Research Report

Stepping Asymmetry Among Individuals With Unilateral Transtibial Limb Loss Might Be Functional in Terms of Gait Stability

Laura Hak, Jaap H. van Dieën, Peter van der Wurff, Han Houdijk

Background. The asymmetry in step length in prosthetic gait is often seen as a detrimental effect of the impairment; however, this asymmetry also might be a functional compensation. An advantage of a smaller step length of the nonprosthetic leg, and specifically foot forward placement (FFP), might be that it will bring the center of mass closer to the base of support of the leading foot and thus increase the backward margin of stability (BW MoS).

Objective. The purpose of this study was to characterize differences in step length, FFP, and the concomitant difference in BW MoS between steps of the prosthetic and nonprosthetic legs (referred to as prosthetic and nonprosthetic steps, respectively) of people after transtibial amputation.

Design. This was an observational and cross-sectional study.

Methods. Ten people after transtibial amputation walked for 4 minutes on a self-paced treadmill. Step length and FFP were calculated at initial contact. The size of the BW MoS was calculated for the moment of initial contact and at the end of the double-support phase of gait.

Results. Step length (5.4%) and FFP (7.9%) were shorter for the nonprosthetic step than for the prosthetic step. The BW MoS at initial contact was larger for the nonprosthetic step, but because of a significant leg × gait event interaction effect, BW MoS did not differ significantly at the end of the double-support phase.

Limitations. All participants were relatively good walkers (score of E on the Special Interest Group in Amputee Medicine [SIGAM] scale).

Conclusions. The smaller step length and FFP of the nonprosthetic step help to create a larger BW MoS at initial contact for the nonprosthetic step compared with the prosthetic step. Hence, step length asymmetry in people after transtibial amputation might be seen as a functional compensation to preserve BW MoS during the double-support phase to cope with the limited push-off power of the prosthetic ankle.

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One of the characteristics of the gait pattern of people after transtibial amputation is an asymmetry in step length. Although the direction of the asymmetry in step length might vary across individuals, the step of the nonprosthetic leg, in which the healthy leg is the leading leg, is generally found to be shorter compared with the step with the prosthetic leg.[1–4] A commonly used explanation for a shorter step with the nonprosthetic leg is the reduced push-off capacity of the prosthetic leg.[4–6] In addition, prosthetic misalignment, stump pain, or discomfort might contribute to step length asymmetry.[7,8]

Gait asymmetry in people after transtibial amputation is often seen as a negative consequence of the disorder. However, although asymmetry may be detrimental from a cosmetic perspective, there is no evidence that step length asymmetry is necessarily detrimental from a functional point of view. A shorter nonprosthetic step could even be beneficial. Modeling studies and experimental studies with healthy controls and people after transtibial amputation[5,9,10] suggest that a shorter step length of the nonprosthetic leg might limit the metabolic cost of the step-to-step transition during the double-support phase of gait. Another possible benefit of a shorter step length of the nonprosthetic leg, which to our knowledge has not been considered explicitly before, could be the regulation of stability during walking.

A basic requirement of walking is that the center of mass (CoM) passes the dorsal border of the stance foot during each single-support phase; otherwise, forward progression will be interrupted, and a backward fall might occur if a recovery response, such as stepping back, fails. The risk of such a backward loss of balance can be quantified in terms of the distance between the kinematic state (position and velocity) of the CoM and the dorsal border of the base of support of the leading foot (Fig. 1).[11–13] In previous studies, we observed that both healthy people and people with a lower limb amputation increase this so-called backward margin of stability (BW MoS) when confronted with challenges of gait stability, despite the concomitant decrease of the margin of stability in a forward direction.[14–18] Hof and colleagues[19,20] provided an analytical expression for this concept in which this CoM state is quantified as the extrapolated center of mass (xCoM). Following this concept, increasing the forward velocity of the center of mass (vCoM) or decreasing step length (and more specifically the *foot forward placement* [FFP], defined as the distance between the leading foot and the CoM) will increase the BW MoS. A larger BW MoS implies that it is easier for the CoM to pass the posterior border of the base of support defined by the new stance leg during the following single-support phase and will decrease the risk of a backward loss of balance.

In a previous study by Hak et al[17] on the regulation of gait stability, it appeared that people after transtibial amputation and healthy people both increased their BW MoS in response to continuous balance perturbations, by maintaining their walking speed and decreasing their step length. However, during both unperturbed and perturbed walking, BW MoS was smaller for the people after transtibial amputation because of their lower walking speed compared with healthy controls.[17] That study, however, focused on average changes in step length, speed, and BW MoS over affected and unaffected steps. It is unknown whether and how differences in step length and CoM velocity between sides influence the BW MoS for the affected and unaffected steps and, therefore, whether step length asymmetry could be understood as a mechanism to preserve BW MoS.

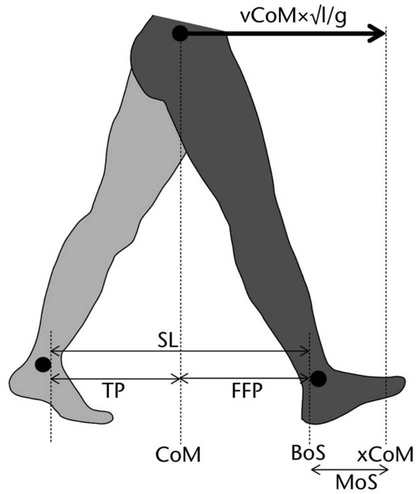

Figure 1.
Schematic representation of the definition of the backward margin of stability (BW MoS), foot forward placement (FFP), and trunk progression (TP). The BW MoS is defined as the distance in the backward direction between the extrapolated center of mass (xCoM; dotted line in the right panel) and the dorsal border of the base of support (BoS) of the leading foot. The xCoM is calculated as the position of the center of mass (CoM) plus its velocity (vCoM) times a factor $\sqrt{(l/g)}$, with l being the length of the pendulum (for which often the leg length is used) and g the acceleration of gravity. The BW MoS can be increased by decreasing FFP or by increasing vCoM. SL=step length, MoS=margin of stability.

Possibly, shortening the nonprosthetic step length can be regarded as a proper strategy to increase the BW MoS for this step. The reduced push-off capacity of the prosthetic leg often results in a limited capacity to deliver mechanical power with the prosthetic leg during the push-off.[5,21–24] This diminished mechanical power might result in a lower vCoM, and thus a smaller BW MoS, during the step-to-step transition from the prosthetic leg to the nonprosthetic leg. A shorter nonprosthetic step, therefore, might serve the purpose of compensating for this detrimental effect of lower vCoM on the BW MoS, as it brings the leading foot closer to the CoM at initial contact, which increases BW MoS. Note, however, that this compensatory effect will only be true when this asymmetry in step length will manifest itself in an asymmetry in FFP, and not in the distance between the trailing foot and the CoM (in this study, called the trunk progression [TP]), because the latter will not influence the size of the BW MoS (Fig. 1).[13]

The purpose of this observational study was to characterize differences in step length and FFP between prosthetic and nonprosthetic steps of people after transtibial amputation and the concomitant difference in BW MoS between these steps. We hypothesized that people after transtibial amputation walk with a shorter intact step length and shorter FFP of the intact step. We expected that vCoM will be reduced during the step-to-step transition from the prosthetic leg to the intact leg as a result of the lack of push-off capacity of the prosthetic leg. Finally, we hypothesized that a smaller FFP for the nonprosthetic step will compensate for the reduction in vCoM to preserve BW MoS.

Method

Participants

Ten adults with a unilateral transtibial prosthesis participated in this observational study. Participants were recruited from the patient population of the National Military Rehabilitation Center Aardenburg, Doorn, the Netherlands. All participants gave their written informed consent in accordance with university policy.

Equipment

During the experiment, the Computer Assisted Rehabilitation Environment (CAREN, Motek Medical BV, Amsterdam, the Netherlands) was used. Participants walked on a treadmill within a virtual environment to create an optical flow during the walking trials. Twelve high-resolution infrared cameras (Vicon, Oxford, United Kingdom) were used to capture kinematic data of 16 reflective markers attached to the pelvis and the lower extremities (lower body plug-in-gait). The treadmill was used in the self-paced mode, which allowed participants to walk at their comfortable walking speed and allowed spontaneous variations in walking speed. This self-paced mode was used to approach overground walking more closely while recording a large number of steps within a single trial. In the self-paced mode, the motor of the treadmill was servo-controlled by a real-time algorithm that took into account the pelvis position in the anteroposterior (AP) direction, as measured by the markers attached to the pelvis, and a reference position on the treadmill, corresponding to the AP midline of the treadmill. A safety harness system suspended overhead prevented the participants from falling but did not provide weight support.

Protocol

Before the experimental trial, participants performed 5 warm-up trials of 3 minutes each to become familiar with walking (self-paced) on a treadmill and with the virtual environment. The experimental trial consisted of 4 minutes walking at self-paced walking speed. During the first 30 seconds of this trial, participants walked at a fixed walking speed, which was substantially lower than their comfortable walking speed, determined during one of the warm-up trials. After 30 seconds, the self-paced mode of the treadmill was started, after which the participants could walk at their comfortable walking speed for the remaining 3.5 minutes.

Data Collection

Data collection took place for the final 3 minutes of the experimental trial. The speed of the treadmill was recorded, and kinematic data of markers attached at the lateral malleoli of the ankles and the pelvis (left and right anterior superior iliac

The Bottom Line

What do we already know about this topic?

People who walk with a lower leg prosthesis often walk with an asymmetry in step length. The step with the prosthesis is usually longer compared with the step with the intact leg.

What new information does this study offer?

This asymmetry in step length is not necessarily a detrimental consequence of the amputation, but might be a functional compensation to regulate gait stability because of limited prosthetic push-off.

If you're a patient or a caregiver, what might these findings mean for you?

Based on the results of this study, the physical therapist might not make regaining gait symmetry the primary aim of prosthetic training.

Table 1.
Characteristics of Participants With Transtibial Amputation

Participant No.	Sex	Age (y)	Weight (kg)	Height (m)	Cause of Amputation	Time Since Amputation (mo)	Foot Type (Name)[a]	Socket Design[b]
1	Female	32	68	1.60	Other	144	Axtion	TSB
2	Male	45	90	1.83	Trauma	48	Elite VT	PTB[c]
3	Male	42	80	1.73	Trauma	96	1C40	PTB[c]
4	Male	21	80	1.88	Trauma	24	Variflex EVO	PTB[c]
5	Male	33	104	1.96	Trauma	27	Variflex EVO	PTB[c]
6	Male	38	86	1.98	Trauma	17	Fusion	PTB[c]
7	Male	66	86	1.76	Dysvascular	12	Celsus	PTB[c]
8	Male	25	84	1.92	Trauma	9	1C40	PTB[c]
9	Male	59	98	1.82	Trauma	120	Propiofoot	PTB[c]
10	Male	27	95	1.82	Trauma	34	1C40	PTB[c]

[a] Prosthetic foot manufacturers: Axtion and 1C40 (Otto Bock, Duderstadt, Germany), Elite VT (Endlolite, Miamisburg, Ohio), Variflex EVO (Ossur, Reykjavik, Iceland), Fusion (Willowwood, Mt Sterling, Ohio), Celsus (College Park, Warren, Michigan), Propiofoot (Ossur, Reykjavik, Iceland).
[b] TSB=total surface bearing, PTB=patellar tendon bearing.
[c] Pin fixation.

spines and left and right posterior superior iliac spines) were collected with the Vicon system. The sample rate of data collection was 120 samples per second. Data were low-pass filtered with a fourth-order bidirectional Butterworth filter with a cutoff frequency of 10 Hz.

Data Analysis

Step length, FFP, and TP. *Step length* was defined as the AP distance between the ankle markers at the instant of heel contact, where heel contacts were detected as the local maxima of the position of the ankle markers in the AP direction. The body's CoM position was approximated by the center of the polygon described by the 4 markers attached to the pelvis.[25] Foot forward placement and TP were respectively calculated as the fore-aft distance between the ankle marker of the leading foot and the CoM and the ankle marker of the trailing foot and the CoM at initial contact. These definitions for FFP and TP were derived from studies by Roerdink and colleagues.[6,26] However, we chose to use a slightly different definition of TP to be able to evaluate asymmetries in FFP and TP independently of each other. In the studies by Roerdink and colleagues,[6,26] TP was defined as the distance traveled by the CoM between subsequent steps, which is partly determined by the length of the FFP of the previous step. Asymmetry indexes were calculated for step length, FFP, and TP as: (value NP/((value NP + value P)/2)) × 100%, where NP represents the nonprosthetic step and P represents the prosthetic step.

vCom. The vCoM was calculated as the first derivative of the position of the center of the pelvis relative to the global reference frame plus the instantaneous velocity of the treadmill. This calculation was done for both initial contact and the moment of toe-off of the contralateral leg. Moments of toe-off were detected as the local minima of the position of the ankle markers in the AP direction.

BW MoS. To calculate the BW MoS, a method derived from the procedure introduced by Hof et al[19] was used. In the current study, the xCoM was estimated by taking the CoM position plus its velocity times a factor $\sqrt{(l/g)}$, with l being the maximal height of the estimated CoM and g being the acceleration of gravity. The BW MoS was calculated as the position of the xCoM minus the position of the lateral malleolus of the ankle of the leading foot (representing the border of the base of support) for the moment of initial contact and the moment of toe-off of the contralateral foot in AP direction. Although basically similar, our method differs from that of Hof et al,[19] who used forceplate data instead of kinematic data[25,27] for calculating the xCoM and BW MoS. Additionally, by using the lateral malleolus of the ankle instead of the heel to define the posterior border of the leading foot, we have made a slight, but systematic, underestimation of the BW MoS.

Statistical Design

To test whether differences between legs (nonprosthetic and prosthetic) in step length, FFP, and TP at initial contact were significant, paired-samples *t* tests were used. To investigate whether BW MoS and vCoM differed between prosthetic and

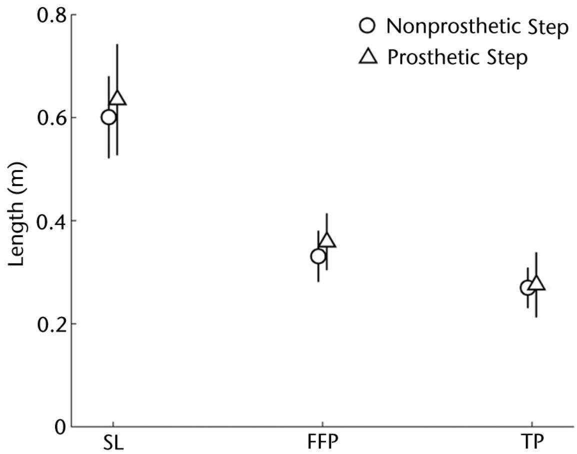

Figure 2.
The average and between-subjects standard deviation (n=10) of the prosthetic (O) and nonprosthetic (Δ) step length (SL), foot forward placement (FFP), and trunk progression (TP) at initial contact.

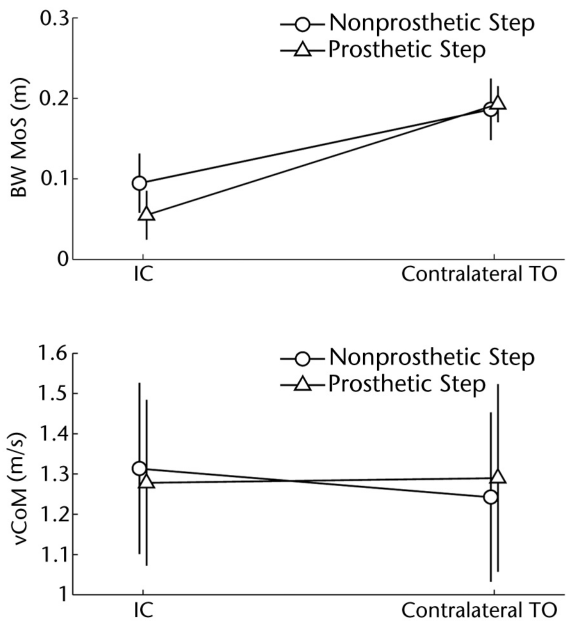

Figure 3.
The average and between-subjects standard deviation (n=10) of the backward margin of stability (BW MoS; upper graph) and the velocity of the center of mass (vCoM; lower graph) at initial contact (IC) and contralateral toe-off (TO) of the prosthetic (O) and nonprosthetic (Δ) steps.

nonprosthetic steps (factor: leg) and whether these variables changed during the double-support phase (ie, between initial contact and toe-off of the contralateral foot) (factor: gait event), 2 × 2 factorial analyses of variance were performed, with the 2 legs and the 2 gait events as within-subject variables. *P* values less than .05 were considered significant. When an interaction effect was present, paired-samples *t* tests with a Bonferroni correction (critical *P* value=.0125) were used to investigate for each leg separately whether the considered variable changed between both gait events and to test for each gait event separately regardless of whether there was a difference between legs. Statistical analyses were performed using SPSS 17.0 (SPSS Inc, Chicago, Illinois).

Role of the Funding Source

We thank Motek Medical BV for their financial support for this study.

Results

Characteristics of the people after transtibial amputation who participated in the study are reported in Table 1. The side of amputation was equally divided between left and right. All participants used their own prosthesis and were able to walk in daily life without any walking device for at least 30 minutes. All participants were classified with a minimum score of E (able to walk more than 50 m) on the Special Interest Group in Amputee Medicine (SIGAM) scale, independent of walking aids, except occasionally for confidence or to improve confidence in adverse terrain or weather).[28]

All participants were able to complete the walking trial on the treadmill and walked on average at a speed of 1.22 m/s (SD: 0.22). Step length (P=.030, t=2.581, df=9) and FFP (P=.015, t=3.006, df=9) were significantly smaller for the nonprosthetic step compared with the prosthetic step (5.4% and 7.9%, respectively), whereas TP (P=.636, t=0.490, df=9) did not differ between legs (Fig. 2). Asymmetry indexes for step length, FFP, and TP were 97.2, 95.9, and 98.9, respectively.

Figure 3 shows the BW MoS (upper panel) and vCoM (lower panel) at initial contact and subsequent contralateral toe-off of the prosthetic and nonprosthetic steps. There were significant main effects of leg (P=.018; F=8.296; df=1,9) and gait event (P=.001; F=395.367; df=1,9) on the size of the BW MoS, with a larger BW MoS for the nonprosthetic step and an increase of the BW MoS between initial contact and contralateral toe-off. In addition, there was a significant leg × gait event interaction effect (P=.009; F=10.844; df=1,9). Post hoc analyses (Tab. 2) showed that BW MoS increased for both legs, but this increase was larger between initial contact and contralateral toe-off for the prosthetic step compared with the nonprosthetic step. Additionally, post hoc analyses showed that the difference in BW MoS between legs was present only for the moment of initial contact.

No significant main effects of leg (P=.668; F=0.196; df=1,9) and gait event (P=.116; F=3.023; df=1,9) on vCoM were found, but there was a significant leg × gait event interaction effect (P=.049; F=5.194; df=1,9). Post hoc tests (Tab. 2) revealed a decrease of vCoM during the double-support phase of the nonprosthetic step, whereas vCoM did not change significantly for the prosthetic step. At initial contact, there was a trend toward a higher vCoM for the nonprosthetic step compared with the prosthetic step. At contralateral toe-off, vCoM was smaller for the nonprosthetic step, but this measurement also did not reach the level of significance.

Table 2.
Analysis of Post Hoc Statistics[a]

Parameter	Between Legs at IC	Between Legs at Contralateral TO	Between IC and Contralateral TO Prosthetic Leg	Between IC and Contralateral TO Nonprosthetic Leg
BW MoS	t=3.723 P=.005[b] df=9	t=−0.883 P=.400 df=9	t=18.151 P<.001[b] df=9	t=8.875 P<.001[b] df=9
vCoM	t=−2.229 P=.053 df=9	t=1.719 P=.120 df=9	t=−0.426 P=.680 df=9	t=3.679 P=.005[b] df=9

[a] IC=initial contact, TO=toe-off, BW MoS=backward margin of stability, vCoM=velocity of the center of mass.
[b] Significant at the .0125 level.

Discussion

The purpose of the current study was to investigate whether differences in step length between prosthetic and nonprosthetic steps, and more specifically in FFP, in prosthetic gait are compatible with a strategy to regulate gait stability in terms of the BW MoS. Step length, on average, was shorter for the nonprosthetic step, which is the step in which the healthy leg was leading. A remarkable result is that the step length asymmetry found in the current study was entirely due to an asymmetry in FFP. At initial contact, the BW MoS was larger for the nonprosthetic step compared with the prosthetic step, but this difference had disappeared at the end of the double-support phase. The average vCoM did not differ between steps, but in contrast to the prosthetic step, vCoM decreased significantly during the double-support phase following the nonprosthetic step.

The results of the current study support our hypothesis that a shorter nonprosthetic step length in people after transtibial amputation contributes to a larger BW MoS at initial contact of the nonprosthetic step. This shorter nonprosthetic step length seems to compensate for the limited capacity of people after transtibial amputation to regulate the size of the BW MoS during the double-support phase following the nonprosthetic step. During this double-support phase, the lack of ankle push-off of the prosthetic leg causes a decrease in the vCoM, which limits the increase of the BW MoS during the double-support phase. A smaller nonprosthetic FFP at initial contact may be necessary to compensate for this limited increase in BW MoS during the following double-support phase and thus decrease the risk of interruption of forward progression and possibly even falling backward after contralateral toe-off. In Figure 4, a schematic overview is given of the effect of the decrease of the nonprosthetic step length on the BW MoS, based on the results of the current study (Fig. 4, II), compared with the BW MoS of the prosthetic step (Fig. 4, I) and compared with the possible consequence of an absence of a shorter nonprosthetic step length on the BW MoS (Fig. 4, III).

The clinical implication of the present results is that regaining a symmetrical step length during the rehabilitation of people after transtibial amputation might result in difficulties with the regulation of BW MoS during walking. Step length symmetry, therefore, should not necessarily be a primary goal in rehabilitation of people after amputation. Pursuing step length symmetry seems to be justified only when sufficient push-off power in the prosthetic leg can be guaranteed to prevent a drop in

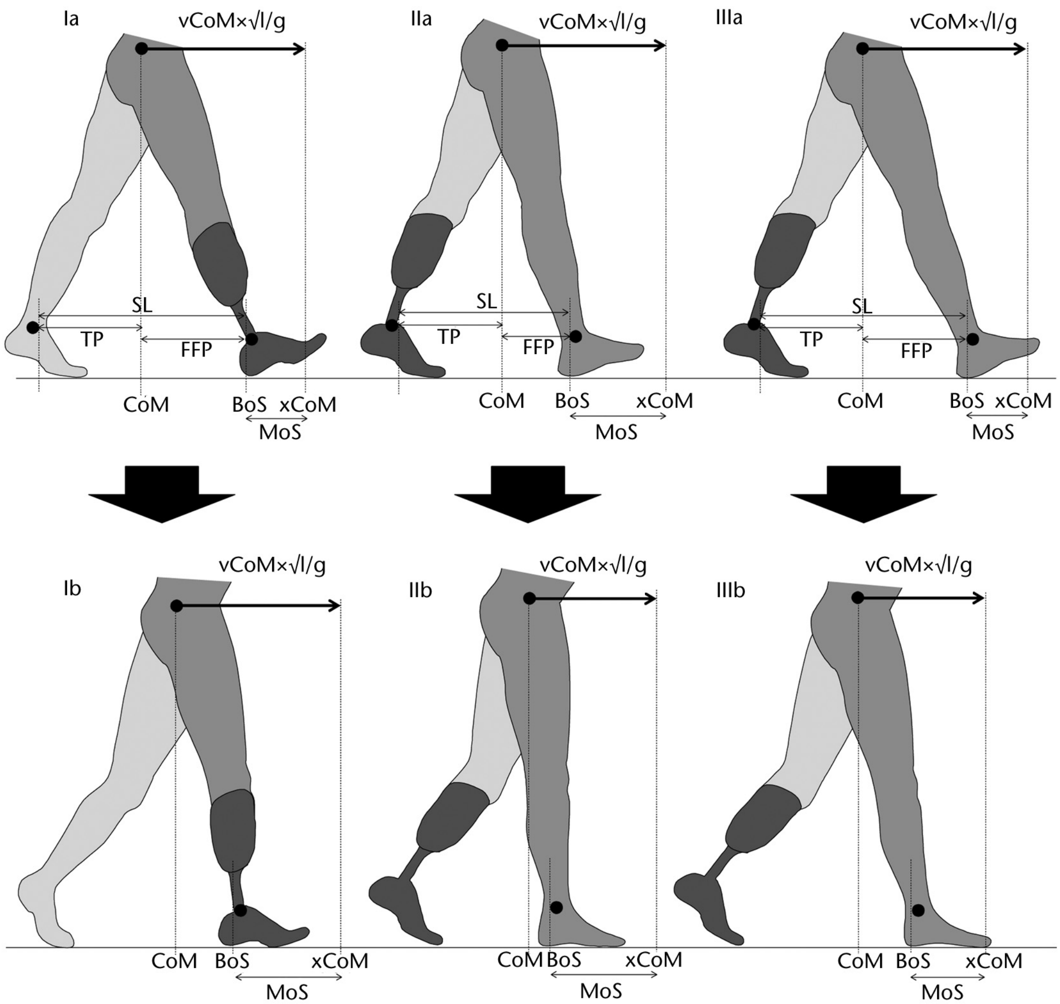

Figure 4.

Schematic overview of possibilities to regulate the backward margin of stability (BW MoS) of the nonprosthetic step (II and III) compared with the prosthetic step (I). For the prosthetic step, the BW MoS increased during the double-support phase as a result of a change in center of mass (CoM) state during this phase (**Ia** vs **Ib**). **IIa**: A shorter nonprosthetic step due to a smaller foot forward placement resulted in a larger BW MoS at initial contact (compared with the prosthetic step in I). **IIb:** A decrease of velocity of the center of mass (vCoM) during the double-support phase of the nonprosthetic step caused a smaller increase of the BW MoS compared with the prosthetic step, resulting in a BW MoS that was about the same size as the BW MoS at the end of the double-support phase of the prosthetic step (compared with the prosthetic step in Ia). **IIIa**: A nonprosthetic step with the same step length as the prosthetic step resulted in a BW MoS of comparable size as for the prosthetic step at initial contact (I). **IIIb**: A decrease of vCoM during the double-support phase of the nonprosthetic step caused a smaller increase of the BW MoS compared with the prosthetic step, resulting in a smaller BW MoS compared with the BW MoS at the end of the double-support phase of the prosthetic step (compared with the prosthetic step in Ia). SL=step length, TP=trunk progression, FFP=forward foot placement, BoS=base of support, xCoM=extrapolated center of mass, MoS=margin of stability.

vCoM during the gait cycle. To restore gait symmetry while preserving BW MoS, the push-off capacity of prosthetic feet should be further enhanced. Currently, advanced dynamic feet are under development that might fulfill this purpose.[29,30] This conclusion, however, is based on an observational study of prosthetic gait. Therefore, some critical reflections are warranted regarding the necessity to regulate BW MoS in the first place, alternative ways to regulate BW MoS, and potential future studies to corroborate our conclusion. These reflections will be discussed below.

Whether the risk of a backward loss of balance is indeed a problem in walking with a lower limb prosthesis can be debated. However, the results of our previous studies indicate that people after transtibial amputation and healthy people increased their MoS in the BW direction instead of forward direction in response to mediolateral platform perturbation.[17,18] Based on these results, we concluded that people prefer to minimize the risk on a backward loss of balance, which would require a reversal of the periodic leg movement (stepping backward) to regain balance instead of a forward loss of balance that requires only a relatively small adaptation of the next steps, such as a temporary increase in step length, to recover.[20] In line with this discussion, one could argue how large the BW MoS should minimally be for stable walking in daily life. A smaller BW MoS at the end of the step-to-step transition from the nonprosthetic step to the prosthetic step might not be problematic as long as it is positive. However, from our previous study, it appeared that at initial contact the BW MoS for the people after amputation was already smaller compared with healthy people.[17] Therefore, a limited increase of the BW MoS during the double-support phase following nonprosthetic step initial contact might result in a safety margin too small to withstand perturbations occurring in daily life, which might cause a backward loss of balance or obstruct forward progression.

In addition to adapting step length (or FFP), people with a lower limb amputation also could use other strategies to preserve BW MoS while walking with symmetrical step length. A possible alternative strategy to enhance BW MoS might be the use of a "controlled falling" strategy in which potential energy is transformed into kinetic energy[24] or through a rotation of the trunk or the arms with respect to the CoM, which will cause a counter-rotation of the whole body.[31] Both strategies could result in an increase of the vCoM. We indeed observed a somewhat higher vCoM at initial contact of the intact limb, indicating that these strategies were partly used, but vCoM did still decline during the subsequent double-support phase. This finding indicates that the use of these strategies was not sufficient to limit the decline in vCoM and preserve BW MoS. Potentially, our participants could have used these alternative strategies to a larger extent if they were forced to walk with a symmetric step length. However, it should be realized that that would also result in gait asymmetry, albeit of a different kind (ie, asymmetric trunk motion), and potentially might impose additional penalties on, for instance, gait economy.

The conclusions drawn in the current study are based on an observation of prosthetic gait. Therefore, it remains difficult to conclude whether the regulation of the BW MoS is the primary reason for the asymmetry in step length. There might be multiple other explanations for gait asymmetry in people after transtibial amputation, such as the minimization of the metabolic cost during the step-to-step transition[5] and pain or an incorrect alignment of the prosthesis.[7,8] Therefore, the apparent effects of a shorter nonprosthetic step length on the MoS also could be a coincidental side effect. To further support our hypothesis that step length asymmetry enhances BW MoS, an experimental approach should be taken. For instance, the use of manipulations that force people after transtibial amputation to walk symmetrically or a comparison between walking with solid versus dynamic feet (that partly restore push-off capacity) might be used to establish a causal effect between step length differences and BW MoS.

Finally, when interpreting the results of the present study, it has to be taken into account that the people after transtibial amputation who participated in this study were all relatively young and generally good walkers (minimum score of E on the SIGAM scale[28]). All participants were able to walk without a device in daily life for at least 30 minutes consecutively and walked at a relatively high walking speed (in comparison, an average walking speed of 1.43 m/s was found in a group of age-matched controls while walking on the self-paced treadmill[17]). All people after transtibial amputation, except 2, underwent amputation following trauma. In this group, the overall walking ability in general is higher than in people after amputation due to vascular disorders.[32] The results of this study also might be applicable for people after transfemoral amputation or people with other unilateral impairments (eg, caused by a stroke); however, generalization of the results needs to be confirmed in these specific groups. It should be noted that there was a lot of heterogeneity in the energy storage and return capacities of the prosthetic feet used by the participants. Despite this heterogeneity, we have found structural differences between the prosthetic step and the nonprosthetic step. However, it can be presumed from our results that increasing push-off capacity of the prosthetic foot and ankle will reduce the decrease in CoM velocity during the step-to-step transition and preserve BW MoS without the need for a shorter step. Because of the relatively small sample size, we could not perform a valid post hoc comparison of the effect of foot type. A future analysis of the effect of foot type will be of interest.

In conclusion, step length asymmetry with a smaller FFP of the nonprosthetic step compared with the prosthetic step can be observed in people after transtibial amputation. This asymmetry preserved the BW MoS at the end of the double-support phase after the nonprosthetic step, despite a reduction of CoM velocity during the double-support phase. Consequently, step length asymmetry in prosthetic gait could potentially be regarded as a functional adjustment to preserve sufficient BW MoS and prevent a backward fall in the presence of limited push-off ability of the prosthetic leg. The results of this study illustrate that the asymmetry in the gait pattern for people after transtibial amputation is not necessarily a detrimental effect of the impairment but could be beneficial in the regulation of gait stability. This finding should be considered in gait training for people after transtibial amputation.

All authors provided concept/idea/research design, writing, and project management. Dr Hak and Dr van der Wurff provided data collection and study participants. Dr Hak provided data analysis. Dr van der Wurff provided facilities/equipment and institutional liaisons. Professor van Dieën and Dr Houdijk provided fund procurement and consultation (including review of manuscript before submission).

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This study was approved by the Medical Ethical Committee of VU Medical Centre Amsterdam (registration number: NL35402.029.11).

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