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Balance control on a moving platform in unilateral lower limb amputees

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Abstract

Objective: To study balance control on a moving platform in lower limb amputees.
Design: Observational cohort study.
Participants: Unilateral transfemoral and transtibial amputees and able-bodied control subjects.
Interventions: Balance control on a platform that moved in the anteroposterior direction was tested with eyes open, blindfolded and while performing a dual task.
Main outcome measures: Weight bearing symmetry, anteroposterior ground reaction force and centre of pressure shift.
Results: Compared to able-bodied subjects, in amputees the anteroposterior ground reaction force was larger in the prosthetic and non-affected limb, and the centre of pressure displacement was increased in the non-affected limb and decreased in the prosthetic limb. In amputees body weight was loaded more on the non-affected limb. Blindfolding or adding a dual task did not influence the outcome measures importantly.
Conclusion: The results of this study indicate that experienced unilateral amputees with a high activity level compensate for the loss of ankle strategy by increasing movements and loading in the non-affected limb. The ability to cope with balance perturbations is limited in the prosthetic limb. To enable amputees to manage all possible balance disturbances in real life in a safe manner, we recommend to improve muscle strength and control in the non-affected limb and to train complex balance tasks in challenging environments during rehabilitation.

Keywords: Dynamic balance; Amputees; Rehabilitation; Centre of pressure

1. Introduction

Maintenance of balance is necessary during activities in daily life. In able-bodied individuals the ankle joint and lower leg musculature play an essential role in maintaining balance by appropriately shifting the centre of pressure (COP) [1,2]. Muscle contractions produce a torque around the ankle, which in turn generates changes in the COP and the direction of ground reaction forces (GRF) and modulates the anteroposterior movements of the centre of mass (COM) [2–4]. Following lower limb amputation, somatosensory input, muscle activity and joint mobility in the amputated part of the limb are compromised. As a consequence, lower limb amputees are unable to use the same motor strategies for balance control as able-bodied subjects and therefore have to adjust the habitual stance control strategies and develop new strategies [3,5,6].

To date, research concerning balance control in amputees has mainly focused on sway in quiet standing. The results of these studies are contradictory. In several studies the postural sway in individuals with recent or long-standing

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amputation was increased compared with able-bodied subjects [2,3,7–10], whereas other studies found no difference [11–14]. A single force plate was often used for balance measurements and the prosthetic and non-affected limbs were analyzed together without taking into account the different properties [2,3,8,10–12]. Studies that analyzed the non-affected and prosthetic limbs separately showed a decrease in weight bearing and COP excursions in the prosthetic limb [9,15–17]. Research has revealed that a good standing balance on the non-affected limb is beneficial for the functional outcome of amputees [18].

Static balance tests may not be sufficiently challenging to detect essential strategies for maintaining balance in daily activities, since balance control is often required during ambulation [3,4,19]. Falls regularly occur when balance control is hindered by an external perturbation [20]. A moving platform is a common method to study perturbations in balance [1,21–24]. Moving the platform displaces the COP away from the projection of the body's COM on the ground. To regain equilibrium, balance control strategies are used to shift the COM in the same direction as the platform displacement [1,25]. The anteroposterior GRF is used to adjust the movements of the COP and COM to the perturbations. It is known that amputees experience most difficulties in balance control in the anteroposterior direction [3].

Apart from the motor control system, balance control in daily life is also dependent on the sensory, visual, cognitive and vestibular systems [13,25]. Humans are able to switch between these balance control systems to compensate for a deficiency in one of the systems or to adjust to the environmental demands. To mimic balance performance in daily life it is important to assess motor skills in combination with other tasks [19,26]. In this way more subtle differences in balance performance between study groups can be detected [27].

In previous studies on quiet standing in amputees, balance control was made more difficult by closing the eyes and adding a dual task. Mean COP sway of both limbs and loading on the non-affected limb in amputees clearly increased when the eyes were closed [8–11,15,17], whereas in able-bodied subjects only a small [8–10] or no [11] effect was found. This would suggest that in amputees an increased contribution of visual control compensated for the impairment within the somatosensory system. Adding a dual task increased postural sway in amputees more than in able-bodied subjects, implicating that the maintenance of balance was not fully automated in amputees and required conscious control [2].

In this study we focused on the performance of more complex balance tasks in amputees by moving a platform, depriving vision and adding a dual task. The first aim was to establish the balance control strategies of the prosthetic and non-affected limbs in amputees compared to able-bodied subjects during standing on a moving platform. We hypothesized that in amputees the vertical and anteroposterior GRF and the anteroposterior COP displacement would increase in the non-affected limb and decrease in the prosthetic limb compared to able-bodied subjects. The second aim was to study the influence of visual deprivation and an acoustic dual task on balance control strategies. We hypothesized that amputees would increase the anteroposterior GRF and COP displacement in both limbs and would shift the vertical GRF more toward the non-affected limb when vision is deprived or a dual task is added compared to the normal condition.

2. Methods

2.1. Subjects

Amputees were approached via a prosthetics workshop. Inclusion criteria were age over 18 years, a unilateral lower limb

Table 1
Patient characteristics of amputees and able-bodied subjects

	Amputees ($n = 8$)	Able-bodied ($n = 9$)	p-Value study group
Sex	Six men, two women	Eight men, one woman	
Age (years)	51.8 ± 12.7	44.8 ± 9.9	*0.239*
Body weight (kg)	83.3 ± 9.7	85.6 ± 9.1	*0.622*
Height (m)	1.78 ± 0.09	1.84 ± 0.07	*0.142*
Level of amp	Three transfemoral, five transtibial		
Time since amp (months)	257.5 ± 195.6		
Cause of amp	Five trauma, one vascular, two tumor		
Prosthetic foot	Three dynamic SACH, two C-walk, one Quantum, one Griessinger multi-axial, one Endolite multiflex		
Prosthetic knee	One Tehlin, one Ottobock 3R60, one total knee		
ABC	84.5 ± 13.0	98.8 ± 1.1	*0.017**
AAS	32.0 ± 31.7		

Mean values and standard deviations of age, body weight, height, time since amputation, AAS and ABC. Sex, level and cause of amputation, and prosthetic feet and knees are provided in absolute numbers. Statistically significant p-values ($p \leq 0.05$) of differences between the study groups are marked with *.

amputation at least 1 year earlier, the use of a prosthesis on a daily basis, and the ability to stand with a prosthesis without walking aids for at least 30 min. A control group of able-bodied subjects was recruited through advertisements at the local blood bank, hospital, and television station. Subjects were excluded if they had any medical conditions that could affect their mobility or balance, such as neurological, orthopaedic or rheumatic disorders, the use of antipsychotic drugs, antidepressants or tranquilizers, otitis media, or impaired vision. Additional exclusion criteria for amputees included reduced sensation of the non-affected limb, ulceration or pain at the stump, or fitting problems of the prosthesis.

Eight amputees and nine able-bodied subjects agreed to participate in the study. The medical ethics committee approved the study protocol and all subjects signed informed consent before testing. Amputees used their own prosthetic limb. The different types of prosthetic feet and knees are detailed in Table 1. To obtain information on functional skills, amputees filled in the modified amputee activity score (AAS), a suitable measure for outpatient amputees with good test–retest reliability and validity [28,29]. The range of the score is −70 to +50 and a higher score represents a higher activity level. Both groups filled out the activities-specific balance confidence scale (ABC), which is designed to assess balance confidence when performing activities such as climbing stairs, reaching above the head, and walking on different surfaces. The maximum score is 100. The ABC is shown to be reliable and there is strong support for validity [30,31].

2.2. Apparatus

Balance measurements were performed on the computer assisted rehabilitation environment (CAREN) system [32] which consists of a 2-m diameter platform that can rotate around three orthogonal axes and translate in three directions along these axes. The platform contains two built-in 0.40 m × 0.60 m Bertec force plates to register GRF with a sampling frequency of 100 Hz. COP data were derived from the GRF and platform moment of force data.

2.3. Procedure

Subjects stood erect on the moving platform with their hands alongside their bodies. For reasons of safety subjects were provided with a safety belt that was connected to the ceiling. The feet were placed in a self-selected position, one on each force plate. Subjects were instructed to stand with both feet on the floor as motionless as possible and to maintain balance while the platform swayed for 60 s in the anteroposterior direction. The platform movements were sinusoidal. During the first 15 s the excursions gradually increased to a maximum amplitude of 0.02 m. Maximum platform excursions were executed from the 15th to the 45th s, after which the excursions slowly diminished towards the end of the test. The frequency of the excursions was 1 Hz. The mean anteroposterior velocity was 0.046 m s^{-1}, the maximum velocity 0.13 m s^{-1} and the maximum acceleration 0.79 m s^{-2}. Between the tests a 60 s break was allowed.

Subjects were tested in three conditions—(1) normal: single task with eyes open, (2) blindfolded: diving goggles with non-transparent black glasses, and (3) dual task: adding the acoustic Stroop test [33,34]. In this test the words "high" and "low" were pronounced in a high or low pitch. Subjects had to name the pitch in which the word was spoken and suppress the tendency to repeat the word they heard. Prior to the balance tests the Stroop test was practiced once. The conditions were presented in random order to avoid learning effects.

2.4. Outcome parameters

MATLAB software was used for data analysis. Since we were interested in the period of maximum platform excursions, the first and last 15 s of the balance recordings were excluded and a period of 30 s remained. Balance control was described by three parameters: (1) the weight bearing index (WBI) as a measure for symmetry in body weight distribution, which was calculated from the vertical component of GRF (GRFz), (2) the anteroposterior component of GRF (GRFy), and (3) the anteroposterior COP displacement (ΔCOPy). WBI in amputees was the ratio of GRFz in the non-affected limb divided by GRFz in the prosthetic limb. A ratio score is more often used to quantify limb asymmetries [35]. To calculate WBI in able-bodied subjects the limb with the largest GRFz was divided by GRFz in the other limb. In amputees the ΔCOPy and GRFy data of the prosthetic and non-affected limbs were analyzed separately, whereas in able-bodied subjects the mean of the right and left limbs was used. ΔCOPy was defined as the sum of absolute values of the COP differences, and GRFy and GRFz as the sum of absolute GRF values.

2.5. Statistical analysis

The Kolmogorov–Smirnov test showed that all data were normally distributed and Levene's test showed that variances in ΔCOPy and GRFy were equal, whereas in WBI heterogeneity of variance was seen. Differences between the groups were analyzed by the independent *t*-test; in ΔCOPy and GRFy with equal variances, in WBI with unequal variances. Differences between conditions (normal–blindfolded and normal–dual task) within the groups were analyzed by the paired *t*-test. The level of significance was set on $p \leq 0.05$.

3. Results

Characteristics of the subjects are presented in Table 1. Apart from a higher score on the ABC in able-bodied subjects, no statistically significant differences in subject characteristics were found between amputees and able-bodied subjects. The results of WBI, GRFy and ΔCOPy are presented in Tables 2–4. All subjects were able to maintain balance during the tests without taking a step. Amputees significantly preferred to bear weight on their non-affected limb in all three conditions. WBI in amputees was significantly more asymmetric than in able-bodied subjects, although in able-bodied subjects weight bearing was not equally divided between the two limbs. In amputees 62–63% of the body weight was loaded on the non-affected limb. GRFy in the non-affected limb of amputees was significantly larger in comparison with able-bodied subjects in all three conditions. GRFy in the prosthetic limb of amputees was also larger than in able-bodied subjects, but only significantly in the normal condition. In the normal and dual task conditions ΔCOPy of the non-affected limb in amputees

Table 2
Mean values and standard deviations of the WBI in amputees and able-bodied subjects

	Amputees ($n = 8$)	Able-bodied ($n = 9$)	p-Value study group
WBI normal	1.65 ± 0.42	1.15 ± 0.14	*0.025**
WBI blindfolded	1.67 ± 0.49	1.17 ± 0.15	*0.008**
p-Value normal–blindfolded	*0.812*	*0.465*	
WBI dual	1.69 ± 0.49	1.19 ± 0.18	*0.010**
p-Value normal–dual task	*0.755*	*0.173*	

Statistically significant p-values ($p \leq 0.05$) of differences between study groups are marked with *.

was significantly larger than in able-bodied subjects. ΔCOPy under the prosthetic limb in amputees was lower than in able-bodied subjects in the blindfolded and dual task conditions. A typical example of GRFy and ΔCOPy in the prosthetic and non-affected limbs of a subject in the amputee group during the normal condition is presented in Fig. 1.

The only significant effect of condition was demonstrated in able-bodied subjects in which ΔCOPy in the blindfolded condition was increased compared to the normal condition. In WBI and GRFy there were no significant differences between the normal, blindfolded and dual task conditions.

4. Discussion

The first aim of this study was to establish balance control strategies on a moving platform in amputees. In our study, amputees loaded 37–38% of their body weight on their prosthetic limb, whereas studies on quiet standing reported that transtibial amputees loaded approximately 45% and transfemoral amputees 40% of their body weight on the prosthetic limb [15,17,36]. Hence, loading of the non-affected limb seems to increase slightly when balance is perturbed, compared to quiet standing. Various explanations have been suggested for the asymmetric weight bearing strategy in amputees; reduced ankle mobility, stump pain, discomfort due to the rigid prosthetic socket or prosthetic alignment, poor hip abductor muscle strength, inadequate sensory information, lack of confidence, poor balance, or habitual stance [11,15,16]. Whereas the advantage of increased weight bearing on the non-affected limb is improved control, the disadvantage is more frequent overloading and arthrosis of the non-affected limb [11,15,37].

In the present study ΔCOPy in the prosthetic limb was limited, which can be explained by the absent ankle musculature, the necessary flexibility of the prosthetic foot and the decreased weight bearing on this limb. As an adjustment strategy, amputees increased ΔCOPy in the non-affected limb, which can be explained by an increased muscle activity in this limb and the trunk. In the normal condition ΔCOPy in the non-affected limb was increased by a factor 2.5 compared to the prosthetic limb, and by a factor

Table 3
Mean values and standard deviations of GRFy of the prosthetic (P) and non-affected (N) limb in amputees and able-bodied subjects

Limb	Amputees ($n = 8$)	Able-bodied ($n = 9$)	p-Value study group
GRFy normal (% BW)			
N	33.9 ± 4.5	23.1 ± 3.3	*0.000**
P	30.9 ± 8.7		*0.022**
GRFy blindfolded (% BW)			
N	36.6 ± 7.8	23.7 ± 4.8	*0.001**
P	33.0 ± 13.5		*0.065*
p-Value normal–blindfolded			
N	*0.119*	*0.763*	
P	*0.388*		
GRFy dual (% BW)			
N	32.1 ± 9.0	22.1 ± 5.1	*0.013**
P	29.7 ± 15.3		*0.188*
p-Value normal–dual task			
N	*0.435*	*0.633*	
P	*0.698*		

GRFy was expressed in % body weight (% BW). Statistically significant p-values ($p \leq 0.05$) of differences between study groups are marked with *.

Table 4
Mean values and standard deviations of ΔCOP of the prosthetic (P) and non-affected (N) limb in amputees and able-bodied subjects

Limb	Amputees ($n = 8$)	Able-bodied ($n = 9$)	p-Value study group
ΔCOP normal (m)			
N	3.38 ± 1.69	1.91 ± 0.62	*0.027**
P	1.36 ± 0.41		*0.053*
ΔCOP blindfolded (m)			
N	4.28 ± 2.18	2.82 ± 0.87	*0.082*
P	1.39 ± 0.41		*0.001**
P-Value normal–blindfolded			
N	*0.063*	*0.001†*	
P	*0.546*		
ΔCOP dual (m)			
N	3.47 ± 1.67	2.14 ± 0.61	*0.043**
P	1.30 ± 0.30		*0.003**
p-Value normal–dual task			
N	*0.689*	*0.222*	
P	*0.352*		

ΔCOP was expressed in m. Statistically significant p-values ($p \leq 0.05$) of differences between study groups are marked with *, and of differences between the normal and blindfolded condition within the groups with †.

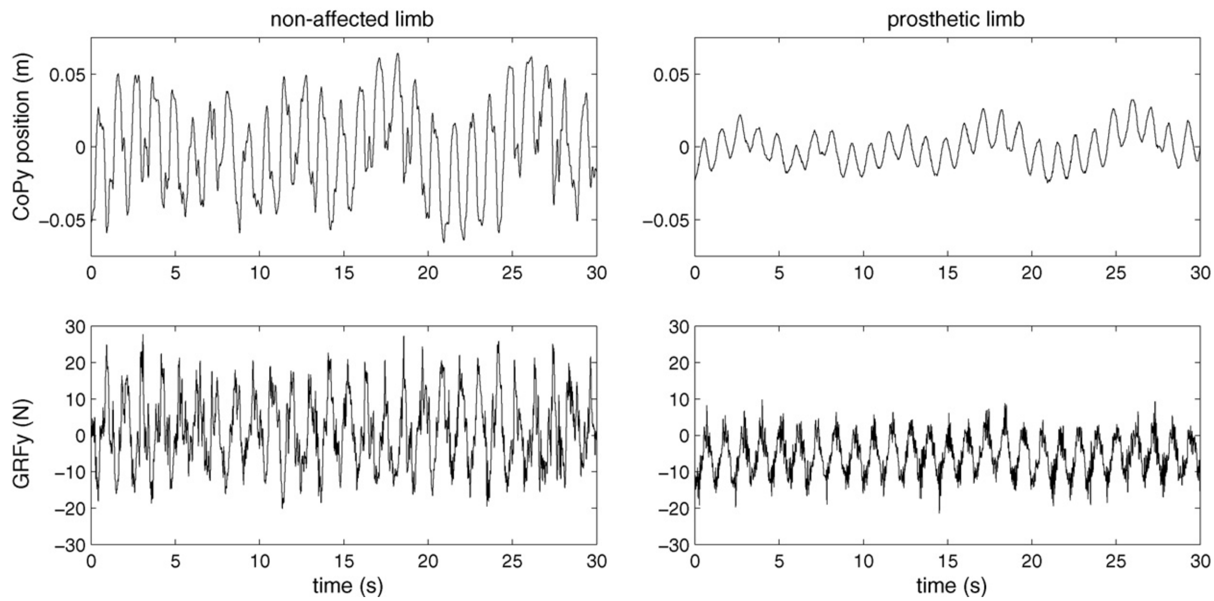

Fig. 1. A typical example of COPy and GRFy in a subject with a transfemoral amputation. COPy and GRFy in the non-affected limb are larger than in the prosthetic limb during the 30-s measuring period.

1.8 compared to able-bodied subjects. Earlier studies on quiet standing in amputees showed similar results in COP; in the non-affected limb of experienced amputees COP excursions were approximately twice as large as in the prosthetic limb [15,17]. In amputees with a recent amputation COP velocity in the non-affected limb was 3.5 times larger than in the prosthetic limb at the end of rehabilitation [2]. From this we may conclude that this adjustment strategy of the non-affected limb does not change when the difficulty of the balance task increases from quiet standing to standing on a moving platform.

Despite the inability of amputees to compensate at the ankle of the prosthetic limb, a larger GRFy was found than in able-bodied subjects. In a study on subjects with somatosensory loss of the lower leg induced by anaesthesia GRFy was also increased in response to platform displacements [25]. By increasing GRFy, more somatosensory input can be received in the prosthetic limb [8,15,17]. Amputees may influence GRFy by using the intact hip musculature in the prosthetic limb. Flexion and extension in the hip shift the COM forward and backward [1,3], and consequently result in a larger GRFy in both limbs. Muscle activity in the prosthetic limb is required to limit the degrees of freedom of the prosthesis during platform perturbations, and in transfemoral amputees to keep the prosthetic knee locked in extension.

The second aim of this study was to determine the effect of visual and conscious control on balance control strategies. In our study only the blindfolded condition had an effect in able-bodied subjects. In amputees balance control was not significantly influenced by the additional task. Several reasons can be found to explain why amputees did not shift to visual or cognitive balance control strategies during standing on a moving platform. The increase of ΔCOPy and weight bearing on the non-affected limb in amputees was already large in the normal condition. Using these adjustment strategies more intensively may have endangered the stability on the platform in amputees. In contrast, able-bodied subjects were not using their balance control strategies to a full extent in the normal condition. They were therefore able to increase muscle activity in the blindfolded condition, resulting in a larger ΔCOPy. Furthermore, managing balance perturbations may have been an entirely automated task, because the amputees were experienced prosthetic users and the platform movements were predictable. The absence of a condition effect may also be explained by the ease of the dual task, the small number of subjects, or an inadequate performance of the dual task.

The present study has a number of limitations. Although clear differences between the study groups were found, the study groups were only small and consisted of subjects with different amputation levels. The results can only be considered indicative for experienced lower limb amputees with a normal to high activity level and may not be generalized to all amputees. Since we wished our study to resemble a real life situation, we did not standardize standing position. As a result, foot position and therefore base of support may have been different in subjects, which influences the outcome parameters. Previous studies reported that able-bodied subjects stood with their feet closer together than amputees [7], and that visual dependency in amputees was reduced when the base of support was wider [11]. Due to technical limitations GRFy and COPy values were not corrected for inertia of the force plate in relation to the sensors. Since this measuring error was only small and similar in groups and conditions, a significant effect on the results would be unlikely.

In order to mimic a real life situation it is important that a task closely simulates an activity which is difficult for amputees [4]. The moving platform in our experiment would simulate standing in a moving bus. The blindfolded and dual task conditions would simulate standing in a bus in the dark, while having a conversation. However, in a bus the perturbations occur more unexpectedly and with a higher velocity. The present study demonstrated the effect of expected perturbations on balance control in amputees. Future research should focus on unexpected balance disturbances.

Most adjustments strategies in amputees occurred in the non-affected limb, whereas the requirements on the prosthetic limb during balance perturbations were only limited. With purpose of enabling amputees to manage all possible balance disturbances in real life in a safe manner, we recommend improvement in muscle strength and control of the non-affected limb and training in complex balance tasks in challenging environments during rehabilitation.

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Conflict of interest

Authors state that no conflict of interest is present in the research.

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