ABSTRACT

The aim of this study was to examine whether impaired balance control is partly responsible for the increased energy cost of walking in persons with a lower limb amputation (LLA). Previous studies used external lateral stabilization to evaluate the energy cost for balance control; this caused a decrease in energy cost, with concomitant decreases in mean and variability of step width. Using a similar set-up, we expected larger decreases for LLA than able-bodied controls. Fifteen transtibial amputees (TT), 12 transfemoral amputees (TF), and 15 able-bodied controls (CO) walked with and without external lateral stabilization provided via spring-like cords attached to the waist. Effects of this manipulation on energy cost, step parameters, and pelvic motion were evaluated between groups. TT (-5%) and CO (-3%) showed on average a small reduction in energy cost when walking with stabilization, whereas TF exhibited an increase in energy cost (+6.5%). The effect of stabilization was only significant between TT and TF. Step width, step width variability, and medio-lateral pelvic displacement decreased significantly with stabilization in all groups, especially in TT. Contrary to expectations, external lateral stabilization did not result in a larger decrease in the energy cost of walking for LLA compared to able-bodied controls, suggesting that balance control is not a major factor in the increased cost of walking in LLA. Alternatively, the increased energy cost with stabilization for TF suggests that restraining (medio-lateral) pelvic motion impeded necessary movement adaptations in LLA, and thus negated the postulated beneficial effects of stabilization on the energy cost of walking.
INTRODUCTION

Regaining walking ability is an important rehabilitation goal for lower limb amputees (LLA). Achieving this goal may be hampered by a significantly elevated energy cost of walking with a lower limb prosthesis, with reported increases between 9-33% for transtibial, and 66-100% for transfemoral amputees\(^2\), \(^25\), \(^98\). While this increased cost of walking is well documented, its underlying causes are still poorly understood.

Previous research has associated the elevated cost of walking in LLA with compensatory strategies related to forward progression of the body. LLA compensate for the lack of ankle push-off power with increased mechanical work produced at the hip, which increases step-to-step transition costs\(^38\). Furthermore, particularly transfemoral amputees show vaulting, hip hiking and circumduction of the prosthetic leg to ensure foot clearance during swing in the absence of active ankle dorsiflexion and knee flexion, which supposedly comes with an extra metabolic cost\(^99\). However, correlations between these adaptations and the elevated energy cost of walking are moderate at best\(^2\), \(^38\), \(^40\), suggesting a role for other factors, possibly not directly related to forward progression. One such factor could be the impaired balance control in LLA\(^100\)-\(^101\). While the energy demand of the motor responses associated with balance control is relatively low in healthy subjects, this cost might rise considerably as a result of compensatory strategies associated with the neuromuscular impairments in LLA, and thus contribute to the elevated energy cost of walking in LLA\(^98\).

Especially in the frontal plane, the most unstable direction during walking, active feedback control appears necessary to ensure stability\(^44\), \(^63\). Primary strategies for medio-lateral balance control are a stepping strategy, a lateral ankle strategy, and a hip strategy\(^46\). The stepping strategy provides gross balance control through adequate foot placement, while fine-tuning is accomplished by ankle inversion/eversion and hip abduction/adduction torques during stance. In LLA, the use of these strategies is hampered by reduced neuromuscular control to correctly place the foot, and a lack of control over the prosthetic ankle joint. Moreover, particularly in transfemoral amputees, the hip strategy is also often impaired due to atrophy and loss of control over the remaining muscles around the hip joint\(^102\).
These impairments can be dealt with by taking wider steps to ensure a sufficient margin of stability\textsuperscript{45}. Indeed, an increase in step width has been observed in LLA compared to controls, with larger increases for transfemoral amputees\textsuperscript{46, 103-105}. Moreover, increased step width variability has been observed in LLA, indicating an increased reliance on the stepping strategy to compensate for the reduced ability to use an ankle and/or hip strategy\textsuperscript{106-107}. While these compensations may help ensure stability, previous work has demonstrated that increasing step width and step width variability adversely affects the energy cost of walking\textsuperscript{18, 53} due to increased mechanical work to redirect the center of mass from side-to-side\textsuperscript{18, 53, 108}, or increased muscle activity to ensure adequate foot placement\textsuperscript{54}.

To estimate the contribution of medio-lateral balance control to the total energy cost of walking, the need for active balance control can be reduced artificially. To this end, Donelan et al.\textsuperscript{47} constructed a set-up to externally stabilize subjects in the medio-lateral direction via stiff spring-like cords attached to the waist. In healthy subjects this resulted in significant reductions in step width and step width variability, with a concomitant reduction in energy cost of 3-7.5\%\textsuperscript{47-49, 75}. Since LLA, and especially transfemoral amputees, naturally take wider and more variable steps, it can be hypothesized that they will benefit more from external lateral stabilization than able-bodied controls, resulting in a substantially larger reduction in energy cost due to stabilization for LLA, particularly for transfemoral amputees.

The aim of the current study was thus to examine whether the increased energy cost of walking in LLA compared to able-bodied people is related to an increased effort for balance control. More specifically, we sought to examine whether external lateral stabilization leads to larger reductions in the energy cost of walking in transfemoral and transtibial amputees compared to able-bodied controls, and expected the largest reductions to occur in transfemoral amputees. Furthermore, we expected concomitant decreases in step width and step width variability.
METHOD

Study population

Thirteen unilateral transfemoral amputees (TF), sixteen unilateral transtibial amputees (TT) and seventeen age-matched able-bodied controls (CO) agreed to participate. All amputees were experienced walkers who had completed their rehabilitation period and were able to walk 5 minutes on a treadmill. Subjects were excluded in case of contraindications for moderate exercise, or co-morbidities or medication use that could interfere with energy expenditure or balance control. Additional exclusion criteria for LLA were improper fitting of the prosthesis and stump problems (e.g., pain, pressure sores). All amputees walked with their custom prosthesis resulting in a heterogeneous group of subjects in terms of prosthetic properties. Subjective balance confidence was assessed with the Activities Specific Balance Confidence Scale (ABC-Scale). Subject characteristics are presented in Table 3. All subjects gave written informed consent prior to participation. This study was approved by the Medical Ethical Committee of the VU University Medical Center, Amsterdam, The Netherlands.

Study protocol

Subjects completed two 5-minute walking trials at their preferred speed on a treadmill. Trials were applied in random order and consisted of normal walking and walking with external lateral stabilization, separated by ~4 minutes of rest. Subjects were allowed handrail support for the first half of the trial if deemed necessary. Prior to the walking trials, resting energy expenditure was recorded for 5 minutes in a seated position after 10 minutes of rest. Thereafter, both experimental conditions were practiced at a comfortable speed for 3 minutes to familiarize subjects with the experimental conditions. Subsequently, the subjects’ preferred walking speed (PWS) was determined without stabilization following a previously described protocol. This PWS was used in both experimental trials.
Experimental set-up

Similar to the set-up of Donelan et al. we used sets of parallel elastic rubber cords to provide external lateral stabilization (Figure 14). To allow normal arm swing the cords were attached on one end to a frame fastened to a hip belt worn tightly around the pelvis\textsuperscript{49}, while the other end was connected to a ball-bearing trolley mounted on a height-adjustable horizontal rail. The trolley moved along with the subject in the anterior-posterior direction, to minimize fore-aft forces of the springs. The rail was adjusted to the subjects’ pelvic height to minimize vertical forces. The springs had an effective spring constant of \(1260\) N·m\(^{-1}\) and negligible damping (\(\sim18.5\) N·s·m\(^{-1}\)). A previous study established that this stiffness is sufficient to stabilize human walking in the sideward direction and maximally reduce the energy cost of walking in healthy subjects\textsuperscript{75}.

![Frame allowing normal arm swing](image)

Data collection

Oxygen consumption was measured breath-by-breath via a pulmonary gas exchange system (Quark \(b^2\), Cosmed, Italy). Optoelectronic markers were attached to the heel of each foot to be able to calculate step parameters and to the four corners of the frame, to estimate pelvic motion. Marker positions were recorded with a 3D motion analysis system (Optotrak 3020, Northern Digital Inc., Waterloo, Canada) at a sampling rate of 100Hz.
Data analysis

Energy cost

Steady state energy expenditure (EE; J·min⁻¹) was calculated from oxygen uptake ($\dot{V}O_2$; ml·min⁻¹) and respiratory exchange ratio (RER) obtained from the pulmonary gas exchange system during the final two minutes of each trial according to the following equation:

$$EE = (4.940 \cdot RER + 16.040) \cdot \dot{V}O_2.$$ 

Resting metabolism was subtracted from EE to obtain nett energy expenditure (EE_{nett}). Nett metabolic cost (EC_{nett}; J·kg⁻¹·m⁻¹) was calculated by dividing EE_{nett} by body mass (kg) and walking speed (m·min⁻¹).

Step parameters

Kinematic data were low-pass filtered with a bi-directional 2nd order Butterworth filter with a cut-off frequency of 10Hz. Only the last 2 minutes of each trial were analyzed, because subjects were allowed handrail support for the first half of the trial. Instants of foot contact were detected based on minima in the vertical heel marker position data and used to calculate mean and variability of stride length and step width. Variability was expressed as the standard deviation of stride length and step width within a trial.

Pelvic motion

Linear and angular motion of the pelvis was estimated based on the four frame markers, which were rigidly connected to the pelvis and therefore defined changes in position and orientation of the pelvis. Again only the final 2 minutes of each trial were analyzed. Peak-to-peak amplitude of the displacement in medio-lateral (ML) and vertical (V) direction, and rotation around the anterior-posterior (pelvic obliquity) and vertical axis (pelvic rotation) of the pelvis with respect to the global coordinate system were calculated for each stride and averaged over a trial.
Statistical analysis

Anthropometric characteristics, balance confidence, PWS, EC_net, step parameters, and pelvic motion during normal (unstabilized) walking were tested for group differences using a one-way ANOVA. A mixed-design ANOVA with Condition ([Normal, Stabilized]) as within-subjects factor and Group ([TF, TT, CO]) as between-subjects factor was used to test for statistically significant differences between conditions and group × condition interaction on energy cost, step parameters and pelvic motion. Significant interaction effects were further examined using a one-way ANOVA on the difference scores between conditions (Normal–Stabilized) with Group as between subjects factor. Level of significance was set at \( p < .05 \). In case of significant effects, post hoc tests were applied with a Bonferroni correction for multiple comparisons.

RESULTS

In each group one subject was unable to complete the protocol, for different reasons: soreness in the stump (TF), difficulty walking on a treadmill (TT) and dizziness (CO). Oxygen data for one subject (CO) were aberrant, therefore this subject was excluded, leaving a total of 12 TF, 15 TT and 15 CO subjects that were included in the analysis.
Table 3 Descriptive characteristics of study population

<table>
<thead>
<tr>
<th></th>
<th>CO (N=15)</th>
<th>TT (N=15)</th>
<th>TF (N=12)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (yrs)</td>
<td>56.7±12.4</td>
<td>58.8±12.7</td>
<td>54.8±13.0</td>
</tr>
<tr>
<td>Gender (male/female)</td>
<td>10/5</td>
<td>13/2</td>
<td>8/4</td>
</tr>
<tr>
<td>Cause (Trauma/Vasc/Other)*</td>
<td>N.A.</td>
<td>10/3/2</td>
<td>9/1/2</td>
</tr>
<tr>
<td>Time since amputation (years)</td>
<td>N.A.</td>
<td>23.3±22.3</td>
<td>23.7±18.9</td>
</tr>
<tr>
<td>Body mass (kg)</td>
<td>76.7±13.1</td>
<td>84.2±14.6</td>
<td>81.0±13.9</td>
</tr>
<tr>
<td>Body height (m)</td>
<td>1.78±0.10</td>
<td>1.78±0.08</td>
<td>1.76±0.08</td>
</tr>
<tr>
<td>BMI</td>
<td>24.3±3.3</td>
<td>26.3±4.1</td>
<td>25.9±4.2</td>
</tr>
<tr>
<td>Trochanter height (m)</td>
<td>0.91±0.06</td>
<td>0.93±0.07</td>
<td>0.91±0.05</td>
</tr>
<tr>
<td>Basal metabolic rate (J·kg⁻¹·min⁻¹)</td>
<td>74.4±18.3</td>
<td>72.4±16.3</td>
<td>68.5±14.3</td>
</tr>
<tr>
<td>ABC score</td>
<td>91.8±6.3</td>
<td>81.3±12.8*</td>
<td>80.3±11.9*</td>
</tr>
<tr>
<td>Walking speed (m·s⁻¹)</td>
<td>1.1±.13</td>
<td>.90±.20*</td>
<td>.71±.19*</td>
</tr>
</tbody>
</table>

* = significantly different from CO; †=significantly different from TT; * = Cause of amputation: Traumatic, Vascular or Other (e.g. bacterial, cancer) in number of subjects.

Group differences

Age and anthropometric characteristics did not differ between groups (Table 3). CO had significantly higher balance confidence (p<.05), and a significantly higher PWS than both amputee groups (p<.05). Also, TT had a higher PWS than TF (p<.05).

Significant group differences were found for normal walking. The energy cost of walking was significantly higher for TF than for CO (77%, p<.01), and TT (34%, p<.01; Figure 15). The difference between TT and CO was borderline significant (33%, p=.086). Step parameters differed significantly between groups, with both amputee groups walking with shorter (p<.05) and wider (p<.001) steps, and a larger stride length variability (p<.01) compared to CO (Figure 16). In addition, TT had a larger step width variability than CO (p<.05). Differences in pelvic motion between groups showed that ML pelvic displacement was larger for both amputee groups than for CO (p<.01 for TT; p<.001 for TF), and also larger for TF compared to TT (p<.05; Figure 17). Finally, TT walked with less peak-to-peak pelvic obliquity and more pelvic rotation compared to CO (p<.01).
Effects of external lateral stabilization

Energy cost

No significant main effect of stabilization ($p=.395$), and no significant group × stabilization interaction ($p=.307$) were found on energy cost (Figure 15). Further inspection of the data showed that one outlier in the energy cost data considerably influenced the significance of these results (Figure 15B). With removal of the outlier, a significant interaction effect on energy cost was observed ($p<.05$). The effect of stabilization was significantly different ($p<.05$) between TT, who on average showed a reduction in cost when walking with stabilization (-5%), and TF, who on average showed an increase in cost with stabilization (+6.5% without outlier). Also, a trend was found between TF, and CO ($p=.104$) who on average showed a reduction in cost (-3%).

Figure 15: Effect of stabilization on net energy cost. A): for CO (blue circles, N=15), TT (green triangles, N=15) and TF (red squares, N=12), error bars indicate SD. B): Boxplot of differences in net energy cost (Stabilized – Normal) for each group, the black line in the box indicates the median difference in energy cost, edges of the box represent the 25th and 75th percentile. In the TF group, one outlier was observed (indicated with the red cross).

Step parameters

Stabilization caused a reduction in step width and step width variability ($p<.001$; Figure 16). In addition, a significant group × stabilization interaction ($p<.01$) revealed that the reduction in step width was larger for TT compared to CO ($p<.01$) and TF
(p<.05). A significant interaction was also found on step length (p<.05), with a reduction in step length for TF compared to a slight increase in TT (p<.05).

**Pelvic motion**

Stabilization caused a decrease in ML pelvic displacement, pelvic obliquity and pelvic rotation (all p-values<.001; Figure 17). Furthermore, a significant group × stabilization interaction was found for pelvic obliquity (p<.01) and rotation (p<.05). Follow-up analysis showed a significantly larger reduction in pelvic obliquity for CO compared to both amputee groups (p<.01). Furthermore, TT showed a trend towards a larger reduction in pelvic rotation than CO (p=.093) and TF (p=.098).
Figure 16: Effect of stabilization on mean and variability of step parameters for CO (blue circles, N=15), TT (green triangles, N=15) and TF (red squares, N=12), error bars indicate SD.
Figure 17: Effect of stabilization on pelvic motion for CO (blue circles, N=15), TT (green triangles, N=15) and TF (red squares, N=12), error bars indicate SD.
DISCUSSION

The goal of this study was to investigate whether the increased energy cost of walking in LLA could partly be explained by impaired balance control, by evaluating the effect of external lateral stabilization in LLA and able-bodied subjects. Our hypothesis that external lateral stabilization would result in larger reductions in energy cost for LLA, especially for TF, was not corroborated by the results. The effect of stabilization on energy cost was small in light of previous studies with healthy subjects (3.1-7.5%\textsuperscript{47-49, 75}), and particularly in light of the large differences in the energy cost of walking between LLA and able-bodied controls (77% and 33% higher than CO for TF and TT, respectively). Moreover, the effects of the stabilization did not differ significantly between TT and CO (-5% vs. -3%), while, after removal of an outlier, TF even showed an unforeseen increase in energy cost when walking with stabilization (+6.5%). These effects on metabolic cost occurred even though as expected, step width, step width variability and medio-lateral pelvic movement decreased with stabilization. These limited effects of stabilization on metabolic cost imply that balance control problems do not have a substantial effect on the energy cost of walking in LLA.

The unexpected increase in cost for TF, however, suggests that potential confounding effects of the stabilization device on the gait pattern of (transfemoral) LLA might have occurred. Potentially, constraints on pelvic motion imposed by the stabilization device interfered with compensatory gait strategies used by TF, unrelated to balance control, resulting in an energetic penalty. With respect to able-bodied controls, TF exhibit excessive medio-lateral displacement of the pelvis and altered patterns of pelvic rotation and obliquity\textsuperscript{46, 109-110}. Together with increased step width, increased ML pelvic displacement is commonly seen as a strategy to help ensure sufficient medio-lateral stability. However, while step width and ML pelvic displacement decreased when walking with external stabilization in TF, they remained more than twice as large as for CO, even though this was no longer necessary for balance control. These findings support a previous suggestion that the excessive ML pelvic movements may not only relate to balance control, but also satisfy other functional requirements for walking with a prosthesis\textsuperscript{46}. The altered patterns of pelvic rotation and obliquity can likewise be regarded as functional
adaptations to walking with a prosthesis. Pelvic rotation has been suggested as a compensation to provide a thrust to the swing leg in the absence of ankle power\textsuperscript{109}, while pelvic obliquity can be used to ensure foot clearance of the prosthetic swing leg (hip-hiking), or result from the trunk leaning over the prosthetic stance leg to reduce the hip abduction moment and relieve weakened hip abductor muscles\textsuperscript{99, 110}. Similar to ML pelvic motion, pelvic obliquity and rotation are opposed by the stabilization device. TF appear to have difficulty dealing with these constraints, as evidenced by the larger ML pelvic displacement, rotation, and obliquity during stabilized walking compared to CO and TT. Working against the springs to maintain these functional compensations, or adapting to the imposed constraints in the face of an already impaired motor system, is likely energetically demanding.

These disadvantageous effects of constraining pelvic motion on the use of compensatory gait strategies are not necessarily restricted to TF alone. TT also exhibit altered patterns of pelvic motion during walking, although the pattern is in general disrupted less than in TF\textsuperscript{99, 110}. The reduction in pelvic displacement, obliquity, and rotation during stabilized walking to values similar to those of the able-bodied group, suggests that they were better able to cope with the imposed constraints. Unfortunately, with the current results we were not able to discern if, and to what extent, these constraints on pelvic motion negate beneficial effects of external lateral stabilization with respect to balance control.

Thus, while we intended to investigate the metabolic effort for balance control, it appears that the external stabilization paradigm as used in this experiment is inappropriate to answer this question for this population. The problems with the constrained pelvic motion might in part be overcome by adapting the set-up to allow pelvic rotations around the vertical and anterior-posterior axis. However, a pilot with such a frame was unsuccessful, suggesting that also medio-lateral pelvis motion is indispensible in amputee gait. Similar problems with constrained pelvic movement may also arise in other patient populations that rely on pelvic movements as compensation strategy, such as stroke survivors and children with cerebral palsy. We know of only one study that used the external stabilization paradigm in a patient population. In this study, involving persons with myelomeningocele, stabilization was claimed to result in a substantial decrease in energy cost, although this claim was
based on heart rate recordings only\textsuperscript{111}. Hence, some reservation about the use of this paradigm in patient populations seems justified.

Besides the limitation with respect to the experimental set-up, two other issues deserve attention when interpreting the results. Firstly, in the TF group, one statistical outlier was found. Although no explanation for this anomaly could be found, this outlier had such a distinct effect on the statistical analysis that we deemed it justified to remove it. Secondly, while all subjects were allowed a habituation period prior to the experiment, it is possible that a more extensive training period would allow LLA to overcome the imposed constraints on the use of their compensatory gait strategies.

In conclusion, external lateral stabilization had no marked effect on the energy cost of walking in LLA, and even appeared to have a detrimental effect for TF. This limited effect potentially emanated from restriction of pelvic motion in the current set-up, which could interfere with functional gait compensations unrelated to balance control. This highlights the importance of pelvic motion as an adaptation strategy to walking with a prosthesis. In all, the energy cost of balance control, and its contribution to the increased energy cost of walking in LLA, remains undetermined and should be unveiled further in future research.