Balance recovery after an evoked forward fall in unilateral transtibial amputees

Carolin Curtze a,b,*, At L. Hof a,c,d, Bert Otten c,d, Klaas Postema a,b,d

Abstract

Falls are a common and potentially dangerous event, especially in amputees. In this study, we compared the mechanisms of balance recovery of 17 unilateral transtibial amputees and 17 matched able-bodied controls after being released from a forward-inclined orientation of 10%. Kinematic analysis revealed statistically significant differences in response time and knee flexion at heel-strike between both groups. However, there were no statistically significant differences in step length of the leading and trailing limb, swing time of the leading limb, and maximal knee flexion during swing.

In the amputees, we found spatial and temporal differences when recovering with the sound versus prosthetic limb first. When leading with the prosthetic limb, they responded faster and also the interval between heel-strike of the leading and trailing limb was shorter. Furthermore, amputees made a longer step and showed less knee flexion at heel-strike when leading with the prosthetic limb. Interestingly, amputees as a group had no specific limb preference, prosthetic or sound, to recover after a forward fall, despite the asymmetry in their locomotor system. Analyses of dynamic stability (extrapolated center of mass) revealed that the amputees were equally efficient in recovering from an impending fall as controls, irrespective whether they lead with their prosthetic or sound limb. We suggest that in amputee rehabilitation, balance recovery after a fall should be trained with both sides, as this can increase confidence in fall-prone situations.

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1. Introduction

In a rapidly ageing society, falls are a particular issue of concern for public health. Some patient populations have an even higher incidence of falls. The annual fall incidence, of one or more falls, in lower limb amputees is approximately 50%, while their elderly able-bodied community-dwelling peers have an incidence of 30–40% [1,2]. While falls in the elderly population have received considerable attention, little is known about the underlying mechanisms of falls and successful balance recovery in lower limb amputees.

A popular approach to study balance recovery after a forward fall without compromising safety is the tether-release method [3,4]; for a review see Hasio-Wechsler [4].

In the event of a forward fall, the center of mass (CoM) gains velocity, away from the base of support. The impending fall can only be prevented by a rapid and spatially well-directed stepping (or reaching) movement. The accurate, time dependent, step length can be predicted by the concept of the "extrapolated center of mass" (XcoM) [7–9]. This concept is based on the inverted pendulum model of balance and allows to define stability in dynamic situations [8,9].

The purpose of this study is to provide an in depth analysis of the mechanisms of balance recovery in lower limb amputees. Furthermore, we hypothesized that unilateral transtibial amputees compensate for the asymmetry in their locomotor system by stepping with their sound limb first, as they lack active control of the ankle joint when stepping with their prosthetic limb first.

2. Methods

2.1. Participants

A group of 17 male unilateral transtibial amputees were included in the study. Their mean age was 55.2 (±9.2) years and they had a height of 1.84 (±0.07) m, and a body weight of 85.1 (±10.7) kg. The mean time since amputation was 12.6 (±13.7) years. Twelve participants had undergone an amputation of their left limb, and five of their right limb. The most frequent reason of amputation was trauma (12), followed by...
vascular disease (4), and limb deficiency (1). The amputees were all experienced walkers, who used their prosthesis on a daily basis. The control group consisted of 17 healthy male participants matched in age (55.0 ± 10.3 years), height (1.85 ± 0.05 m) and weight (87.1 ± 9.1 kg).

2.2. Apparatus

The ground reaction forces during standing in a forward-inclined orientation were measured by an AMTI force plate, sampled at a rate of 1000 Hz (Fig. 1). A second force plate in front of the participant was used in some of the trials only, when a participant happened to recover balance with both feet on this second force plate. One of these trials is presented in the results section for illustration purposes. To record full-body kinematics, 35 reflective markers were attached to the participant’s anatomical bony landmarks as specified in the Vicon Plug-in Gait full-body model. On the prosthetic limb the markers were placed at the corresponding positions. The reflective markers were tracked by an eight-camera Vicon motion capture system at a sampling rate of 100 Hz. Anthropometric measurements were taken for each individual according to the Vicon requirements and fed into the model. The lean-control cable was equipped with an electromagnet as release mechanism. The participant’s instantaneous lean angle was determined by means of a force transducer. Finally, all measurement data were further processed using Matlab.

2.3. Procedure

To evoke a forward fall, participants were suddenly released from a fixed forward-inclined orientation of 10% (±6°; Fig. 1). The participants were held back by a cable fixed to a full-body safety harness. The magnitude of the forward lean angle was controlled by adjusting the lean-control cable length until the force transducer attached to the cable indicated that it supported 10% of the participant’s body weight. The forward fall was initiated by the experimenter by pressing a button releasing the electromagnet. The participants were verbally instructed to prevent themselves from falling. No specifications were given regarding a desired balance recovery technique. Participants were given one practice trial. After three trials they were requested to initiate the recovery by stepping with the other, non-preferred limb first.

To prevent the participants from falling to the ground in the event of failed recovery, they were secured by a safety rope attached between the harness and the ceiling.

Prior to testing, all amputees completed the activities-specific balance confidence (ABC) scale [10–12], a self-efficacy measure assessing balance confidence across 16 specific activities on an analogue scale (0–100%).

The experimental protocol was approved by the local Medical Ethics Committee. All participants gave their informed consent.

2.4. Data analysis

We determined the leading limb preference (sound/prosthetic, left/right) from the first three trials of the fall experiment. Furthermore, trials were classified as single stepping, if the length of the trailing step did not exceed the length of the leading step by more than half a foot length. All other trials were classified as multiple stepping. The step length was normalized by the leg length resulting in a dimensionless number [13].

Fig. 1. Experimental setup. The participant was held in a forward-inclined orientation (10%). The force transducer was used to determine the magnitude of forward-inclination which could be adjusted by changing the length of the cable. The participant was positioned on a force plate. A safety rope was attached to the full-body harness. The electromagnet was used as release mechanism, which could be triggered by a switch ceasing the current. After release the center of mass (CoM) gained velocity, a ground reaction force (GRF) acted on the CoM (real coordinates).
Table 1

<table>
<thead>
<tr>
<th>Group</th>
<th>Spatial and temporal characteristics.</th>
<th>Controls (n = 9)</th>
<th>Amputees (n = 17)</th>
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<tbody>
<tr>
<td></td>
<td>Controls</td>
<td>Amputees 1</td>
<td>Amputees 2</td>
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<td></td>
<td>Left (n = 5)</td>
<td>Prosthetic</td>
<td>Sound</td>
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<td>Right (n = 4)</td>
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<td>Leading</td>
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<td></td>
<td>Step length (leading)</td>
<td>Prosthetic</td>
<td>Sound</td>
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<td>Response time (ms)</td>
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<td>Swing time (ms)</td>
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<td>Maximal knee flexion during swing</td>
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<td>Knee flexion at heel-strike</td>
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<td></td>
<td>Heel-strike interval (ms)</td>
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<tr>
<td>Response time</td>
<td>196 (22)</td>
<td>238 (21)</td>
<td>273 (24)</td>
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<tr>
<td>Swing time</td>
<td>260 (18)</td>
<td>316 (20)</td>
<td>341 (28)</td>
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<td>Maximal knee</td>
<td>30.8 (6.4)</td>
<td>28.1 (8.7)</td>
<td>27.4 (10.8)</td>
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<td>flexion during</td>
<td>20.2 (6.4)</td>
<td>20.3 (6.4)</td>
<td>20.5 (6.4)</td>
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<td>swing</td>
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<tr>
<td>Heel-strike</td>
<td>238 (40)</td>
<td>243 (10)</td>
<td>239 (31)</td>
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<td>interval (ms)</td>
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<td>Mean (S.D.)</td>
<td>220 (40)</td>
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<td>* p &lt; 0.05</td>
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<td>** p &lt; 0.01</td>
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The following temporal characteristics were determined: (1) response time (ms), defined as the interval between release \( t_0 \) and toe-off of the leading limb, (2) swing time (ms), defined as the interval between toe-off and heel-strike of the leading limb, and (3) heel-strike interval (ms), defined as the interval between heel-strike of the leading and trailing limb. The different events were estimated form the force transducer, the force plate data and the marker data of the heel and toe. Additionally, the maximal knee flexion of the leading limb during swing and the knee flexion at heel-strike were determined from the kinematic data.

The XCoM [8] was used to determine the function of the leading limb in braking the forward fall. Here, the forward position of the XCoM \( (\xi) \) is defined as

\[
\xi = x - \frac{v}{a_0}
\]

where \( l \) is the effective pendulum length (trochanteric height times 1.24) [14]. When the CoP is placed on the XCoM a stable posture will result. When the CoP is placed at some distance from the XCoM, the XCoM will move away from the CoP.

In a successful trial the CoP under the leading limb is positioned slightly beyond the XCoM. As a result the XCoM will start to move backward, the CoM movement will slow down and the fall will eventually be arrested. When balance will be successfully restored with the leading limb the XCoM thus shows a maximum, i.e. its first derivative is zero. Stability can be regained either with the leading limb, before heel-strike of the trailing limb, or after. Each trial was classified accordingly.

2.5. Statistical analysis

Pearson’s chi-square test was used to control for differences in leading limb preference in the group of amputees (prosthetic, sound) and controls (left, right). Subsequently, a one-way ANOVA was performed to test if amputees with prosthetic and sound limb leading preference differed in balance confidence.

For each leading limb condition the mean values of the spatial and temporal characteristics were calculated (Table 1). Separate two-way ANOVAs for repeated measures with leading limb (prosthetic/left, sound/right) as within-subject factor leading limb preference (prosthetic/left, sound/right) as between-subject factor were run on the different outcome parameters, first for the group of amputees and controls individually, and then over all participants. Furthermore, the relation of stepping strategy (single, multiple) and leading step length was determined (Pearson correlation).

Normality of data distribution (Kolmogorov–Smirnov test) was not given for one outcome measure, the percentage of trials in which balance was recovered with the leading limb. Here, non-parametric testing (Wilcoxon signed-rank test, Mann–Whitney test) was applied.

The level of significance was set to \( p \leq 0.05 \). All statistical analyses were performed using SPSS 16.0.

3. Results

All participants, amputees and healthy controls, were able to recover balance with any of both limbs leading after being released from a forward-inclined orientation of 10%. A failed recovery occurred only in one amputee participant who slipped on the floor after release when leading with his sound limb. No statistically significant differences in leading limb preference were found for the amputees (41.2% prosthetic, 58.8% sound) and the controls (52.9% left, 47.1% right) (\( \chi^2(1, N = 34) = .119, p = .73 \)). The group of amputees with prosthetic limb leading preference as well as the sound limb leading preference group scored high on the ABC score (92.9 ± 7.2 and 89.5 ± 11.3 respectively). No statistically significant differences in balance confidence were found between these two amputee groups (\( F(1,15) = .461, p = .508 \)).

Finally, we analyzed the contribution of the leading limb in arresting the forward fall. Fig. 2 shows a single step recovery of a participant. While leaning in a forward-inclined orientation the CoP was posterior to the CoM, and the participant was just being held back by the lean-control cable. Upon release the CoM started to gain velocity, i.e. the XCoM and the CoM moved apart. By stepping the participant brought his base of support over the XCoM, thereby breaking the forward fall. Shortly after heel-strike of the leading limb, the forward motion of the XCoM reached its maximum. In this study, this was taken as a marker that balance...
After heel-strike of the trailing limb, the participant was standing upright and CoM, XcoM and CoP coincided. However, this illustrates some minor inaccuracies. The concept of the XcoM is based on the inverted pendulum model, simplifying the human body as a pendulum with a mass, but recovering from a forward fall also involves accelerations of the swing leg. In theory the CoP should be positioned beyond the XcoM in order to move it back, while in Fig. 2 the CoP is only close to the XcoM.

The percentage of trials in which balance was recovered with the leading limb is given in Fig. 3. The group of amputees, who prefer leading with their prosthetic limb, recovered balance in a single step, when leading with their preferred prosthetic limb in 53.6% of the trials and when leading with their non-preferred sound limb in 42.9% of the trials. Interestingly, amputees who had a sound limb leading preference recovered balance in the leading step more often when leading with their non-preferred prosthetic limb (78.0%), then when leading with their preferred sound limb (65.0%). Non-parametric testing revealed no statistically significant differences in balance recovery for leading with the prosthetic versus sound limb in amputees (\( z = - .89 , p = .374 \)), and left versus right limb in controls (\( z = - .21 , p = .833 \)). Furthermore, the leading limb preference did not have a statistically significant effect on balance recovery in amputees (\( U = 25.5 , n_1 = 10 , n_2 = 7 , p = .346 \)), nor in controls (\( U = 26.5 , n_1 = 9 , n_2 = 8 , p = .352 \)). In addition, amputees and controls did not differ on the extent to which they recovered balance with the leading limb (\( U = 117 , n_1 = 17 , n_2 = 17 , p = .336 \)).

In depth comparative analysis of the mechanisms of balance recovery revealed the following mean values for the special and temporal parameters given in Table 1. In lower limb amputees the step length of the leading limb was statistically significant longer when leading with the prosthetic limb (\( F(1,15) = 9.71 , p < .001 \)), while no such step length asymmetries were found in controls (\( F(1,15) = .28 , p = .606 \)). Besides, there was no statistically significant correlation between stepping strategy, multiple or single, and the leading step length (\( r = -.085 , p = .492 \)).

In amputees, the step length of the trailing limb was statistically significant longer, when leading with the prosthetic and trailing with the sound limb (\( F(1,15) = 4.86 , p = .043 \)). Moreover, amputees, who had a preference for leading with their sound limb, made longer steps with the trailing limb, irrespective of the leading limb condition (\( F(1,15) = 13.25 , p = .002 \)). In the control group, however, the step length of the trailing limb was not affected by their leading limb preference (\( F(1,15) = .21 , p = .651 \)), nor the leading limb condition (\( F(1,15) = 1.71 , p = .212 \)).

With respect to the temporal dynamics of balance recovery, amputees showed longer response times when leading with the sound limb (\( F(1,15) = 14.03 , p = .002 \)). Furthermore, leading limb preference had a statistically significant effect on response time, amputees with a prosthetic limb leading preference responded faster (\( F(1,15) = 5.89 , p = .028 \)). In the group of controls, response time was neither affected by the leading limb condition (\( F(1,15) = .01 , p = .907 \)), nor by the leading limb preference (\( F(1,15) = .02 , p = .891 \)). Comparison between groups revealed that amputees responded statistically significant faster than controls (\( F(1,32) = 4.53 , p = .041 \)).

Analyses of the swing time in amputees yielded no marked differences between leading with the sound or prosthetic limb (\( F(1,15) = 2.61 , p = .127 \)). In addition, no statistically significant differences in swing time were found between amputees who preferred leading with their sound limb and those who preferred leading with their prosthetic limb (\( F(1,15) = .26 , p = .617 \)). Just like for the group of amputees, no effects of leading limb condition (\( F(1,15) = 2.81 , p = .114 \)) and leading limb preference (\( F(1,15) = .02 , p = .898 \)) on swing time were found for the group of controls.

In amputees, the heel-strike interval was shortened when leading with the prosthetic limb (\( F(1,15) = 10.84 , p = .005 \)). However, the leading limb preference, prosthetic versus sound, did not have a statistically significant effect on the heel-strike interval (\( F(1,15) = .05 , p = .835 \)). Again, no statistically significant differences on the heel-strike interval for leading with the left or right limb (\( F(1,15) = 2.63 , p = .127 \)) and leading limb preference (\( F(1,15) = 1.13 , p = .307 \)) were found in controls.

Analyses of the kinematics of the maximal knee flexion during swing in amputees revealed no statistically significant differences for leading with the prosthetic or sound limb, which agrees with the intact knee function in the prosthetic limb (\( F(1,15) = .26 , p = .624 \)).
controls who had a leading preference for the left or right limb (24, 25) differences when leading with their right or left limb during swing (86, 87, 88). Furthermore, the maximal knee flexion during swing was not statistically significant different between controls who had a leading preference for the left or right limb (24, 25) = .353). Furthermore, the maximal knee flexion during swing was not statistically significant different between controls who had a leading preference for the left or right limb (24, 25) = .347).

At heel-strike amputees showed a smaller knee flexion when leading with the prosthetic limb (24, 25) = 9.08, p = .009). Moreover, amputees with a preference of stepping with the prosthetic limb first, had a smaller knee flexion at heel-strike (24, 25) = 4.81, p = .044). The controls did not show statistically significant differences when leading with their right or left limb (24, 25) = .44, p = .517) or having a preference to step with the left or right limb first (24, 25) = .15, p = .704). Comparison between both experimental groups revealed, that amputees had an overall smaller knee flexion at heel-strike than controls (24, 25) = 8.83, p = .006).

4. Discussion

In this study we investigated the effect of lower limb amputation on balance recovery after an evoked forward fall. Despite the asymmetry of their locomotor system the amputees as a group had no specific preference for recovering with their prosthetic or sound limb. Interestingly, about 40% of the amputees had a preference to initiate recovery with their prosthetic limb. This is even more striking, considering that the fall itself was not unexpected, but only the moment of release. This means that they could plan their stepping response beforehand. In post-experimental interviews most participants indicated that the amputation did not affect their leading limb preference and stepping with the non-preferred limb first would feel unnatural. In the present set-up, amputees were just as capable to recover balance with their sound as their prosthetic limb; making a change in leading limb preference unnecessary. Furthermore, we found that amputees and controls were equally efficient in regaining stability with the leading limb, after being released from a forward-inclined orientation of 10%. In about 60% of the trials, amputees regained/initiated stability with the leading limb. But, as we were interested in studying the natural recovery behavior, participants were not limited in the number of steps taken or instructed to regain balance with the leading step already. This however implies that we may have underestimated their actual ability to recover balance with a single step.

Kinetametrical analysis of balance recovery in amputees revealed less knee flexion at heel-strike when leading with the prosthetic limb. This reduced knee flexion may have been associated with the larger step length of the leading limb. When stepping with the prosthetic limb first, amputees needed to take a longer step, as they were not able to actively shift the CoP forward under their prosthetic foot after heel-strike. A correction of the CoP position after heel-strike is only possible with active ankle control.

In order to recover balance with the leading step the CoP under the leading limb and the XCOM need to coincide, bringing the base of CoP under the position of the CoM is not sufficient, because the CoM would still have velocity and move away from the CoP.

When leading with the prosthetic limb, amputees shortened the heel-strike interval, hereby increasing stability. One of the amputees even performed jump like movements, nearly landing in synchrony with both limbs. To further minimize the impact forces acting on the stump, he bent his knee and quickly lowered his arms, thereby the vertical trunk acceleration was reduced and the GRF became smaller.

With respect to movement planning Do et al. [3] have suggested an invariant preparation phase when recovering balance. However, our results did not support this idea. In our study amputees responded faster when leading with their prosthetic limb, irrespective of their leading limb preference. Moreover, no interaction between leading limb preference and leading limb condition were found.

Previous studies have addressed the effect of instructions limiting the number of steps on the kinematics and kinetics of balance recovery [15, 16]. They showed that the leading step was nearly identical, irrespective of the stepping strategy. In this study we also did not find a systematic statistically significant effect of stepping strategy, single versus multiple, on the different outcome parameters.

A limitation of the tether-release method is its difference to stumbling in daily life, where tripping usually comes as a surprise. Furthermore, stumbling due to unplanned foot contact involves more complex dynamics. However, the observed behavior appeared to be surprisingly natural.

It is remarkable that our experienced amputee participants performed quite well in this challenging test, even with their prosthetic limb leading. Several of them experienced this as a pleasant surprise, which added to their confidence. This experience suggests that amputees may benefit from bilateral fall training in rehabilitation.

Suppliers

(a) Advanced Mechanical Technology, Inc., 176 Waltham Street, Watertown, MA 02472–4800, USA.
(b) Vicon Motion System, 14 Minus Business Park, West Way, Oxford, OX20JB, UK.
(c) Brosa AG, Dr. Klein Straße 1, 88069 Tettnang, Germany.
(d) The MathWorks, Inc., 3 Apple Hill Drive, Natick, MA 01760-2098, USA.
(e) SPSS Inc., 233 S. Wacker Drive, 11th floor, Chicago, IL 60606-6307, USA.

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Conflict of interest

The authors declare to have no conflict of interest in this work.

References