TEMPORAL-DISTANCE AND KINEMATIC ADAPTATIONS TO A NOVEL WALKING TASK

by

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ABSTRACT

The process of relearning locomotor skills is a complex one for the person with a lower-limb amputation and difficult to track in the rehabilitation setting. An in-house designed prosthetic simulator (PS) was created to allow able-bodied individuals to walk in a prosthetic-like situation. The purpose of this study was to follow the changes in selected gait variables during a novel walking task. Kinematic data were collected for ten able-bodied individuals during 30-minutes of continuous walking with the PS. Walking speed and selected gait characteristics and the vertical orientation of body segments were computed every 5% of the total walking distance during the first lab visit and walking speed again during a second lab visit. Separate repeated measures ANOVAs were conducted with $p < 0.01$.

Participants were immediately able to walk unassisted with the PS. Walking speed on the first test session was initially slow ($0.27\ m \cdot s^{-1}$) but significantly increased over distance walked (to $0.70\ m \cdot s^{-1}$). Initial time in stance was significantly greater on the intact limb (86 %) than on the prosthetic limb (68 %). Prosthetic step length was significantly longer (0.52 m) than intact step length (-0.10 m). Lower-limb segments were significantly less vertically oriented at prosthetic/intact foot contact during the walking task. Initial walking speed on the second session ($0.58\ m \cdot s^{-1}$) was significantly higher than on the first session. Variability of the measured gait variables was initially high but decreased within the first 5% of the total distance walked. Walking speed during the first five strides after removing the PS ($1.13\ m \cdot s^{-1}$) was significantly slower than the control condition ($1.30\ m \cdot s^{-1}$).

Participants were able to adapt quickly to the new constraints imposed by a PS by modifying kinematic variables. Changes occurred during the first 5%-10% of total walking distance suggesting adaptive strategies were developed early in the task. The presence of a short-term speed after effect suggested that adaptation had occurred. The findings from this project provide a novel outlook for rehabilitation strategies with the potential of tracking able-bodied individuals as they learn to walk in a prosthetic-like situation.

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CHAPTER 1: INTRODUCTION

The incidence of lower-limb amputation is increasing due to ageing and illness, as well as trauma (Armstrong et al., 1997). Lower-limb amputation can be performed either below (transtibial) or above (transfemoral) the knee. Regardless of the level of amputation, neuromechanical aspects of locomotion have been altered and learned motor patterns must be adapted. A person who has experienced a unilateral lower-limb amputation must deal with anatomical, physiological, neuromuscular and psychological issues resulting from the structural, motor and sensory asymmetry. However, with modern prostheses and treatment programmes, individuals who have had a lower-limb amputation can learn to perform most of the activities they could do prior to the amputation.

As a result of a unilateral amputation, skeletal links and joints, muscles, nerve connections and receptors signalling joint position of the lower-limb have been lost and replaced with prosthetic components on one side of the body. Many studies have found that individuals with a lower-limb amputation exhibit asymmetrical gait and modified motor patterns (Donker & Beek, 2002; Isakov et al., 1996; Jaegers et al., 1995; Nolan et al., 2003; Sanderson & Martin, 1997; Sanderson & Tokuno, 2001; Winter & Sienko, 1988). Previous research has also shown that persons with a lower-limb amputation adapt or modify their locomotor pattern for walking and that these modifications occur on both the prosthetic and intact sides (Sanderson & Tokuno, 2001; Winter & Sienko, 1988). The following sections explain the different types of adaptations (temporal-distance, kinematic, kinetic and neuromuscular) that individuals with a lower-limb amputation make when walking with a prosthesis.

1.1 Characteristics of unilateral prosthetic gait

1.1.1 Temporal-distance characteristics

Locomotor symmetry and walking velocity are often used to describe prosthetic gait in clinical settings (Donker & Beek, 2002). It has been well documented in the literature that amputees exhibit asymmetrical temporal-distance walking patterns and walk at a slower self-selected walking speed than able-bodied individuals (Hale, 1991; Isakov et al., 1996; Jaegers et al., 1995). Walking speed is the product of step length and frequency and can be attained by different combinations of these two variables. Some studies have reported that amputee walkers increase walking speed by increasing step length more so than step frequency (James & Oberg, 1973; Jaegers et al., 1995). However, as velocity increases, the number of possible combinations is reduced given the biomechanical constraints of the musculoskeletal system to further increase step length and/or frequency during walking (Donker & Beek, 2002).

Other features specific to amputee walking include longer step length on the prosthetic side compared to the intact side and more time spent in stance and double support on the intact limb and in swing on the prosthetic limb. Longer stance and increased loading on the intact limb has been attributed to the fact that the amputee tries to protect their stump, which may serve as a compensatory action for increasing gait stability (Donker & Beek, 2002; Jaegers et al., 1995; Nolan et al., 2003). The stump refers to the part of the limb that is left after the amputation is performed. Furthermore, James and Oberg (1973) found that time spent in stance and step length for the prosthetic and intact limbs remained asymmetrical as walking speed increased from preferred (0.99 ± 0.14 m $\cdot$ s^{-1}) to rapid (1.34 ± 0.16 m $\cdot$ s^{-1}). Similarly, Sanderson & Martin (1997) reported significantly different stance and swing times on the prosthetic and intact limbs as speed increased from 1.2 m $\cdot$ s^{-1} to 1.6 m $\cdot$ s^{-1} indicating that asymmetrical gait patterns were observed despite increased walking speed.

Several studies have also reported that amputees walk with a larger step width compared to able-bodied individuals (James & Oberg, 1973; Jaegers et al., 1995). Step width reflects the possibility of stabilising the pelvis in the frontal plane during the prosthetic stance phase (James & Oberg, 1973).

Regardless of the asymmetrical nature of their walking style, individuals with a lower-limb amputation can walk successfully with a prosthesis. The loss of the ankle complex and ankle plantarflexor musculature can be partially compensated for through the use of modern designed prostheses. Although these devices return a portion of the stored energy during the propulsive phase in mid- to late-stance, compensations for the loss of power generation must still occur (Sanderson & Martin, 1997). These modifications are made at the sites proximal to the amputation and therefore, examining the role of the knee and hip during prosthetic gait is especially important.

1.1.2 Kinematic characteristics

Despite the altered physical constraints, amputees can learn to walk proficiently and do so by reportedly showing similar kinematic profiles to able-bodied individuals. Sanderson & Martin (1997) noted that amputees walked with patterns of motion that were not easily distinguishable from non-amputees. Specifically, they found that ankle, knee and hip angular position and velocity patterns exhibited only minor differences between limbs at preferred ($1.2\ m \cdot s^{-1}$) and fast ($1.6\ m \cdot s^{-1}$) walking speeds. The knee and hip angles on the prosthetic side were in a more extended position during the first two-thirds of stance than these joint angles on the intact side. The prosthetic limb was maintained more vertically aligned throughout the stance phase which prevented the knee from flexing. Also, reducing the demands on the musculature and knee joint loading assisted in protecting the stump. The authors concluded that while amputees and non-amputees walked with comparable lower-limb kinematic profiles, they achieved these movements by employing different kinetic profiles. Therefore, an investigation into the kinetic modifications that occur on both the prosthetic and intact limbs would reveal the internal control strategies that produce the movement outcome.

1.1.3 Kinetic modifications

Even with the loss of a limb, amputees must still solve the problem of moving with relative comfort, while maintaining stability and with acceptable appearance (Eberhart et al., 1954). A prosthesis may provide the structural support needed to ambulate however, dynamic control from the lost musculature is difficult to replace. Therefore, kinetic adaptations are essential because of the reduced propulsive capacity in the prosthetic limb due to the loss of ankle plantarflexor musculature.

Kinetic adaptations reveal the level of internal adjustment to the altered lower-limb mechanics as a result of amputation. The prosthetic limb has smaller push-off or propulsive forces (anterior-posterior ground reaction force component) and decreased vertical ground reaction force components (Mattes et al., 2000; Nolan et al., 2003; Sanderson & Martin, 1997). Sanderson and Martin (1997) reported that amputees generated similar support functions while employing different motor strategies in the intact and prosthetic legs. These authors also reported that the neuromuscular reorganisation on the prosthetic limb led to a net knee moment that was flexor in orientation throughout stance. Therefore, internal adjustments occurred on both the affected (prosthetic) and unaffected (intact) limbs.

1.1.4 Neuromuscular adaptations

Studies examining dynamic electromyographic (EMG) muscle activation patterns reveal some of the mechanisms for muscular control during amputee gait. Winter and Sienko (1988) found increased co-contraction of the knee musculature (biceps femoris, semitendinosus, rectus femoris and vastus lateralis) on the prosthetic limb during early stance. They also reported hyperactive hip extensors (gluteus maximus) during early and mid-stance as a result of the absence of the energy-generating plantarflexors at push-off. Sanderson and Tokuno (2001) concluded that the reduced knee extensor moment during stance seen among transtibial amputees occurred as a result of the shift in the degree of activation of the vastus lateralis and semitendinosus. More co-contraction served to stabilise the knee joint among transtibial amputees.

Czerniecki and Gitter (1996) suggested that neuromuscular control mechanisms at the proximal joints to the site of amputation (i.e. knee and hip musculature) may assist the individual to moderate the applied forces and rate of lower extremity loading. They reported that the hip extensor muscles of the prosthetic limb partially compensated for the reduced knee extensor activity and reduced ability to generate mechanical power with the prosthetic foot at push-off. The hip extensors become the primary source of energy absorption and generation (Latash, 1998). Therefore, studies assessing the development of modified locomotor strategies, when the propulsive capacities of the ankle plantarflexor musculature have been lost, should focus on the body segments surrounding the knee and hip joints.

1.2 Prosthetic simulator (PS) as a tool for studying prosthetic gait

The compensatory movements and internal adaptation described above exemplify the adaptive nature of the central nervous system that allows individuals with a lower-limb amputation to walk. Yet amputee patients gain proficiency using their prostheses at different rates and therefore, acquiring information on how specific groups respond to gait training is important for therapists (Baker & Hewison, 1990).

The relevant literature does not adequately discuss the nature of change or the time course of gait adaptations an individual makes following a lower-limb amputation. A review of the published research has found one group of authors that have designed and studied the use of a prosthetic simulator (PS) among able-bodied participants (Lemaire et al., 2000). The PS was a custom-made prosthetic device that provided a prosthetic-like experience for non-amputees. The PS consisted of a prosthetic foot and orthotic knee joint while the participant's knee was secured in a flexed position with a lower-leg sling attachment.

The purpose of their study was to help rehabilitation instructors recognise and evaluate the exertion, balance requirements and insecurities experienced by those who must learn to walk with a prosthesis. Once the prosthetic socket was properly fitted, participants began weight-bearing between parallel bars. They completed two gait training sessions, in which they progressed from walking between parallel bars to walking with a cane to unassisted walking, before kinematic and kinetic data were collected. Participants were able to walk unassisted with the PS. Their decreased stride length and increased stride time reflected a slower walking speed ($0.82 \pm 0.11 \text{ m} \cdot \text{s}^{-1}$) than the average walking speed ($0.99 \pm 0.05 \text{ m} \cdot \text{s}^{-1}$) for experienced prosthetic users reviewed in their study. Furthermore, kinematic and kinetic profiles were similar to those for individuals with a transfemoral lower-limb amputation showing that a PS could be used to simulate prosthetic gait among non-amputees. The results from Lemaire et al.'s study provided interesting information about the use of a PS to better understand prosthetic gait in a rehabilitation setting. How individuals adapt to altered lower-limb mechanics and learn to walk successfully with a PS remains to be explored.

1.3 Prosthetic simulator (PS) to create a novel walking task

Little is known about the time course of the adaptation as an individual learns to walk with a prosthetic device within their new level of physical constraints. The complex issues associated with tracking individuals who have experienced a lower-limb amputation, from the moment they enter the rehabilitation setting until they leave as competent walkers, make this a difficult task. To counter this difficulty, an in-house design unilateral lower-limb PS was created for this study to allow able-bodied individuals to walk in a unique situation. Thus, the nature and development of changes in gait variables could be monitored during this novel walking task. The experiment provided a challenge for otherwise healthy, able-bodied individuals to develop modified locomotor adaptations when lower-limb mechanics were altered. This could be accomplished from the moment the participant was fitted with the PS and following practise as they learned to ambulate more comfortably.

1.3.1 Pilot studies to investigate the whole body system in movement when walking with a PS

The displacement of the body's centre of mass (CoM) during the cycle of motion is often considered to simplify the motions describing human locomotion (Saunders et al., 1953). By applying the concept of the CoM, it is assumed that all of the body's mass is concentrated at that point. The movements of the body's limbs are co-ordinated in such a way as to minimise the trajectory of the whole body CoM to conserve energy (Saunders et al., 1953). At each step, the CoM is successively behind or in front of the point of contact with the foot on the ground. It is displaced twice in the vertical direction during one stride. Each peak corresponds to the middle of the stance phase of the supporting limb when the opposite limb is in the middle of swing. The CoM falls to its lowest point when both feet are in contact with the ground during the double support phase.

A series of pilot studies were conducted to investigate the motion of the lower limbs and CoM in the sagittal plane when walking with a PS. The motion of the lower-limb segments influences the vertical oscillation of the CoM such that walking over a flexed knee reduces the vertical displacement of the CoM. This also has implications for the metabolic costs associated with walking because large displacements of the CoM result in high energy costs.

The aim of the pilot studies was to gain a better understanding of the biomechanical factors that influenced the motion of the CoM when walking with the PS, because one limb consisted of a thigh and a shank that could move in synchrony to flex and extend throughout the gait cycle, while the other limb consisted of a rigid link with no knee flexion.

When an experienced participant (i.e. one who has walked with the PS on many occasions over a period of six months) walked with the PS at a preferred walking speed of $0.88\ m \cdot s^{-1}$, the peak vertical position of the CoM occurred during single stance and was higher on the prosthetic leg than on the intact side. This corresponded to when the prosthetic limb was vertical and was attributed to the rigid tubing that provided the means of structural support on the prosthetic side and absence of an artificial knee joint. Moving the CoM over a supporting limb that did not allow any flexion produced a 'vaulting sensation'. During able-bodied walking at the same speed, peak vertical position of the CoM occurred when both the thigh and shank segments were also vertically oriented in mid-stance. Furthermore, the peak vertical position of the CoM was the same during right and left single stance and was lower than when walking with the PS. This was because the summit of the arc representative of the motion of the CoM is reduced when walking over a flexed knee (Saunders et al., 1953). The results from the pilot studies showed that the vertical orientation of the lower-limb segments influenced the peak vertical position of the CoM.

During walking, maintaining balance is a challenging task for the postural control system because a single limb supports the body throughout approximately one third of the stance phase, with the body's CoM passing outside the base of support. Individuals with a lower-limb amputation flex their trunk forwards early in the stance phase in order to maintain the body's CoM above the supporting limb (Donker & Beek, 2002). Results from the pilot studies also showed greater forward trunk lean when walking with the PS than when walking with two intact limbs. This helped the participant in our pilot study maintain body equilibrium when walking with altered lower-limb mechanics imposed by the PS.

1.4 Variability

Analysis of specific gait variables and motion of body segments is used to characterise and improve gait patterns in a clinical setting. However, the analysis would be incomplete without the investigation of how the variability of these specific gait features changes over time when individuals learn a complex task and adapt to altered lower-limb mechanics. Variability in the sensorimotor system is typically operationally defined as the standard deviation (SD) of a specific parameter from repeated measures over successive attempts during the execution of a particular motor task (Newell & Corcos, 1993).

Variability is an integral part of human motion. Evidence of this lies in the fact that several attempts at the same task lead to different patterns of performance, including kinematics, kinetics and patterns of muscle activation (Latash et al., 2002). With regards to human walking, within-subject variability of gait variables has often been considered an indicator of unsteady gait and increases the risk of falling (Hausdorff et al., 1997; Hausdorff et al., 2001). However, it has now been recognised that movement variability plays a functional role in the detection and exploitation of stability boundaries during bipedal stance and locomotion (van Emmerik & van Wegen, 2002). For example, in the literature on postural control, the movement of an individual's centre of pressure during quiet standing on a force platform has been associated with postural stability and more movement generally indicates greater instability. However, increased postural sway could arise from exploratory movement that in fact contributes to postural stability and typically exhibits patterns of motion that are higher in frequency and lower in amplitude (van Emmerik & van Wegen, 2002). The higher-frequency components of the centre of pressure, reflective of both exploratory and compensatory activity, are necessary to control or correct the position of the CoM.

Stable patterns are low in variability. When variability is large, the system is characterised as unstable and one in transition (Clark, 1995). Therefore, the variability associated with learning a new and challenging task would presumably be higher than during a skilled task. Exploratory behaviour contributes to the predicted higher variability. Therefore, for individuals with an altered motor system as a result of different lower-limb mechanics imposed by a PS, exploring their modified stability boundaries is central to developing a pattern of movement that is functionally stable and yet adaptive. Thus, following the change in variability of specific gait variables complements data pertaining to the description of body segments in motion during a novel walking task.

1.5 Motor learning

Studies that have investigated walking with a prosthesis have not conducted clinical gait analysis on patients with a lower-limb amputation in the rehabilitation setting as they take their first steps with a prosthesis and learn to walk proficiently over time. Undoubtedly, these individuals must deal with a variety of complex factors that makes taking part in biomechanical studies difficult. Therefore, tracking able-bodied individuals as they learn to walk within a new level of physical constraints with a PS could have important implications for the rehabilitation community.

Motor learning is a set of processes that underlie the changes in the capability for movement (Schmidt & Lee, 1999). Although not directly observable, motor learning is relatively permanent and can be inferred from changes in motor behaviour. One example is a change in performance with practice that causes a change in the capability for moving. In addition to improvements in performance, the existence of motor after effects reinforces the assumption that adaptation has occurred (Reynolds & Bronstein, 2003). An after effect is the continued occurrence of a response to a perturbation following removal of the perturbation. The continued adaptive strategies to a perturbation that has been removed indicate the role of feedforward mechanisms.

1.5.1 Feedforward mechanisms

Feedforward mechanisms prepare the motor system for upcoming motor commands in advance of the movement (Schmidt & Lee, 1999). For example, lifting the arm in front of the body, from a relaxed position by the side, will cause the CoM to be shifted forward and balance may be lost unless some kind of compensatory movement accompanies the action itself. Research has shown that, even before the first signs of muscle activity were observed in the shoulder muscles, EMG muscle activity onset occurred in the contralateral leg (Schmidt & Lee, 1999). Similarly, a backward balance loss can be avoided through the use of feedforward control to improve stability at the onset of a slip by increasing the CoM forward velocity and/or shifting the CoM forward (Pai & Iqbal, 1999).

1.5.2 After effect

Several studies have examined adaptive strategies to sustained changes during locomotion that were not induced by nerve injury (Jensen et al., 1998; Lam et al., 2003). In theses studies, compensations were evident in the presence of the initial perturbation but long-term compensatory strategies were also observed following its removal. Jensen et al. (1998) reported the adaptational effects that occurred during split-belt walking and the factors that influenced these effects. These authors reported a speed after effect when participants were unable to adjust to a control speed (3 km $\cdot$ h^{-1}) after completing a training period with split-belt walking (1.5 and 4.5 km $\cdot$ h^{-1}). After a perceived adjustment to the control speed, a significant speed difference between the two legs was found. The authors concluded that a newly learned locomotor pattern could exert a continuing and over-riding influence over adaptational mechanisms.

Individuals can make adaptive modifications following altered sensory input. Lam et al. (2003) investigated whether infants had the ability to adapt to sustained changes in sensory input during stepping movements. The focus was on how infants responded to an extra load applied to their leg during the swing phase of treadmill stepping. They also addressed the issue of whether adaptation had occurred by measuring the maximum height of toe trajectory in the first step with the weight off. These authors reported that all infants showed the ability to respond to the extra weight on the leg by generating greater flexor muscle torque at the knee and hip indicating that the response was mediated at the spinal level. Some infants also showed a short-term but continued high-stepping response, as measured by the height of the toe trajectory, immediately upon removal of the weight indicating the presence of an after effect and that adaptation had occurred. The group of infants that exhibited an after effect upon removal of the extra load appeared to adapt their stepping well to the added weight on their leg.

Thus, the presence of an after effect is one method of inferring that adaptation to a novel motor task has occurred. A relatively permanent change of a performance variable, determined by a lasting change in motor behaviour measured over repeated attempts at the same task, would indicate skill retention and motor learning. Given the difficulties associated with following patients in a rehabilitation setting, the potential for using a PS, to determine how individuals adapt to walking with different lower-limb mechanics takes on an important role. Assessing changes of a gait variable, such as walking speed, before and after walking with a PS and over successive attempts, could easily be measured in a clinical setting.

1.6 Specific objectives and hypotheses

When an individual loses a lower limb, as a result of an accident or disease, they typically must learn to walk with a prosthesis and within a new level of physical constraints as a result of altered lower-limb mechanics. The previous sections have identified that amputees can perform many of the activities they could do prior to the amputation, and one of the most common activities is walking. These individuals can learn to walk with a prosthesis by making adaptations in different aspects of gait mechanics. Much of the existing research has described gait patterns of proficient amputee walkers. However, we do not know the time course of the adaptation as individuals learn to walk with a prosthesis and become proficient walkers.

Therefore, the purpose of this study was to assess how individuals adapted to altered lower-limb mechanics, imposed by wearing a prosthetic simulator (PS), during walking. To accomplish this, this study examined three specific objectives.

The first objective was to determine the changes in temporal-distance and selected kinematic variables of a novel walking task with a specially-designed PS during 30 minutes of continuous walking. Temporal-distance gait variables included walking speed, step characteristics (step width, length and frequency), and body support times (time spent in stance, swing and double support) on both the intact and prosthetic limbs. Kinematic variables included the absolute vertical orientation of the prosthetic limb at prosthetic contact (PS-ON), intact shank and thigh segment angles at intact foot contact and anterior-posterior trunk lean at instances of both PS-ON and intact foot contact. The dependent variables selected in this study were considered important in characterising walking patterns among individuals with altered lower-limb mechanics. All measured variables were analysed throughout 30 minutes of continuous walking with the PS.

Specifically, it was hypothesised that, walking speed would initially be very slow but would increase over distance walked as an expression of adapting to walk with the PS. Walking speed would also increase as a function of increasing step length and step frequency on both the prosthetic and intact limbs. It was hypothesised that initial walking speed during the second test session would be higher than initial walking speed on the first session. When walking with a PS, it was hypothesised that step width would increase to stabilise the pelvis as a consequence of a faster walking speed. Also a result of faster walking speed over distance walked, it was hypothesised that time in stance would decrease, and concurrently time in swing would increase, and time in double support would decrease for both limbs.

In order to keep the body's centre of mass above the supporting limb at PS-ON and intact foot contact, it was hypothesised that the prosthesis and intact lower-limb segments would initially be more vertically oriented. To reduce the horizontal and vertical displacement of the CoM during the stance phase, it was hypothesised that the trunk segment would initially show greater forward lean at PS-ON and intact foot contact. Since the prosthetic limb consisted of a rigid link while the shank and thigh segments on the intact limb could move in synchrony to flex and extend throughout the gait cycle, it was hypothesised that the peak vertical position of the whole body CoM would be higher during prosthetic compared to intact single stance.

The second objective was to assess the change in variability of the temporal-distance and kinematic variables throughout the walking task with the PS.

It was hypothesised that the variability for all gait parameters would initially be high but decrease over distance walked indicating the system had developed a functional walking style with the PS as the participants adapted to altered lower-limb mechanics.

The third objective was to determine whether a speed after effect occurred upon removal of the PS. The presence of a speed after effect would be one method of inferring that participants had adapted to walking with altered lower-limb mechanics.

It was hypothesised that self-selected walking speed during the first 5 strides immediately upon removal of the PS would be slower than before walking with the PS (control condition). However, this effect would disappear within the subsequent five strides such that walking speed would return to the control level during the first ten strides after removing the PS. The presence of a short-term speed after effect would support the notion that adaptation to walking with the PS had occurred.

CHAPTER 2: METHODS

2.1 Participant characteristics

Ten healthy, able-bodied individuals (four males, six females; age: 22.6 ± 2.3 yrs; height: 173.0 ± 6.4 cm; mass: 66.3 ± 6.8 kg) volunteered for this study (Table 1). Inclusion criteria were that participants had no known neuromuscular or balance disorders, recent musculoskeletal injury to either lower limb or any previous experience walking with a prosthetic simulator (PS). Once fitted with the PS, they must have been able to walk unassisted. Individuals within a height range of 165–185 cm were selected for the study because of the dimensions of the PS. All participants gave informed consent to take part in this study in accordance with the University of British Columbia's Clinical Research Ethics Board.

Table 1. Participant characteristics

Participant	Age (yrs)	Height (cm)	Mass (kg)	Gender
1	27	180	70	Male
2	20	170	70	Male
3	25	176	65	Female
4	21	174	75	Female
5	23	182	71	Male
6	23	167	52	Female
7	19	165	69	Female
8	23	166	56	Female
9	23	181	69	Male
10	22	172	57	Female
Female mean (SD)	22.2 (2.0)	169.7 (4.5)	62.3 (8.6)	
Male mean (SD)	23.3 (2.9)	178.0 (5.6)	69.9 (0.7)	
Mean (SD)	**22.6 (2.3)**	**173.0 (6.4)**	**66.3 (6.8)**	**4 males; 6 females**

2.2 Prosthetic simulator

The PS consisted of a rigid plastic cuff into which the individual placed their dominant leg (Figure 1). Dominant leg was determined by asking the participant which leg they would use to kick a soccer ball and for all participants this was their right leg. The knee was flexed at 90° and securely maintained in this position with Velcro straps running across the posterior side of the thigh and shank. A rigid telescoping aluminium tube provided the means of support at the lower extremity and was easily adjusted according to the individual's lower-limb length. A slip-resistant rubber cap was placed at the end of the aluminium tube.

Figure 1. Participant fitted with in-house designed unilateral lower limb prosthetic simulator

2.3 Testing protocol

Each participant took part in two testing sessions one week apart. For the first session, kinematic data were recorded while the participant first walked without the PS (control condition) and then with the PS fitted to their leg. The control condition was when participants walked barefoot with two intact legs before walking with the PS. Reflective markers were placed on specific anatomical locations according to the Helen Hayes full body marker set (Figure 2). The lab was equipped with a high-speed 8-

camera system (Motion Analysis Corporation, Santa Rosa, CA, USA). The ExpertVision (EVaRT) software programme (Motion Analysis Corporation, Santa Rosa, CA, USA, 2001) was used to collect three-dimensional data during the session at a sampling rate of 60 Hz.

1. L. shoulder
2. L. shoulder offset *
3. L. elbow
4. L. wrist
5. L. ASIS
6. L. thigh
7. L. anterior thigh
8. L. knee
9. L. medial knee
10. L. lateral shank
11. L. shank
12. L. ankle
13. L. toe
14. L. medial ankle
15. L. heel
16. Sacral *

Above markers are listed for left side only with corresponding markers for the right side

* no corresponding bilateral marker

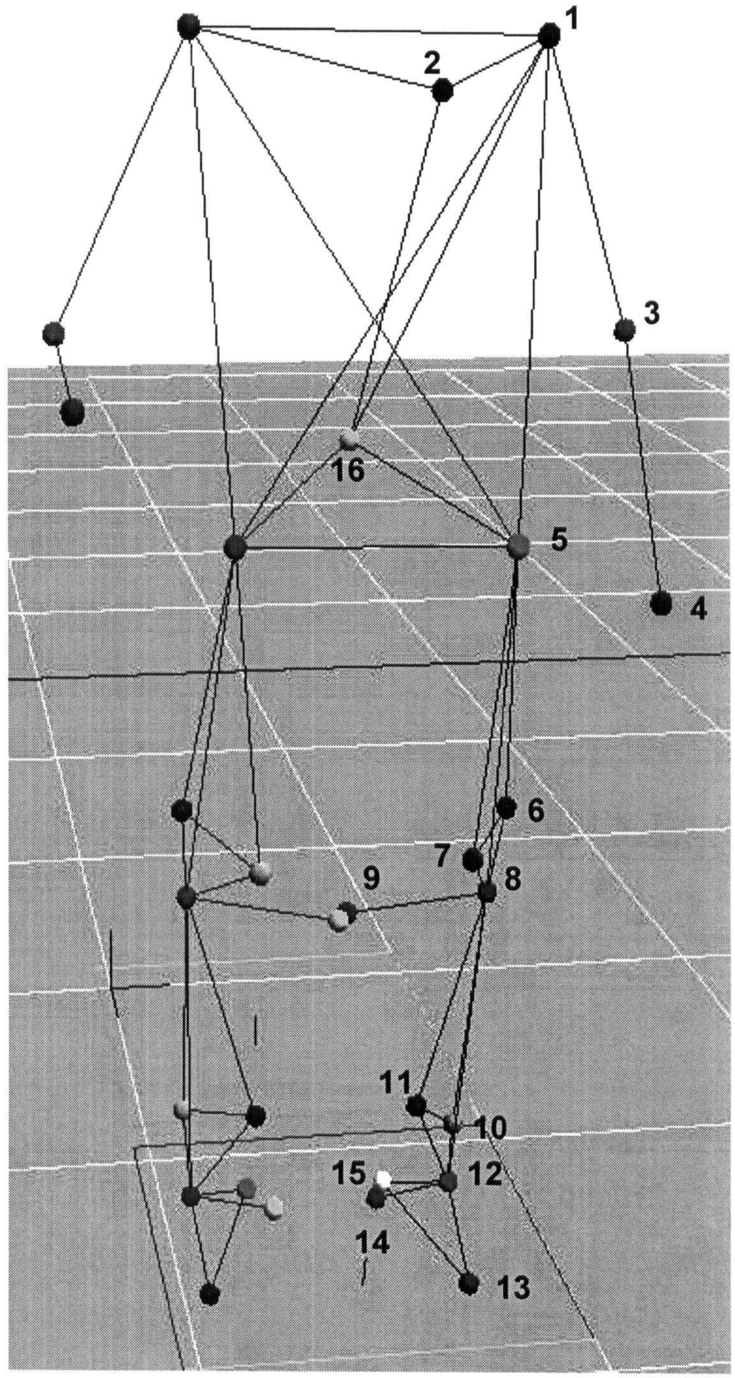

Figure 2. Helen Hayes full-body marker set for stationary data acquisition (additional medial knee and ankle markers are placed bilaterally)

During the first testing session participants first completed several control trials at their self-selected walking speed along a 10-metre walkway while kinematic data were collected. The PS was then fitted to the participant's right leg and the telescoping tube of the PS was adjusted such that the distance from the position of the lateral knee marker on the PS to the ground was 2 cm shorter than the recorded lower leg length. Adjusting the length of the prosthetic segment 2 cm shorter was considered sufficient for ground clearance given the prosthetic segment did not allow for knee flexion. Right intact lower leg length was measured from the lateral knee marker to the ground while the participant was standing. Due to the absence of a prosthetic foot, a three-coordinate marker set with markers for the heel, toe and lateral aspect of the ankle was fastened to the end of the aluminium tube. These markers were used by the OrthoTrak software system to indicate instances of prosthetic contact and prosthetic off.

Once fitted with the PS, participants were asked to walk continuously, unassisted along a 30-metre loop for a total of 30 minutes. The loop was configured with two approximately 10-metre straight sections and two circular sections with a radius of approximately five metres. They were not given any prior instruction on how to walk with the PS. Three-dimensional coordinate data were recorded for 10 seconds, or approximately five consecutive strides, every time the participant completed a loop and came into the cameras' field of view. After 30 minutes, the PS was removed. Walking speed was also measured during the first five strides and then the first ten strides immediately upon removal of the PS when participants walked with two intact legs.

For the second testing session, participants again completed several control trials at their self-selected walking speed along a 10-metre walkway. Walking speed was determined with the use of photocells located two metres apart. The PS was then fitted onto their right leg and the telescoping tube was adjusted to the same lower leg length as during the first session. Participants were instructed to walk continuously along a 30-metre loop for 30 minutes. Walking speed was monitored every time the participant completed a loop. After 30 minutes, walking speed was also measured during the first five and ten strides immediately after removing the PS.

2.4 Phases of data analysis using the OrthoTrak system

The first phase of data analysis involved video equipment calibration and participant preparation. A 4-point calibration and wand calibration of the capture space were performed prior to the participant's arrival. Upon their arrival, participants changed into tight-fitting clothing and reflective markers were placed on specific anatomical landmarks using the Helen Hayes marker set.

The second phase ensured that data were successfully collected during the testing session. This involved two short (0.5 second) data acquisitions when the person was standing stationary. Data were collected for the right and left legs using medial and lateral knee and ankle markers that were subsequently used to establish knee and ankle joint centres, respectively. Once the acquisitions for each leg were captured and reconstructed in the tracking programme, the medial knee and ankle markers were removed and dynamic, walking trials began.

Stationary data acquisition was also repeated once the participants were fitted with the PS. The location of thigh and shank markers on the right leg were marked on the skin surface and then removed. These were then re-located onto the corresponding locations of the plastic cuff (thigh) and aluminium tubing (shank). Due to the absence of a prosthetic foot on the PS, a three-coordinate marker set, with markers for the heel, toe and lateral aspect of the ankle, was fastened to the end of the aluminium tube. These markers served to identify gait events in OrthoTrak.

Data for walking trials were acquired in the third phase. The participants walked along the walkway in the laboratory and data were collected every time they entered the video field. The video field was large enough to capture approximately five consecutive strides. An external trigger was used to activate the video data collection. The number of trials collected depended on the number of laps completed during 30-minutes of walking and varied for participants. When walking with the PS, participants followed a 30-metre loop traced on the floor. This was to satisfy the requirement of not turning on the spot but walking continuously in the laboratory.

The final phase involved data analysis using data from the motion capture system (Eva/EVaRT). Data analysis was done using the OrthoTrak software package.

2.5 Data processing

All data were processed using the EVaRT software programme. Three-dimensional coordinate data from each completed lap were interpolated using a cubic spline algorithm and filtered using a 4^{th} order Butterworth filter with a cut-off frequency of 6 Hz. The processed data were then exported into OrthoTrak 5.0 Gait Analysis Software (Motion Analysis Corporation, Santa Rosa, CA, USA, 2001) where gait events were determined. Instances of foot/prosthetic contact and foot/prosthetic off were identified from video coordinate data. Before the coordinate data were exported for analysis, all gait events were individually verified in OrthoTrak with reference to video data. Any trials in which the participant tripped were disregarded. A trip occurred when the participant did not successfully clear the ground with the end of the PS during the swing phase and consequently abducted their hips, causing them to hop on their intact leg to regain balance. Five participants tripped at least once when passing through the video field and these trials were excluded from data analysis. However, no participants fell when walking with the PS.

2.5.1 Identification of gait events

The filtered three-dimensional coordinate data were exported into OrthoTrak. Gait events, such as foot/prosthetic contact and foot/prosthetic off were determined from video coordinate data. OrthoTrak automatically calculated all of the foot/prosthetic contact and foot/prosthetic off events based on a Gait Event Extraction Algorithm. The instance of foot/prosthetic contact was determined from the resultant linear velocity of the heel and toe markers. A central difference velocity calculation was applied and foot/prosthetic contact occurred when the velocity of either the heel or toe marker reached zero. Once the marker that first attained a velocity of zero was determined, the time of foot/prosthetic contact was established as the frame in which the marker's velocity fell below a set threshold relative to that marker's maximum velocity. The occurrence of foot/prosthetic off was determined when the linear velocity of the toe marker reached a set threshold relative to its maximum.

2.5.2 Joint centre calculations

OrthoTrak calculated the position of the joint centres using data collected during the stationary data acquisition trials. The shoulder centre was determined from coordinate data of the left and right shoulder markers. The pelvic coordinate system was determined from left and right anterior superior iliac spine (ASIS) and sacral markers. The left and right hip joint centres were calculated from the pelvic coordinate system. They were predicted as displacements along the three-dimensional pelvic axes, which were expressed as fixed percentages of the ASIS distances (22% in the posterior/X-direction; 32% in the lateral/Y-direction and 34% intact the inferior/Z-direction). The percentages were determined previously from measurements of 31 skeletons. A special measurement device was designed that would indicate the distances from the origin of the pelvic coordinate system to the hip joint centres (OrthoTrak 5.0 Gait Analysis Software, Motion Analysis reference manual, 2001). The left and right knee joint centres were calculated as the midpoints between the lateral and medial knee markers. The left and right ankle joint centres were calculated as the midpoint between the lateral and medial ankle markers. The lateral knee and ankle markers remained on the legs during the walking trials while the locations of medial knee and ankle markers were determined during the stationary data acquisition trials.

2.5.3 Temporal-distance variable definitions

Temporal-distance variables including walking speed, body support times (stance, swing and double support) and step width, length and frequency for both the left/intact and right/prosthetic limbs were calculated in OrthoTrak. Calculations of all body support times were dependent upon the gait events of foot/prosthetic contact and foot/prosthetic off.

Walking speed was calculated as the average anterior/posterior velocity of the sacral marker during each gait cycle. Stance, swing and double-support duration, expressed as a percentage of the full gait cycle (100%), were calculated from the gait events of foot/prosthetic contact and foot/prosthetic off. Stance was the total amount of time spent during the support phase of gait when the foot/prosthesis was in contact with the ground. Swing was calculated as the time the foot/prosthesis spent swinging in preparation for the subsequent foot/prosthetic contact. Double-support occurred at the beginning and the end of each gait cycle. Initial double support indicated the double support when both limbs were in contact with the floor at the beginning of the cycle and was calculated as the percentage of the gait cycle between foot/prosthetic contact of one limb and foot/prosthetic off of the opposite limb. Step length was reported as the anterior/posterior distance of the heel maker from foot/prosthetic contact of one limb to foot/prosthetic contact of the opposite limb. Step width was the difference between the ankle centres on the room medial/lateral axis during intact foot/prosthetic contact of each limb. Step frequency was extrapolated as the number of steps taken per minute of time.

2.5.4 Kinematic data definitions

Vertical orientation of the prosthetic, thigh, shank and trunk segments was calculated based on coordinate data from OrthoTrak. Projected two-dimensional segment angles of the lower limbs were determined in the sagittal plane. The prosthetic limb was defined as the segment between the hip and ankle joint centres. Prosthetic angle was defined as the vertical orientation of the prosthetic limb with respect to the vertical-axis through the virtual ankle joint centre (Figure 3A). On the intact side, the shank angle was defined as the vertical orientation of the shank segment with respect to the vertical-axis through the ankle joint centre (Figure 3B), while the thigh angle was defined as the vertical orientation of the thigh segment with respect to the vertical-axis through the knee joint centre (Figure 3C). The trunk segment was calculated as the segment from the shoulder centre to the pelvic centre. The trunk angle was defined as the vertical orientation of the trunk segment with respect to the vertical-axis through the pelvic centre (Figure 3D). Values for the shank and thigh angles and prosthetic angle were determined at foot and prosthetic contact, respectively. The trunk angle was determined at both foot and prosthetic contact. Segment angles behind the vertical were positive and negative angles were in front of the vertical. Peak vertical displacement of the whole body CoM was determined during single stance for each limb.

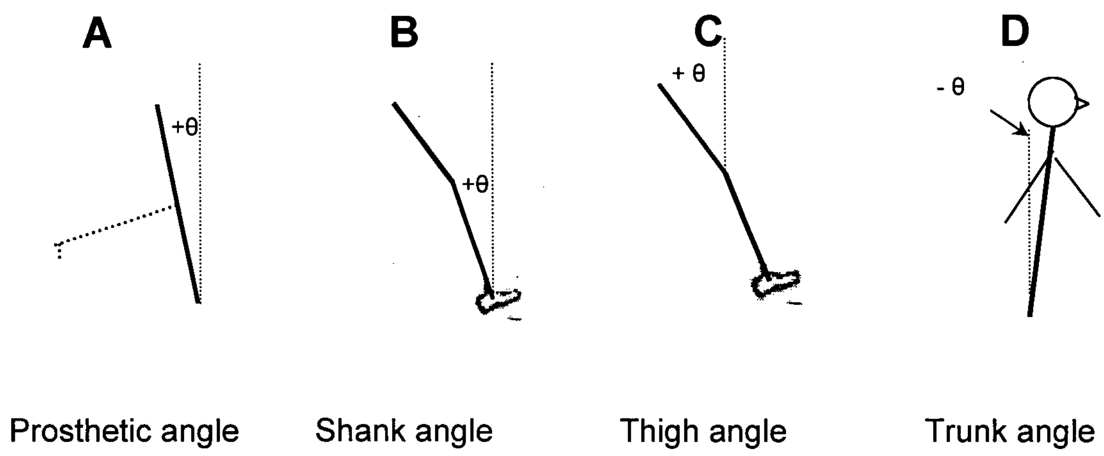

Figure 3. Body segment angles

2.6 Statistical analysis

2.6.1 Temporal-distance and kinematic variables

Mean (SD) values were found for all temporal-distance and kinematic variables for three consecutive strides for each lap the subject walked. Each subject completed a different number of total laps during the course of the 30-minute walking period and each lap corresponded to a specific percentage of total walking distance which varied among the participants. Therefore, to compare across participants, mean (SD) values for the dependent variables were determined at every 5% of total walking distance by linear interpolation, creating a total of 21 distance points. To determine the changes in measured variables between the two limbs (i.e. intact vs. prosthetic) during 30 minutes of continuous walking with the PS, separate 2 x 21 (limb x distance) repeated measures ANOVAs were used for body support times, step length and step frequency and trunk angle. A 2 x 21 (limb x distance) repeated measures ANOVA was used to determine the change in peak vertical position of the CoM. Separate 1 x 21 (dependent variable x distance) repeated measures ANOVAs were used for walking speed, step width and prosthetic, shank and thigh angles. Walking speed during 30 minutes of continuous walking with the PS from the first test session was compared with the second test session using a 2 x 21 (session x distance) repeated measures ANOVA.

2.6.2 Variability

The mean standard deviation of all T-D and kinematic variables was determined at every 5% of total walking distance by linear interpolation, creating a total of 21 distance points. To determine the changes in variability between the two limbs (i.e. intact vs. prosthetic) during 30 minutes of continuous walking with the PS, separate 2 x 21 (limb x distance) repeated measures ANOVAs were used for body support times, step length and step frequency and trunk angle. Separate 1 x 21 (dependent variable x distance) repeated measures ANOVAs were used for walking speed, and prosthetic, shank and thigh segment angles.

2.6.3 Speed after effect

Mean (SD) self-selected walking speed for the ten participants was determined prior to walking with the PS on both the first and second test sessions. Self-selected walking speed during able-bodied gait was measured during the first five strides and then the subsequent five strides immediately upon removal of the PS. Two separate 1 x 3 (walking speed x time) repeated measures ANOVAs were used to determine the change in self-selected walking speed for both test sessions. The three levels of time were before, five strides and ten strides after removal of the PS.

2.6.4 Statistical significance

Huynh-Feldt adjustments were made across all statistical tests using SPSS v. 11.0. The Bonferroni adjustment technique was used to reduce the Type I error and results were considered statistically significant at $p < 0.01$. Tukey's HSD post-hoc analyses ($p < 0.01$) were performed if significant main effects were found. Visual inspection of these data showed that the greatest change in variables occurred early for total distance walked. Therefore, mean values for significant main effects were compared by comparing the first distance measure to the second. If this difference exceeded a critical difference, then the second measure was compared to the third. This was repeated until no further critical differences were found.

CHAPTER 3: RESULTS

3.1 Temporal-distance and kinematic adaptations

3.1.1 Walking speed

All participants were immediately able to walk unassisted with the prosthetic simulator (PS) and did so continuously for 30 minutes without falling. The total distance walked varied considerably among participants, ranging from 540 to 1350 metres (mean 1017 ± 278 metres). A significant main effect of distance was found for walking speed with the PS ($F_{(11.65,\ 104.89)} = 22.47$, $p = 0.000$). Speed was initially low at $0.26 \pm 0.13\ m \cdot s^{-1}$ but increased throughout the 30 minutes to $0.70 \pm 0.21\ m \cdot s^{-1}$ at the end of the first test session which was just over half of the mean self-selected walking speed during the control condition (Figure 4). The control condition was when participants walked barefoot with two intact legs before walking with the PS. A significant increase in speed was seen during the first 10% of the walking distance ($p < 0.01$) during the first test session. A significant main effect of distance was also found for walking speed during the second test session one week later ($F_{(12.07,\ 108.65)} = 4.69$, $p = 0.000$). Initial walking speed ($0.58\ m \cdot s^{-1}$) on the second test session was significantly higher than on the first test session ($p < 0.01$) and then levelled off during the latter half of the walking task at approximately $0.75\ m \cdot s^{-1}$.

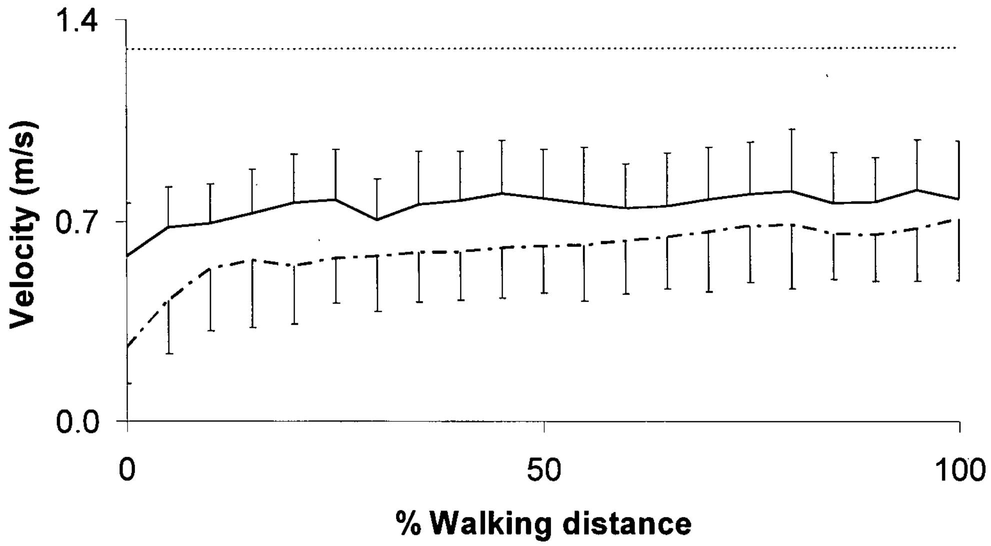

Figure 4. Mean and standard deviation for walking velocity with the PS during the first (dashed line) and second (solid line) test sessions. Dotted line represents average walking velocity in the control condition.

3.1.2 Step characteristics

Step length on the prosthetic side was significantly greater than on the intact side ($F_{(1,\ 9)} = 60.79$, $p = 0.000$) and step length of the collapsed means significantly increased during the walking task ($F_{(9.08,\ 81.74)} = 25.01$, $p = 0.000$) (Figure 5). Post-hoc analysis revealed the increase in step length of the collapsed means was significant during the first 5% of the walking distance ($p < 0.01$). Step length on the intact side was initially negative indicating that the intact foot did not step past the PS. Step length on the prosthetic side remained relatively constant throughout the duration of the walking task while step length on the intact limb increased over time. At the end of 30 minutes, step length on the intact limb when walking with the PS was 73% of step length in the control condition. Furthermore, six of ten participants showed an initial reactive compensatory stepping response of taking two steps with the intact limb for one step with the prosthetic limb to maintain body equilibrium.

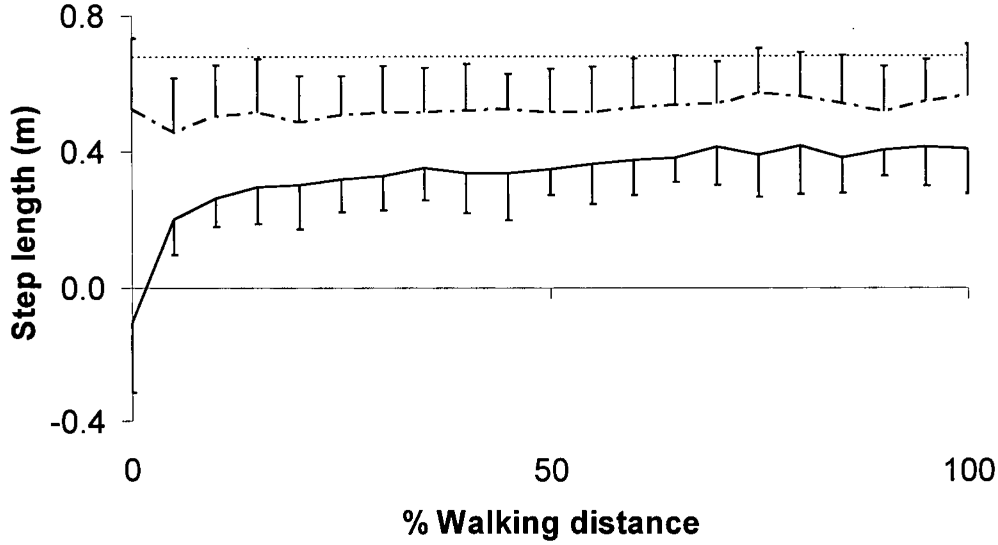

Figure 5. Mean (SD) step length for the intact (solid line) and prosthetic (dashed line) limbs. The dotted line presents the average value in the control condition.

A significant interaction was found for step frequency ($F_{(3.47,\ 31.24)} = 9.68$, $p = 0.000$) (Figure 6) and was attributed to the stepping response seen during the initial steps with the PS. Post-hoc analysis showed that the interaction was significant for both the prosthetic and intact limbs during the first 5% of the walking task ($p < 0.01$). When walking with the PS, step frequency was only 75% of that in the control condition. Decreased step frequency, coupled with shorter step lengths on both limbs when walking with the PS, resulted in a slower self-selected walking speed with the PS than in the control condition. No significant main effect of distance was found for step width when walking with the PS ($F_{(10.51,\ 94.56)} = 0.44$, $p = 0.929$), which was larger than in the control condition (Figure 7).

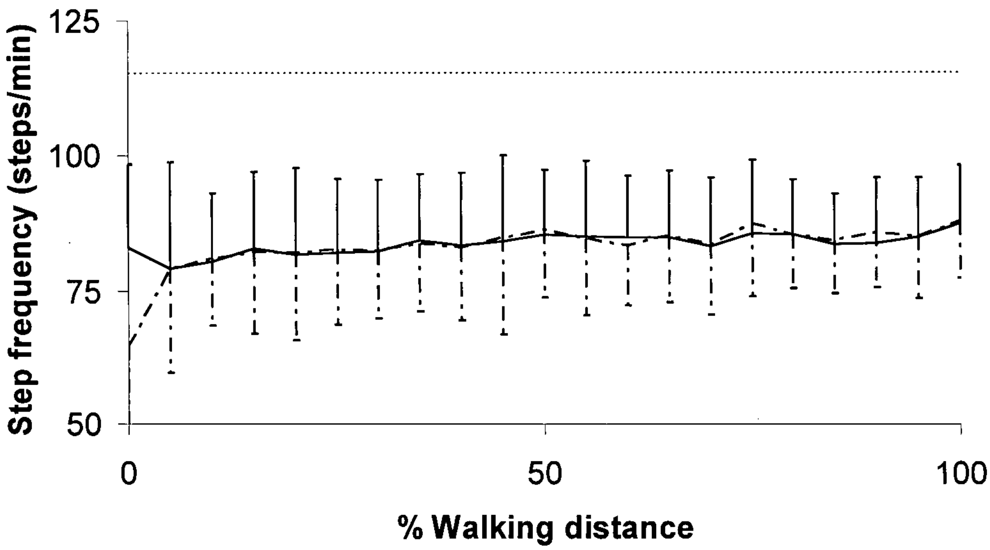

Figure 6. Mean (SD) step frequency for the intact (solid line) and prosthetic (dashed line) limbs. The dotted line presents the average value in the control condition.

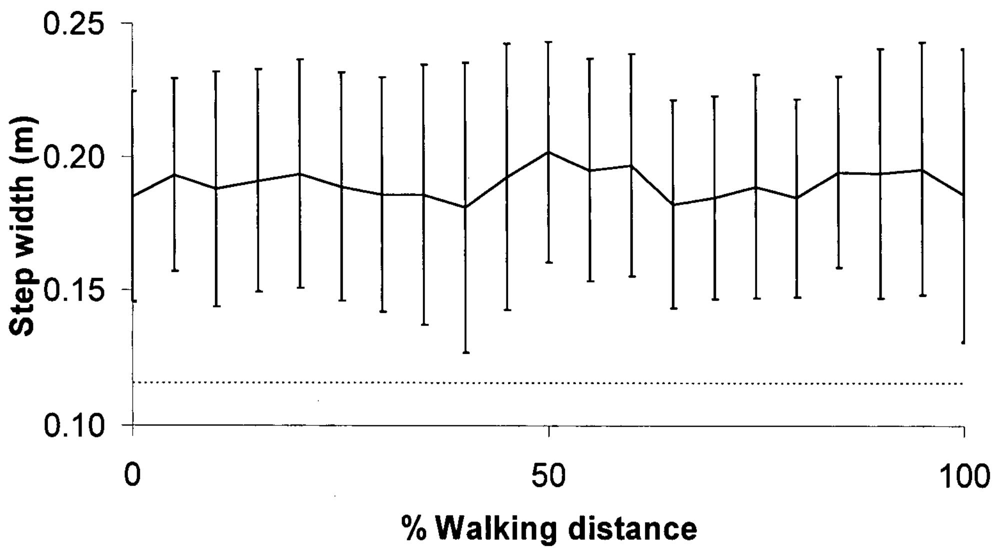

Figure 7. Mean (SD) step width (solid line). The dotted line presents the average value in the control condition.

3.1.3 Body support times

Differences were found in relative (to the gait cycle) body support times. Overall, a significant main effect of distance was found for time spent in stance ($F_{(6.25, 56.21)}$ = 26.99, *p* = 0.000), swing ($F_{(5.98, 53.82)}$ = 22.822, *p* = 0.000) and double support ($F_{(7.79, 70.19)}$ = 18.05, *p* = 0.000) for the prosthetic and intact limbs. Post-hoc analysis revealed these differences were all significant during the first 5% of the walking distance (p < 0.01). Time in stance was initially 86% of the gait cycle compared to 68% on the prosthetic limb, but significantly decreased over distance for both limbs. The intact limb spent a significantly longer time in stance compared to the prosthetic limb throughout the walking task ($F_{(1, 9)}$ = 190.39, *p* = 0.000) (Figure 8). Conversely, time in swing increased over distance for both limbs and was significantly higher on the prosthetic limb compared to the intact limb ($F_{(1, 9)}$ = 207.40, *p* = 0.000) (Figure 9). The main effect of distance showed a significant decrease of time spent in double support during the walking task for both limbs (Figure 10). No main effect for limb was found for time in double support. Although it was initially higher on the intact limb compared to the prosthetic limb (26.5% vs. 21.4%), this difference was not significant ($F_{(1, 9)}$ = 0.72, *p* = 0.419).

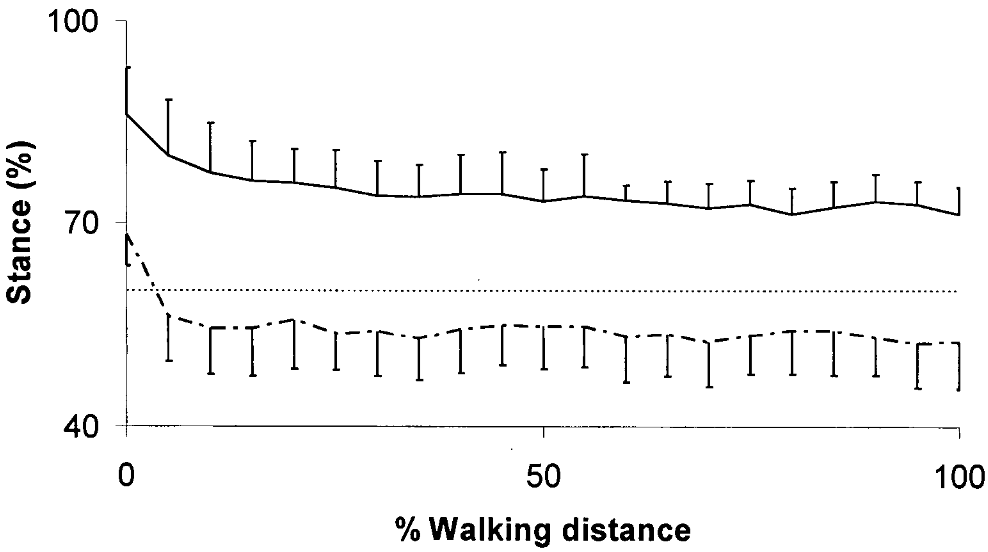

Figure 8. Mean (SD) time spent in stance for the intact (solid line) and prosthetic (dashed line) limbs. The dotted line presents the average value in the control condition.

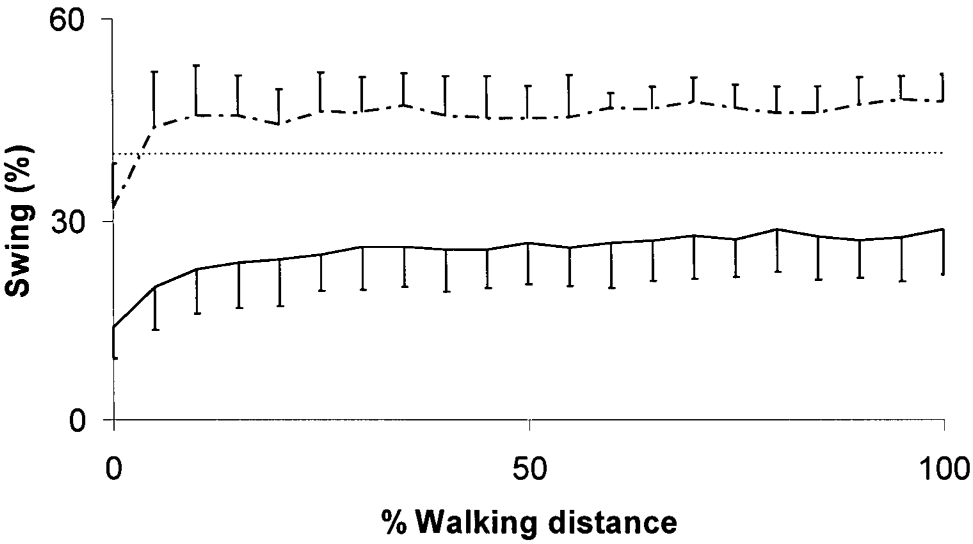

Figure 9. Mean (SD) time spent in swing for the intact (solid line) and prosthetic (dashed line) limbs. The dotted line presents the average value in the control condition.

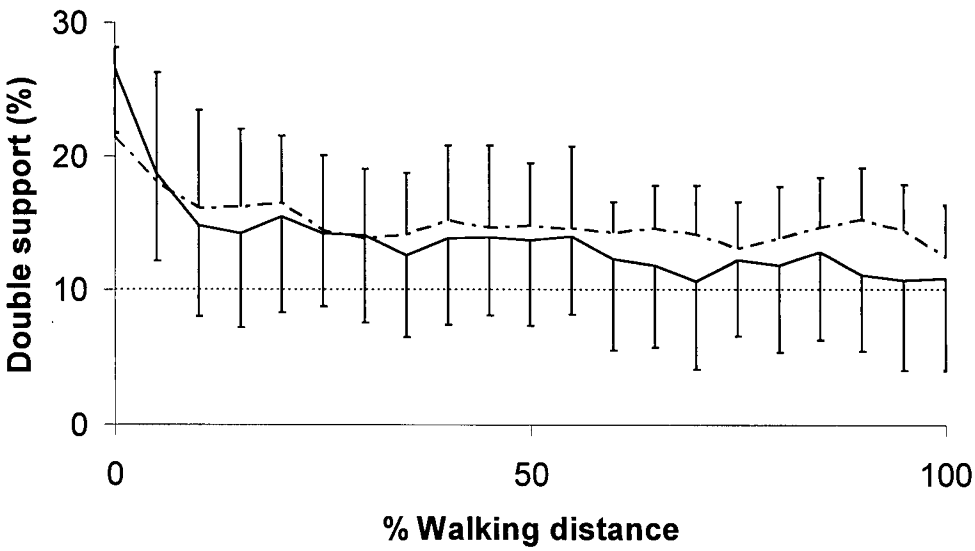

Figure 10. Mean (SD) time spent in double support for the intact (solid line) and prosthetic (dashed line) limbs. The dotted line presents the average value in the control condition.

3.1.4 Body segment angles

The prosthetic angle at the moment of prosthetic contact with the ground, PS-ON, showed a significant increase over distance ($F_{(10.25, 92.29)} = 4.05$, $p = 0.000$), indicating that the prosthetic limb became less vertically oriented at PS-ON over distance walked (Figure 11A). A post-hoc test found this increase to be significant between 5 and 10% of total walking distance only ($p < 0.01$).

Similarly, the shank angle at the moment of intact foot contact significantly increased over time ($F_{(4.64, 41.72)} = 14.04$, $p = 0.000$) and post-hoc analysis showed it significantly increased during the first 5% of total walking distance. The initial negative shank angle indicated that the shank was oriented in front of the vertical and thus, an increase in shank angle implied the shank segment was positioned behind the vertical at intact foot contact. No significant difference over distance was found for the thigh angle ($F_{(7.52, 67.67)} = 1.73$, $p = 0.112$) (Figure 11B) at intact foot contact. Considering the virtually constant absolute thigh angle of approximately 13° behind the vertical at the moment of intact foot contact (Figure 11C), these data showed that the knee on the intact leg was initially flexed and positioned in front of the ankle at the instant of intact foot contact. This initial response could only be seen on the intact limb, since the PS did not have an artificial knee joint that would allow for prosthetic limb knee flexion.

Trunk angle at intact foot contact and PS-ON significantly increased in a positive direction over distance walked ($F_{(7.72, 69.47)} = 5.81$, $p = 0.000$). Trunk angle was initially negative indicating forward lean; therefore, an increase indicated that there was a reduction in forward lean during the walking task. Although not statistically different, a trend could be seen between the intact and prosthetic limbs. Trunk angle at PS-ON remained relatively constant near the vertical, while there was greater forward lean at intact foot contact than at PS-ON ($F_{(1, 9)} = 9.03$, $p = 0.015$) during the 30-minute walking task (Figure 11D).

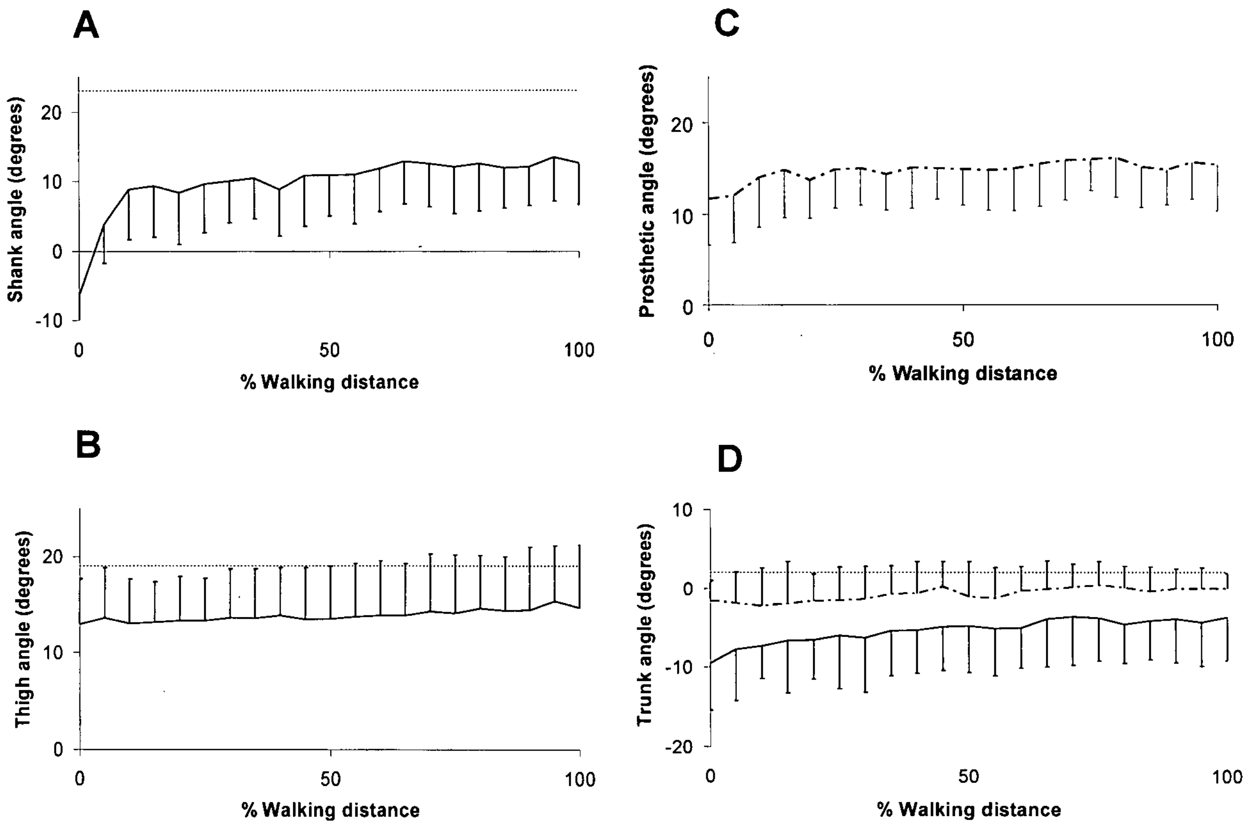

Figure 11. Mean (SD) shank angle (A) and thigh angle (B) at intact foot contact; prosthetic angle (C) at prosthetic contact. Mean (SD) trunk angle (D) at intact foot (solid line) and prosthetic (dashed line) contact. The dotted line presents the average value in the control condition.

3.1.5 Peak vertical position of the centre of mass

Overall, the peak vertical position of the CoM during single stance remained relatively constant throughout the walking task (Figure 12). No significant difference was found in the peak vertical position of the CoM during prosthetic and intact single stance over distance walked ($F_{(7.33,\ 66.01)} = 1.03$, $p = 0.419$). Although no main effect for limb was found ($F_{(1,\ 9)} = 2.546$, $p = 0.145$), the peak vertical position of the CoM was slightly higher during intact single stance than during prosthetic single stance. When walking in the control condition, the peak vertical position of the participants' CoM during right and left right single stance was the same. However, when walking with the PS, the peak vertical position of the CoM was slightly greater (less than 1 cm) than in the control condition throughout the entire walking task for the intact limb and for the first 50% of distance walked for the prosthetic limb.

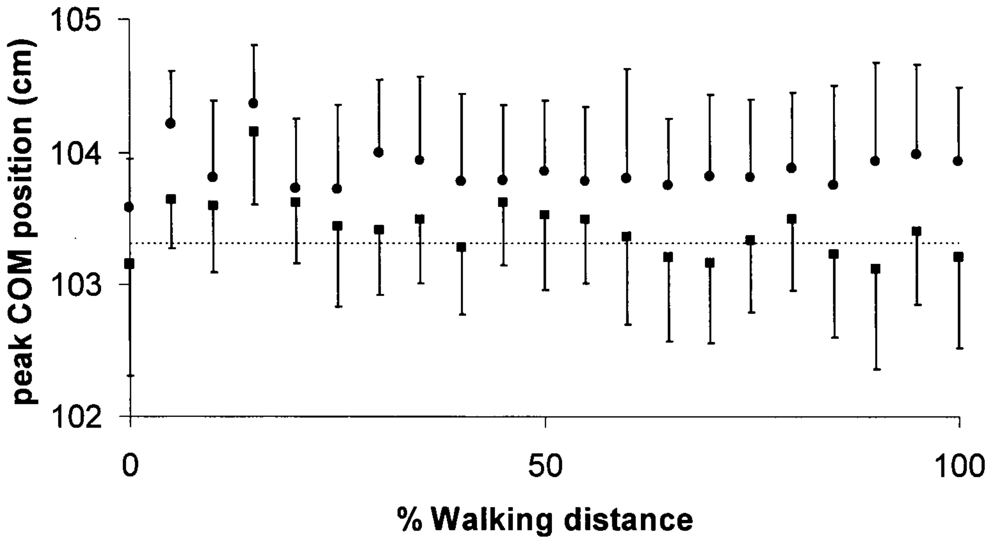

Figure 12. Mean (SD) peak vertical position of the whole body CoM during intact (●) and prosthetic (■) single stance. The dotted line represents the mean peak vertical position of the CoM during single stance in the control condition.

3.2 Variability

3.2.1 Temporal-distance and kinematic variables

Overall, the variability of specific gait variables when walking with the PS, as measured by their standard deviation, was higher than in the control condition. When walking with the PS, variability of walking speed did not significantly change over distance walked ($F_{(8.33, 75.01)} = 1.13$, $p = 0.353$). The variability pattern was very similar for step length and frequency. A significant main effect of distance was found for step length ($F_{(4.18, 37.65)} = 5.94$, $p = 0.001$) and step frequency ($F_{(7.55, 67.91)} = 7.29$, $p = 0.000$) such that both variability of step length and frequency decreased over distance (Figure 13). Post hoc analysis revealed that this difference was significant during the first 5% of distance walked ($p < 0.01$). No main effects of limb for variability of step length and frequency were found ($F_{(1, 9)} = 2.83$, $p = 0.127$; $F_{(1, 9)} = 0.100$, $p = 0.759$, respectively).

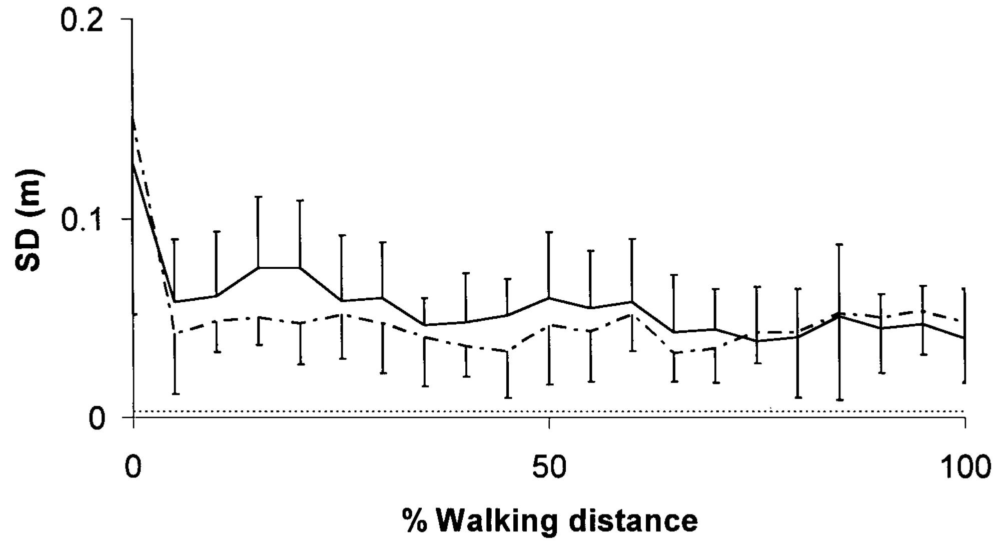

Figure 13. Variability of step length for the intact (solid line) and prosthetic (dashed line) limbs. The dotted line represents the average value in the control condition.

The variability of time spent in stance and swing was the same for both variables and significantly decreased over distance walked ($F_{(13.46,\ 121.11)} = 2.93$, $p = 0.001$) (Figure 14). This decrease was only significant during the first 5% of total distance for both variables ($p < 0.01$). A significant main effect of limb showed that there was significantly greater variability on the prosthetic compared to the intact limb ($F_{(1,\ 9)} = 17.46$, $p = 0.002$) for both stance and swing. Variability of time spent in double support significantly decreased over distance ($F_{(5.48,\ 49.28)} = 3.66$, $p = 0.005$). Although not statistically significant, a trend could be seen between the intact and prosthetic limbs such that time in double support on the intact limb was greater than on the prosthetic limb ($F_{(1,\ 9)} = 8.27$, $p = 0.018$).

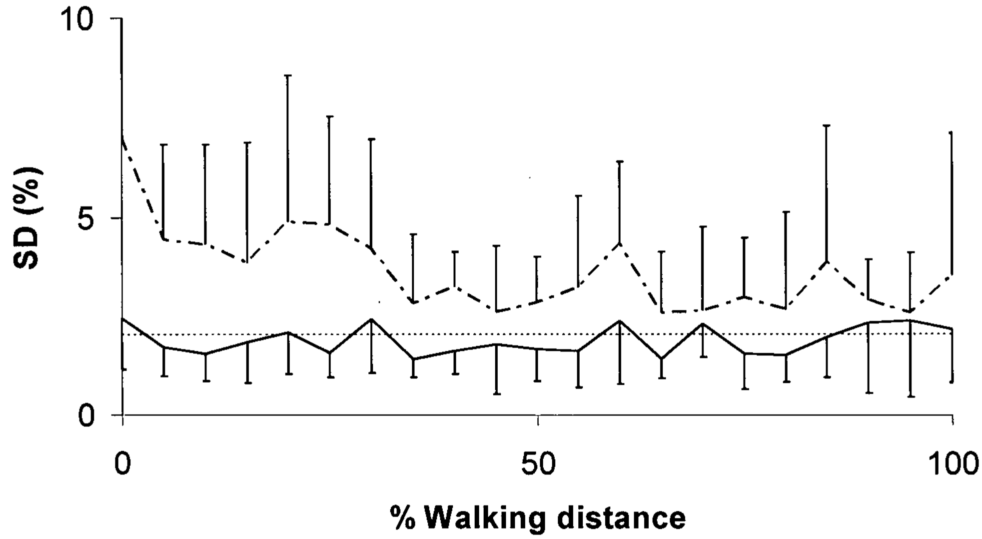

Figure 14. Variability of time spent in stance and swing for the intact (solid line) and prosthetic (dashed line) limbs. The dotted line represents the average value in the control condition.

3.2.2 Body segments and centre of mass

No significant changes of variability were found over distance for either the vertical orientation of intact shank and thigh segments or for the vertical alignment of the prosthesis at instances of intact foot contact and PS-ON, respectively. Although there was no main effect of distance, a main effect of limb was found for variability of trunk lean at instances of intact foot contact and PS-ON. There was significantly more variability of trunk lean at intact foot contact than at PS-ON ($F_{(1, 9)} = 19.71$, $p = 0.002$). No significant changes of variability were found over distance or between limbs for the peak vertical position of the whole body centre of mass during single stance.

3.3 Speed after effect

Mean group self-selected walking speed in the control condition was 1.30 (0.08) $m \cdot s^{-1}$ on the first test session and 1.33 (0.10) $m \cdot s^{-1}$ on the second test session. For the first test session, a significant main effect of time was found for walking speed before and after removal of the PS indicating the occurrence of a speed after effect ($F_{(1.80, 16.18)} = 11.88$, $p = 0.001$) (Figure 15). Post-hoc analysis revealed that participants walked with a significantly slower self-selected walking speed of 1.13 (0.16) $m \cdot s^{-1}$ during the first five strides immediately upon removal of the PS ($p < 0.01$). Walking speed increased in the first ten strides after removing the PS however, it did not increase to the mean walking speed established in the control condition. No significant difference was found for walking speed between the first five and ten strides after removal of the PS. For the second test session, walking speed during the first five strides immediately upon removal of the PS was 1.27 (0.15) $m \cdot s^{-1}$. Although this speed was slower than in the control condition, no significant main effect of time was found ($F_{(1.63, 14.70)} = 2.753$, $p = 0.104$).

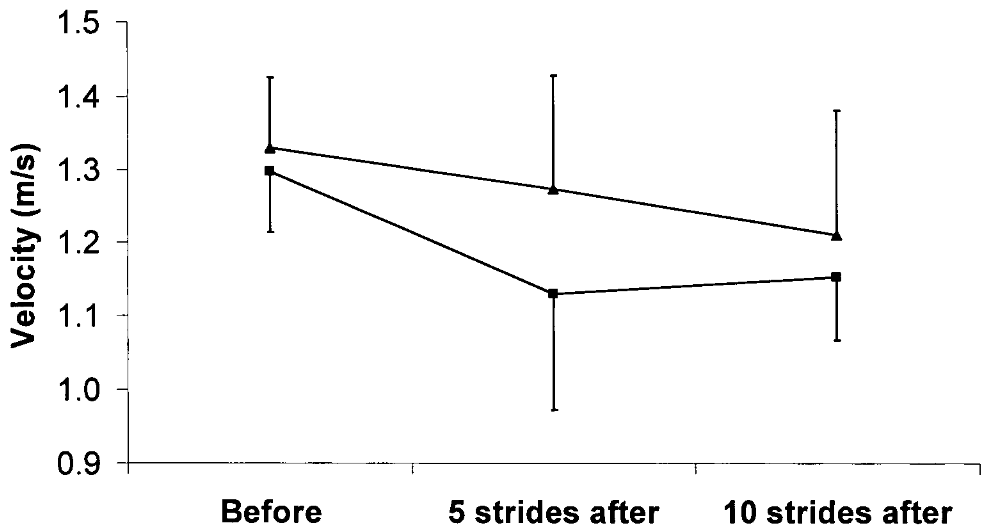

Figure 15. Occurrence of a speed after effect upon removal of the PS following the first test session (■) and second test session (▲) one week later. Graph shows group mean self-selected walking speed before walking with the PS (control condition), the first five strides after and subsequently, ten strides after removal of the PS following the 30-minute walking task.

CHAPTER 4: DISCUSSION

4.1 Temporal-distance and kinematic adaptations

The results from this study show that participants were able to adapt successfully and quickly to the new set of physical constraints imposed by a prosthetic simulator (PS) by adapting kinematic variables. Significant changes occurred early in the task, during the first 5 to 10% of total walking distance. This observation is an important one because it showed that these individuals could quickly adjust to a challenging walking task despite a loss of speed and marked change in stride parameters between the two limbs invoked by wearing the PS. Furthermore, the fact that the greatest changes occurred within the first 10 % of walking distance showed that participants did not need to walk for 30 minutes. Rather, a step by step analysis over a shorter period of time would have allowed us to assess how individuals immediately changed their walking style to surmount the difficulty imposed by the PS.

4.1.1 Walking speed

Participants initially walked at just 37% of their final walking speed (0.70 ± 0.21 m $\cdot$ s^{-1}) on the first test session. However, after having covered only 10% of total walking distance, they had already reached 76% of their final walking speed, showing that the changes occurred rapidly.

Although it might be expected that walking at slower vs. faster velocities allowed for greater control and stability among prosthetic walkers, results from Donker and Beek (2002) suggest that the instability associated with the prosthesis decreased as walking velocity increased. They explained that, at relatively higher velocities, walking consists of a series of controlled falls while at low velocities, walking risks of becoming a sequence of standing postures with less dynamic stability. As well, prosthetic walkers may be more susceptible to falling because their walking pattern is less stable across a range of different walking speeds (Donker & Beek, 2002). In the present study, it was postulated that dynamic stability increased as participants gained experience walking with the PS and walked with a faster walking speed that resembled a sequence of controlled falls rather than of standing postures. Most of the measured variables reached a plateau and did not further increase or decrease after 10% of the total walking distance. Therefore, as participants became more accustomed to the altered sensations of wearing the PS and adapted a modified walking style, their perceived feeling of steadiness and confidence in their ability to ambulate with the PS was greater than compared to the initial steps.

The results from this study showed that participants adopted a walking speed of approximately half of the group mean self-selected walking speed in the control condition. The goal of this study was not to attain a specific walking speed with the PS. However, this finding showed that participants walked considerably slower with the PS than in the control condition. By definition, walking speed is the product of step length and step frequency. That walking speed in the latter half of the first session, as well as between test sessions, remained relatively constant was due to the fact that any subsequent increase in speed would have required the participants to further increase step length and/or frequency beyond a level that had adopted as comfortable or preferred. In return, this would have compromised their perceived level of dynamic stability.

Learning is a set of internal processes associated with practise or experience leading to relatively permanent changes in the capability of a skill (Schmidt & Lee, 1999). When participants walked with the PS a second time, their initial walking speed was significantly higher than initial walking speed on the first session. Although learning cannot be directly observed, the results from this study suggest that learning to walk in a novel situation had occurred during the first test session. This could be seen by the relatively permanent change in walking speed on the second test session in which participants very quickly regained the final walking speed achieved on the first day and maintained it throughout 30 minutes of additional practise.

4.1.2 Step characteristics

The results from this study showed that significantly longer time was spent in swing on the prosthetic limb compared to the intact limb. Also, the results show that prosthetic step length was longer and remained relatively constant throughout the walking task compared to intact step length. Based on these observations, an explanation for these findings is that participants selected a prosthetic step length and modulated their intact limb to their prosthetic side because step length changed on the intact side but remained relatively stable on the prosthetic side. Furthermore, the significant increase in intact step length, from a negative to positive value, reflected a stepping response on the intact side. A stepping response is a type of reactive compensatory movement the participants used to avoid balance loss, when body equilibrium was compromised due to significantly different step lengths between the two limbs. Step length can be negative when one foot does not pass the other foot (Lamoureux, 1971). Considering the initial longer prosthetic step length and shorter prosthetic single stance compared to the intact limb, 6 out of 10 participants took two intact steps for the intact foot to pass the prosthetic limb. This compensatory response was also reflected in the initial higher step frequency observed on the intact limb, after which step frequency remained relatively constant throughout the walking task. Therefore, increases in walking speed were mainly attributed to increases in intact step length rather than step frequency.

The wider step width when walking with the PS was attributed to balance maintenance. Hip abductor function plays an important role in pelvic stabilisation. When pelvic stabilisation was not satisfactorily achieved, individuals with a lower limb amputation often responded by walking with a greater step width (James & Oberg, 1973; Jaegers et al., 1995). Other factors that influence step width include prosthetic alignment and socket fit. When persons with a lower-limb amputation adopt a broader gait pattern, there is a tendency towards greater lateral bending of the trunk, especially over the prosthesis when this is the supporting limb (James & Oberg, 1973). The participants in this study modified step width to maintain the CoM over the base of support and achieve sufficient stabilisation of the pelvis during the stance phase. Proper alignment of the PS was not taken into consideration and may have contributed to the greater step width.

4.1.3 Body support times

Differences in body support times for the prosthetic and intact limbs when walking with the PS were consistent with data reported in the literature for lower-limb amputees. For example, Jaegers et al. (1995) reported mean (SD) stance phase of the prosthetic leg at a comfortable walking speed of 1.01 (0.18) $m \cdot s^{-1}$ of 58% and 63% on the intact leg. Similarly, Sanderson & Martin (1997) reported significantly different values for mean stance on the prosthetic limb (61.8%) and intact limb (65.3%) at a preferred walking speed of 1.2 $m \cdot s^{-1}$. In this study, a significantly greater amount of time was spent in stance on the intact limb which was likely due to the perceived instability associated with the prosthesis. It has been suggested that prosthetic stance time is shortened to avoid pain and minimise the need for controlling the mechanical interaction between the prosthetic limb and the environment (Donker & Beek, 2002). Other factors that have been suggested as contributing to shorter prosthetic stance include reduced confidence in supporting body weight on the prosthetic leg and lack of proprioceptive feedback regarding foot position (Mattes et al., 2000). In this study, the differences in support times between the limbs were attributed to the fact that the participants were trying to protect their prosthetic side by spending less time in prosthetic stance and consequently, loading their intact limb more by spending longer time in intact stance. The participants protected their prosthetic limb due to the discomfort associated with weight-bearing on it. This protective component has previously been reported in the literature (Donker & Beek, 2002; Nolan et al., 2003; Sanderson & Martin, 1997).

Participants reported feeling considerable discomfort when walking with the PS. Immobilisation of the shank and thigh to maintain a 90°-knee angle for 30 minutes induced altered sensory feedback compared to when walking in the control condition. During prosthetic stance, the flexed knee joint, immobilised by the PS, was loaded and supported the participant's body weight during intact stance since the knee was the distal portion of the thigh segment that was connected to the aluminium tubing. This thin tubing (approximately 4 cm in diameter) provided the means of structural support when the prosthetic limb was in contact with the ground. The discomfort associated with the PS was also due to the fact that the rigid plastic cuff had been designed to fit the dimensions of the thigh and shank of one particular individual and thus, was not individualised for each participant. Although measures where taken to improve participants' comfort level with padding, this remained one limitation of the study that had a negative impact on their walking style and the changes in gait variables they made throughout the task. These adverse effects could have included a slower walking speed, shorter intact step length and loading the intact limb more by spending more time in intact stance.

Although time in stance on the intact limb (over 70% stride cycle throughout the entire walking task) was higher than values reported for lower limb amputees (Isakov et al., 1996; Jaegers et al., 1995; Sanderson & Martin, 1997) this was likely due to the novelty of the walking task with the PS. If walking at low velocities is considered a series of standing postures, then the time when one or both limbs are in contact with the ground would expectedly be longer than at higher velocities. Furthermore, data reported in the literature have often been recorded for proficient amputee walkers. Studies have not assessed walking patterns among individuals who are re-learning to walk shortly after a lower-limb amputation in the rehabilitation setting. Therefore, the relative time spent in stance and swing during the gait cycle as the person weight-bears on their prosthesis for the first time has not been reported in the literature. Thus, whether the amount of time spent in intact stance when learning to walk with a prosthesis for the first time is actually as large for individuals with a lower-limb amputation as the support times observed for the participants learning to walk with the PS, remains unclear. The causes of long intact stance in the current study reflect a combination of factors including the novelty of walking with the PS, altered kinaesthetic feedback, discomfort and slow initial walking speed.

4.1.4 Body segment angles

The shank and prosthetic limb segments were more vertically oriented at instances of intact foot contact and PS contact (PS-ON), respectively, during the walking task with the PS than during the control condition. On the intact side, participants modulated the vertical orientation of the shank segment but not the thigh. The shank was initially positioned in front of the vertical at intact foot contact. Thus, with the thigh angle relatively constant throughout the walking task, the knee on the intact limb was initially flexed and positioned in front of the ankle at foot contact. This strategy maintained the CoM *above* the supporting limb and minimised its displacement in the direction of walking progression during the intact stance phase. However, within just 10% of the total walking distance, the knee became more extended and was positioned behind the ankle at intact foot contact. Therefore, the CoM was *behind* the supporting limb at intact foot contact and needed to be elevated over top of the intact limb during the stance phase. This change in knee joint angle and position on the limb at intact foot contact indicated that participants had achieved a level of perceived stability associated with greater vertical and horizontal displacement of the CoM during walking without compromising their balance requirements.

The participants also modified their posture such that, by keeping the prosthetic limb more vertically aligned at PS-ON, they maintained the CoM above the supporting limb. This way their CoM was closer to their base of support resulting in a smaller displacement of the CoM during the prosthetic stance phase. By abducting the hip on the prosthetic side during swing, the participants could gain a longer step length while maintaining the limb more vertically oriented at PS-ON. The orientation of the intact lower limbs enhanced stability by shifting the CoM anteriorly. The greater forward trunk lean (measured as the negative absolute trunk angle) on the intact side helped to maintain the CoM above the intact supporting limb at foot contact. Considering a significantly greater amount of time was spent in single support on the intact limb, this was an important balance strategy when walking with the PS. The vertical orientation of the intact lower-limb and prosthetic segments at intact foot contact and PS-ON, respectively, were quite similar after the first 10% of total distance walked (mean shank: 11°; mean thigh: 14°; mean prosthetic: 15°). This may provide some information regarding the optimal vertical alignment of lower limb segments at intact foot/prosthetic contact as the participants in this study learned to walk with a PS.

The fact that both the intact and prosthetic limbs were very similarly oriented indicated that participants tried to maintain a symmetrical walking style, despite the inherent asymmetry imposed by the PS. Large vertical displacement of the CoM during the stance phase of one limb, relative to the other limb, could have implications for the metabolic costs associated with walking with the PS. Comparable movements produced by both the prosthetic and intact limbs would lead to similar CoM motion profiles during the stance phase of the gait cycle. Had the shank segment angle been much greater (positioned farther behind the vertical) at intact foot contact than the prosthetic angle at PS-ON, then the CoM would have been positioned behind the supporting limb (as opposed to above) and would have been displaced more in the horizontal and vertical directions than during prosthetic stance. Thus, the similar kinematic profiles of the lower limb body segments helped maintain more similar CoM profiles during the intact and prosthetic stride cycle.

4.1.5 Peak vertical position of the centre of mass

The findings of this study did not support the hypothesis that the peak vertical position of the CoM would be greater during prosthetic single stance than during intact single stance. This was hypothesised because the prosthetic limb consisted of a rigid link rather than two segments that could move in synchrony to flex and extend throughout the gait cycle as the thigh and shank segments of the intact limb could.

Data from several pilot studies have shown that peak vertical position of the CoM was achieved when both the shank and thigh segments were vertical during single stance. Walking with the PS did not allow for artificial knee flexion on the prosthetic side. Therefore, in order to safely clear the ground during prosthetic swing, participants maintained their intact lower limbs more vertically aligned during intact single stance. Consequently, this likely resulted in a somewhat greater peak vertical position of the CoM during intact single stance reported in the present study.

During walking, the body's CoM must be kept above the supporting limb. More time was spent in intact vs. prosthetic stance therefore, maintaining the CoM above the intact limb was an important strategy to control body equilibrium when walking with the PS. Considering the shortened time in prosthetic single support and temporal asymmetry between the two limbs, peak vertical position of the CoM may not have been achieved during prosthetic single support but rather in the following double support phase in the middle of the gait cycle.

4.2 *Variability*

Variability of various parameters was examined by computing and analysing the standard deviation of each variable. The results from this study showed that the variability of specific gait parameters decreased over distance walked. Overall, there was more variability in gait parameters when walking with a PS than for the participants in this study when they walked in the control condition (over ground at a preferred speed of $1.30 \pm 0.08\ m \cdot s^{-1}$).

4.2.1 *Temporal-distance and kinematic variables*

It has been reported that walking speed influences movement variability (Brisswalter & Mottet, 1996). Speeds near the walk-run and run-walk transition (i.e. $2.0 - 2.2\ m \cdot s^{-1}$) and very low velocities ($0.2 - 0.6\ m \cdot s^{-1}$) showed a corresponding increase in the variability of stride characteristics (Brisswalter & Mottet, 1996). In our study, mean walking speed with the PS throughout the first two thirds of the distance walked on the first test session was between $0.2 - 0.6\ m \cdot s^{-1}$. The degree of variability of gait characteristics when walking with the PS, such as step length and frequency and body support times, was greatest at the initial walking speed of $0.26 \pm 0.13\ m \cdot s^{-1}$. Subsequently, variability decreased throughout the first two thirds but levelled off during the last third of the walking task (Figures 13 & 14). Given the association of slow walking speeds and increased variability of stride characteristics, a speed effect contributed to the increased variability observed during the initial stages of the walking task.

Significantly greater variability of time spent in stance and swing was observed on the prosthetic compared to the intact limb. Given the novelty of the task constraints, this finding is an expression of exploratory behaviour on the prosthetic side, as well as reflects the instability and discomfort associated with the PS.

4.2.2 Body segments and centre of mass

It has been reported that greater variability of body segment orientation would likely be observed before and at the instant of foot contact, thereby allowing for more flexibility and adaptability to unexpected perturbations (Heiderscheit et al., 2002). Although the participants in the present study were walking in the secure environment of a clinical gait laboratory, free of unexpected perturbations, the initial novelty of the task led to some level of uncertainty prior to intact foot/prosthetic contact. The variability associated with the vertical orientation of the lower limbs at intact foot contact did not significantly change over distance walked. However, it was much larger than during the control condition. The average standard deviation of the intact shank and thigh angles were more than threefold of that observed during the control condition. Additionally, forward trunk lean was more variable at intact foot vs. prosthetic contact when walking with the PS.

The increased variability associated with intact foot contact provided the flexibility to make compensatory postural adjustments for maintaining the CoM within the base of support at that instant. The observation of increased variability on the intact or unaffected leg has previously been reported in the literature. Heiderscheit et al (2002) investigated variability of stride characteristics and body segment patterns among individuals with unilateral patellofemoral pain. These authors suggested that the increased variability reported on the unaffected limb may have provided the adaptability to compensate for the more constrained patterns of the affected limb. Therefore, the results of this study support the notion that greater variability on the intact limb played a role in the participant's ability to adapt to the constraints imposed by the PS by providing the flexibility to make postural adjustments when dynamic stability was perturbed and lower-limb mechanics were altered.

4.3 Speed after effect

The findings from this study indicated the occurrence of a short-term speed after effect following continuous walking in a novel situation with a PS. A significant decrease in group mean self-selected walking speed was seen during the first five strides immediately upon removal of the perturbation. Although walking speed increased somewhat during the subsequent five strides and was not significantly different from the control condition, it did not reach the baseline level within ten strides. Additionally, with subsequent practise of walking with altered lower-limb mechanics during a second test session, no significant difference in walking speed between the control condition and after removing the PS was found.

Previous studies have reported adaptational after effects upon removal of a perturbation, providing strong evidence that learning had occurred (Jensen et al., 1998; Lam et al., 2003). The main reason for the occurrence of a speed after effect in our study was not attributed to overall fatigue following 30 minutes of continuous walking with the PS. Rather, when learning to walk with altered lower-limb mechanics, participants adapted to altered sensory feedback from the lower limbs by adapting their existing locomotor programme. Slower walking speed was an expression of the adapted motor behaviour of walking with a PS. As well, altered kinaesthetic feedback when walking with the PS was the result of an abrupt change in the pattern of afferent input from the immobilisation of the lower limbs with the PS. Therefore, when asked to revert back to walking with two intact limbs following removal of the perturbation, participants had to switch back to an able-bodied locomotor pattern and reintegrate input from afferent sources that had been modified because of the PS. This adaptation manifested itself as a short-term speed after effect and indicated the role of feedforward mechanisms to adapt to the perturbation caused by the PS. The presence of an after effect reinforced the assumption that adaptation had occurred.

4.4 Neurophysiological considerations

The neuronal circuitry evoked during walking is always aimed at maintaining the body's CoM within the base of support (Dietz & Duysens, 2000). During bipedal gait, specific neuronal mechanisms maintain the body in an upright position and these must correspond to the requirements of the task (Dietz, 2002). A combination of sensory inputs can provide the central nervous system (CNS) with the required information to control body equilibrium. Individuals who have altered lower-limb mechanics and sensory feedback imposed by wearing a PS likely depend on a different combination of afferent input to signal the location of the CoM with respect to the ground during walking. Adjusting to walking with altered lower-limb mechanics and sensory feedback was accomplished presumably by adaptations made at both the spinal levels and from signals from higher brain centres.

Afferent information comes from several different sources including visual, vestibular and somatosensory input. The following discussion will mainly focus on proprioceptive feedback since altered sensory input is one consequence of walking with the PS. At the spinal level, several receptor pathways are responsible for providing sensory input to the spinal cord during walking. Muscle spindles provide information regarding muscle stretch or length, Golgi tendon organs provide information regarding muscle tension developed at the tendons, mechanoreceptors in the skin, especially of the foot, provide load-related afferent input and extensor load receptors in the hip joints contribute to the activation patterns of leg muscles during walking (Dietz, 2002; Dietz & Harkema, 2004).

When walking with the PS, the foot of the limb fitted with the PS was not in contact with the ground and therefore, did not provide information signalling body loading. However, when participants made prosthetic ground contact, they loaded their knee which was the distal part of the thigh that was connected to the aluminium tubing that provided the structural means of support. Therefore, mechanoreceptors in the skin surrounding the knee joint likely provided load information during walking with the PS and may have contributed to the rapid adaptation of walking with altered sensory feedback.

It has been documented in the literature that individuals make modifications for the loss of the plantarflexor muscles by making compensations at joints distal to the site of the amputation, especially at the hip joint (Czerniecki & Gitter, 1996; Latash, 1998; Winter & Sienko, 1988). When individuals walked with the PS, compensations for the loss of energy generation by the plantarflexors and absence of knee flexion were made by the muscles surrounding the hip joint. Evidence of these compensations was also supported by the fact that participants abducted their hip during swing on the prosthetic side for sufficient ground clearance given the absence of knee flexion. During walking, afferent input from hip joint muscles is important for shaping the activation pattern of leg muscles and controlling phase transitions from stance to swing (Dietz & Harkema, 2004). It has also been well-documented that individuals who have lost the function of the ankle plantarflexors have a greater reliance on the hip extensor muscles to generate energy during late stance (Czerniecki & Gitter, 1996; Latash, 1998). Therefore, proprioceptive afferent input from the hip extensor muscles could have played an important role in signalling load information that allowed for the quick adaptation to walking with altered lower-limb mechanics imposed by the PS.

At the subcortical level, the cerebellum controls movement indirectly by regulating the output of the major descending motor systems from the brain. The cerebellum compensates for deviations between the intended movement and the actual movement by comparing internal and external feedback signals. Because the cerebellum receives input from the periphery and from all levels of the CNS, it can modify central motor programmes so that subsequent movements can accomplish their goals with fewer errors (Kandel et al., 1995). The cerebellum plays an important role in the learning of motor tasks. Walking with the PS was a novel task and required participants to form adaptive locomotor strategies to compensate for the altered lower-limb mechanics and sensory feedback. This adaptation of shaping and refining movement patterns to accomplish walking with the PS most likely occurred in the cerebellum.

The role of higher centres during voluntary movement is important because movements organised by the brainstem and spinal cord would have a stereotypical pattern and non-adaptable quality if they could not be modified to accomplish specific tasks by higher brain centres (Kandel et al., 1995). A combination of somatosensory information, including proprioceptive input from mechanoreceptors in the knee and extensor load receptors in the hip, would be logical candidates for the type of information provided to the cerebellum to adapt to walking with the PS.

4.5 Conclusion

The purpose of this study was to assess how individuals respond to altered lower-limb mechanics imposed by a PS. This was assessed by following the changes in temporal-distance and kinematic variables throughout 30-minutes of continuous walking with the PS; by tracking variability over distance walked and by detecting the presence of a speed after effect.

In summary, the results from this project have shown that individuals learned to adapt to altered lower-limb mechanics and sensory input during walking with a PS and did so relatively quickly. By modifying gait variables, such as walking speed, step and body support characteristics and the orientation of body segments, the participants in this study adapted to walking with a PS. Although variability decreased initially, it remained higher over total distance walked than during the control condition. Increased variability is a feature of a system in transition and was probably an expression of a combination of compensatory (on the intact limb) and exploratory (on the prosthetic limb) behaviour given the altered lower-limb mechanics. Finally, learning to walk in a novel situation was inferred from the relatively permanent change in walking speed that was measured on the first and second test sessions. Furthermore, the occurrence of a short-term speed after effect reinforced the assumption that learning had occurred.

CHAPTER 5: FUTURE CONSIDERATIONS

This project has shown that individuals can learn to walk with modified lower-limb mechanics relatively quickly and develop a locomotor pattern that allows them to ambulate continuously for 30 minutes. The results from this project also provide a novel outlook for rehabilitation strategies. The potential of tracking changes in gait variables among able-bodied individuals, as they learn to walk in a prosthetic situation, may be extended to individuals who have actually experienced the amputation of a lower limb.

In light of these findings, rehabilitation programmes should strongly encourage patients to walk as soon as possible. Although one might be inclined to suggest that learning to walk with altered limb mechanics and sensory input occurs relatively quickly, and therefore patients who have actually lost a lower limb should be forced to immediately walk as part of a rehabilitation programme, caution should be taken with this approach for several reasons.

Firstly, the participants in this study were young, able-bodied individuals who did not have to deal with many of the personal issues that impact the successful rehabilitation of actual patients. These include factors such as phantom-limb pain, cardiovascular function, psychological factors, logistical and financial concerns. Such factors influence the patient's ability to deal with the demanding task of relearning locomotor skills such as walking. The amputee population is diverse and recommendations from studies examining young, healthy patients may not be applicable for the elderly, dysvascular patient.

Secondly, whether elderly, able-bodied patients would adapt to walking with the prosthetic simulator, and as quickly as the young participants in this study did, remains to be investigated. Given that no walking aids and/or harness were used in this study, implications associated with the fear of falling would influence elderly individuals in a negative way when walking in a novel situation more so than young participants.

Thirdly, creating a simulator that more closely resembles the prosthesis of a lower-limb amputee, by incorporating an artificial knee joint and prosthetic foot, would create an even more challenging task. The necessity of controlling artificial knee and ankle joints are part of the real prosthetic walking experience. Therefore, testing a more varied able-bodied population in a more realistic prosthetic experience creates the potential for more specific implications for a group of lower-limb amputees.

Considering the greatest improvements in walking speed were made early in the task and that the gait variables appeared to have achieved a plateau after the first 10% of the total distance walked, indicated that adaptive strategies occurred early in the task. Future analysis should especially focus on the initial steps. Changes in variables should be compared between steps rather than between completed laps, as in the current study. Participants may therefore not be required to walk for 30 minutes, perhaps eliminating some of the discomfort associated with wearing the PS over an extended period of time.

During the initial stage of walking with the prosthetic simulator, participants probably acquired new neuromuscular patterns required to successfully ambulate with the device. Future research can also incorporate the use of electromyography to observe neuromuscular compensations and muscle pattern reorganisation on both limbs. The use of force platforms would provide additional insight regarding internal strategies that cause the movement and may reveal the level of internal adjustment. Therefore, kinetic analysis of ground reaction force components and joint moments on the prosthetic and intact limbs should be included as part of subsequent analyses. Investigating balance strategies as participants walk with the prosthetic simulator should include analysis of upper body motion.

This study provides an interesting and novel perspective on the use of a prosthetic simulator that allows non-amputees walk in a prosthetic experience. Future studies can use the information gained from this project to form a more complete picture and assess the gait adaptations individuals make when lower-limb mechanics have been altered.

CHAPTER 6: GLOSSARY

Centre of mass: With the concept of the CoM, it is assumed that all of the body's mass is concentrated at that point. During able-bodied gait, the body's CoM follows a rhythmic upward and downward pattern in the sagittal plane of progression. Two vertical peaks occur at approximately 25% and 75% of the stride cycle, when the limb is in single support and the other limb is swinging forward. The CoM is also displaced laterally in the horizontal direction. Two summits also occur, but this time to the right and left extremities during the support phases as weight-bearing occurs on the right and left limbs. When the vertical and horizontal displacements of the CoM are projected onto the coronal plane, they describe a figure of eight (Saunders et al., 1953).

Double support: Support is transferred from the trailing leg to the leading leg during an intermediate period where both feet are in contact with the ground. In able-bodied gait, typically 16-22% of the stride cycle duration (Winter, 1987).

Dynamic stability: Reflects the ability to ambulate in a functional manner that does not compromise body equilibrium even with the occurrence of a perturbation.

A **perturbation** may be in many forms including a slip or trip, presence of an obstacle along the path of progression, or may be imposed by altered lower-limb mechanics, such as with a prosthetic simulator.

Gait cycle: The period of time from one event (typically initial intact foot/prosthetic contact) of one limb until the occurrence of the same event with the same limb.

Intact foot contact: When the intact foot makes initial contact with the ground.

Kinematic: Kinematic analysis describes the linear and angular motion of markers, bodies or body segments.

Kinetic: Kinetic analysis provides information into the internal strategies that cause the change in movement and may reveal the level of internal adjustment. Typically examines components of ground reaction forces, moments of force, mechanical power work and energy.

Prosthesis: A prosthesis is the artificial substitute of a body part. For an individual with a lower-limb amputation, the prosthesis generally consists of a socket, shank and foot and provides the necessary structural support following the loss of a limb.

Prosthetic contact (PS-ON): When the end of the tubing of the prosthetic simulator makes initial contact with the ground.

Prosthetic simulator (PS): A prosthetic device designed to allow able-bodied individuals walk in a prosthetic experience. An in-house design unilateral lower-limb PS was created for this study. The PS consisted of a plastic cuff into which the knee was flexed at 90° and securely maintained in this position with Velcro straps. A rigid telescoping aluminium tube provided the means of support at the lower extremity.

Sagittal plane: The anterior-posterior vertical plane that passes through the body from front to back, dividing the body or body segments into right and left parts.

Single support: The period of time when only one foot is in contact with the ground.

Stance: The period of time when the leg that is not swinging forward is providing the support. In able-bodied gait at a preferred speed, typically 58-61% of the stride cycle duration (Winter, 1987).

Step frequency: The rate at which a person walks. Expressed in steps per minute ($\text{steps} \cdot \text{min}^{-1}$).

Step length: The distance from the point of contact with the ground of the intact foot/prosthesis to the subsequent occurrence of the same point of contact with the other foot/prosthesis. A comparison of right and left step length provides an indication of gait asymmetry (Lamoureux, 1971). If one foot comes even with the other, step length will be zero; if one foot does not pass the other foot, step length will be negative. Expressed in metres (m).

Step width: The side to side distance between the foot and prosthetic end. Typically measured from the ankle joint centre. Expressed in metres (m).

Swing: The period of time when the leg that is not providing the support is swinging forward. In able-bodied gait at a preferred speed, typically 39-42% of the stride cycle duration (Winter, 1987).

Symmetry: Symmetry in walking has been defined as the perfect agreement between the actions of the lower limbs such that both limbs behave identically (Sadeghi et al., 2000). In healthy, able-bodied individuals, kinematic and kinetic gait patterns deviate by only a small percentage from perfect symmetry (Kim & Eng, 2003).

Walking: The process of human locomotion in which the erect moving body is supported alternately by one leg, followed by the other (Lamoureux, 1971).

Walking speed: The rate of change of linear displacement of the whole body centre of mass along the direction of progression and measured over one or more strides. Expressed as metres per second ($m \cdot s^{-1}$).

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APPENDIX A: PARTICIPANT DATA

The following pages contain data from each of the participants in this study. The data include:

- walking velocity for both test sessions
- relative body support times
- step characteristics
- vertical orientation of lower-limb segments
- vertical orientation of the trunk
- peak vertical position of the whole body centre of mass during single stance
- vertical displacement of the whole body centre of mass in the sagittal plane

APPENDIX A1: WALKING VELOCITY

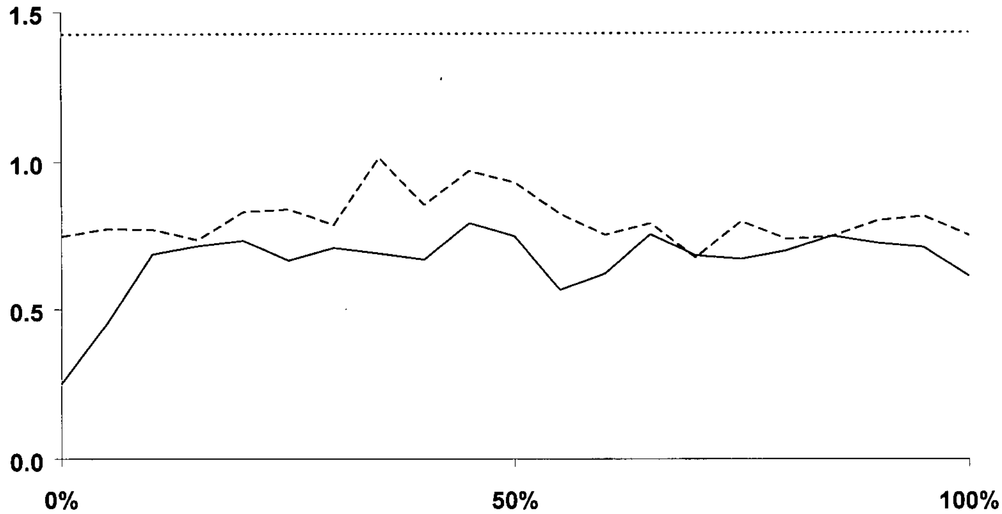
The data in the following pages contain walking velocity for each participant as they walked with the prosthetic simulator during the first and second test sessions. The solid line represents the first session; the dashed line represents the second session; the dotted line represents self-selected walking speed during the control condition (able-bodied gait).

Units are expressed as a percentage (%) of total distance walked for the horizontal axis and walking speed ($m \cdot s^{-1}$) for the vertical axis.

Abbreviations

P = participant

P1



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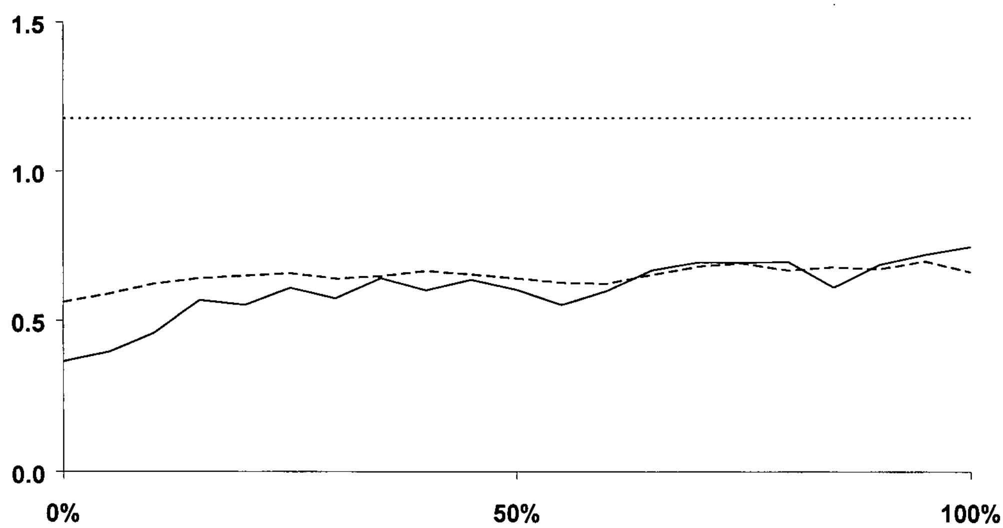

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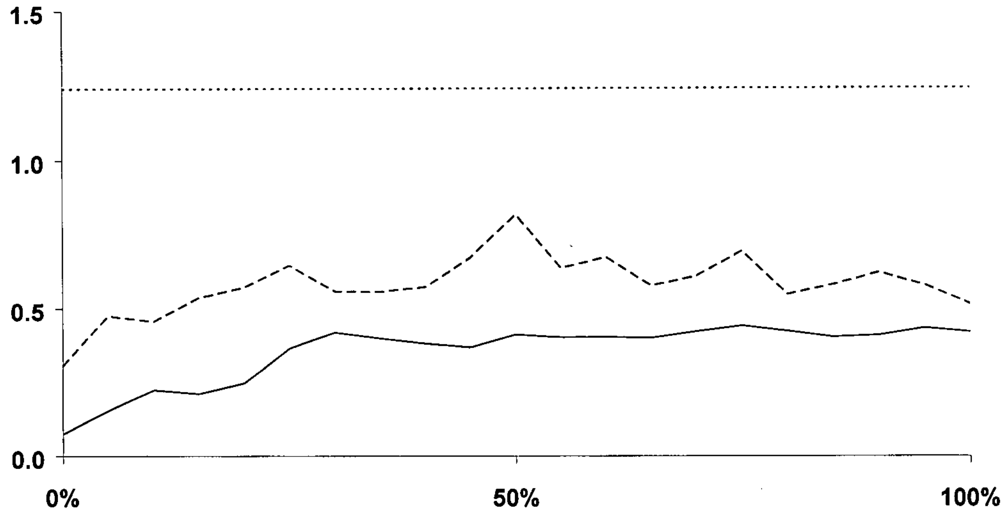

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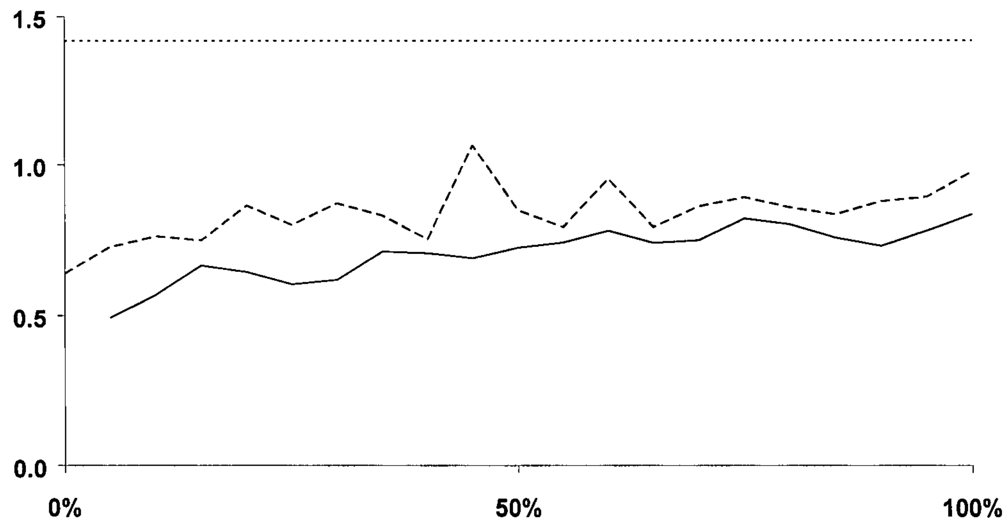

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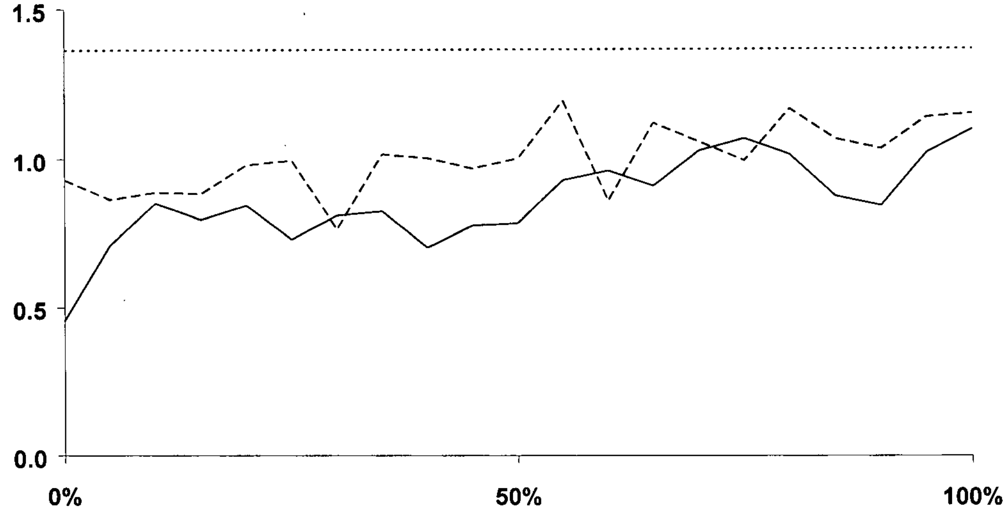

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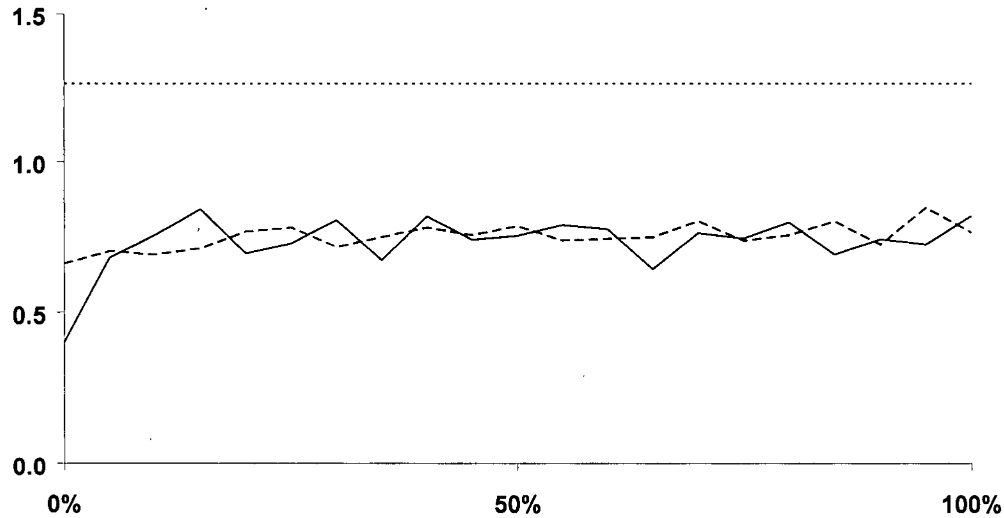

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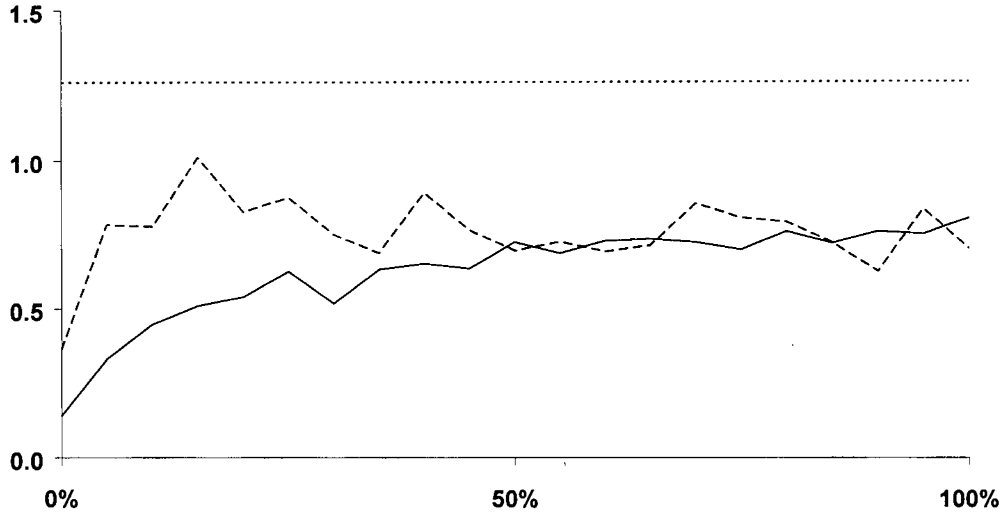

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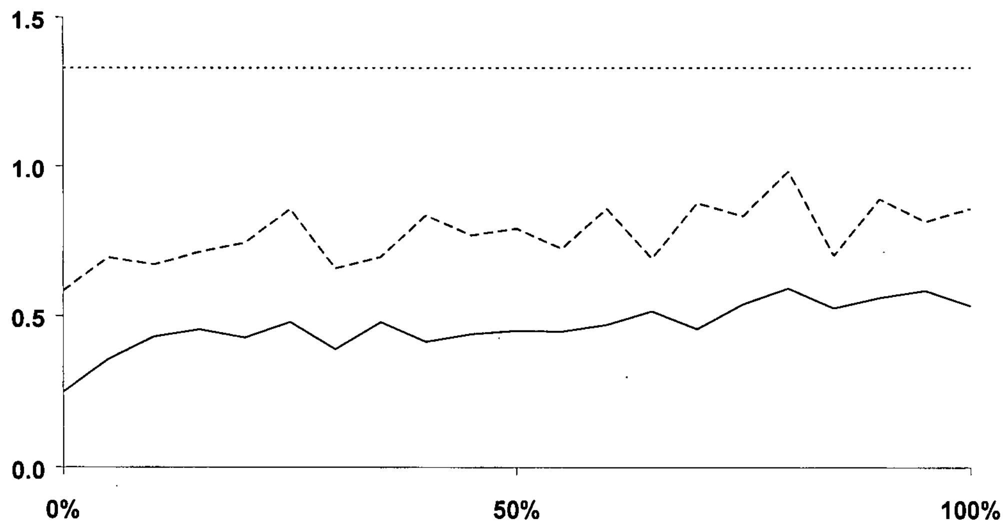
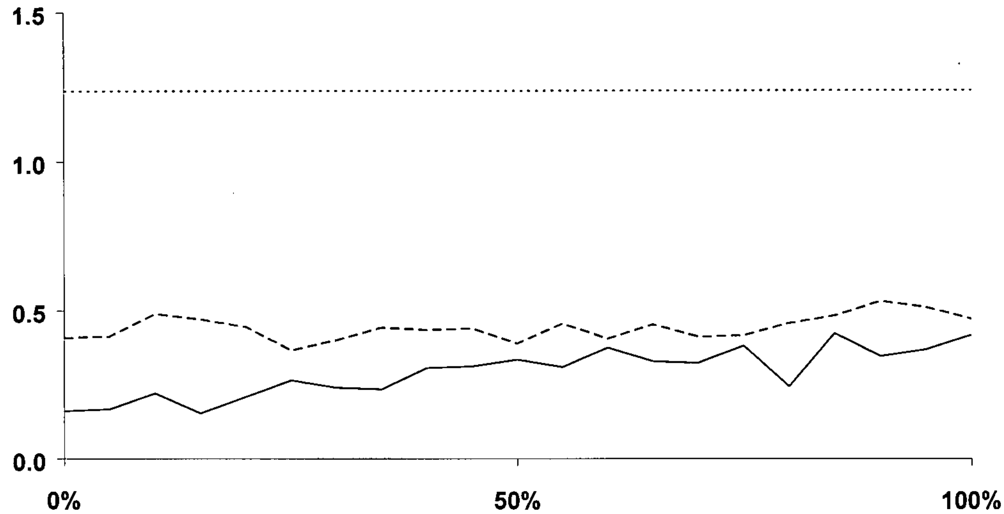


P9



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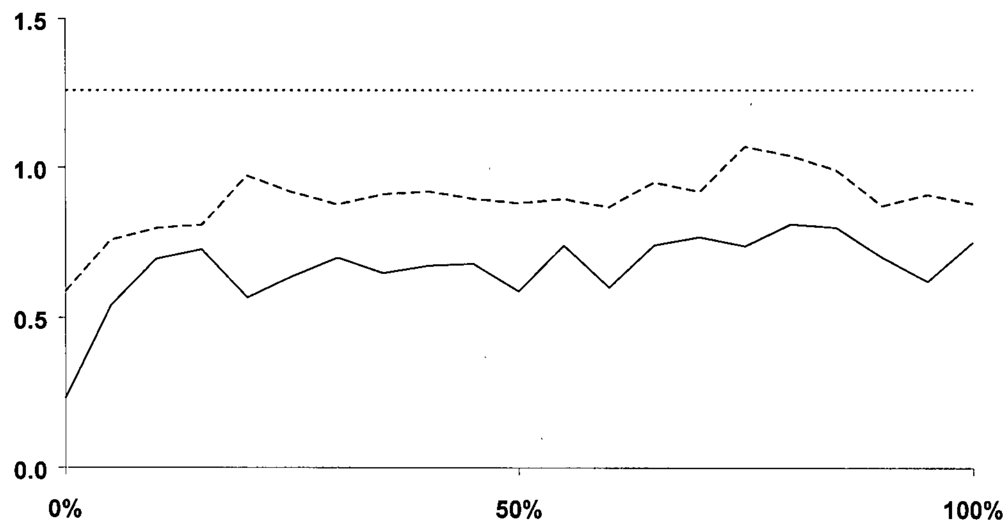


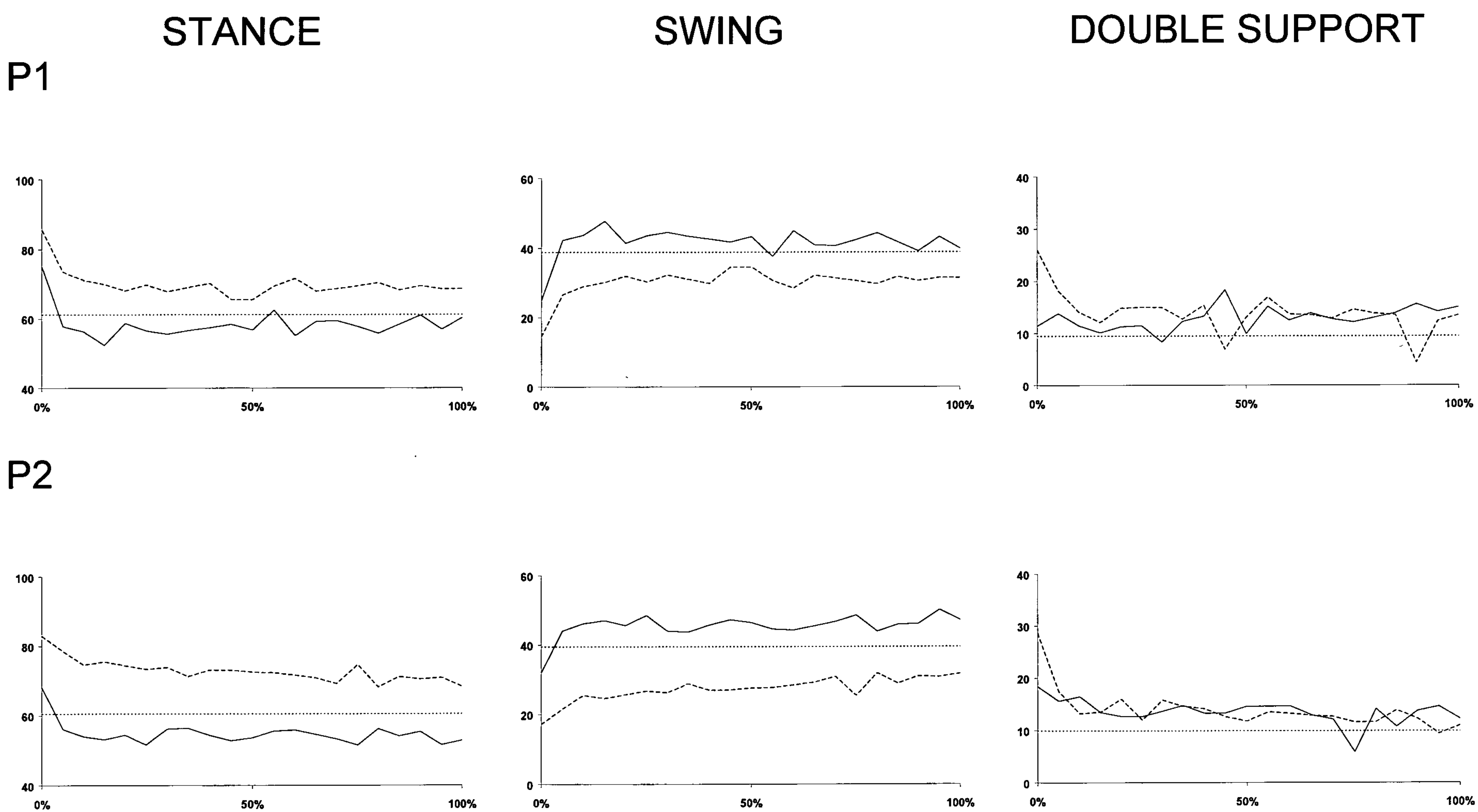
APPENDIX A2: BODY SUPPORT TIMES

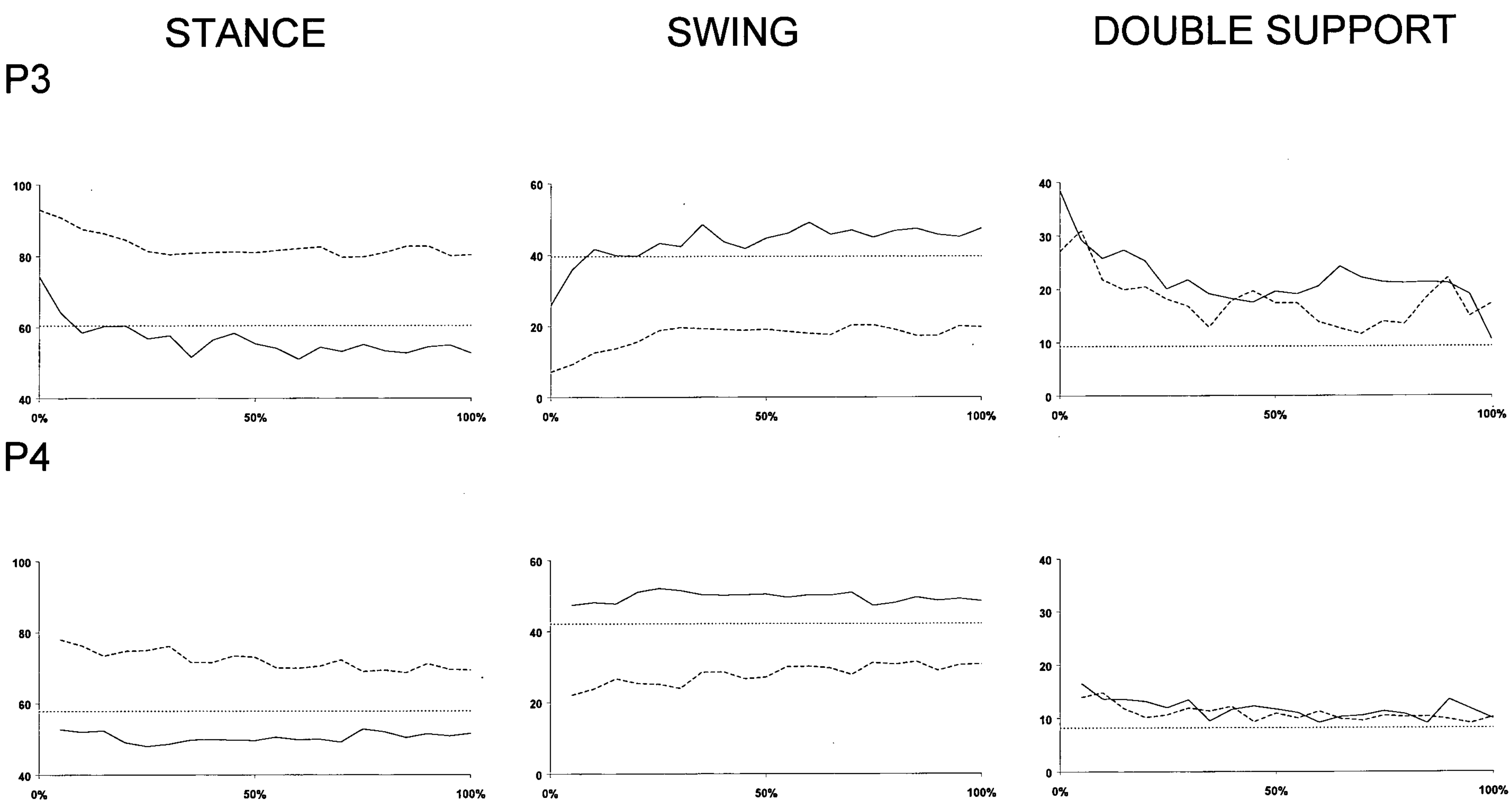
Data presented in the following pages contain body support profiles for each participant as they walked with the prosthetic simulator. The solid line represents the prosthetic limb; the dashed line represents the intact limb; the dotted line represents the control condition (able-bodied gait).

Units are expressed as a percentage (%) of total distance walked for the horizontal axis and as relative body support time (%) for the vertical axis.

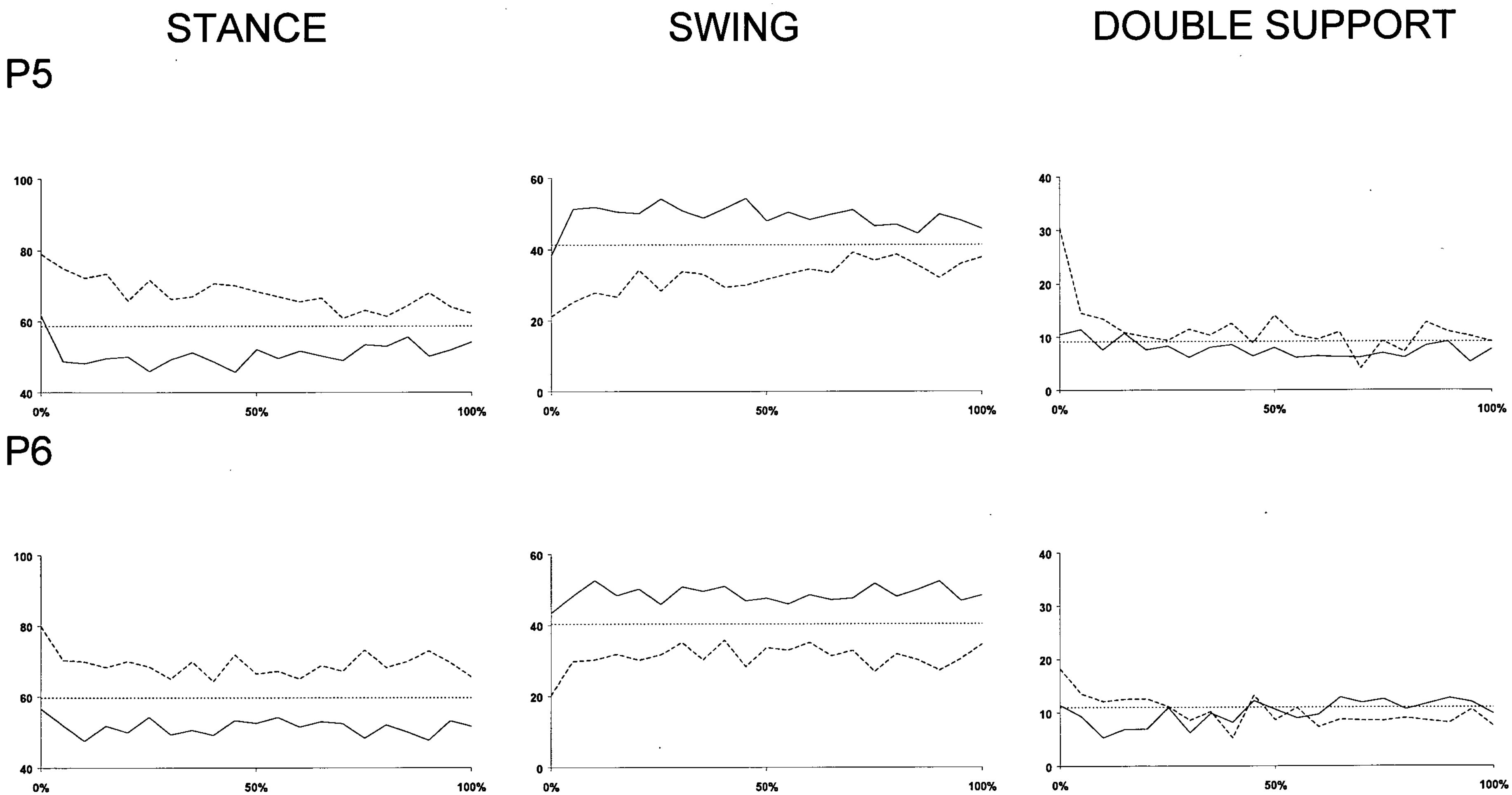
Abbreviations

P = participant

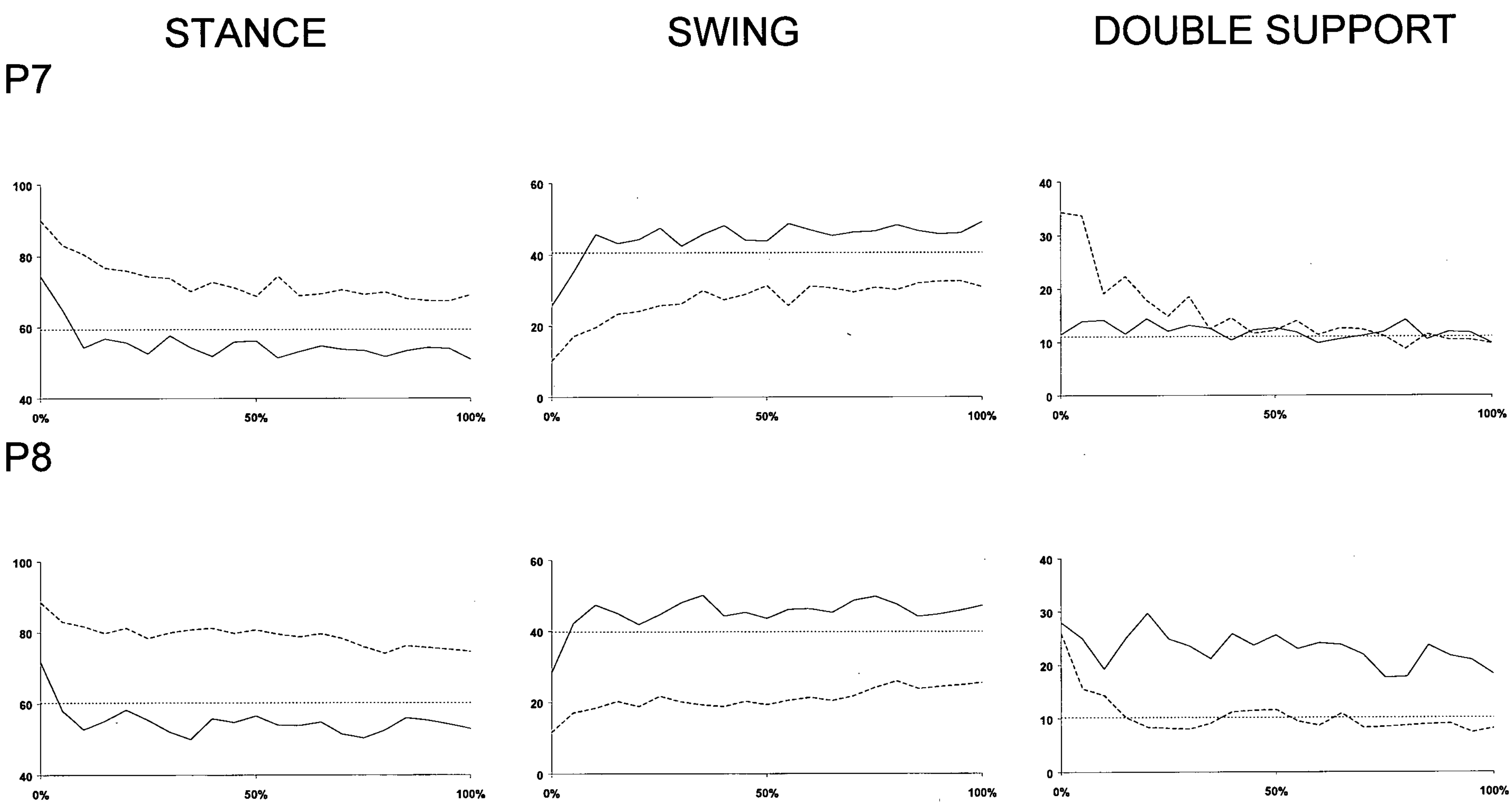

STANCE
SWING
DOUBLE SUPPORT
P1
P2
100
80
60
40
0%
50%
100%
60
40
20
0
30
10



STANCE
SWING
DOUBLE SUPPORT
P3
P4
100
80
60
40
0%
50%
100%
60
40
20
0
30
10



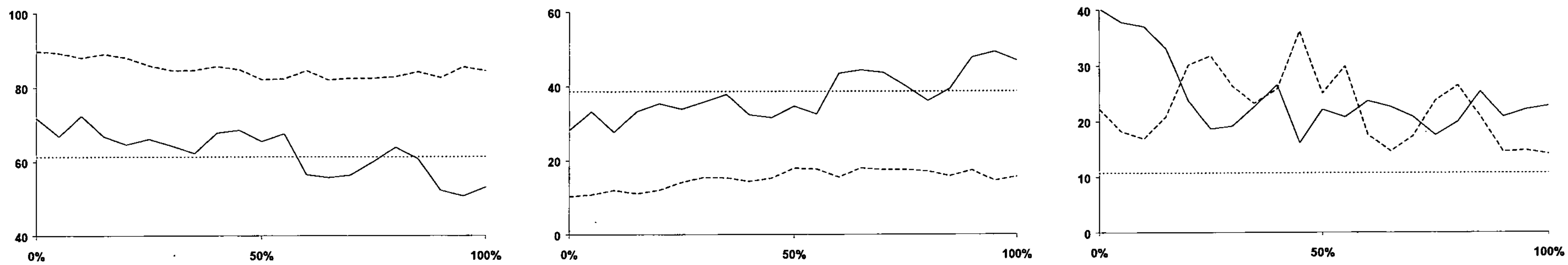
STANCE
SWING
DOUBLE SUPPORT
P5
P6
100
80
60
40
0
20
30
0%
50%
100%



STANCE
SWING
DOUBLE SUPPORT
P7
P8
100
80
60
40
0%
50%
100%
60
40
20
0
30
10

STANCE SWING DOUBLE SUPPORT

P9



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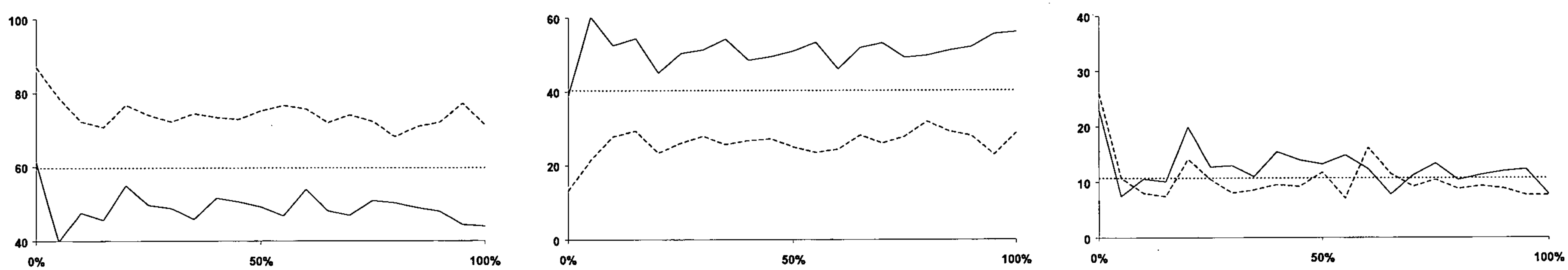


APPENDIX A3: STEP CHARACTERISTICS

Data presented in the following pages show step length, frequency and width profiles for each participant as they walked with the prosthetic simulator. For step length and frequency, the solid line represents the prosthetic limb; the dashed line represents the intact limb; and the dotted line represents the control condition (able-bodied gait). Step width is represented by the solid line when walking with the prosthetic simulator, while the dotted line represents the control condition (able-bodied gait).

For the horizontal axis, units are expressed as a percentage (%) of total distance walked. For the vertical axis, units for step length and width are in metres (m), while units for step frequency are in steps/min.

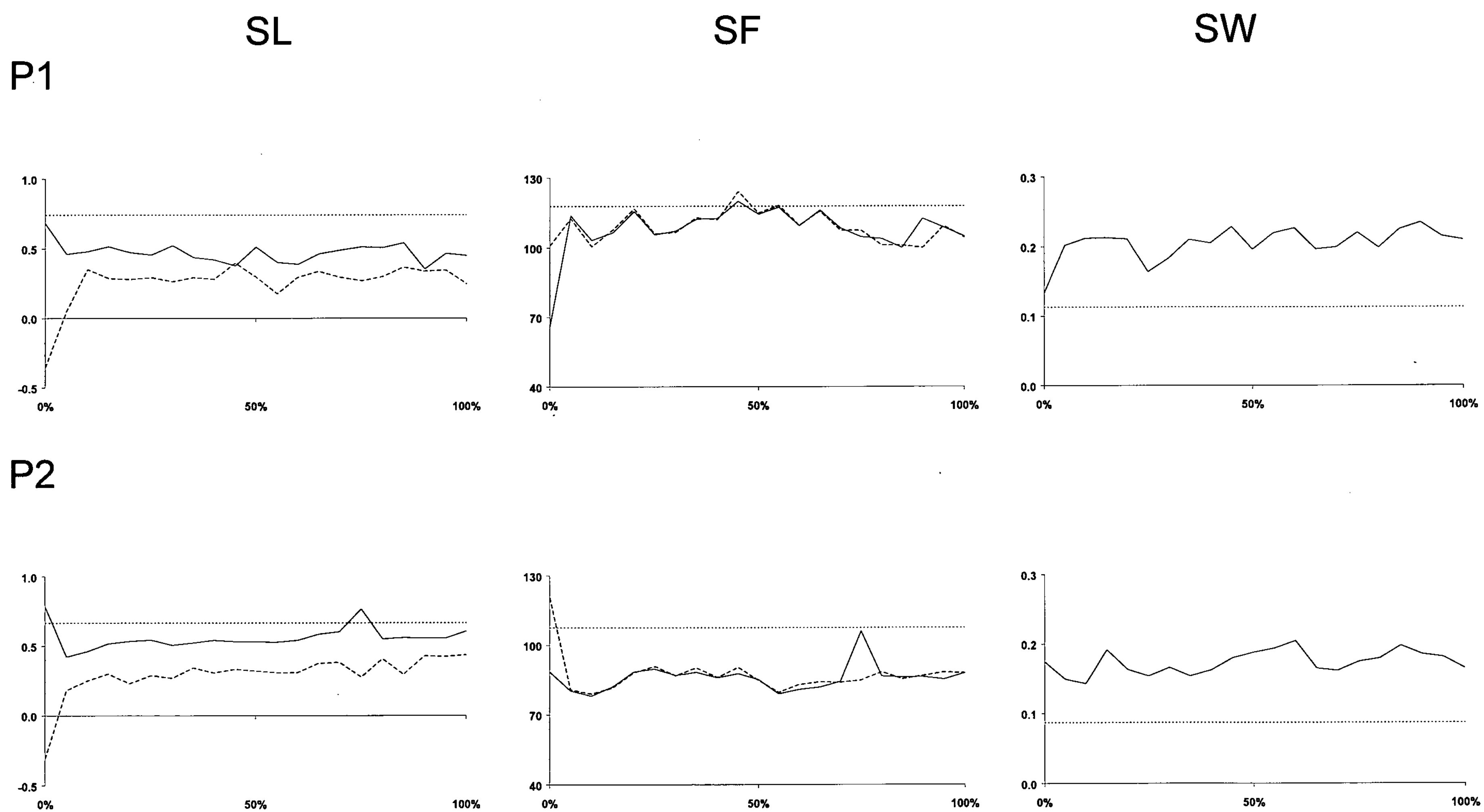
Abbreviations

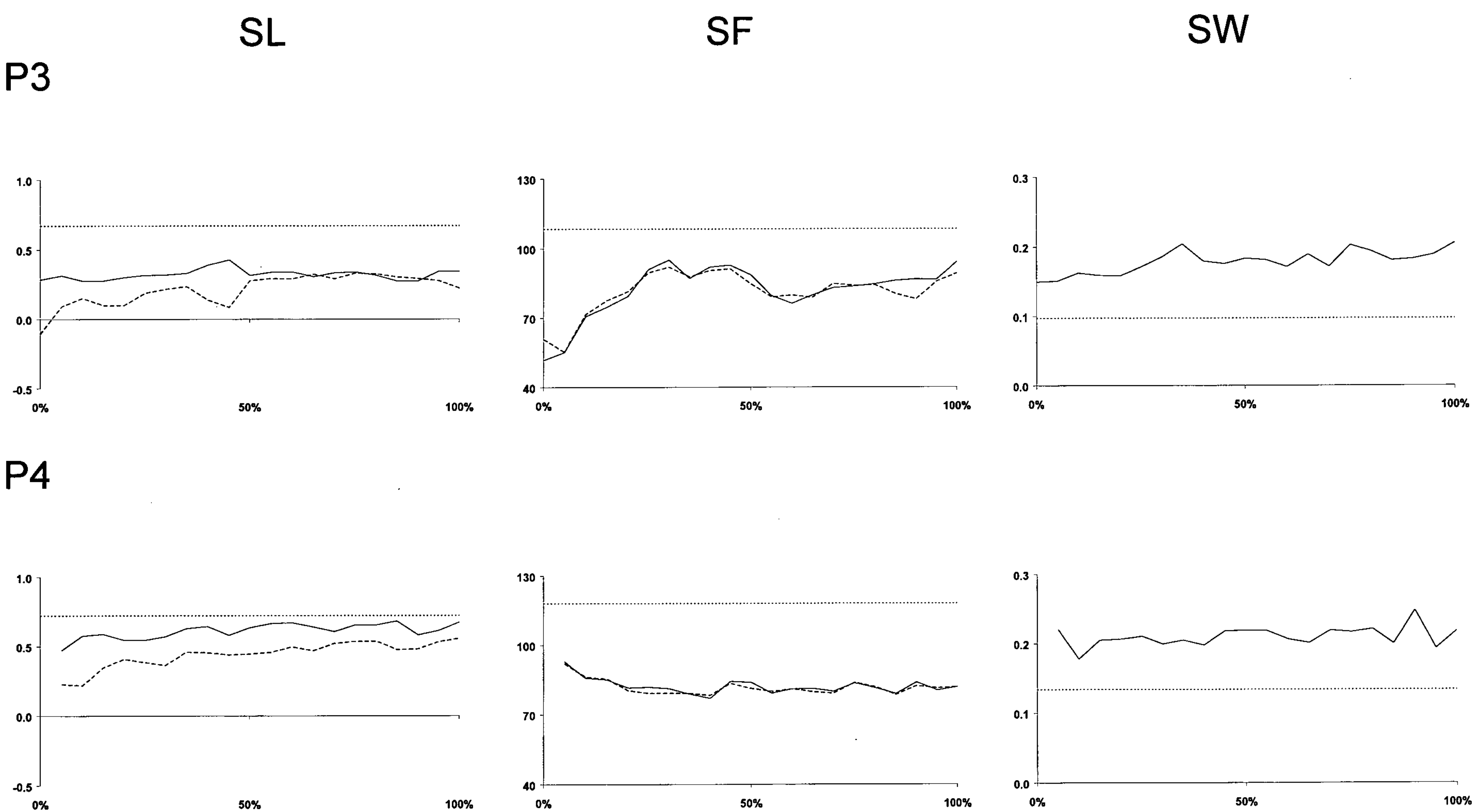
P = participant

SL = step length

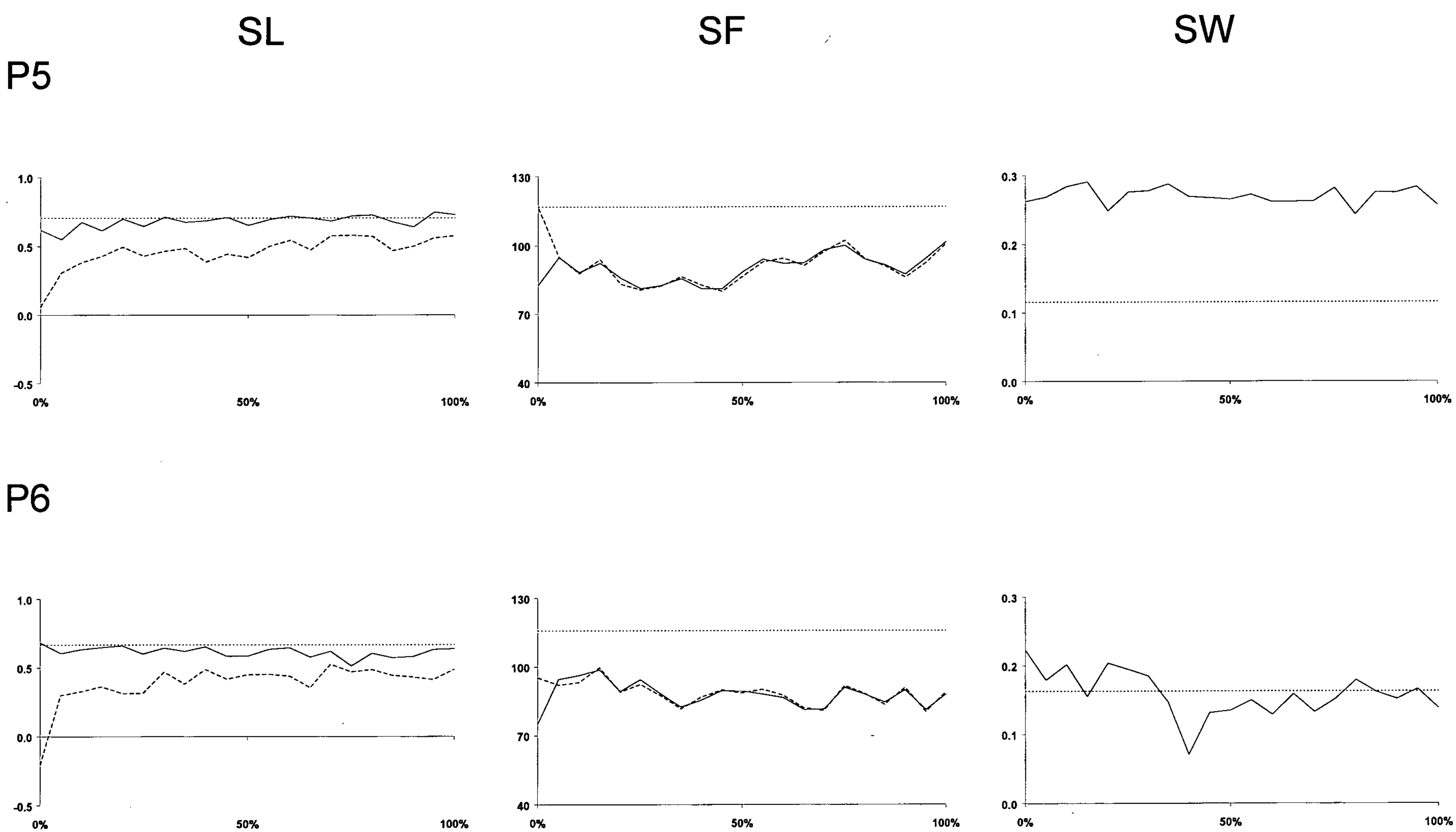
SF = step frequency

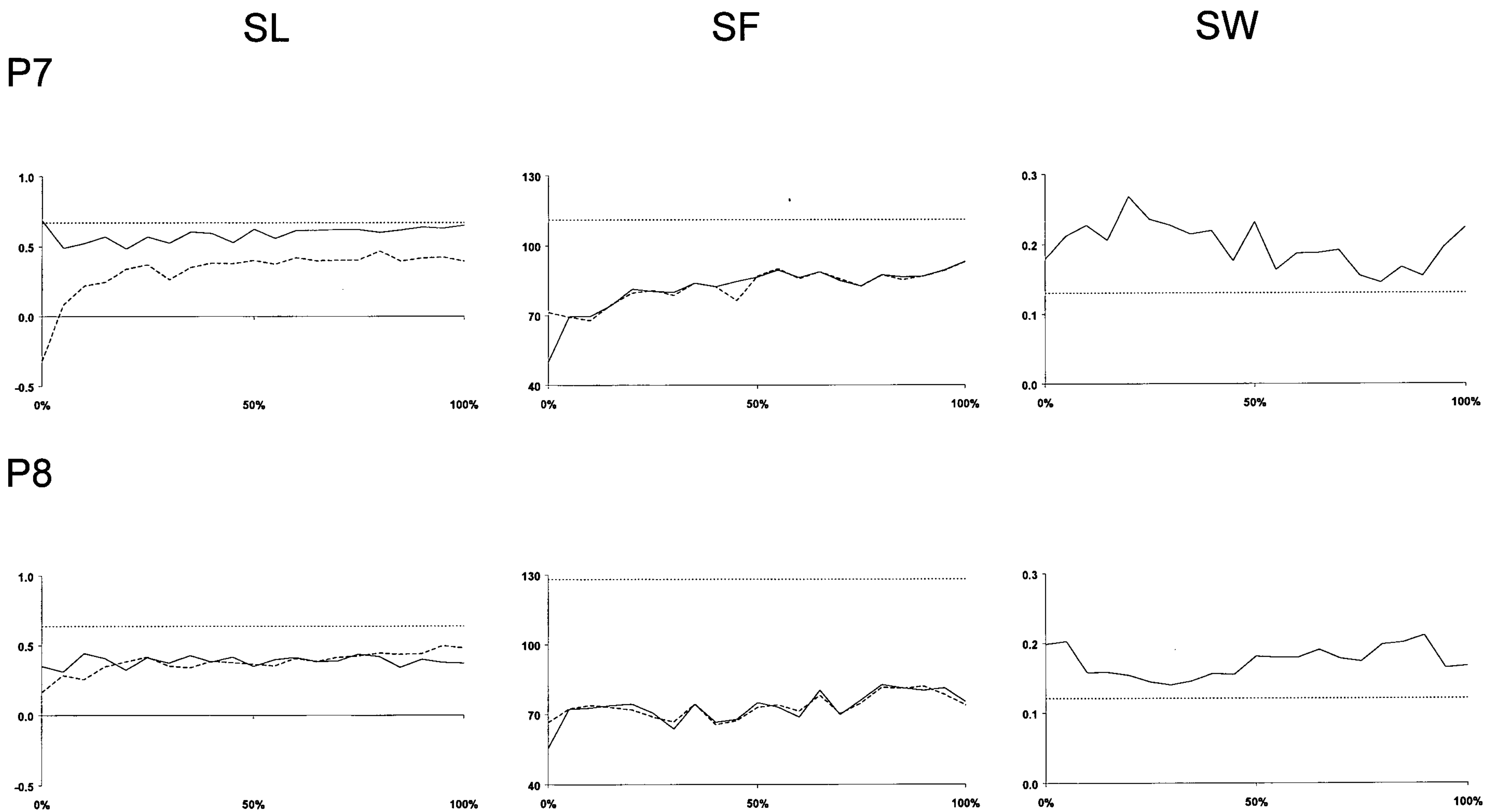
SW = step width

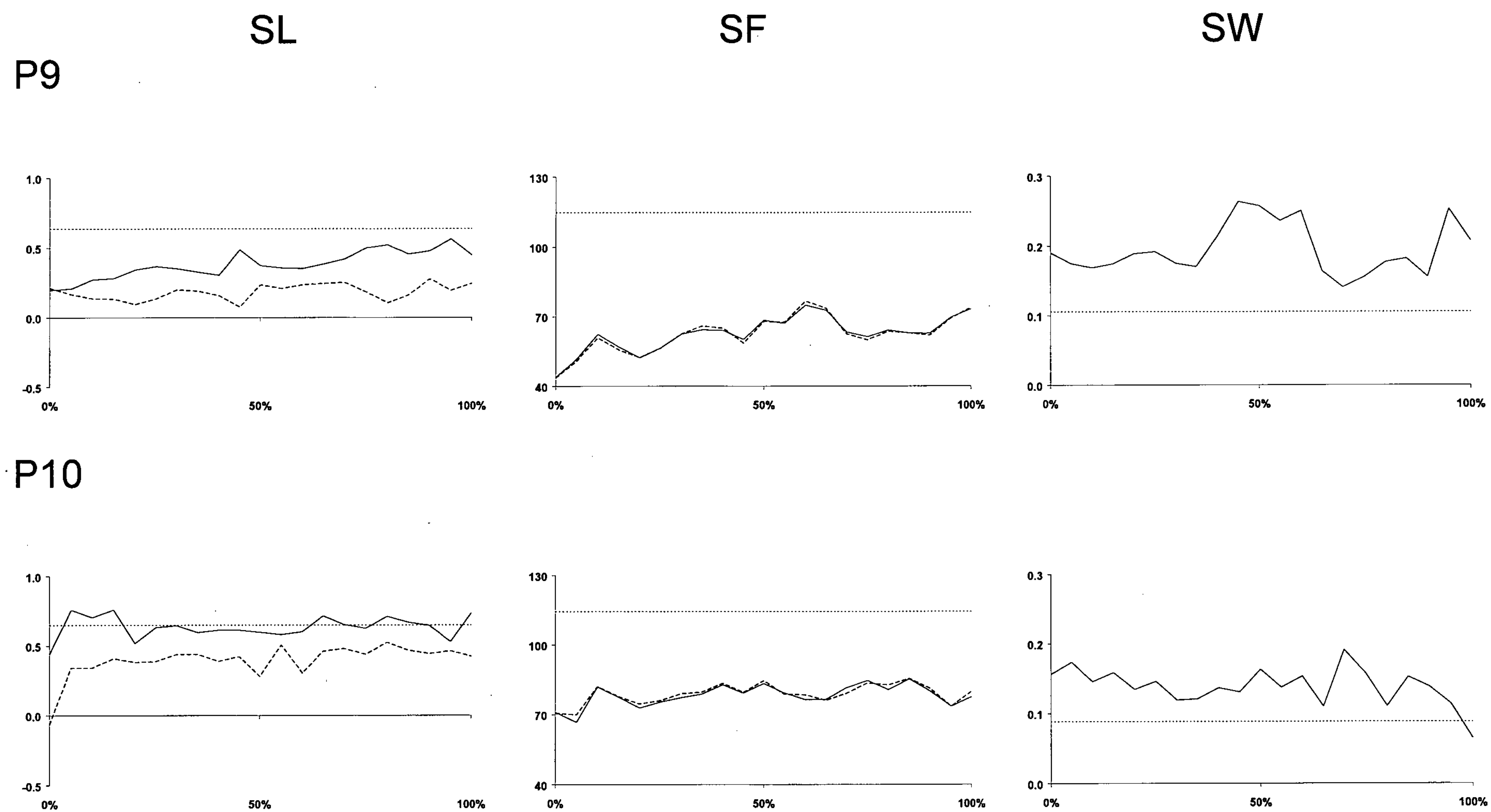

SL
SF
SW
P1
P2
1.0
0.5
0.0
-0.5
130
100
70
40
0.3
0.2
0.1
0.0
0%
50%
100%



SL
SF
SW
P3
P4
1.0
0.5
0.0
-0.5
130
100
70
40
0.3
0.2
0.1
0.0
0%
50%
100%


SL
SF
SW
P5
P6
1.0
0.5
0.0
-0.5
130
100
70
40
0.3
0.2
0.1
0.0
0%
50%
100%


SL
SF
SW
P7
P8
1.0
0.5
0.0
-0.5
130
100
70
40
0.3
0.2
0.1
0.0
0%
50%
100%


SL
SF
SW
P9
P10
1.0
0.5
0.0
-0.5
130
100
70
40
0.3
0.2
0.1
0.0
0%
50%
100%

APPENDIX A4: VERTICAL ORIENTATION OF LOWER-LIMB SEGMENTS

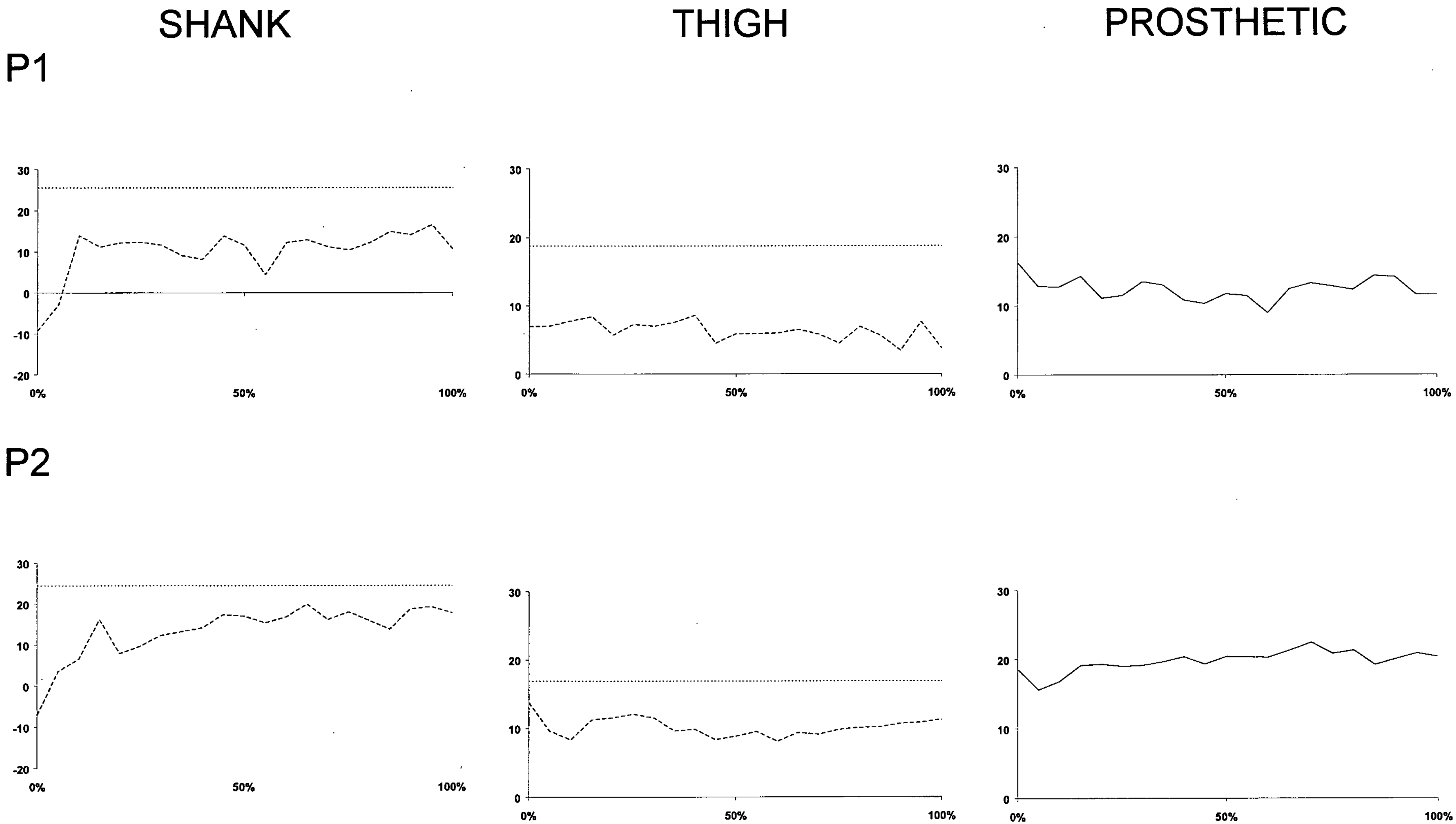
Data presented in the following pages show the vertical orientation of the shank and thigh on the intact limb at intact foot contact and prosthetic segment at prosthetic contact for each participant as they walked with the prosthetic simulator. The dashed line represents the lower-limb segments on the intact limb; and the dotted line in the control condition (able-bodied gait). The solid line represents the prosthetic segment.

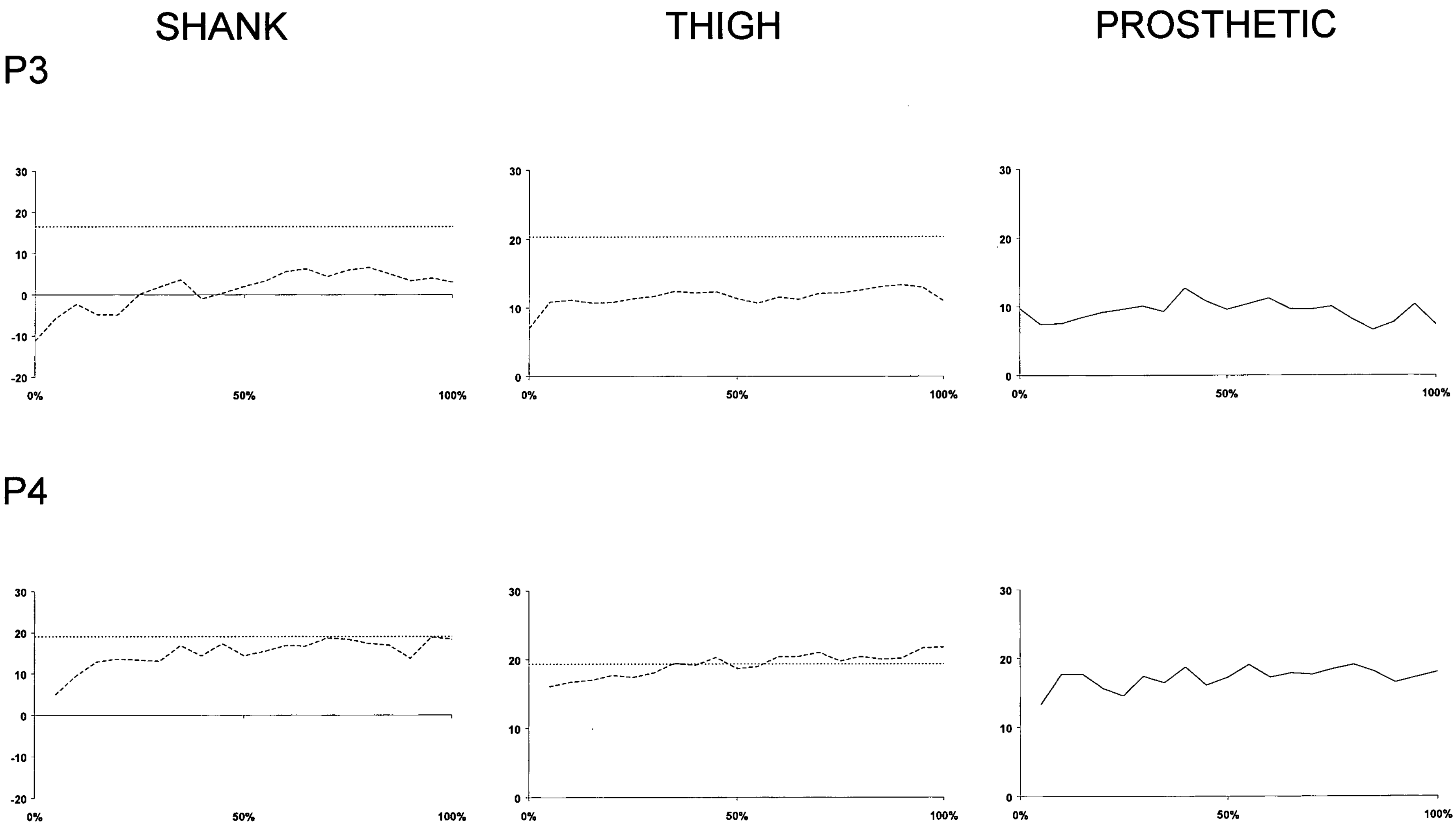
Units are expressed as a percentage (%) of total distance walked for the horizontal axis and in degrees (°) for the vertical axis.

A positive angle indicates the segment is behind the vertical, while a negative angle indicates the segment is in front of the vertical.

Abbreviations

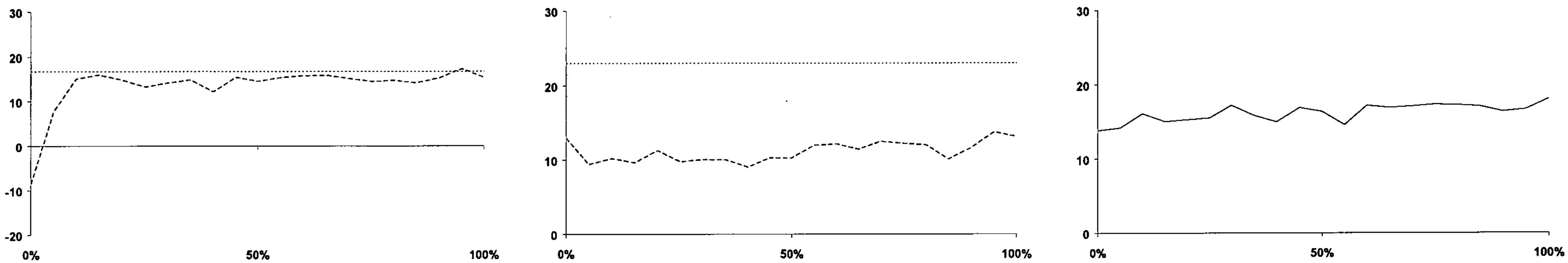
P = participant


SHANK
THIGH
PROSTHETIC
P1
P2
30
20
10
0
-10
-20
0%
50%
100%

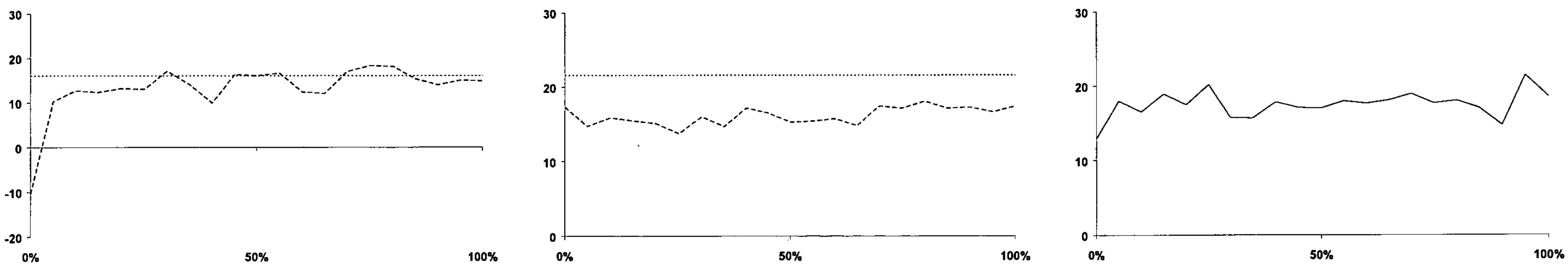

SHANK
THIGH
PROSTHETIC
P3
P4
30
20
10
0
-10
-20
0%
50%
100%

SHANK THIGH PROSTHETIC

P5

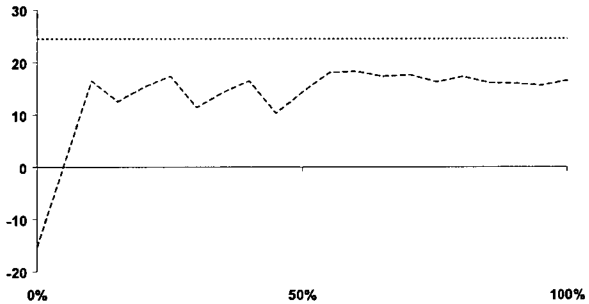


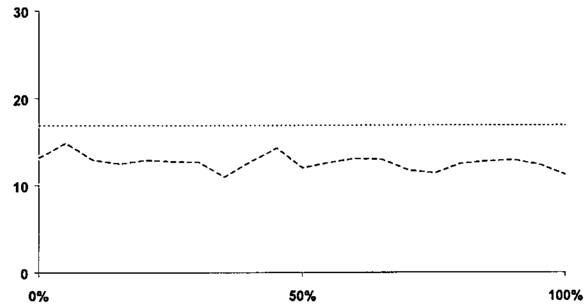
P6

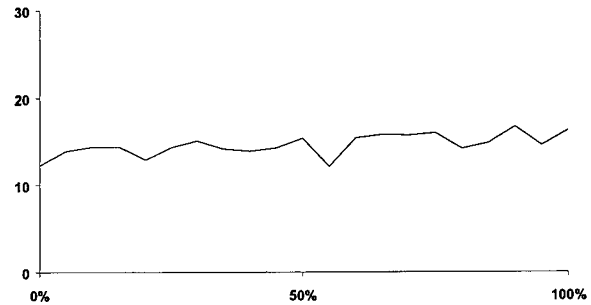


SHANK THIGH PROSTHETIC

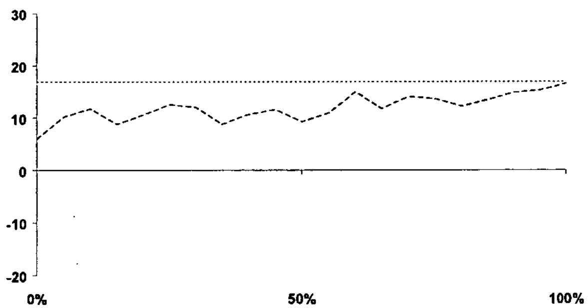
P7

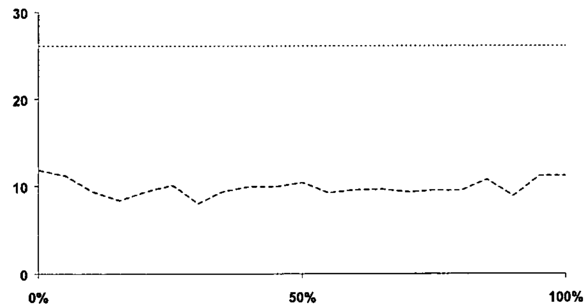


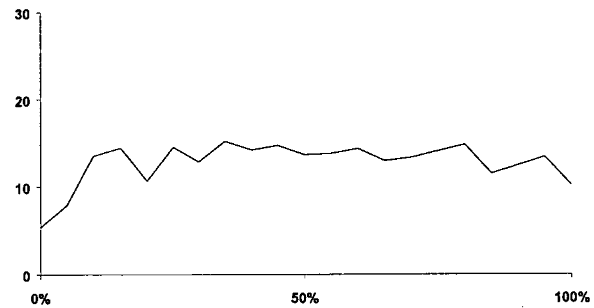




P8

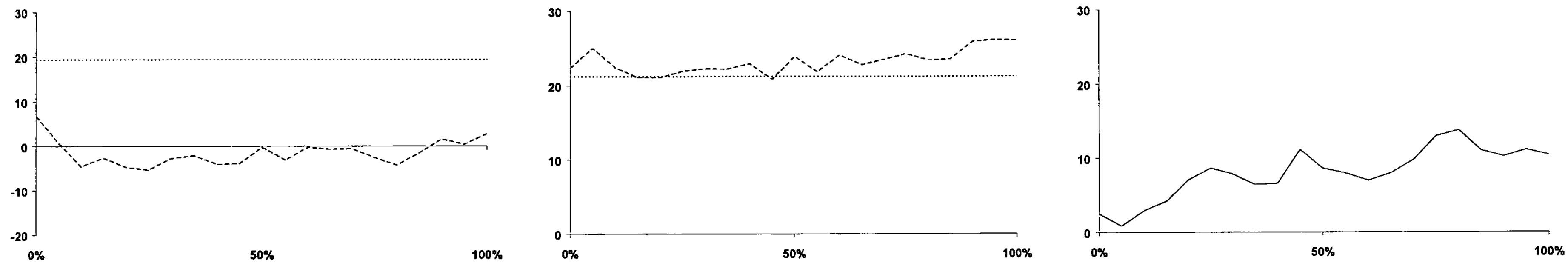






SHANK THIGH PROSTHETIC

P9



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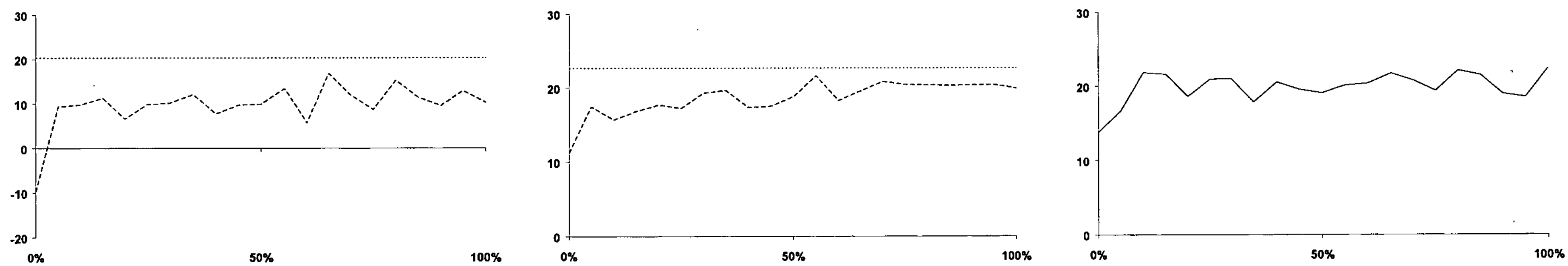


APPENDIX A5: TRUNK ANGLE AT FOOT AND PROSTHETIC CONTACT

Data presented in the following pages show the vertical orientation of the trunk at intact foot and prosthetic contact for each participant as they walked with the prosthetic simulator. The solid line represents trunk angle at prosthetic contact; the dashed line at intact foot contact and the dotted line in the control condition (able-bodied gait).

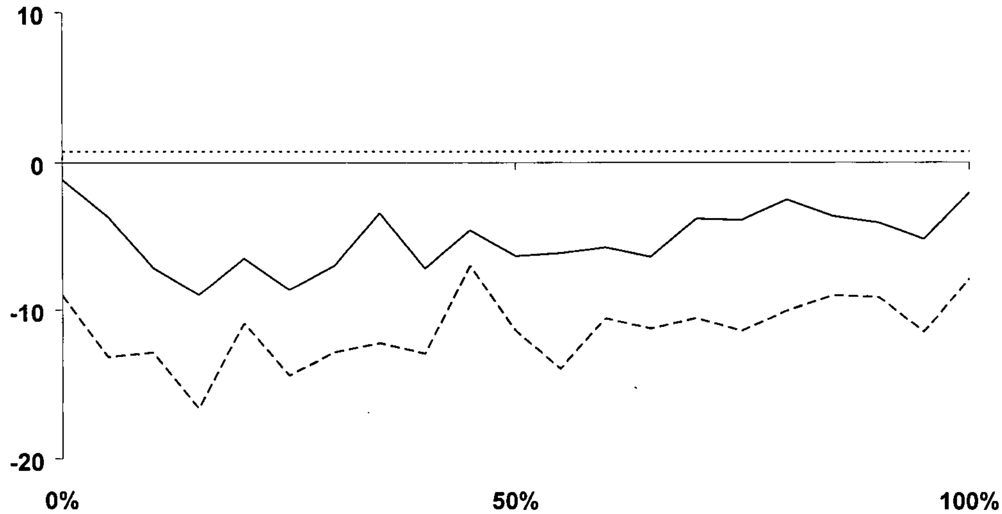
Units are expressed as a percentage (%) of total distance walked for the horizontal axis and in degrees (°) for the vertical axis.

A positive angle indicates the segment is behind the vertical, while a negative angle indicates the segment is in front of the vertical. Therefore, a negative trunk angle indicates forward trunk lean.

Abbreviations

P = participant

P1



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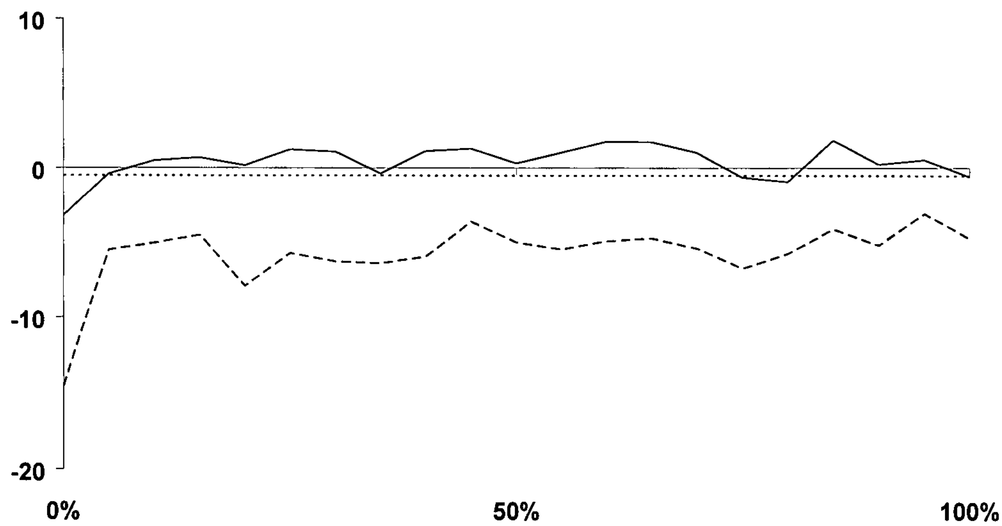

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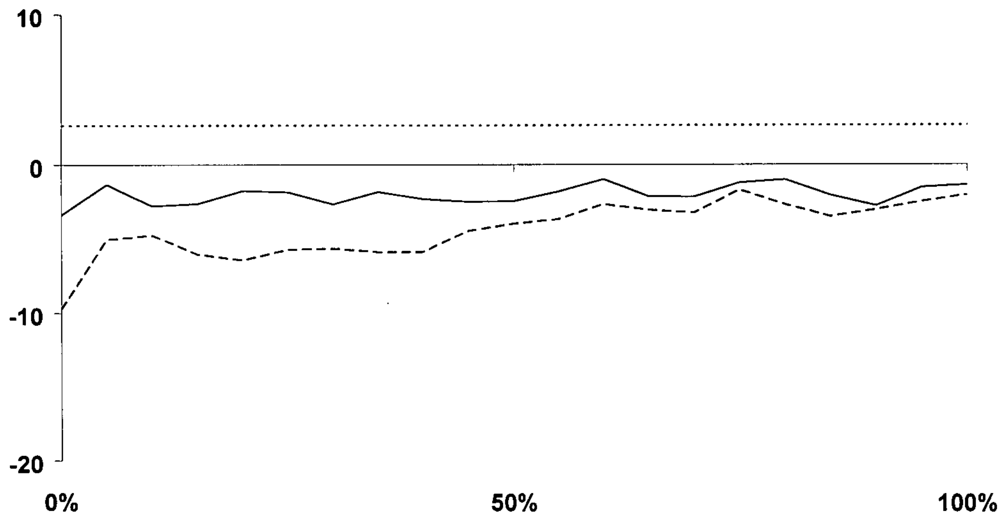

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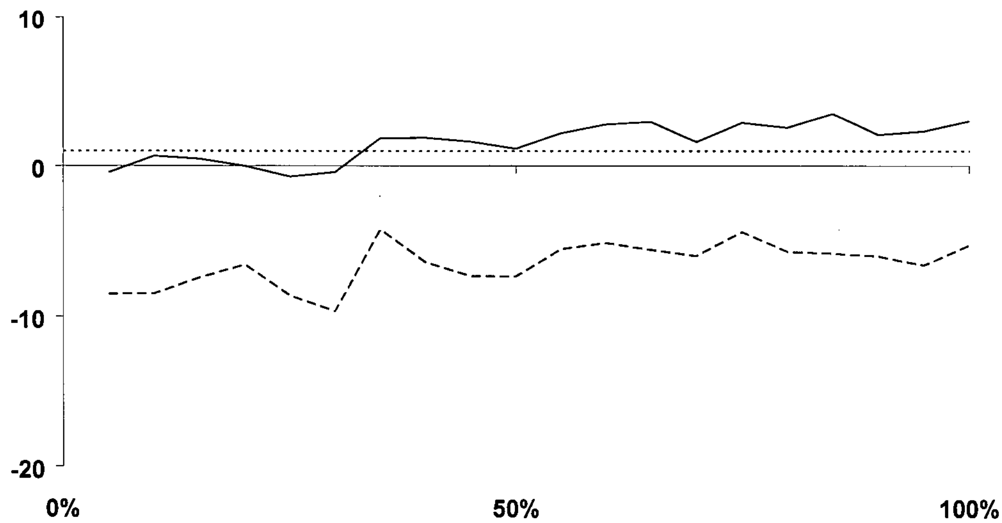
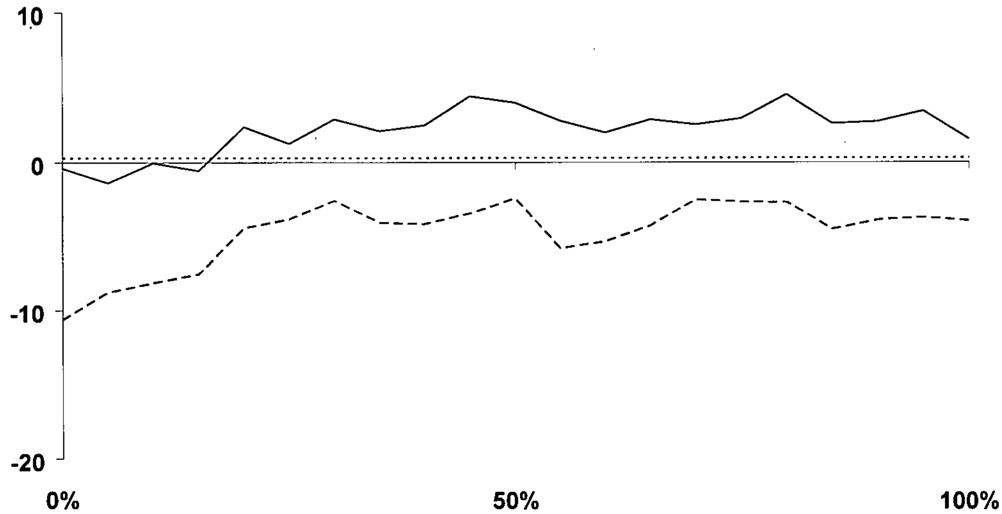


P5



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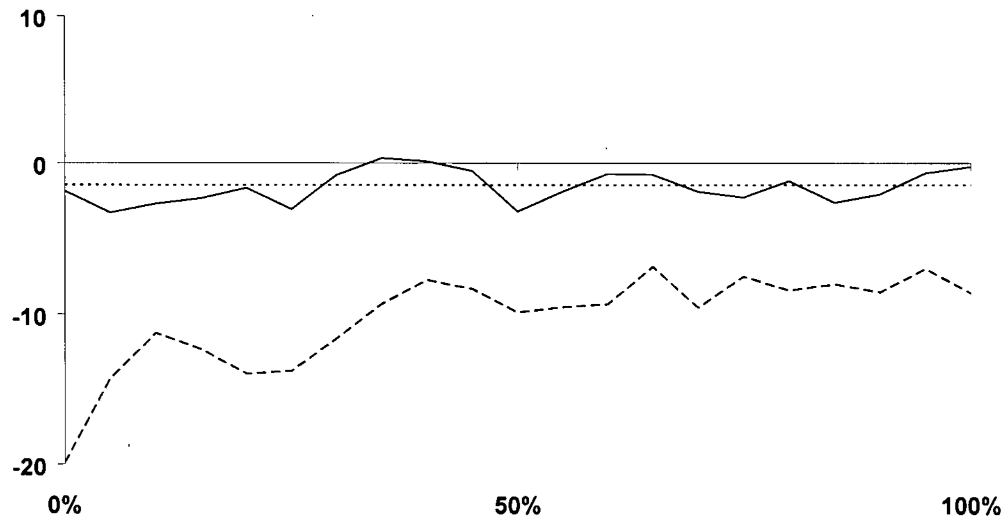
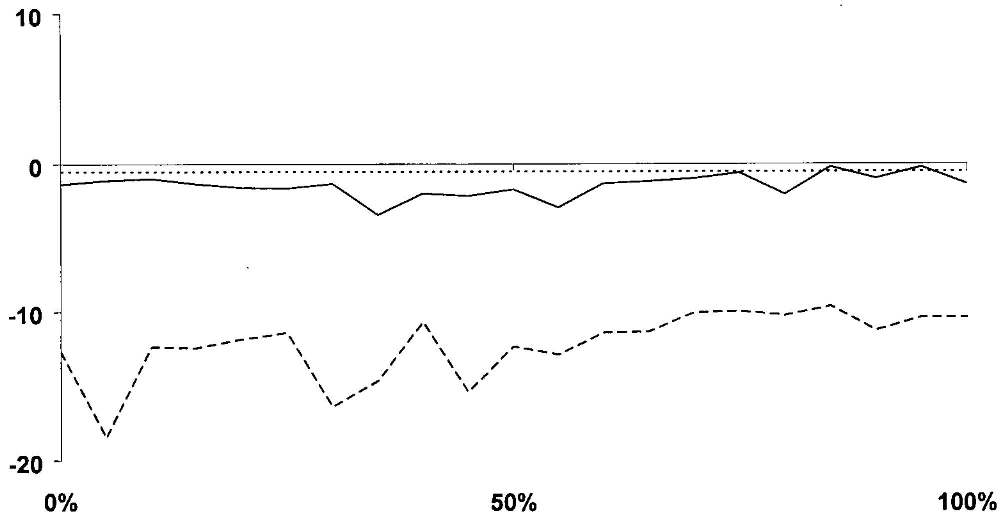
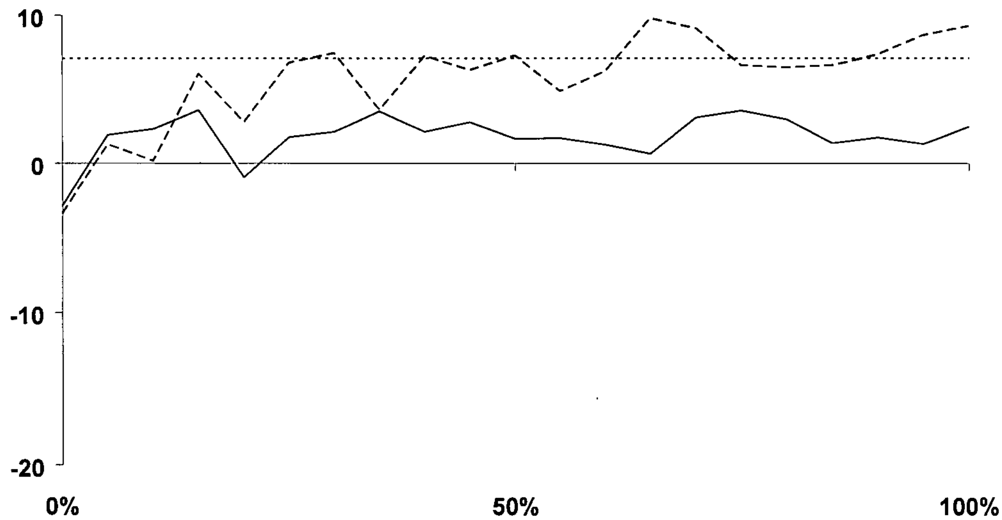


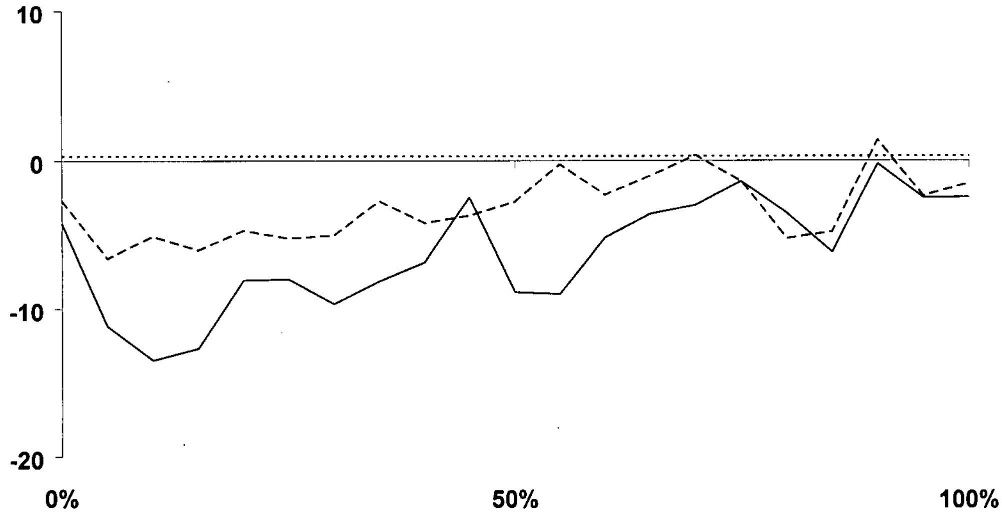
P7



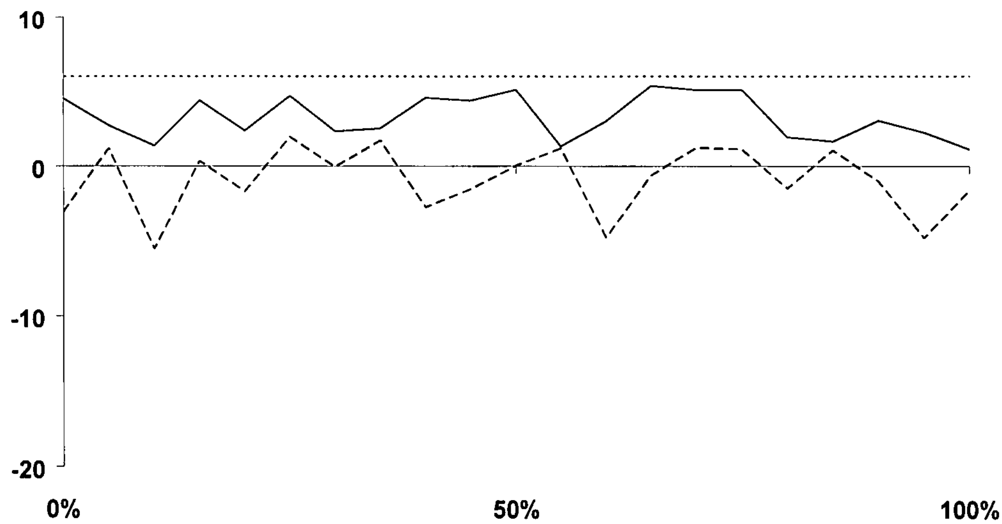
P8



P9



P10



APPENDIX A6: PEAK VERTICAL POSITION OF THE WHOLE BODY CENTRE OF MASS

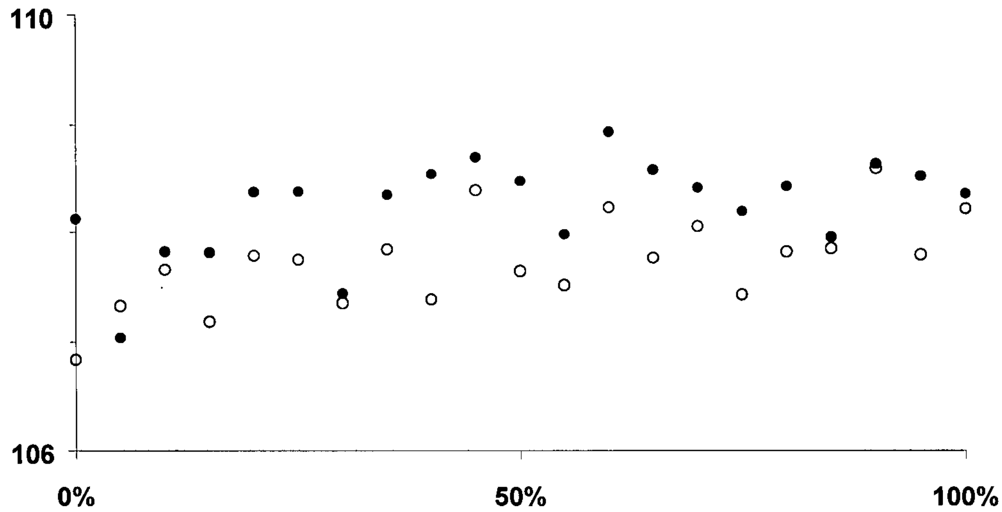
Data presented in the following pages show the peak vertical position of the whole body centre of mass during prosthetic and intact single stance for each participant as they walked with the prosthetic simulator. Filled circles represents intact single stance; unfilled circles represent prosthetic single stance.

Units are expressed as a percentage (%) of total distance walked for the horizontal axis and in centimetres (cm) for the vertical axis.

Abbreviations

P = participant

P1



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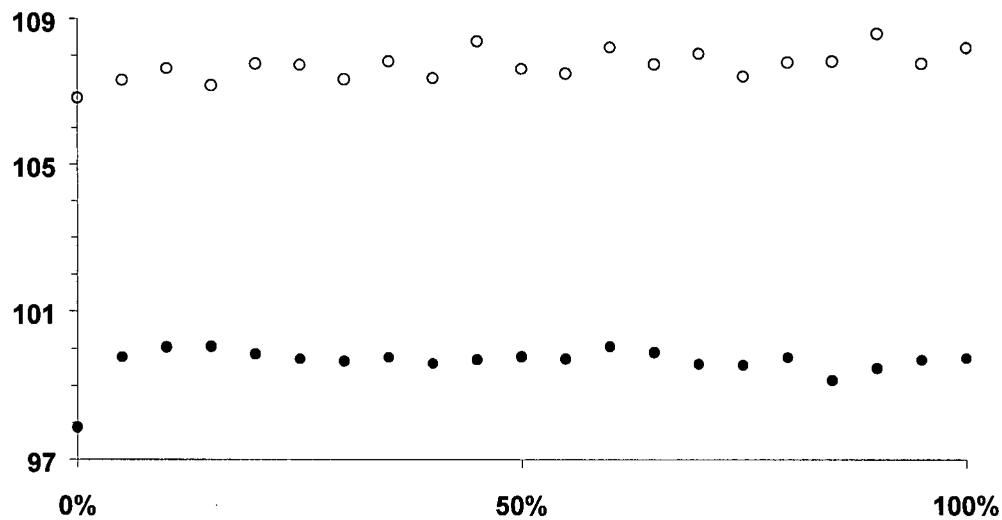
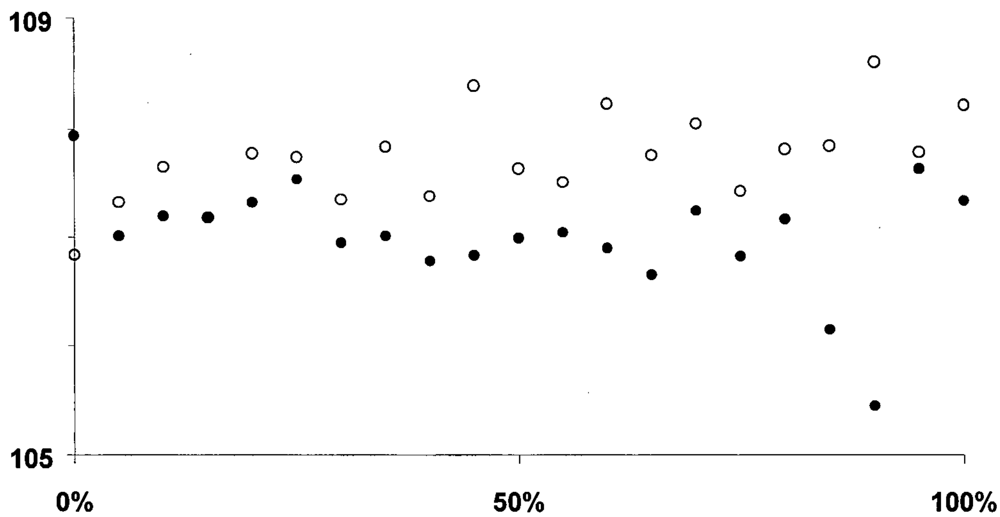


P3



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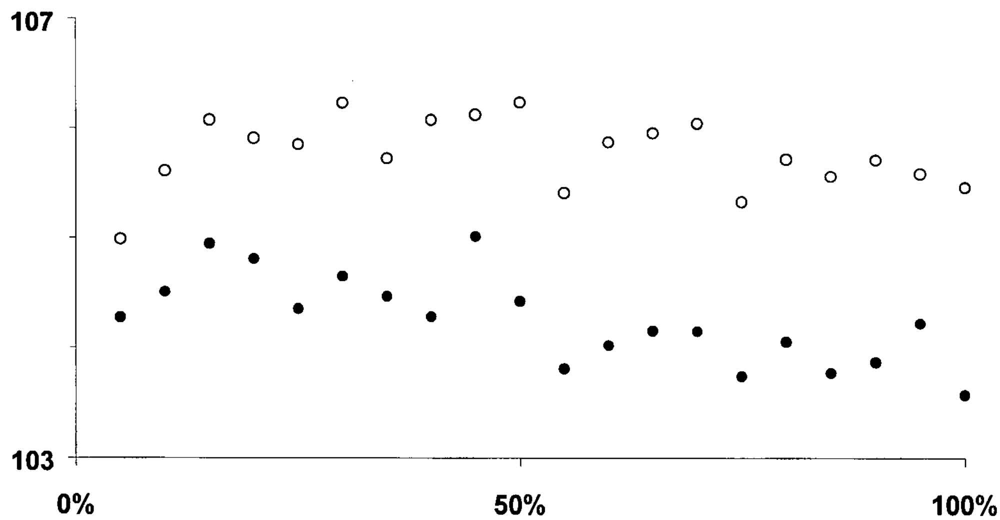
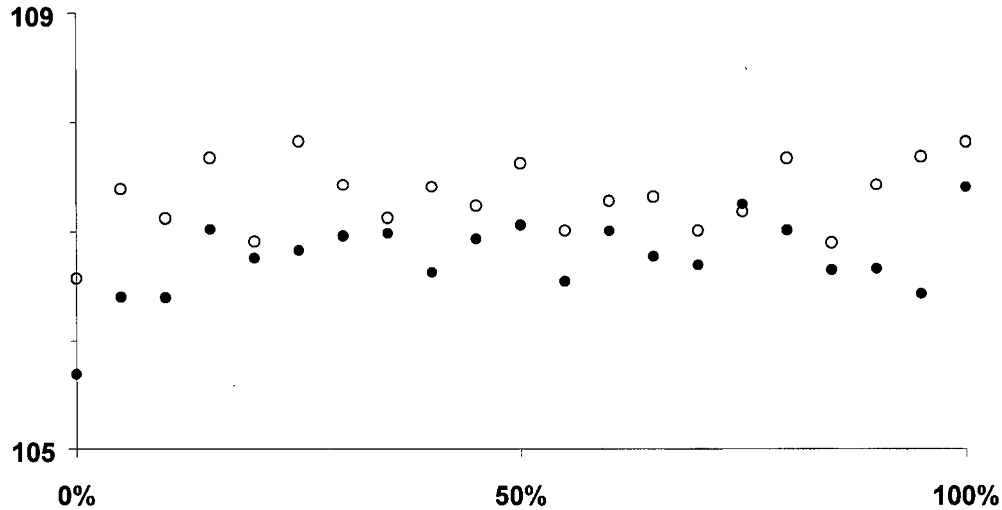
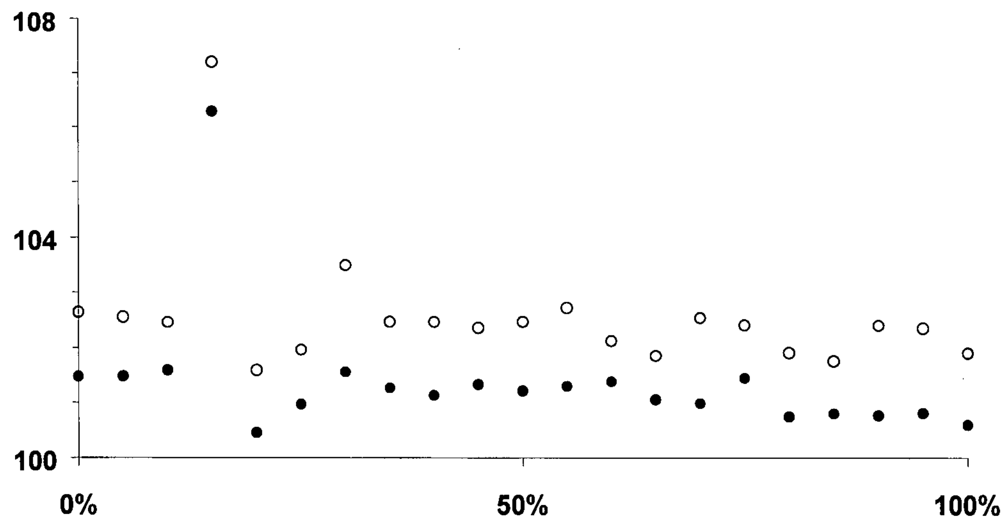


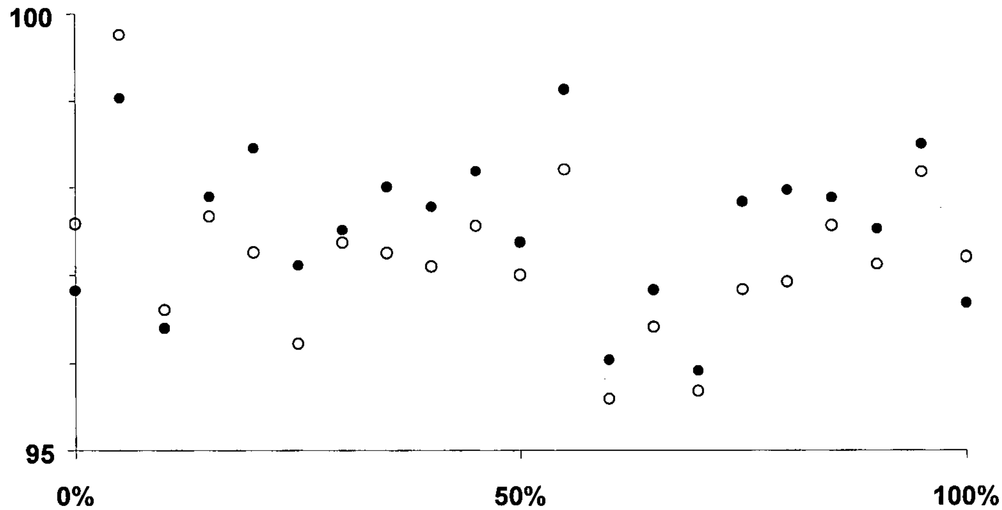
P5



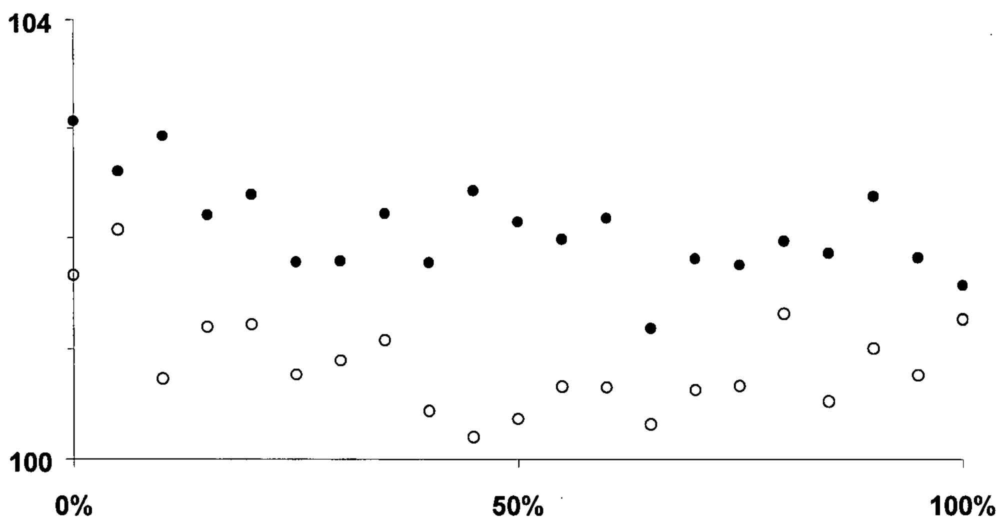
P6



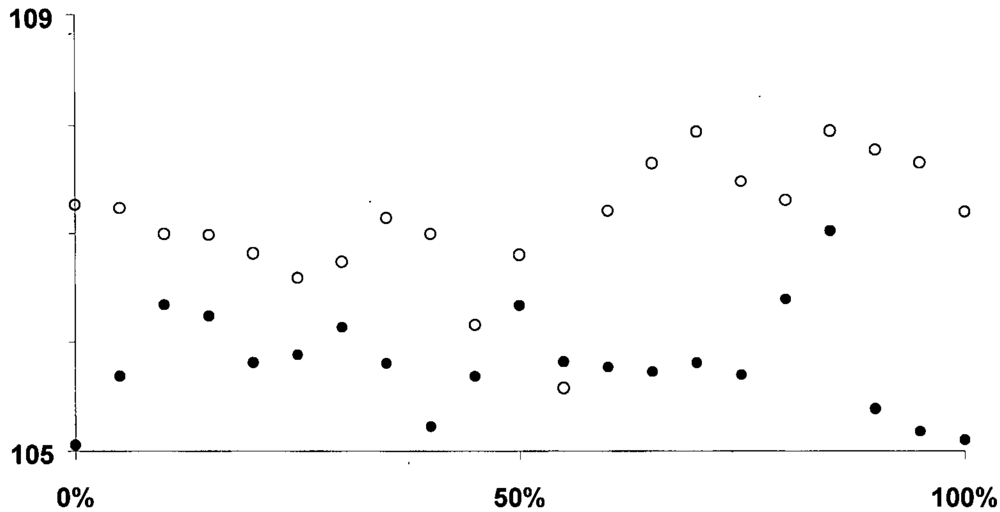
P7



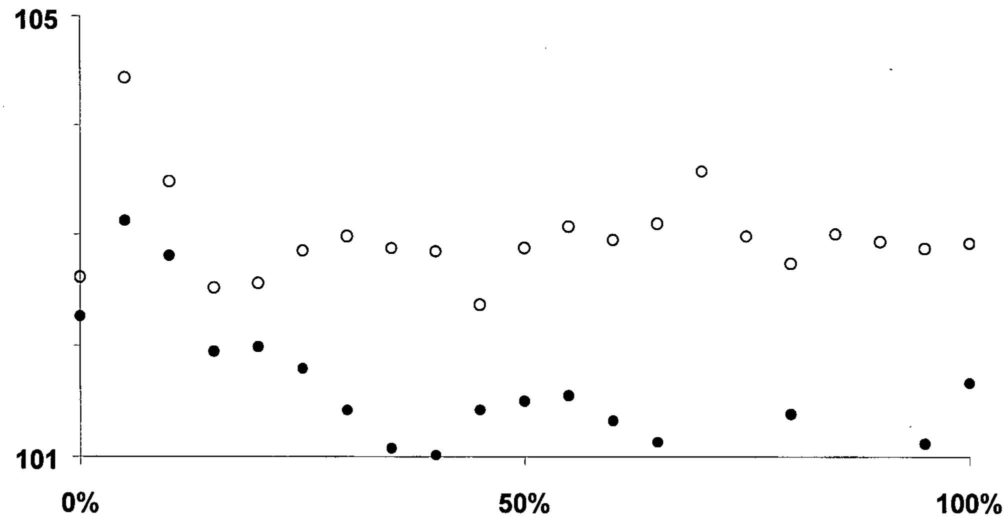
P8



P9



P10



APPENDIX A7: VERTICAL DISPLACEMENT OF THE WHOLE BODY CENTRE OF MASS IN THE SAGITTAL PLANE

Data presented in the following pages show the vertical displacement of the whole body centre of mass in the sagittal plane for each participant as they walked with the prosthetic simulator.

Units are expressed as stride time (sec) for the horizontal axis and in centimetres (cm) for the vertical axis.

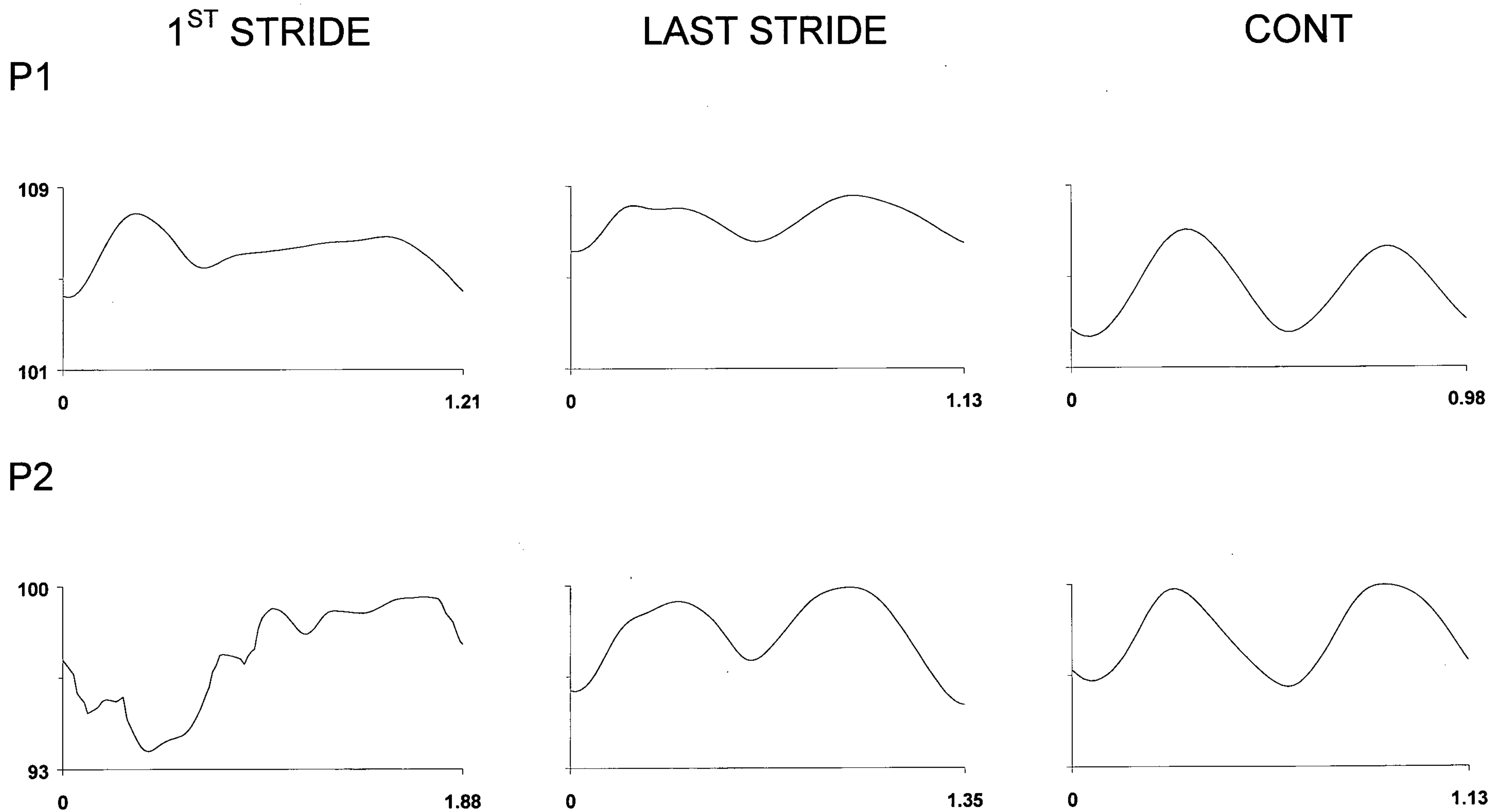
Abbreviations

P = participant

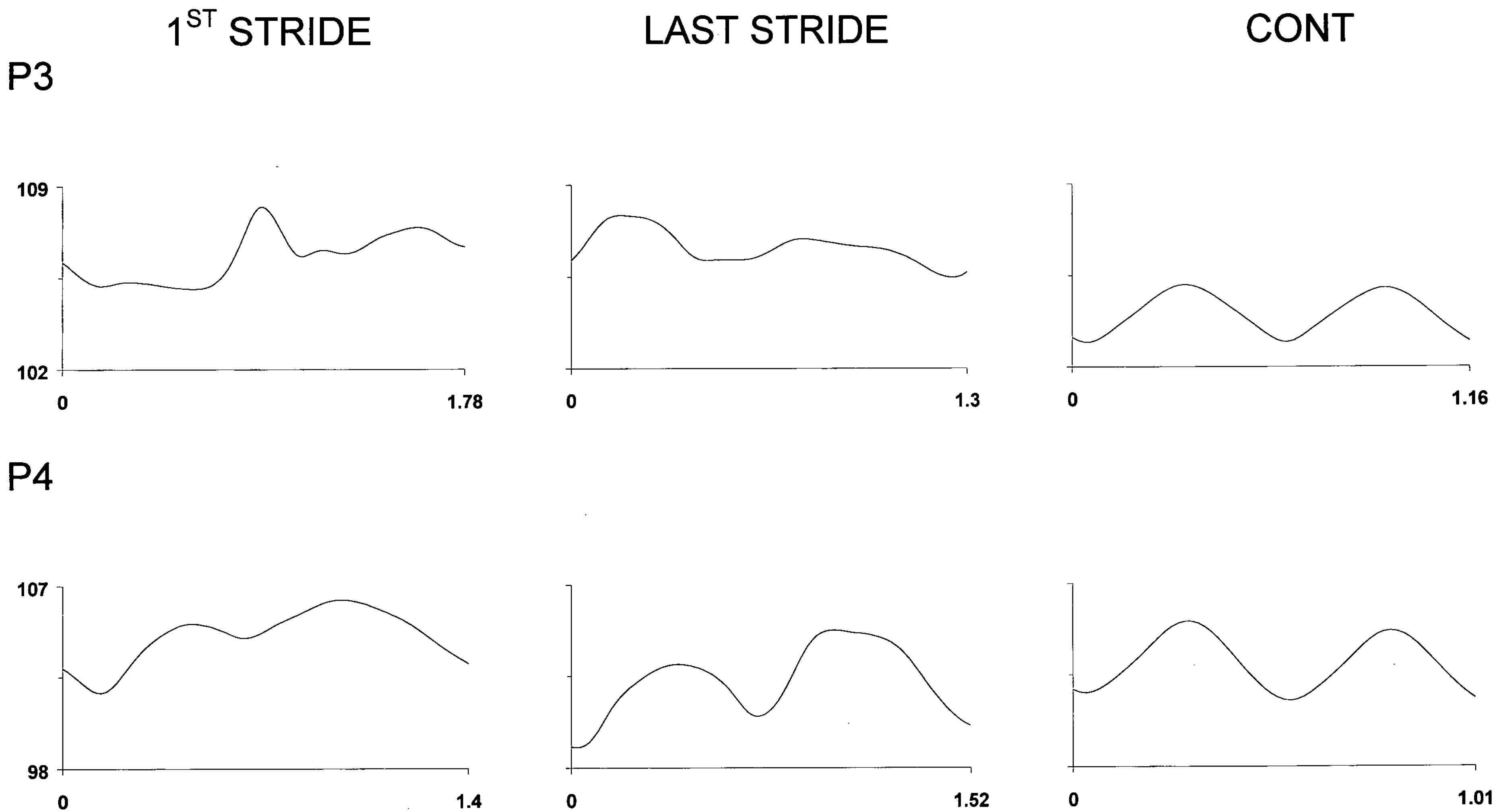
1st STRIDE = 1st right stride with the prosthetic simulator (right stride begins with the first prosthetic contact to the subsequent prosthetic contact)

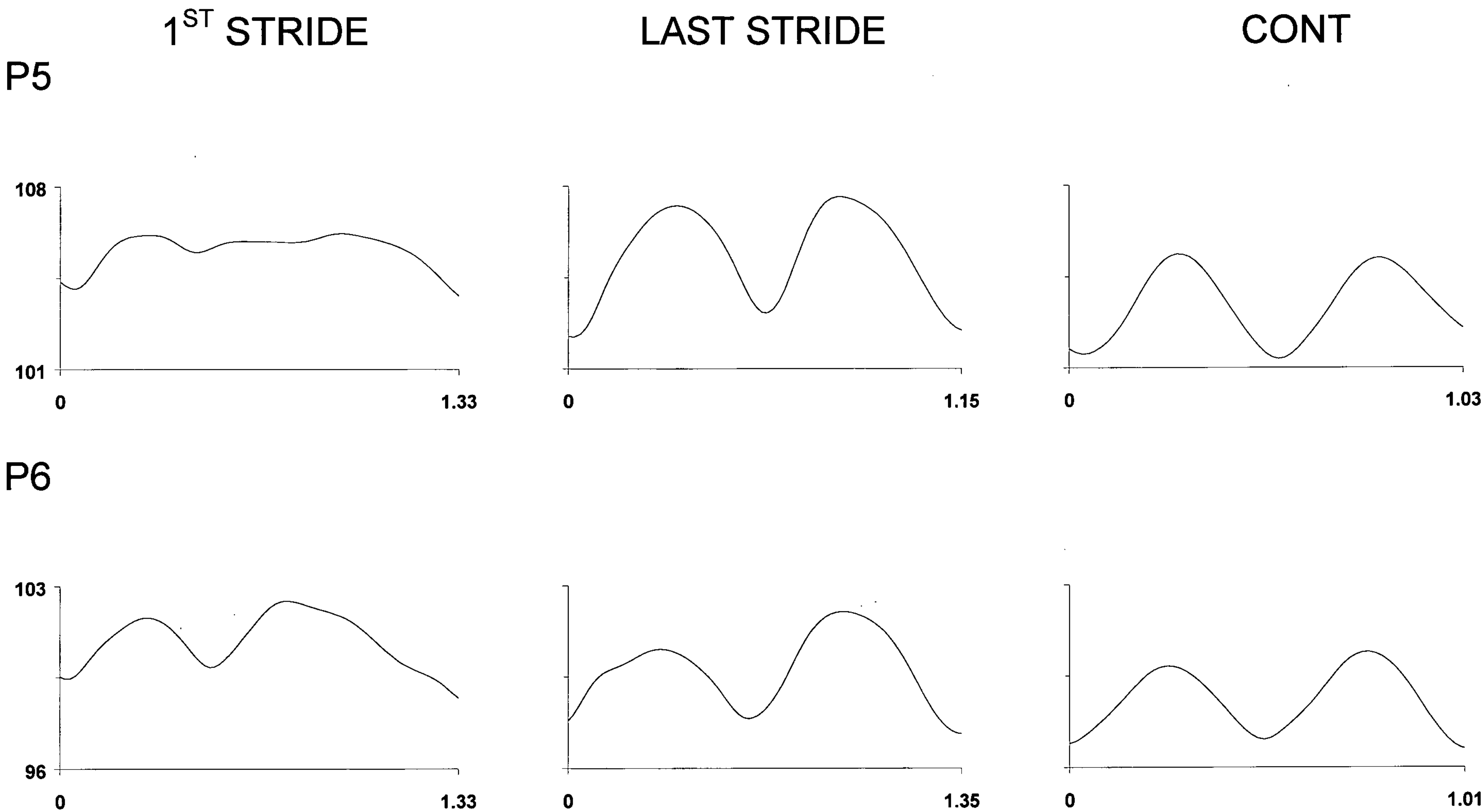
LAST STRIDE = last right stride with the prosthetic simulator

CONT = control condition

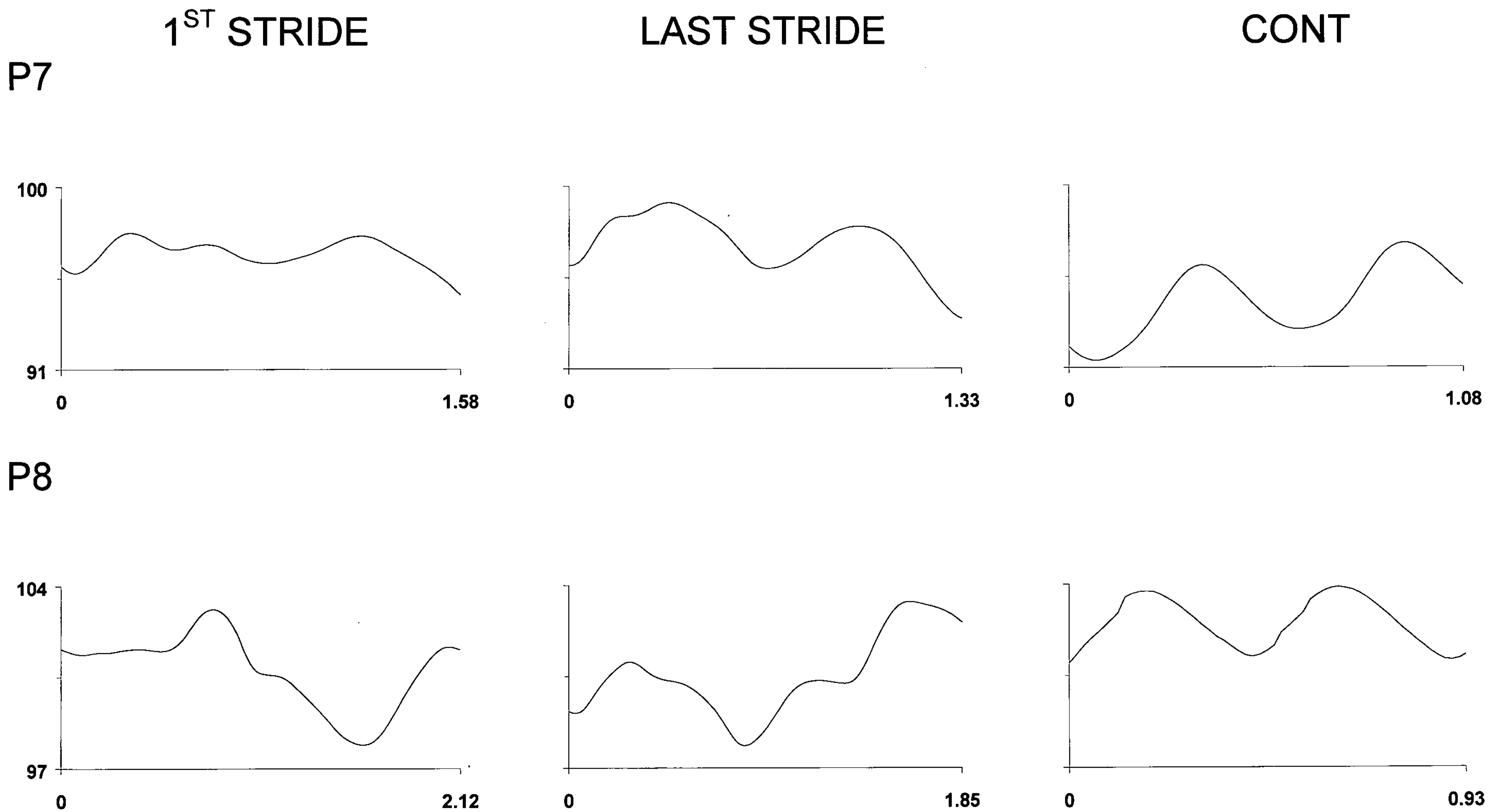


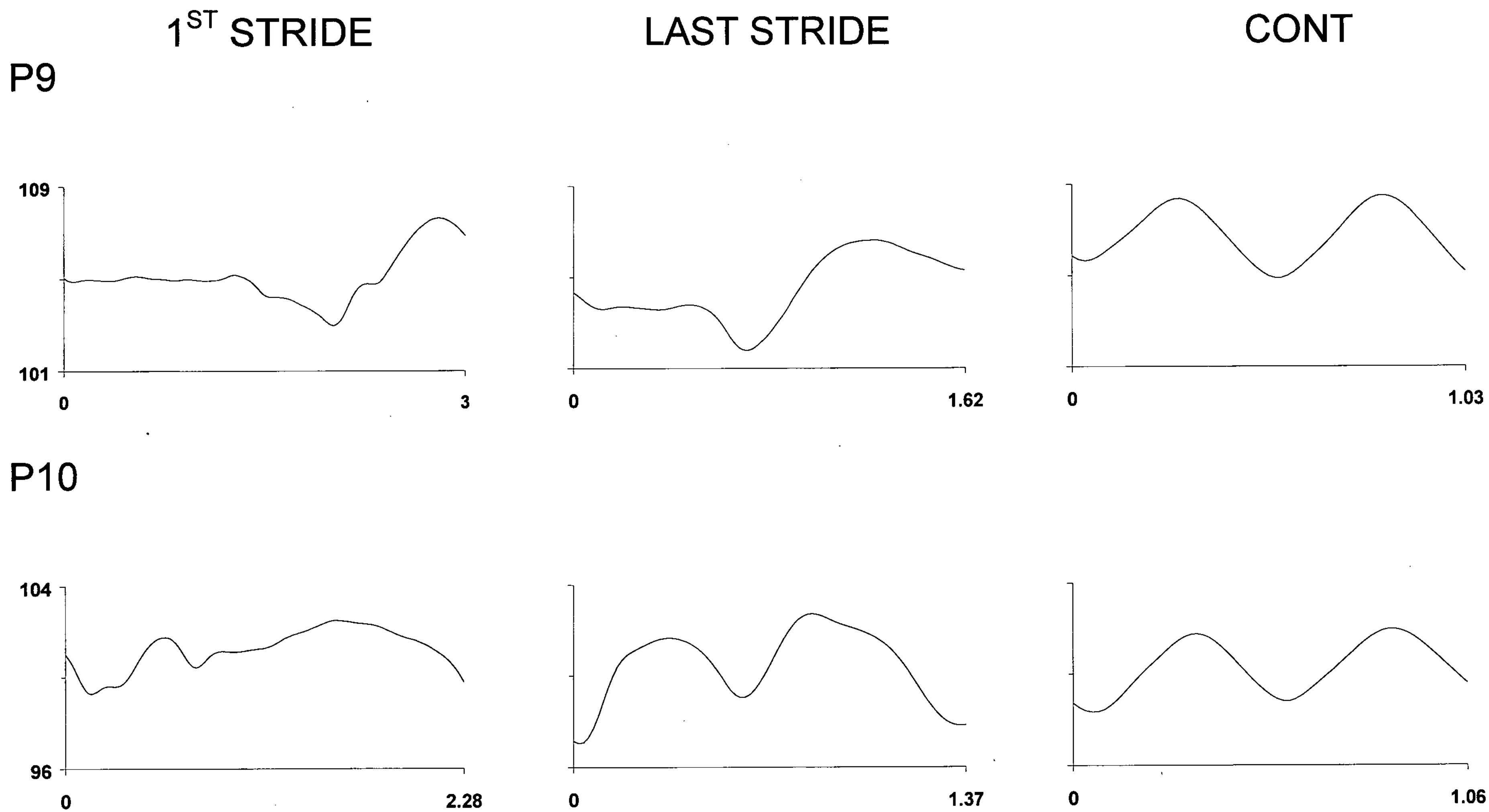
1ST STRIDE
LAST STRIDE
CONT
P1
109
101
0
1.21
0
1.13
0
0.98
P2
100
93
0
1.88
0
1.35
0
1.13


1ST STRIDE
LAST STRIDE
CONT
P3
109
102
0
1.78
0
1.3
0
1.16
P4
107
98
0
1.4
0
1.52
0
1.01



1ST STRIDE
LAST STRIDE
CONT
P5
108
101
0
1.33
0
1.15
0
1.03
P6
103
96
0
1.33
0
1.35
0
1.01


1ST STRIDE
LAST STRIDE
CONT
P7
100
91
0
1.58
0
1.33
0
1.08
P8
104
97
0
2.12
0
1.85
0
0.93


1ST STRIDE
LAST STRIDE
CONT
P9
109
101
0
3
0
1.62
0
1.03
P10
104
96
0
2.28
0
1.37
0
1.06

The total time for your involvement will be approximately 2-3 hours in two separate testing sessions.

Exclusion:

If you meet any of the following criteria, you will be excluded from this study:

- If you have any known neuromuscular and/or balance disorder
- If you have experienced a musculoskeletal injury to either lower limbs within the past 18 months
- If you have any prior experience using a unilateral lower limb prosthetic simulator
- If your dominant leg is your left leg
- If you are unable to walk unassisted with the prosthetic simulator

Risks:
You are likely to experience instability at first when walking with the prosthetic simulator. The Investigator will walk along your side. You may also experience fatigue walking with the prosthetic device and muscle soreness following testing. If you feel any pain or intense discomfort, and feel unable or do not wish to continue, you may quit at any time. Reflective markers will be fixed to the skin surface using adhesive tape. Skin reactions to the adhesive tape are rare.

Benefits:
You will not receive any direct benefits by participating in this study.

Remuneration/Compensation:
You will not receive any reimbursement for expenses, gifts-in-kind and/or payment by participating in this study.

Confidentiality:
Any information regarding subject identification resulting from this study will be kept strictly confidential. All documents will be identified only by code number and kept in a locked and secured filing cabinet. Data will be kept up to a maximum of ten years and will only be accessible to Dr. Sanderson and Natalie Vanicek. Data files stored on computer will be labelled using code numbers on a computer in the Biomechanics Laboratory. You will not be identified by name in any reports of the completed study.

"Your confidentiality will be respected. No information that discloses your identity will be released or published without your specific consent to the disclosure. However, research records and medical records identifying you may be inspected in the presence of the Investigator or his or her designate by representatives of Health Canada, and the UBC Research Ethics Board for the purpose of monitoring the research. However, no records which identify you by name or initials will be allowed to leave the Investigators' offices."

APPENDIX C: DATA MAP

All raw data can be found in C:\My Documents\P055 on the PLUTO computer in the Biomechanics lab.

All raw data for each participant can be found in the P055 folder. Within this folder, there are two types of Excel files which have been used for data analysis.

- Files with a .TRBCoord ending contain raw three-dimensional coordinate data.
- Files with an .XLS ending contain T-D and kinematic data

All files within a specific condition (i.e. walking with the PS) were named in sequential order. For example, when Participant #4 walked with two intact legs before walking with the PS, file names began with SUBJECT_4_ for as many trials as were recorded. When Participant #4 walked with the PS, file names began with PEG_SUBJECT_4_. Upon removal of the PS, when walking with two intact limbs, file names began with AFTER_SUBJECT_4_

(*NB* a trial is used to signify each time data were recorded when the participant came into the cameras' filed of view when they completed a lap)

* Exceptions

- Participant #1, where all trials when walking with the PS start with PEG_SUBJECT_
- Participant # 5, ONLY trials when walking with two intact limbs before walking with the PS were named SUBJECT_4_. All PEG and AFTER trials are labelled as Participant #5.

All processed data can be found in C:\My Documents\P055\ALL SUBJECTS' DATA. This folder contains a total of 20 sub-folders (2 folders for each participant). One folder contains T-D and kinematic data, the other coordinate data.

1. <u>**Coordinate folder**</u> - contains 2 files

for Participant #1, for example, **PEG_SUBJECT_1_COORD PAGES** and **SUBJECT_1_COORD PAGES**. They are structured in exactly the same way except that one is for PEG data, the other when walking without the PS.

All raw coordinate data for all recorded trials when walking with the PS can be found within **PEG_SUBJECT_1_COORD PAGES**. Calculations of segment angles (shank, thigh, prosthetic and trunk) and the vertical position of the CoM can also be found in this workbook. A sheet entitled **CoM TEMPLATE** found the maximum values of the CoM during single stance for 3 strides during each trial. A mean (SD) value was then calculated for each trial (starting row 34). "Normalised values" were calculated every 5% walking distance (row 48) based on the slope between known data points (row 58).
(*NB* "Normalised" is used to signify interpolated data, which was then used to calculate all group mean values)

Sheets entitled, **R_PROS**, **L_SHANK**, **L_THIGH**, **L_TRUNK** AND **R_TRUNK** are linked to the PEG sheets that contain the calculated segment angles for all the captured frames (number of frames varies for each trial).

The last sheet entitled **KINEMATIC TEMPLATE** found the values of the segment angles at their instant of foot or prosthetic contact for 3 strides during each trial. A mean (SD) value was then calculated for each trial (starting row 35). "Normalised values" were then calculated every 5% walking distance (row 58) based on the slope between known data points (row 46).

All "Normalised values" for all segment angles for all 10 participants were then linked to a new worksheet entitled **PEG_SUMMARY_KINEMATIC** (located in C:\My Documents\P055\ALL SUBJECTS' DATA). Data from each participant is contained in sheets entitled **S_1**, **S_2**, ... **S_10**. Group mean values are calculated in **S_1** (starting row 15). All other sheets within this workbook are graphs of either group data or individual raw data.

2. <u>**T-D and kinematic data folder**</u> – contains 2 files

for Participant #1, for example, **PEG_SUBJECT_1_SUMMARY** and **SUBJECT_1_SUMMARY**. They are structured in exactly the same way except that one is for PEG data, the other when walking without the PS.

All the Orthotrak files with calculated T-D data and 3-D angles for all recorded trials when walking with the PS can be found within **PEG_SUBJECT_1_SUMMARY**. Column C4:C17 identifies the variables of interest, while columns D-G show the calculated values for each individual stride for that trial. Frame numbers for gait events (right and left foot contact and toe off – right side is prosthetic side) can be found starting C18:C21. 3-D angles calculated from Orthotrak can be found in row 24. A sheet entitled **SUMMARY 1** is linked to all the Orthotrak files containing the T-D variables for each trial.

T-D data for all participants are linked to a new workbook entitled **PEG_SUMMARY PAGES_b_DVs** (located in C:\My Documents\P055\ALL SUBJECTS' DATA) in sheets entitled **SUMMARY 1**, **SUMMARY 2**, ... **SUMMARY 10** †. Mean (SD) values for T-D variables for each trial for each participant can be found (starting row 18) in their respective worksheet (i.e. SUMMARY 1 for participant 1, etc ...). Normalised values for all T-D variables (starting row 85) were calculated based on the slope between known data points (starting row 114). Group mean values were calculated in SUMMARY 1 (starting row 145). All other sheets within this workbook are graphs of either group data or individual raw data.

† (*NB* T-D data for all participants with two intact legs before and after walking with the PS are linked to a new workbook entitled **SUMMARY PAGES** located in C:\My Documents\P055\ALL SUBJECTS' DATA)