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Gait termination in lower limb amputees

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Abstract

Objective: To study the limitations in function and adjustment strategies of lower limb amputees in gait termination.
Design: Observational cohort study.
Setting: University Medical Centre.
Participants: Unilateral transfemoral and transtibial amputees, and able-bodied control subjects.
Main outcome measures: Leading limb preference, temporal variables, lower limb joint angles, ground reaction forces, and centre of pressure shift.
Results: Compared to able-bodied subjects, amputees showed a decreased peak braking ground reaction force in the prosthetic limb, no anterior centre of pressure shift during leading with the prosthetic limb and an increased mediolateral centre of pressure shift. Amputees used several adjustment strategies to compensate for the limitations in function; leading limb preference for the non-affected limb, longer production of braking force in the non-affected limb, decreased gait termination velocity and more weight-bearing on the non-affected limb.
Conclusion: Limitations in function and adjustment strategies were mainly similar in transfemoral and transtibial amputees. Due to the lack of active ankle function, amputees were not able to increase the braking force and to shift the centre of pressure anteriorly. Leading with the non-affected limb is favourable for adequate deceleration and balance control, but in daily life not always applicable. It is important that amputees are trained in gait termination during rehabilitation and prosthetic design should focus on a more active role of the prosthetic foot and knee.

Keywords: Gait; Balance; Amputation; Prosthetics; Rehabilitation

For safe and independent walking, it is necessary that a person is able to adapt the gait pattern to various situations. One example of such an adaptation occurs in the transition from walking to standing, also called gait termination. In everyday life, gait termination is a common movement which is often performed in daily activities [1]. Compared to normal walking, gait termination places a bigger demand on the control of postural stability and requires a complex integration and cooperation in the neuromuscular system [2–4].

For safe gait termination, forward movement of the body has to be slowed down to achieve a stable upright position [2,5]. In able-bodied persons the leading limb, which is the limb that stands still first, is mainly responsible for the production of the necessary braking ground reaction force (GRF). Compared to normal walking the braking GRF is increased in the final stance phase [6–8]. A large burst of soleus muscle activity and reduced activation in the tibialis anterior muscle of the leading limb bring the foot flat to the ground [5,7,9,10]. The vasti and gluteus medius muscles are activated, respectively, to extend the knee and to prevent the trunk from bending forward. In the trailing limb the tibialis anterior, biceps femoris and gluteus medius muscles increase activity to bring the body down and backwards with the foot flat to the ground, resulting in a further decrease

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in forward movement [9]. The muscle activity moves the centre of pressure (COP) anteriorly and keeps the centre of mass (COM) behind the leading limb [5,9]. The position of the COP in front of the COM and the increased braking GRF lead to deceleration of the body [11]. Also stability requirements have to be fulfilled for safe gait termination. In the final bipedal standing position the COM closely coincides with the COP and lies within the base of support [12,13]. The leading limb has the task to create a stable landing placement at the end of the gait termination process.

Lower limb amputees are not able to use an ankle strategy and in transfemoral amputees an active knee function is also absent. In addition, after a lower limb amputation balance control is reduced [14–16]. Due to the loss of nerves, muscles and joints, it is to be expected that gait termination may lead to difficulties in lower limb amputees. No studies on gait termination in amputees have been published so far. Our first objective was to determine which functions were limited in amputees during planned gait termination. We formulated three hypotheses: (1) the braking GRF in the leading prosthetic limb will be decreased due to absence or inefficiency of ankle plantar flexors and knee extensors, (2) the anterior shift of the COP in the prosthetic limb will be reduced as the result of a deficient ankle strategy, and (3) the mediolateral COP shift will be increased owing to reduced balance control.

To carry out gait termination in a safe manner, amputees may have to adjust their gait pattern. The second purpose of this study was to assess which adjustment strategies amputees use during gait termination in order to compensate for the limitations in function. We hypothesized four possible compensation strategies: (1) the production of a larger braking GRF in the non-affected limb will compensate for the reduced braking GRF in the prosthetic limb, (2) gait termination velocity will be reduced so that less braking GRF is needed, (3) a preference for the non-affected limb as leading limb will be seen as a result of the larger braking GRF in this limb, and (4) swing phase duration of the non-affected limb will be shortened to minimize single-limb stance duration on the prosthetic limb, which improves stability.

1. Methods

1.1. Subjects

Amputees who were regularly attending the local prosthetics workshop were invited to participate in this study. Inclusion criteria were: uni-lateral transfemoral amputees (TF) and transtibial amputees (TT), amputation at least 12 months before inclusion, the use of a prosthesis on a daily basis, and the ability to walk more than 50 m without walking aids. We recruited a control group of able-bodied

Table 1
Patient characteristics, leading limb preference and temporal variables in the leading prosthetic (LP), the trailing prosthetic (TP), the leading non-affected (LN), and the trailing non-affected (TN) limb condition in TF, TT, and AB

Group	TF ($n = 7$)	TT ($n = 12$)	AB ($n = 10$)
Sex (men/women)	6/1	10/2	9/1
Age (years)	44.0 ± 14.1	49.6 ± 11.6	45.2 ± 9.4
Body weight (kg)	81.4 ± 12.4	84.2 ± 8.2	86.5 ± 9.1
Height (cm)	182.6 ± 6.2	180.9 ± 8.5	184.4 ± 6.7
Time since amputation (months)	210.7 ± 158.1	207.8 ± 69.4	–
Side amputation (right/left)	5/2	6/6	–
Cause of amputation			
Trauma	4	6	–
Vascular	0	2	–
Oncology	3	4	–
AAS	35.9 ± 26.9	33.8 ± 26.1	–
ABC	83.5 ± 15.9*a	88.4 ± 5.4*b	98.7 ± 1.0
LLP (%)	42.9 ± 9.5	47.4 ± 6.0	54.2 ± 11.5
GTV (m/s)			
LP	0.74 ± 0.14*a	0.85 ± 0.21*b	–
LN	0.75 ±0.12*a	0.89 ± 0.23*b	1.10 ± 0.28
SwD (s)			
LP	0.55 ± 0.07*a,c †	0.44 ± 0.08*c	–
TP	0.43 ± 0.06*a,c †	0.34 ± 0.06*c	–
LN	0.44 ± 0.09	0.39 ± 0.05	0.43 ± 0.04
TN	0.27 ± 0.05	0.30 ± 0.06	0.31 ± 0.04

Mean values and standard deviations of age, body weight, height, time since amputation, amputation activity scale (AAS), activities-specific balance confidence scale (ABC), gait termination velocity (GTV) and swing phase duration (SwD). Mean values and standard error of leading limb preference (LLP), in amputees for the prosthetic limb and in AB for the right limb. Sex, side and cause of amputation are provided in absolute numbers. Statistically significant p-values ($p \leq 0.05$) of between group differences are marked with *; *a for differences between AB and TF, *b for differences between AB and TT, and *c for differences between TF and TT. Statistically significant p-values ($p \leq 0.05$) of differences between the prosthetic and non-affected limb within TF and TT are marked with †.

subjects (AB) through advertisements at the local radio and television, blood bank and hospital. We excluded subjects who suffered from medical conditions that could affect their mobility or balance: neurological, orthopaedic or rheumatic disorders, cognitive problems, significantly impaired vision, reduced sensation of the non-affected leg, or use of antipsychotic drugs, antidepressants or tranquillisers. Amputees with pain or wounds of their amputation limb or fitting problems of the prosthesis were excluded as well.

Prior to the start of the study we obtained approval from the medical ethics committee. Seven TF, 12 TT and 10 AB agreed to participate in the study. Before testing, all subjects signed informed consent. The subject characteristics are provided in Table 1. The amputees used different types of prosthetic feet and knees. All TF used free moving prosthetic knees.

1.2. Apparatus

The study was performed in a motion analysis laboratory, which is equipped with an 8 m long aluminium walkway and a force plate[1] of 40 cm × 60 cm. We recorded the gait pattern with two video cameras[2] in the coronal and sagittal plane. The frame frequency was 25 Hz. We collected data on leading limb preference, temporal variables and joint angles. We used six electro-goniometers, in which high accuracy and repeatability were demonstrated, to measure the joint angles.[3] The goniometers were placed on the (prosthetic) ankle, (prosthetic) knee and hip of both limbs. We calibrated the goniometers by placing the subject in an erect position with hips and knees in extension and the feet in a plantigrade position. Subjects walked with their own shoes. The soles were provided with flexible aluminium strips at the heel and forefoot. Contact of the strips with the conductive walkway indicated the timing of initial contact and toe-off. The signals of the goniometers and foot contacts were recorded by a portable data acquisition system[4] at a sampling frequency of 800 Hz. The runs on the force plate were used for the assessment of the GRF and COP. Recording, analysing and synchronizing of all measurements was performed by using a custom developed Gait Analysis System[5] at 100 Hz.

1.3. Procedure

Amputees filled in two questionnaires to determine their activity level and balance confidence, respectively, the modified amputee activity score (AAS) [17,18] and the activities-specific balance confidence scale (ABC) [19–21]. AB only filled out the latter questionnaire.

Subjects performed eight runs on the walkway. They were instructed to walk at their self-selected velocity, to stop walking on their own initiative approximately halfway the walkway and to stand still for at least 2 s. The timing and exact position of stopping were voluntary chosen. No instructions were given on which limb should be used as leading limb.

For the force plate runs, we instructed the subjects to terminate walking by stepping with the leading limb on the force plate, followed by placing the trailing limb next to the leading limb. The subjects performed repeated runs, until the prosthetic and non-affected limb were both used twice as leading limb in an adequate manner. The subjects performed at least three steps prior to the gait termination step, to achieve steady-state gait [22–24]. Adjustment of the step length in order to hit the force plate was avoided by practicing the task in advance to select an appropriate distance from the starting point to the force plate. The subjects were instructed to look at the end of the walkway instead of at the force plate.

1.4. Outcome parameters

In amputees we analysed the outcome parameters in two limb conditions: (1) leading with the prosthetic limb and trailing with the non-affected limb and (2) trailing with the prosthetic limb and leading with the non-affected limb. Consequently, data on four

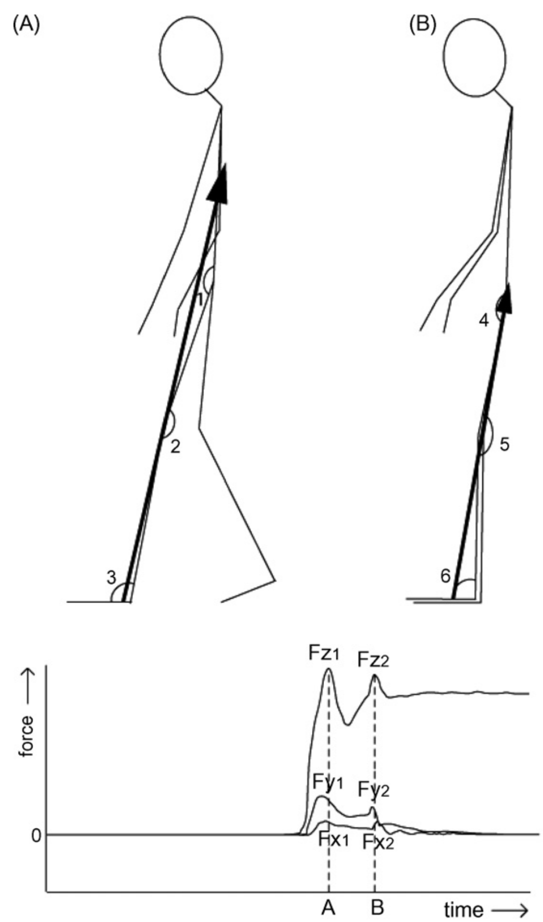

Fig. 1. Stick diagram showing the analysed joint angles and resultant GRF, and schematic illustration of the analysed peak components of the GRF of an AB subject. Fz represents the peak vertical component of the GRF, Fy the peak anteroposterior component and Fx the peak mediolateral component. (A) The hip, knee and ankle joint angles of the leading limb at the moment of trailing limb toe-off, which coincides with Fz1. (B) The hip, knee and ankle joint angles at the moment of trailing limb initial contact, which coincides with Fz2. The arrows represent the direction and amplitude of the resultant GRF at these events in the gait cycle.

[1] Bertec Corporation, 6171 Huntley Road Suite J, Columbus, OH 43229, USA.

[2] Panasonic F15 HS, Panasonic Info Centre, Postbus 236, 5201 AE's-Hertogenbosch, The Netherlands.

[3] Penny & Giles SG 150, Penny & Giles Biometrics Ltd., Unit 25 Nine Mile Point Industrial Estate, Cwmfelinfach Gwent NP1 7HZ, UK.

[4] PORTI, Twente Medical Systems International BV, H. ter Kuilestraat 181, 7547 SK Enschede, The Netherlands.

[5] GAS, University Medical Center Groningen, Hanzeplein 1, 9700 RB Groningen, The Netherlands.

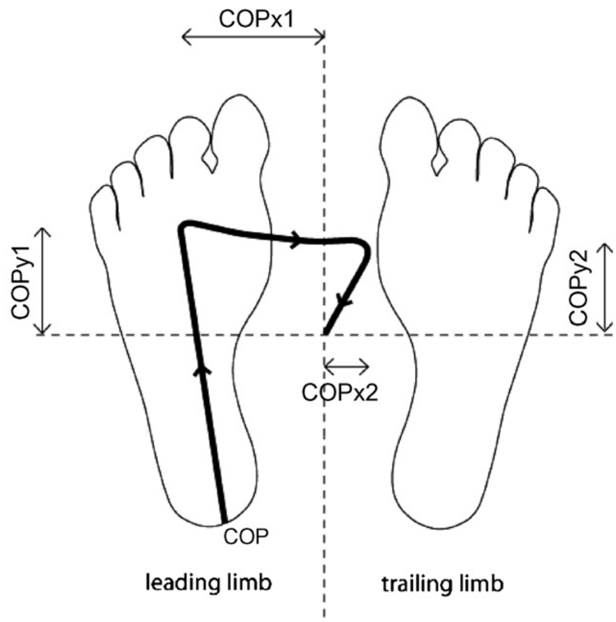

Fig. 2. Schematic representation of the four measuring points of the COP trajectory in an AB subject. The COP moves anteriorly and towards the side of the leading limb during single-limb stance. The most anterior and lateral position is reached at midstance. Just prior to initial contact of the trailing limb the COP moves towards trailing side, followed by a small posterior shift until the final bipedal stance position is reached. COPy1: distance between the final bipedal stance position and the most anterior position on the leading side, COPx1: distance between the final bipedal stance position and the most lateral position on the leading side, 3: COPx2: distance between the final bipedal stance position and the most lateral position on the trailing side, and COPy2: distance between the final bipedal stance position and the most anterior position on the trailing side.

lower limbs were obtained: leading prosthetic, trailing prosthetic, leading non-affected, and trailing non-affected. When amputees used the same limb as leading limb in all walkway runs, temporal variables and joint angles of the opposite leading limb could not be determined. In AB we used the mean outcome of the right and left limb in the data analysis to compensate for any asymmetry in the limbs. Similar to amputees, the results for the leading and trailing limb in AB were analysed separately.

The leading and trailing limb were determined by using the video images. To obtain a leading limb preference the percentage of prosthetic leading limb runs was determined in amputees and TT, while in AB the percentage of right leading limb runs was scored. The analysed temporal variables were swing phase duration and gait termination velocity in the final stride. Gait termination velocity was assessed at leading limb initial contact and was derived from the anteroposterior component of the GRF by integration. The joint angles of the hips, knees and ankles were analysed; for the leading limb at the moment of toe-off of the trailing limb and for the trailing limb at the moment of initial contact of the trailing limb (Fig. 1). These events of the gait cycle are critical in gait termination and coincide with the maximum peaks of the vertical GRF component.

We assessed the maximum amplitude of the GRF in the vertical (Fz), anteroposterior (Fy) and mediolateral (Fx) direction (Fig. 1). The first peak GRF (Fx, y, z1) represented the maximum exerted force by the leading limb in single-stance and the second peak GRF (Fx, y, z2) the maximum produced force in bipedal stance when the trailing limb was placed. To exclude the influence of body weight we expressed GRF as a percentage of body weight. The trajectory of the COP was described by using four measuring points (Fig. 2). These points were related to the final bipedal stance position at the end of gait termination. Due to the use of a single force plate, prior to initial contact of the trailing limb the COP shift under the leading limb was measured, whereas after trailing limb initial contact the resultant COP shift under both limbs was assessed.

1.5. Statistical analysis

For each limb condition the mean value of the outcome variables was calculated. Normality of the variables within the groups was tested with the Kolmogorov–Smirnov test. Differences in outcome parameters among the three groups were analysed by using an ANOVA followed by post hoc analysis according to the least-significant difference (LSD) method. Differences in time since amputation, AAS and leading limb preference were only tested between TF and TT. The paired *t*-test was used to analyse the differences between the non-affected and the prosthetic limb within TF and TT. The level of significance was set on $p \leq 0.05$.

2. Results

Unless otherwise mentioned, only statistically significant differences are presented. Data on activity level, balance confidence, leading limb preference, gait termination velocity and swing phase duration are shown in Table 1. TF and TT demonstrated a lower ABC score than AB did. Activity level in TF and TT was quite similar. Leading limb preference revealed that TF and TT used the non-affected limb more often (but non-significantly) as leading limb than the prosthetic limb. All subjects alternated the leading limb and used both limbs at least once as leading limb in the walkway runs. In TF and TT gait termination velocity was lower than in AB. Whether gait termination was led with the prosthetic or non-affected limb did not influence gait velocity. In TF swing phase duration was longer in the prosthetic limb compared to their non-affected limb, TT and AB.

The joint angles are shown in Fig. 3. In TF hip flexion of the leading prosthetic limb and knee flexion on the leading and trailing prosthetic side were decreased compared to TT, AB, and the non-affected limb in TF. In TT hip flexion in the leading prosthetic limb was reduced compared to the non-affected limb within the TT group. Knee flexion in the trailing prosthetic limb of TT was lower than in the non-affected limb and compared to AB.

In Fig. 4 the results of the GRF are provided. In TF and TT Fz1 of the leading prosthetic and non-affected limb was reduced compared with Fz1 in AB. In TT Fz2 in trailing with the prosthetic limb was also lower than in AB. Fz2 in trailing with the non-affected limb was larger in TF than in TT. In AB Fz1 in the leading limb and Fz2 in the trailing limb were similar, whereas in TF and TT Fz2 was larger than Fz1 in both limbs. Fy in leading and trailing with the prosthetic limb in TF and TT was decreased compared to Fy in AB and in the non-affected limb in TF and TT. Fy2 in trailing with

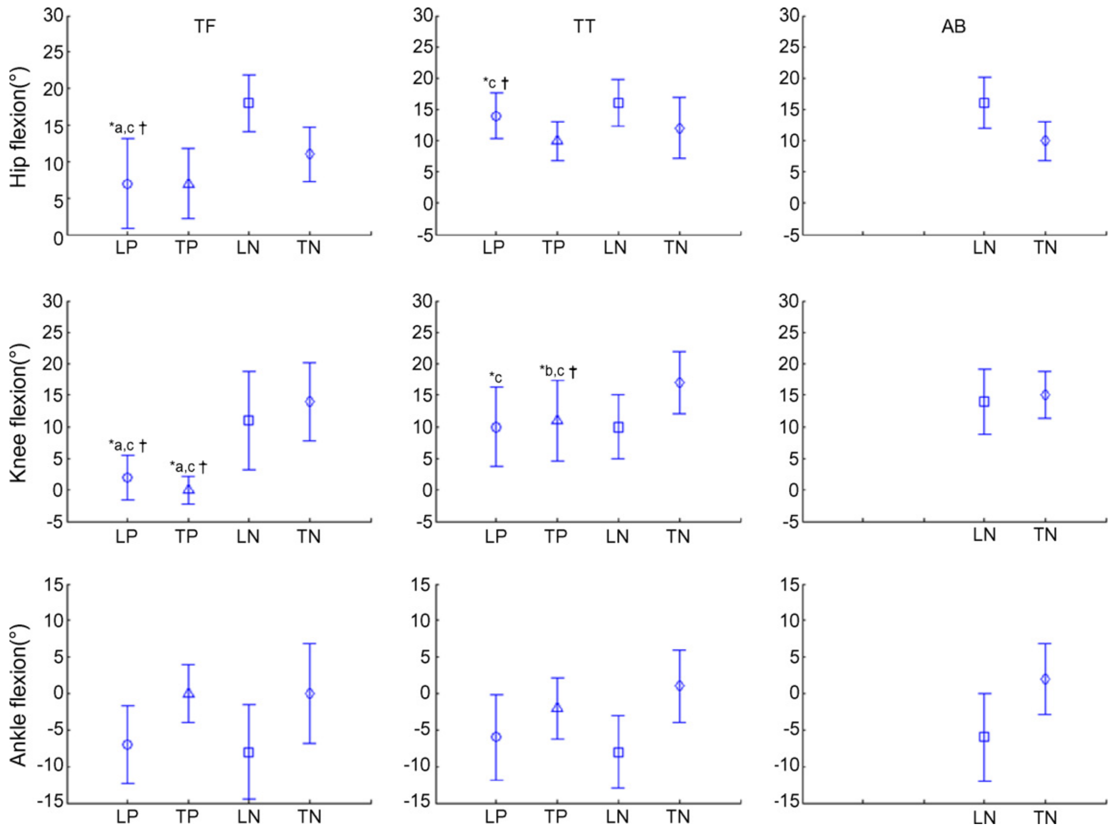

Fig. 3. Mean values and standard deviations of hip, knee and ankle joint angles in the leading prosthetic (LP), the trailing prosthetic (TP), the leading non-affected (LN), and the trailing non-affected (TN) limb condition in TF, TT and AB. Statistically significant p-values ($p \leq 0.05$) of between group differences are marked with *; *a for differences between AB and TF, *b for differences between AB and TT and *c for differences between TF and TT. Statistically significant p-values ($p \leq 0.05$) of differences between the prosthetic and non-affected limb within TF and TT are marked with †. Ankle dorsal flexion is positive, ankle plantar flexion is negative.

the prosthetic limb in TF was smaller than in TT. The leading non-affected limb in TF and TT demonstrated a decreased Fy1 compared to AB. In all groups the leading limb exerted a larger Fy than the trailing limb. In TF and TT the trailing prosthetic limb showed a decreased Fx2 compared to the non-affected limb. In the trailing non-affected limb of TF Fx2 was larger than in AB.

The results of the COP trajectory are presented in Fig. 5. In comparison with AB, COPx1 and COPx2 were increased in TF and TT when leading with the prosthetic limb. In TF COPx2 in leading with the non-affected limb was also larger than in AB. COPy1 of the leading prosthetic limb in TF and TT was directed posteriorly, whereas in leading with the non-affected limb in TF, TT and AB COPy was located anteriorly. Posterior COPy in the leading prosthetic limb of TT was smaller than in TF. Finally, COPy2 did not show significant differences in both limb conditions. Fig. 6 shows a typical example of the COP trajectory of a subject in the TF group.

3. Discussion

The most important requirements for gait termination are the production of sufficient Fy and the anterior displacement of COPy [11]. From our study we can conclude that amputees are only able to produce a limited amount of Fy in the prosthetic limb. Fy in the prosthetic limb of TF was reduced by at least 50%, and in TT by approximately 33% compared to AB. In the leading prosthetic limb COPy remained near the heel, whereas in trailing with the prosthetic limb and in AB the COPy was moved anteriorly.

The limitations in function of the prosthetic limb can be explained by the deficient lower limb musculature and the different properties of a prosthetic device compared to a non-affected limb. The stiffness of a prosthetic foot impedes a smooth shift of the COPy toward the forefoot. In a prosthetic knee a locking mechanism ensures knee extension during weight-bearing in early stance. Knee flexion in the trailing prosthetic limb was reduced in both amputee groups, especially TF, which impedes positioning of the body behind the leading limb and lowering of the body. The reduced weight and more proximally located COM in the prosthetic limb compared to a non-affected limb may have contributed to the smaller Fy in the prosthetic limb. However, in normal walking no significant effects of inertial prosthetic properties on GRF in amputees are found [25].

In amputees, adjustment strategies were seen that benefit deceleration. First, in TF the hip and knee in the leading

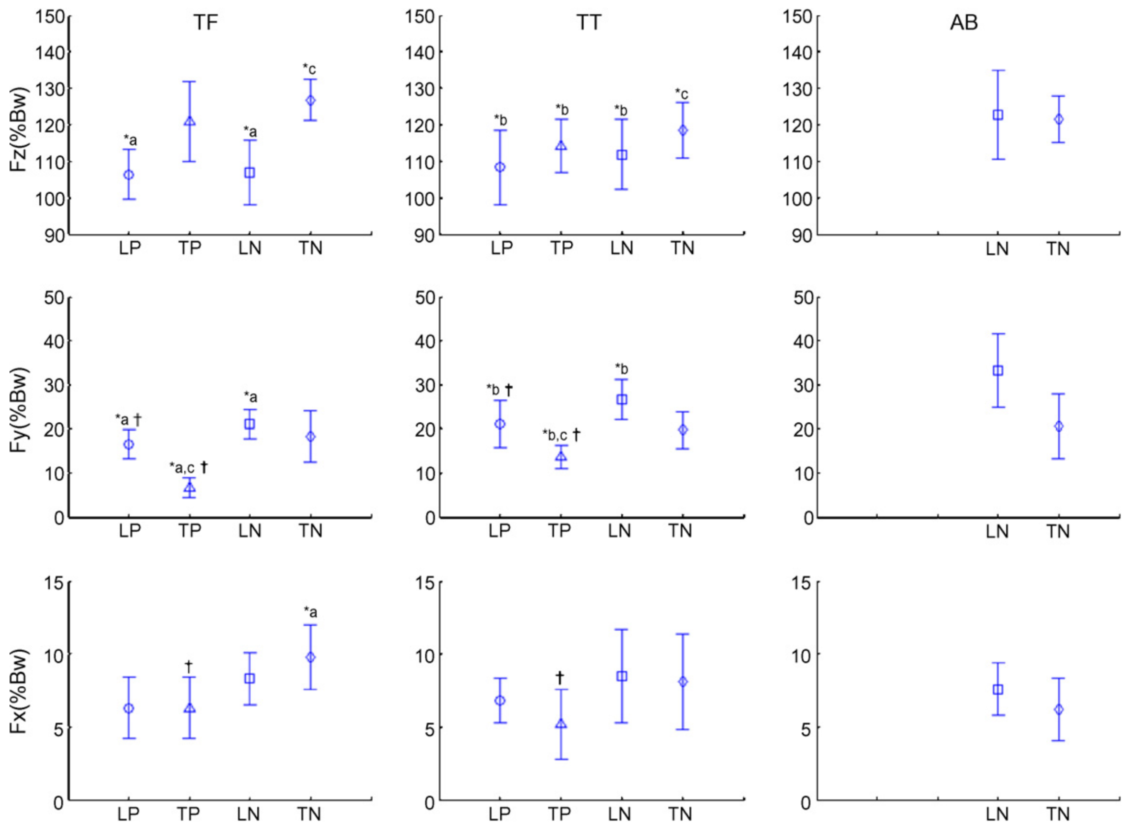

Fig. 4. Mean values and standard deviations of the GRF components in the vertical direction (Fz), the anteroposterior direction (Fy) and the mediolateral direction (Fx) of the leading prosthetic (LP), the trailing prosthetic (TP), the leading non-affected (LN) and the trailing non-affected (TN) limb condition in TF, TT and AB. Statistically significant *p*-values ($p \leq 0.05$) of between group differences are marked with *; *a for differences between AB and TF, *b for differences between AB and TT and *c for differences between TF and TT. Statistically significant *p*-values ($p \leq 0.05$) of differences between the prosthetic and non-affected limb within TF and TT are marked with †. Fz is positive in an upward direction, Fy in a backward direction and Fx the trailing limb direction.

prosthetic limb were more extended. Second, the lower Fy in the prosthetic limb was compensated by a longer period of force production. Fy is mainly executed in single-limb stance duration of the leading limb, which is similar to swing phase duration of trailing limb. In our study TF stood longer on their non-affected leading limb and were able to increase Fy in this way. Third, amputees lowered gait termination velocity, resulting in a decrease in the required Fy. We allowed the subjects to walk at their own self-selected velocity to pursue a true to nature observation. Consequently, the results may be influenced by gait velocity since joint angles and GRF depend on gait velocity [26–28]. Our hypothesis of larger production of Fy in the non-affected limb was not confirmed in the study.

Apart from slowing down the forward movement, the leading limb has to provide stability as well. In TF swing phase of the prosthetic limb was prolonged, thus TF spent more time in the single-limb stance on the non-affected limb. Literature has shown that amputees mainly experience difficulties in stability in single-limb stance on the prosthetic limb [14]. Consequently, a longer period of single-limb stance in the non-affected limb will not endanger balance control seriously. The larger COPx shift in amputees may be the result of decreased balance control. In leading with the prosthetic limb the COPx was increased in both amputee groups, whereas in the other limb condition COPx was only larger in TF. The larger COPx shift can also be caused by an increase in stride or stance width. COPx shift was most clearly seen in leading with the prosthetic limb during single-limb stance, which supports the hypothesis of reduced balance control in amputees being the main problem. Stance width among subjects was not standardized, because we chose to investigate self-selected gait termination.

Amputees also used an adjustment strategy to improve balance control. Amputees loaded the non-affected limb more than the prosthetic limb. Fx in the trailing non-affected limb was larger than in the prosthetic limb, resulting in a more lateral COPx shift in trailing with the non-affected limb. As soon as the non-affected limb was placed on the ground, amputees loaded their weight on this limb to enhance stability. The results of the Fz confirm the preference for weight-bearing on the non-affected limb in amputees; Fz in the non-affected limb was larger than in the prosthetic limb at the moment of trailing limb initial contact.

In addition, the preference for the use of the non-affected limb as leading limb in amputees may represent an adjustment strategy. Amputees profit from leading with

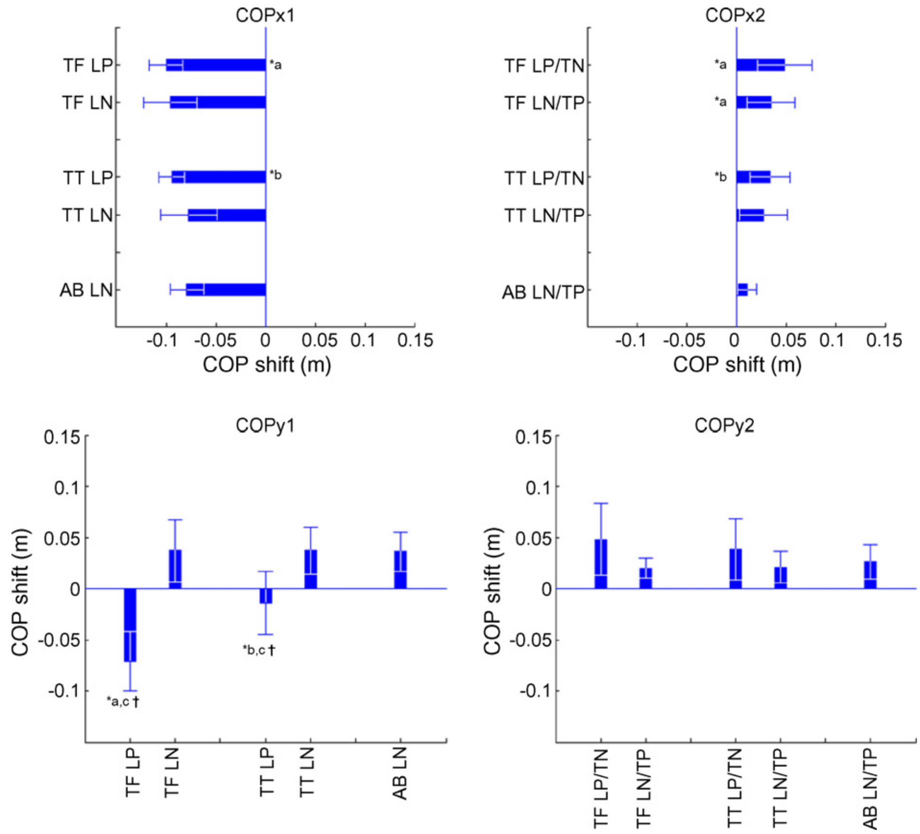

Fig. 5. Mean values and standard deviations of mediolateral COP shift in the leading limb direction (COPx1) and the trailing limb direction (COPx2), and of anteroposterior COPy shift in the leading limb direction (COPy1) and the trailing limb direction (COPy2) when leading with the prosthetic (LP) and the non-affected (LN) limb, and when trailing with the prosthetic (TP) and the non-affected (TN) limb in TF, TT and AB. Statistically significant p-values ($p \leq 0.05$) of between group differences are marked with *; *a for differences between AB and TF, *b for differences between AB and TT and *c for differences between TF and TT. Statistically significant p-values ($p \leq 0.05$) of differences between the prosthetic and non-affected limb within TF and TT are marked with a †. COPx is positive in the direction of the leading limb, COPy in the anterior direction.

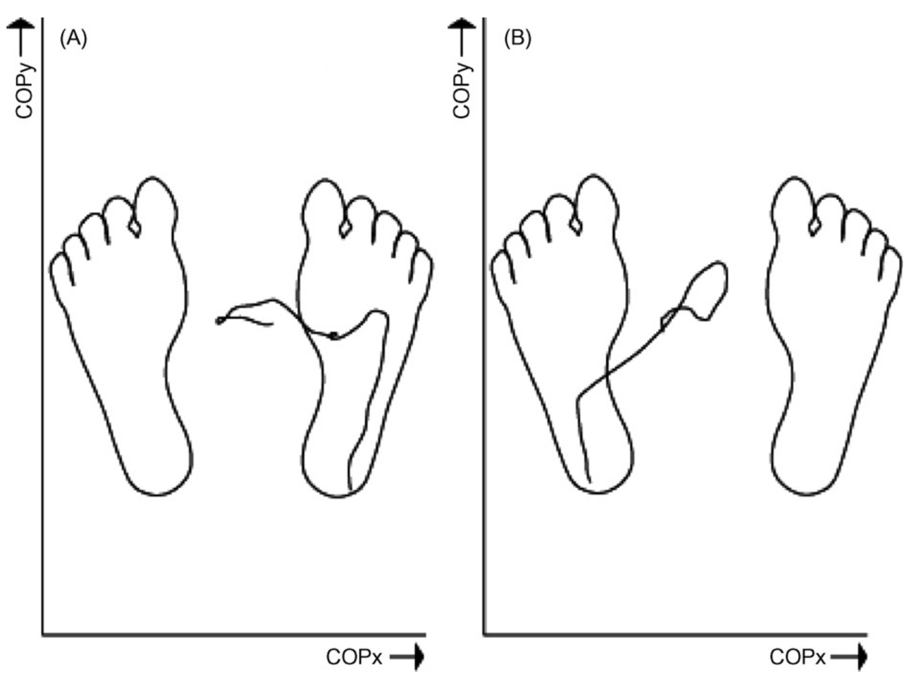

Fig. 6. Examples of trajectories of the COP in a subject of the TF group. (A) The COP trajectory during leading with the right non-affected limb. The COP on the leading side is shifted towards the forefoot. (B) The COP trajectory during leading with the left prosthetic limb. During single-limb stance the COP of the prosthetic limb does not move anteriorly.

the non-affected limb, because Fy is larger, COPy moves in front of the COM, and COPx shift is smaller.

However, in real life the choice of the leading limb in self-selected gait termination will often coincide with reaching the destination. When in daily life the stopping location is at an exact position, such as a door, chair, or wall, amputees will terminate gait with the limb that reaches that location first. Another possibility is that amputees adjust their step length prior to reaching the stopping location to emerge with the preferred limb. Otherwise, amputees only have a choice in leading limb when gait termination occurs at a self-selected place and time. Therefore, it is important to train gait termination during rehabilitation, but specific advice on leading limb preference is of minor importance.

A limitation of the present study was the lack of information on leg dominance. In most subjects the amputation was already performed years ago, and therefore we could not determine the leg dominance prior to the amputation in a reliable way. Another limitation was that only outcome variables in the last step were assessed. Although most deceleration occurs in this step [7,22,23], other studies have shown important adjustments in the trailing limb in the step prior to termination, namely a decrease in push-off GRF [5,8,9,11]. Finally, due to technical limitations the data of leading limb preference, temporal variables and joint angles were collected in different runs than the GRF and COP data. Since the walking pattern of the subjects was consistent, we assumed it was justified to analyse the data together.

4. Conclusion

AB adjust their gait pattern in gait termination by increasing the braking GRF and shifting the COPy anteriorly. In the prosthetic limb of both amputee groups Fy is decreased, COPx shift enlarged, and in leading with the prosthetic limb COP is not moved toward the forefoot. Amputees used several adjustment strategies to compensate for the limitations in function. They preferred leading with the non-affected limb, prolonged the production of Fy in the non-affected limb, decreased gait termination velocity and loaded more weight on their non-affected limb. It is important that amputees are trained in a gait termination task during rehabilitation. Leading with the non-affected limb is favourable for adequate deceleration and balance control, but in daily life not always applicable. In the future, technology that can assist in a more active role of the prosthetic foot and knee may ease gait termination in amputees.

Conflict of interest statement

Authors state that no conflicts of interest are present in the research.

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