	1.58	1.32	61.50	59.80	0.81	0.79
EAB3	1.19	103.00	1.40	1.17	62.60	62.10	0.73	0.73
EAB4	1.15	108.00	1.28	1.12	63.20	60.20	0.71	0.67
EAB5	0.87	84.90	1.23	1.41	61.80	62.60	0.87	0.88
EAB6	0.99	107.00	1.11	1.13	64.90	65.50	0.73	0.74
EAB7	0.66	110.00	0.72	1.09	66.20	64.80	0.72	0.71
EAB8	1.03	89.30	1.38	1.34	64.30	64.70	0.86	0.87
EAB9	0.84	87.80	1.14	1.37	63.90	65.10	0.88	0.89
mean	1.03	100.58	1.23	1.21	63.31	62.94	0.77	0.76
std dev	0.21	13.06	0.24	0.15	1.66	2.17	0.09	0.10

Harness 20% BWS								
EAB1	0.88	97.40	1.08	1.23	60.30	63.40	0.74	0.78
EAB2	0.96	79.80	1.44	1.50	61.40	61.20	0.92	0.92
EAB3	0.82	88.40	1.11	1.36	60.20	60.80	0.82	0.83
EAB4	0.90	87.80	1.22	1.37	62.30	59.50	0.85	0.82
EAB5	0.60	70.20	1.02	1.71	58.60	61.50	1.00	1.05
EAB6	0.87	102.00	1.02	1.18	65.10	66.10	0.77	0.78
EAB7	0.53	99.50	0.64	1.21	66.00	64.50	0.80	0.78
EAB8	0.74	68.10	1.29	1.76	64.40	64.90	1.13	1.14
EAB9	0.62	70.10	1.06	1.71	66.10	66.20	1.13	1.13
mean	0.77	84.81	1.10	1.45	62.71	63.12	0.91	0.91
std dev	0.15	13.35	0.22	0.23	2.78	2.46	0.15	0.15
Free-walking 2								
EAB1	1.69	133.00	1.52	0.90	61.10	62.20	0.55	0.56
EAB2	1.13	106.00	1.30	1.14	63.20	62.50	0.72	0.71
EAB3	1.44	107.00	1.61	1.12	62.60	62.00	0.70	0.69
EAB4	1.36	116.00	1.41	1.03	63.40	60.40	0.65	0.62
EAB5	1.23	102.00	1.45	1.18	62.60	62.80	0.74	0.74
EAB6	1.22	115.00	1.27	1.04	64.20	65.60	0.67	0.68
EAB7	1.06	124.00	1.03	0.97	66.70	65.80	0.65	0.64
EAB8	1.41	109.00	1.55	1.10	66.50	64.40	0.73	0.71
EAB9	1.21	115.00	1.27	1.04	63.90	63.40	0.66	0.66
mean	1.31	114.11	1.38	1.06	63.80	63.23	0.67	0.67
std dev	0.19	9.68	0.18	0.09	1.82	1.76	0.06	0.06

Lower Limb Amputee Temporo-spatial Data								
Ambulation Type	**Velocity (m/s)**	**Cadence (steps/ min)**	**Stride Length (m)**	**Gait Cycle (sec)**	**Sound Stance (%GC)**	**Pros. Stance (%GC)**	**Sound Stance (sec)**	**Pros. Stance (sec)**
Free-walking 1								
LLA1	0.82	81.20	1.22	1.48	72.50	61.60	1.073	0.912
LLA2	1.06	100.00	1.27	1.20	68.70	64.70	0.824	0.776
LLA3	1.21	100.00	1.45	1.20	67.60	63.70	0.811	0.764
LLA4	0.89	93.10	1.15	1.29	68.80	66.70	0.888	0.860
LLA5	0.84	103.00	0.98	1.17	66.00	71.70	0.772	0.839
LLA6	1.33	101.00	1.58	1.19	67.50	65.10	0.803	0.775
LLA7	0.97	82.50	1.41	1.46	68.20	61.50	0.996	0.898
LLA8	0.33	49.10	0.81	2.44	76.40	65.20	1.864	1.591
LLA9	1.02	92.80	1.32	1.29	69.10	65.60	0.891	0.846
mean	0.94	89.19	1.24	1.41	69.42	65.09	0.991	0.918
std dev	0.28	16.99	0.24	0.40	3.15	3.04	0.342	0.258
Harness 0% BWS								
LLA1	0.71	81.50	1.22	1.47	74.80	61.60	1.100	0.906
LLA2	1.14	101.00	1.36	1.19	67.30	64.10	0.801	0.763
LLA3	1.07	98.10	1.31	1.22	72.60	61.30	0.886	0.748
LLA4	0.72	94.50	0.91	1.27	68.10	69.10	0.865	0.878
LLA5	0.79	98.20	0.96	1.22	68.30	70.50	0.833	0.860
LLA6	1.36	107.00	1.52	1.12	63.50	63.20	0.711	0.708
LLA7	0.90	79.40	1.35	1.51	68.80	59.60	1.039	0.900
LLA8	0.36	51.50	0.84	2.33	75.40	66.00	1.757	1.538
LLA9	1.06	97.90	1.29	1.23	68.90	67.00	0.847	0.824
mean	0.90	89.90	1.20	1.40	69.74	64.71	0.982	0.903
std dev	0.29	16.93	0.24	0.37	3.83	3.70	0.314	0.248
Harness 10% BWS								
LLA1	0.62	69.60	1.07	1.73	74.30	62.80	1.285	1.086
LLA2	0.84	87.30	1.15	1.37	69.80	64.50	0.956	0.884
LLA3	0.87	84.10	1.25	1.43	68.40	61.10	0.978	0.874
LLA4	0.68	93.70	0.87	1.28	66.70	68.30	0.854	0.874
LLA5	0.58	89.50	0.78	1.34	73.90	73.10	0.990	0.980
LLA6	1.30	104.00	1.51	1.16	63.50	63.00	0.737	0.731
LLA7	0.74	71.50	1.25	1.68	69.90	58.90	1.174	0.990
LLA8	0.34	48.10	0.84	2.49	78.90	68.10	1.965	1.696
LLA9	0.86	85.80	1.19	1.40	70.10	66.30	0.981	0.928
mean	0.76	81.51	1.10	1.54	70.61	65.12	1.102	1.005
std dev	0.26	16.34	0.24	0.40	4.55	4.32	0.361	0.277

Harness 20% BWS								
LLA1	0.55	62.70	1.06	1.91	74.60	62.00	1.425	1.184
LLA2	0.66	76.00	1.05	1.58	70.20	59.40	1.109	0.939
LLA3	0.57	62.00	1.09	1.93	69.60	59.50	1.343	1.148
LLA4	0.55	88.80	0.74	1.35	66.80	68.80	0.902	0.929
LLA5	0.46	81.60	0.68	1.47	71.70	70.40	1.054	1.035
LLA6	0.90	84.30	1.28	1.42	68.20	61.20	0.968	0.869
LLA7	0.62	58.50	1.27	2.05	66.50	57.90	1.363	1.187
LLA8	0.25	51.10	0.58	2.35	81.50	71.10	1.915	1.671
LLA9	0.70	79.50	1.05	1.51	72.70	64.70	1.098	0.977
mean	0.58	71.61	0.98	1.73	71.31	63.89	1.242	1.104
std dev	0.18	13.24	0.25	0.34	4.66	5.07	0.311	0.243

APPENDIX F

YOUNG ABLE-BODIED GRF DATA						
	Vertical GRF				Antero-posterior GRF	
	1st Max (%BW)	**Min (%BW)**	**2nd Max (%BW)**	**Impulse (%BW)**	**Braking (%BW)**	**Propulsion (%BW)**
Free Walk 1 - Right						
YAB1	104.80	74.30	109.60	61.64	16.60	17.80
YAB2	110.30	70.00	120.80	85.07	22.70	26.35
YAB3	122.50	67.00	114.90	81.81	22.80	24.25
YAB4	133.90	49.50	129.10	76.41	35.70	31.35
YAB5	115.10	72.70	109.50	80.97	24.35	22.65
YAB6	114.00	73.60	109.70	76.35	21.00	22.20
YAB7	126.10	58.30	123.70	78.87	28.50	26.60
YAB8	121.00	63.80	129.00	84.53	23.65	28.60
YAB9	123.30	66.60	106.00	79.10	19.90	21.80
mean	119.00	66.20	116.92	78.30	23.91	24.62
std dev	8.84	8.10	8.94	6.98	5.49	4.07
Free Walk 1 - Left						
YAB1	100.50	78.80	106.90	62.13	14.10	18.50
YAB2	110.20	73.10	117.20	85.57	23.45	27.50
YAB3	118.60	68.00	118.60	78.51	23.00	23.90
YAB4	132.20	55.80	119.40	76.48	36.60	29.70
YAB5	122.30	69.50	111.80	82.03	22.45	21.85
YAB6	117.20	69.80	105.90	78.98	25.15	19.25
YAB7	118.50	64.40	120.00	78.28	24.20	25.00
YAB8	123.90	65.10	123.20	86.09	31.00	25.55
YAB9	122.40	63.90	99.20	73.67	18.50	20.10
mean	118.42	67.60	113.58	77.97	24.27	23.48
std dev	8.95	6.46	8.06	7.19	6.52	3.85
Harness 0% - Right						
YAB1	92.30	86.50	93.60	72.41	9.35	10.50
YAB2	100.10	90.60	103.10	79.66	10.65	16.45
YAB3	106.40	81.60	108.00	73.46	16.45	17.80
YAB4	119.70	63.40	124.90	79.65	27.55	30.80
YAB5	106.40	88.80	100.50	80.90	15.25	16.40
YAB6	101.60	83.20	97.10	79.66	15.10	18.50
YAB7	109.20	77.20	109.00	76.45	18.75	18.60
YAB8	113.40	77.40	113.10	86.50	20.90	21.80
YAB9	113.90	73.50	100.30	90.56	16.90	22.05
mean	107.00	80.24	105.51	79.92	16.77	19.21
std dev	8.28	8.49	9.50	5.78	5.42	5.52

Harness 0% - Left						
YAB1	94.60	84.20	92.00	66.00	6.70	12.00
YAB2	100.70	89.50	104.70	78.97	10.75	16.35
YAB3	103.90	83.40	107.00	75.50	13.90	17.65
YAB4	112.20	65.40	113.80	76.39	22.50	26.40
YAB5	108.30	83.90	102.70	77.04	12.35	19.65
YAB6	100.60	80.70	98.00	78.69	14.00	16.85
YAB7	112.40	78.70	112.50	80.96	23.50	18.40
YAB8	116.60	70.20	115.30	80.00	20.25	21.55
YAB9	106.30	74.70	99.10	83.82	13.85	19.85
mean	106.18	78.97	105.01	77.49	15.31	18.74
std dev	6.97	7.62	7.92	4.99	5.62	3.95
Harness 10% - Right						
YAB1	95.20	82.10	92.60	67.91	8.05	10.30
YAB2	93.40	86.70	89.60	76.54	9.75	14.35
YAB3	100.50	79.50	94.40	70.87	12.30	14.25
YAB4	100.20	66.50	97.60	68.15	17.10	24.55
YAB5	100.80	81.00	86.70	63.61	12.25	16.40
YAB6	95.50	76.80	86.80	79.74	11.35	13.45
YAB7	99.20	76.50	90.40	74.08	11.75	15.85
YAB8	105.10	72.70	103.30	75.56	14.55	21.80
YAB9	101.80	69.20	89.90	78.23	11.85	18.50
mean	99.08	76.78	92.37	72.74	12.11	16.61
std dev	3.71	6.44	5.38	5.42	2.59	4.40
Harness 10% - Left						
YAB1	93.00	83.60	89.60	66.85	6.90	11.45
YAB2	94.10	85.10	90.10	73.58	8.20	14.05
YAB3	99.40	78.80	95.30	68.94	12.30	13.65
YAB4	106.10	63.30	97.50	68.74	14.30	25.50
YAB5	99.80	82.30	90.20	67.55	11.80	12.45
YAB6	96.30	78.50	87.90	77.27	9.65	15.45
YAB7	96.40	75.40	88.10	69.35	9.40	16.95
YAB8	104.50	68.50	105.00	75.04	14.95	18.00
YAB9	97.90	70.40	89.00	74.92	9.70	15.35
mean	98.61	76.21	92.52	71.36	10.80	15.87
std dev	4.41	7.44	5.72	3.83	2.72	4.16

Harness 20% - Right						
YAB1	84.70	73.50	77.00	55.83	5.80	9.05
YAB2	79.90	69.20	70.30	65.29	4.95	12.90
YAB3	88.80	74.50	81.60	56.73	7.45	10.85
YAB4	86.00	62.90	77.80	68.47	8.60	21.00
YAB5	83.30	78.30	81.50	65.14	7.40	11.85
YAB6	89.60	73.00	83.40	69.87	8.50	12.35
YAB7	90.10	70.80	83.50	70.03	7.25	15.30
YAB8	90.20	72.30	85.40	65.44	7.15	17.80
YAB9	93.10	71.20	77.70	73.57	9.15	15.35
mean	87.30	71.74	79.80	65.59	7.36	14.05
std dev	4.13	4.20	4.62	5.96	1.34	3.71
Harness 20% - Left						
YAB1	81.50	75.10	77.20	55.06	3.40	10.75
YAB2	79.50	70.60	76.20	65.38	5.25	12.10
YAB3	87.40	75.20	80.10	52.55	5.85	10.60
YAB4	85.60	64.00	74.40	74.85	8.70	18.50
YAB5	85.50	77.60	82.70	65.54	7.10	10.35
YAB6	90.40	72.80	80.80	66.91	7.30	13.75
YAB7	88.70	70.90	86.20	69.54	7.55	18.90
YAB8	89.10	69.90	81.90	68.84	7.60	16.40
YAB9	93.70	70.40	81.60	69.85	9.20	16.70
mean	86.82	71.83	80.12	65.39	6.88	14.23
std dev	4.39	3.96	3.64	7.18	1.79	3.46
Free Walk 2 - Right						
YAB1	103.70	79.10	106.10	79.40	16.60	16.75
YAB2	100.90	84.80	113.70	84.14	16.30	22.45
YAB3	115.50	72.20	112.80	76.36	21.55	20.75
YAB4	136.70	47.90	130.80	82.77	37.60	31.30
YAB5	110.60	81.80	105.10	88.04	20.60	19.00
YAB6	120.10	65.30	110.30	82.81	25.60	21.45
YAB7	118.10	67.80	118.10	75.25	21.65	25.40
YAB8	123.60	64.80	123.70	83.31	25.45	25.20
YAB9	122.70	60.40	110.80	88.41	23.70	25.90
mean	116.88	69.34	114.60	82.28	23.23	23.13
std dev	10.92	11.60	8.34	4.59	6.34	4.34

Free Walk 2 - Left						
YAB1	101.90	81.90	104.30	79.67	15.20	17.85
YAB2	100.50	85.60	111.90	84.20	13.40	23.90
YAB3	113.70	72.80	105.00	75.95	21.30	19.45
YAB4	126.40	56.20	120.30	81.12	33.40	28.10
YAB5	116.10	78.90	105.10	85.98	17.05	19.10
YAB6	116.40	69.90	104.60	84.59	24.05	21.80
YAB7	119.00	65.40	119.50	73.75	20.00	26.65
YAB8	123.00	61.10	123.30	83.39	26.35	24.65
YAB9	125.30	64.10	98.30	75.67	20.20	23.00
mean	115.81	70.66	110.26	80.48	21.22	22.72
std dev	9.33	9.96	8.83	4.46	6.12	3.50

ELDERLY ABLE-BODIED GRF DATA						
	Vertical GRF				Antero-posterior GRF	
	1st Max (%BW)	**Min (%BW)**	**2nd Max (%BW)**	**Impulse (%BW)**	**Deccn. (%BW)**	**Accn. (%BW)**
Free Walk 1 - Right						
EAB1	128.40	57.00	116.70	78.91	29.00	25.25
EAB2	133.20	67.10	106.90	83.51	23.20	23.20
EAB3	117.20	67.40	109.10	80.08	20.95	21.20
EAB4	119.70	63.90	103.30	75.43	25.80	24.05
EAB5	112.90	78.10	110.30	83.82	17.40	21.50
EAB6	100.70	82.00	102.60	74.86	15.80	19.85
EAB7	100.90	80.50	107.10	76.10	14.45	13.00
EAB8	104.10	88.00	101.30	86.55	14.75	15.95
EAB9	104.70	80.00	113.00	83.64	17.30	21.45
mean	113.53	73.78	107.81	80.32	19.85	20.61
std dev	11.99	10.23	5.06	4.27	5.19	3.91
Free Walk 1 - Left						
EAB1	134.90	57.20	119.10	78.22	28.80	25.55
EAB2	129.90	62.20	116.10	85.35	21.75	26.15
EAB3	113.90	67.90	111.60	80.93	22.55	19.90
EAB4	119.90	63.70	108.40	77.62	24.55	25.65
EAB5	113.40	78.70	106.50	79.92	18.30	19.45
EAB6	107.10	82.30	103.80	73.96	16.35	16.85
EAB7	97.70	86.00	104.90	83.93	11.10	10.80
EAB8	98.20	83.40	105.30	85.80	9.20	14.90
EAB9	108.50	78.00	108.90	82.00	19.40	19.20
mean	113.72	73.27	109.40	80.86	19.11	19.83
std dev	12.83	10.60	5.27	3.89	6.26	5.27
Harness 0% - Right						
EAB1	111.00	73.00	114.60	82.20	17.95	21.85
EAB2	124.80	77.70	102.50	94.09	21.65	11.50
EAB3	107.00	80.80	100.50	85.86	13.95	17.25
EAB4	119.70	73.80	100.10	81.93	29.40	18.95
EAB5	98.20	90.30	100.80	94.81	12.20	16.60
EAB6	97.30	92.00	105.50	117.66	12.55	21.50
EAB7	102.20	90.00	100.50	92.55	9.90	8.80
EAB8	103.80	88.60	97.50	91.02	15.85	12.85
EAB9	100.80	91.30	105.60	91.16	13.25	16.90
mean	107.20	84.17	103.07	92.37	16.30	16.24
std dev	9.60	7.82	5.05	10.65	6.02	4.44

Harness 0% - Left						
EAB1	118.20	71.30	119.30	85.03	20.95	21.85
EAB2	119.30	77.00	102.20	88.47	20.95	19.60
EAB3	103.80	80.90	106.60	87.06	15.65	17.30
EAB4	111.40	76.70	96.10	82.44	27.15	17.60
EAB5	98.10	90.10	97.20	96.08	13.35	16.80
EAB6	96.80	92.60	102.80	108.50	13.00	17.95
EAB7	100.20	90.30	100.88	98.19	9.95	9.00
EAB8	99.40	85.70	104.00	95.74	9.90	13.20
EAB9	100.20	92.80	103.10	95.40	13.10	15.30
mean	105.27	84.16	103.58	92.99	16.00	16.51
std dev	8.75	7.94	6.74	8.07	5.83	3.72
Harness 10% - Right						
EAB1	103.00	76.00	102.90	81.38	12.15	18.00
EAB2	109.40	72.90	82.80	80.05	14.35	16.10
EAB3	106.40	78.60	93.30	79.64	14.20	14.15
EAB4	86.00	80.60	85.70	70.07	14.55	12.35
EAB5	94.30	78.90	84.60	79.56	7.40	14.95
EAB6	93.90	81.40	89.40	95.36	12.25	18.00
EAB7	92.30	83.60	89.80	89.47	7.50	7.95
EAB8	94.90	83.40	87.10	89.69	14.75	11.75
EAB9	94.10	87.60	92.70	81.71	10.10	15.95
mean	97.14	80.33	89.81	82.99	11.92	14.36
std dev	7.51	4.38	6.05	7.44	2.95	3.25
Harness 10% - Left						
EAB1	109.80	73.80	109.60	80.27	15.10	19.00
EAB2	106.90	70.00	79.40	80.10	14.15	18.40
EAB3	102.40	79.60	94.40	82.39	15.45	13.45
EAB4	87.30	76.20	90.50	67.82	7.75	17.55
EAB5	92.90	79.00	86.30	84.37	9.00	13.40
EAB6	94.00	81.80	88.40	87.97	10.75	15.80
EAB7	90.90	85.20	91.90	104.82	7.40	9.10
EAB8	94.00	83.20	91.20	91.84	7.15	12.85
EAB9	93.30	87.40	92.60	76.68	10.05	14.15
mean	96.83	79.58	91.59	84.03	10.76	14.86
std dev	7.67	5.56	8.07	10.35	3.34	3.16

Harness 20% - Right						
EAB1	89.90	71.80	81.60	68.87	4.90	15.00
EAB2	89.80	63.80	71.90	60.59	7.75	13.20
EAB3	86.7	70.4	79.8	73.02	6.75	11.85
EAB4	81.60	71.70	73.60	62.40	8.05	9.10
EAB5	80.00	69.70	70.60	77.84	4.10	10.40
EAB6	83.60	70.40	81.50	79.30	7.85	15.60
EAB7	84.80	78.10	82.60	76.35	4.00	8.40
EAB8	82.80	68.20	73.80	58.43	6.25	10.45
EAB9	80.80	77.60	79.80	73.81	5.95	12.35
mean	84.44	71.30	77.24	70.07	6.18	11.82
std dev	3.68	4.42	4.70	7.85	1.58	2.49
Harness 20% - Left						
EAB1	91.00	65.90	95.20	60.00	9.20	15.90
EAB2	94.90	59.80	64.10	62.31	10.50	19.00
EAB3	84.60	70.50	82.30	70.37	8.05	10.95
EAB4	83.60	71.20	72.50	62.78	7.10	9.90
EAB5	82.90	70.20	71.60	69.58	5.10	11.55
EAB6	80.60	71.50	82.00	80.26	7.80	12.55
EAB7	84.30	76.40	80.90	72.56	4.85	8.60
EAB8	82.90	69.50	78.10	62.82	2.85	13.15
EAB9	79.80	76.70	80.60	70.76	6.80	12.80
mean	84.96	70.19	78.59	67.94	6.92	12.71
std dev	4.90	5.13	8.71	6.49	2.35	3.15
Free Walk 2 - Right						
EAB1	122.40	61.60	116.50	83.27	25.25	26.45
EAB2	130.80	70.30	100.40	84.17	23.50	17.90
EAB3	111.50	74.00	109.80	82.17	17.55	19.90
EAB4	127.20	64.80	96.00	72.40	29.20	18.15
EAB5	111.20	81.00	107.00	86.20	15.90	21.85
EAB6	100.40	80.10	108.40	88.47	19.75	24.35
EAB7	99.90	83.60	106.40	89.03	13.25	14.20
EAB8	103.90	84.70	103.60	84.38	15.65	16.45
EAB9	101.80	76.50	114.50	76.54	15.65	21.25
mean	112.12	75.18	106.96	82.96	19.52	20.06
std dev	11.96	8.21	6.45	5.41	5.35	3.87

Free Walk 2 - Left						
EAB1	131.50	60.30	122.80	84.46	30.05	24.65
EAB2	121.80	70.00	107.50	88.36	21.00	22.45
EAB3	110.50	77.30	100.80	85.73	17.35	17.60
EAB4	130.20	59.50	102.10	77.12	30.25	24.45
EAB5	109.60	80.50	105.10	89.34	19.40	19.50
EAB6	104.90	78.70	107.40	83.14	20.25	18.90
EAB7	104.70	84.60	106.70	98.59	14.70	11.50
EAB8	103.40	80.50	108.10	84.24	10.25	16.35
EAB9	104.90	77.20	110.70	78.91	17.10	20.10
mean	113.50	74.29	107.91	85.54	20.04	19.50
std dev	11.31	9.04	6.36	6.29	6.58	4.15

LOWER LIMB AMPUTEE GRF DATA						
	Vertical GRF				Antero-posterior GRF	
	1st Max (%BW)	**Min (%BW)**	**2nd Max (%BW)**	**Impulse (%BW)**	**Braking (%BW)**	**Propulsion (%BW)**
Free Walk - Sound						
LLA1	103.00	85.50	107.30	75.21	15.70	13.55
LLA2	131.60	71.80	100.90	77.43	22.60	16.55
LLA3	106.00	80.50	103.10	76.70	18.00	20.25
LLA4	107.90	89.80	99.10	84.01	11.10	10.65
LLA5	108.10	86.50	93.40	74.87	15.75	10.80
LLA6	119.10	75.70	108.70	77.21	19.25	24.40
LLA7	106.00	87.60	108.50	93.67	11.60	19.80
LLA8	100.10	85.90	90.80	82.62	6.80	8.05
LLA9	124.80	80.40	98.40	81.37	21.95	15.65
mean	111.84	82.63	101.13	80.34	15.86	15.52
std dev	10.75	5.97	6.44	5.97	5.26	5.32
Free Walk - Prosthetic						
LLA1	67.10	63.30	65.40	38.05	2.35	4.65
LLA2	116.70	72.10	95.00	72.16	14.45	14.35
LLA3	105.40	79.70	97.00	72.38	12.70	14.35
LLA4	101.20	89.10	98.80	77.56	6.65	11.50
LLA5	140.60	74.00	98.40	70.44	19.85	9.55
LLA6	105.30	75.70	100.00	73.55	14.85	14.75
LLA7	59.30	48.90	51.50	45.77	8.45	6.65
LLA8	71.80	62.20	75.30	52.23	5.40	5.80
LLA9	104.70	82.40	97.60	74.35	9.70	16.10
mean	96.90	71.93	86.56	64.05	10.49	10.86
std dev	26.11	12.16	17.95	14.60	5.46	4.34
Harness 0% - Sound						
LLA1	99.60	89.60	103.10	76.64	10.30	11.95
LLA2	114.10	80.00	95.50	79.15	16.15	15.50
LLA3	103.40	86.60	100.40	79.23	9.30	17.70
LLA4	101.60	94.50	101.50	81.97	4.65	4.75
LLA5	99.90	91.10	93.80	79.11	10.90	8.40
LLA6	117.10	69.40	113.80	80.59	19.20	33.00
LLA7	102.70	88.00	106.10	84.31	11.30	17.70
LLA8	100.80	85.10	88.70	99.83	4.45	7.70
LLA9	117.50	85.50	98.30	82.41	17.40	15.30
mean	106.30	85.53	100.13	82.58	11.52	14.67
std dev	7.60	7.30	7.32	6.85	5.23	8.29

Harness 0% - Prosthetic						
LLA1	76.20	61.50	63.50	48.01	1.45	6.00
LLA2	121.10	75.10	97.60	74.57	13.45	11.25
LLA3	100.60	87.00	100.00	89.71	13.10	9.55
LLA4	100.30	91.50	101.60	75.28	5.25	10.75
LLA5	103.80	87.60	96.70	74.42	8.55	9.60
LLA6	99.40	71.90	95.30	65.40	10.60	11.45
LLA7	61.20	55.50	61.80	51.00	10.15	6.90
LLA8	81.50	76.50	84.90	65.80	3.15	7.45
LLA9	100.00	86.00	98.80	72.45	7.40	14.90
mean	93.79	76.96	88.91	68.52	8.12	9.76
std dev	17.77	12.43	15.63	12.88	4.21	2.74
Harness 10% - Sound						
LLA1	99.30	80.90	84.40	64.75	9.10	10.65
LLA2	96.50	78.90	82.40	68.31	10.60	10.20
LLA3	94.00	82.00	89.20	77.20	7.10	13.45
LLA4	95.80	86.50	89.10	90.16	6.25	9.20
LLA5	93.10	87.30	89.60	81.11	7.20	7.00
LLA6	115.70	68.10	99.50	79.78	16.10	27.10
LLA7	90.80	80.80	86.80	81.09	7.70	15.45
LLA8	92.10	87.20	92.40	88.55	3.70	5.05
LLA9	105.70	75.20	84.60	67.89	12.85	11.65
mean	98.11	80.77	88.67	77.65	8.96	12.19
std dev	7.97	6.26	5.13	9.03	3.74	6.40
Harness 10% - Prosthetic						
LLA1	77.70	60.00	60.90	42.73	0.60	5.65
LLA2	108.90	74.90	82.20	62.33	10.40	7.25
LLA3	87.10	82.70	86.50	77.23	5.50	7.50
LLA4	93.90	84.50	86.60	85.47	7.40	11.50
LLA5	100.00	83.20	91.00	83.67	6.60	9.85
LLA6	100.70	71.30	89.50	66.35	10.40	11.15
LLA7	69.50	61.00	64.30	52.12	5.65	6.95
LLA8	67.90	63.30	76.00	48.00	2.90	6.90
LLA9	95.20	79.90	86.90	61.42	5.50	11.80
mean	88.99	73.42	80.43	64.37	6.11	8.73
std dev	14.46	9.94	11.04	15.36	3.17	2.34

Harness 20% - Sound						
LLA1	95.70	67.10	73.90	56.70	10.80	9.85
LLA2	86.90	66.00	73.80	61.05	2.95	7.40
LLA3	81.70	72.90	75.30	57.11	3.30	10.35
LLA4	86.30	76.30	78.90	91.13	6.90	7.50
LLA5	82.50	77.10	80.50	75.05	5.70	7.20
LLA6	105.00	56.20	94.40	57.64	8.50	28.30
LLA8	76.30	73.30	77.40	81.67	1.40	4.75
LLA7	80.6	72.2	74.1	69.77	3.2	12.6
LLA9	93.30	70.10	73.60	71.86	9.30	10.60
mean	87.59	70.13	77.99	69.11	5.78	10.95
std dev	8.95	6.41	6.64	12.15	3.28	6.91
Harness 20% - Prosthetic						
LLA1	70.60	57.60	60.00	38.85	0.60	5.20
LLA2	95.70	70.00	76.00	60.28	6.25	6.10
LLA3	79.00	73.20	75.50	56.97	1.25	6.85
LLA4	84.60	75.50	76.90	76.43	3.35	9.50
LLA5	83.50	74.30	79.80	75.94	3.90	8.15
LLA6	87.50	64.90	71.90	46.49	2.80	5.25
LLA7	69.20	59.50	61.40	47.68	5.10	6.55
LLA8	69.50	65.10	73.30	78.82	0.95	4.20
LLA9	88.60	76.40	79.30	60.59	4.45	9.60
mean	80.91	68.50	72.68	60.23	3.18	6.82
std dev	9.48	7.02	7.25	14.44	1.96	1.91