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ORIGINAL RESEARCH

Effects of Handrail and Cane Support on Energy Cost of Walking in People With Different Levels and Causes of Lower Limb Amputation

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Abstract

Objective: The energy cost of walking with a lower limb prosthesis is higher than able-bodied walking and depends on both cause and level of amputation. This increase might partly be related to problems with balance control. In this study we investigated to what extent energy cost can be reduced by providing support through a handrail or cane and how this depends on level and cause of amputation.

Design: Quasi-experimental study.

Setting: Rehabilitation gait laboratory.

Participants: Twenty-six people with a lower limb amputation were included: 9 with vascular and 17 with nonvascular causes, 16 at transtibial, and 10 at transfemoral or knee disarticulation level (N=26).

Interventions: Participants walked on a treadmill with and without handrail support and overground with and without a cane.

Main Outcome Measures: Energy cost was assessed using respirometry.

Results: On the treadmill, handrail support resulted in a 6% reduction in energy cost on average. This effect was attributed to an 11% reduction in those with an amputation attributable to vascular causes, whereas the nonvascular group did not show a significant difference. No interaction with level of amputation was found. Overground, no main effect of cane support was found, although an interaction effect with cause of amputation demonstrated a small nonsignificant decrease in energy cost (3%) in the vascular group and a significant increase (6%) in the nonvascular group when walking with a cane. The effect of support was positively correlated with self-selected walking speed.

Conclusions: This study demonstrates that providing external support can contribute to a reduction in energy cost in people with an amputation due to vascular causes with reduced walking ability while walking in the more challenging condition of the treadmill. Although it is speculated that this effect might be related to problems with balance control, this will need further investigation.

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Regaining walking ability is a challenging rehabilitation goal for people with a lower limb amputation. Walking with a prosthesis can require up to twice as much metabolic energy as in able-bodied walking.[1] This may cause severe mobility limitations, especially because the aerobic capacity of people with a lower limb amputation is generally lower than that of their able-bodied peers.[2-5] The combination of a low capacity and high load can result in a markedly increased relative aerobic load during walking[2] and contribute to the reduced ambulatory activity in this population. To facilitate ambulation, rehabilitation programs can aim to improve aerobic capacity via exercise training or to decrease the aerobic load by trying to reduce the energy cost of walking (ie, the energy expenditure per distance traveled).[2]

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Studies investigating the energy cost of walking in people with lower limb prosthesis typically divide this population based on the level of amputation (with amputation at transtibial and transfemoral as main levels) and on the etiology of amputation (generally separated between amputation due to vascular deficiency and nonvascular problems such as trauma or tumors). Both level and cause of amputation have an effect on the energy cost of walking; on average, people with a transfemoral amputation have a higher energy cost of walking than those with a transtibial amputation, and people with an amputation due to vascular causes have a higher energy cost than those with nonvascular causes.[1]

The effect of the level of amputation on the energy cost of walking is regularly attributed to the biomechanical constraints of the prosthesis. The lack of an active foot and ankle joint that can generate push off has been shown to explain part of the increased energy cost in people with a transtibial amputation.[6-8] Additionally, the higher energy cost of people with a transfemoral amputation relative to a transtibial amputation can be partly related to the mechanical constraints of the passive knee joint, predominantly in the swing phase.[9-11]

The mechanisms accounting for the effect of etiology of amputation on the energy cost of walking are, however, not easily explained by such biomechanical factors. People after vascular or nonvascular amputation experience similar mechanical deficits as a consequence of losing part of their biological leg. A factor that could potentially account for the difference in energy cost between people with different causes of amputation might be found in their capacity to control their movement and specifically their capacity for balance control.[12] Balance control and balance confidence are reduced in people with lower limb amputation and are generally worse in people after vascular amputation compared with traumatic amputation.[13,14] Balance deficits have also been related to level of amputation because balance control is more impaired in people with a transfemoral compared with transtibial amputation.[15] Differences in the effort for balance control and associated energy requirement[16] might therefore contribute to the difference in energy cost of walking between subgroups of amputees.

Assuming that balance problems contribute to the energy cost of walking, providing patients with a support device might help to reduce this energy cost. This has previously been shown for people after stroke who were provided with a handrail or cane when walking on a treadmill and overground, respectively. The effect of support was largest in patients with limited levels of walking ability, who experienced a reduction in energy cost up to 19% when walking on a treadmill with a handrail hold.[17,18] For people after lower limb amputation, the effect of external support on energy cost has not been established. A previous study using a set of elastic springs to provide external lateral stabilization forces to the pelvis to support balance[19-22] rendered inconclusive results because the concomitant restriction on pelvic motion imposed by the system seemed to prevent useful gait compensations required in the gait pattern of amputees.[23] Providing people with a lower limb amputation with a support aid such as a handrail or cane imposes less restriction on their gait pattern and could therefor elucidate the potential effect of support on the energy cost of walking in people with a lower limb amputation.

In this study we investigated the potential effect of handrail (on a treadmill) and cane (overground) support on energy cost of walking in people with lower limb amputation and the potential differential effect on people with different levels and etiology. We hypothesized that both handrail and cane support would have a significant effect on the energy cost of walking in people with lower limb amputation and that this effect would be larger for people with lower levels of walking ability. Specifically, we hypothesized that this effect is larger for people with vascular compared with nonvascular cause of amputation and for those with transfemoral compared with transtibial level of amputation.

List of abbreviations:

ANOVA	**analysis of variance**
EE	**energy expenditure**
KD	**knee disarticulation**
PWS	**preferred walking speed**
TF	**transfemoral**
TT	**transtibial**
$\dot{V}o_2$	**oxygen consumption**
vasc	**vascular etiology**
non-vasc	**non-vascular etiology**

Methods

Participants

A convenience sample of 26 people with a unilateral lower limb amputation was included in this study. This sample size was calculated based on the main effect size previously found by Ijmker et al[17] for a comparable study on patients after stroke (Gpower, 3.1; effect size, 0.3; α, 0.05; β, 0.8; correlation among repeated measures, 0.5). All participants were experienced walkers who had completed their rehabilitation program and were able to walk on a treadmill and overground without support for at least 4 minutes. Participants were excluded in cases of (1) contraindications for moderate exercise; (2) comorbidity or medication use that could affect walking performance, energy cost, or balance control; (3) improper fitting of the prosthesis and residual limb problems; or (4) problems with understanding and following instructions and study protocol. This study was approved by the local ethical committee of Heliomare Rehabilitation and the medical ethical committee of Vrije Universiteit Medical Centre Amsterdam (VUmc 2016.151). All participants were fully informed about the aim and content of the study and signed a written informed consent before participation.

Procedure

Prior to the experiment, participants filled out the Activities-specific Balance Confidence scale.[24] Thereafter, resting energy expenditure was recorded in a seated position for a duration of 4 minutes.

The experimental protocol consisted of 2 support conditions (support, no support), which were executed both on a treadmill and overground, resulting in 4 walking trials. During overground walking, support was provided using a standard, height-adjustable cane on the nonprosthetic side. During treadmill walking, support was provided through a sideward-adjustable instrumented handrail on the nonprosthetic side. Participants were instructed not to lean heavily on the devices nor to use the device for propulsion. As such we intended for the participants to use the devices mainly for balance corrections in either direction and a general feeling of safety.

Participants were randomly assigned to start on the treadmill or overground and completed the 2 support conditions in random order. Each trial had a duration of 4 minutes, and trials were separated by a resting period of at least 10 minutes to avoid fatigue. During all trials, participants walked at their preferred walking speed (PWS). Before the treadmill block, PWS was established on the treadmill while walking without support following a previously described protocol.[17] This speed was subsequently used in both treadmill trials. For the overground trials, participants were asked to walk back and forth on a 40-m even level indoor track at their PWS. This PWS could differ between supported and unsupported trials in the overground conditions.

Instrumentation

During all walking trials, oxygen consumption ($\dot{V}o_2$) and respiratory exchange ratio were measured breath by breath using a portable open circuit respirometry,[a] which was strapped around the trunk of the participant (fig 1). During treadmill trials, participants walked on an instrumented treadmill mounted flush into the floor and equipped with an embedded force plate[b] (size: 1m×1.5m; sampling rate 100Hz) from which step parameters were derived (supplemental appendix S1, available online only at http://www.archives-pmr.org/). An instrumented sideward-adjustable handrail, equipped with two 6 degrees-of-freedom force sensors[c] (sampling rate 100Hz), was placed on the nonprosthetic side of the participant during treadmill walking to measure the forces exerted on the handrail (supplemental appendix S2, available online only at http://www.archives-pmr.org/).

Data analysis

Energy cost

Steady state energy expenditure (EE; J/min) was calculated from the average $\dot{V}o_2$ (mL/min) and respiratory exchange ratio during the final 2 minutes of each trail according to the following equation[25]:

$$EE = (4.940 \cdot \textit{respiratory exchange ratio} + 16.040) \cdot \dot{V}O_2$$

Data were visually inspected to ensure steady-state $\dot{V}o_2$ was reached in that time period (see fig 1). For overground walking, instances of turning remained included in the data. Resting energy expenditure was subtracted from energy expenditure during the walking trials to obtain net energy expenditure (EE_{net}). Net energy cost (EC_{net}; J/kg$^{/}$m) was calculated by dividing EE_{net} by body mass (kg) and walking speed (m/min).

Statistical analysis

Participants were classified regarding level and cause of amputation to allow subgroup analyses. Level of amputation was split in 2 categories: below the level of the knee (transtibial [TT]) and at or above the level of the knee (transfemoral [TF]+knee disarticulation [KD]). Etiology of amputation was split in 2 categories: vascular etiology (including diabetes) and nonvascular etiology. Descriptive characteristics of the study population were obtained, and differences between the subgroups were analyzed using a *t* test.

Energy cost data were checked for normality using visual inspection and the Shapiro-Wilk test. To determine the effect of balance support on the energy cost of walking and the influence of the cause and level of amputation on these effects, we used a mixed methods analysis of variance (ANOVA) with support (unsupported, supported) as within-participants factor and level (TT, TF+KD) and etiology (nonvascular, vascular) as between-participants factor. A priori, it was decided to only assess the following effects in this statistical model: the main effect of support and the support times level and support time etiology interaction effects. Significant interaction effects were followed up by post hoc paired *t* tests to evaluate which groups (eg, based on level or etiology) showed a significant effect of support. The ANOVA was executed for treadmill and overground walking separately. Partial Eta squared (η_p^2) was used as estimate of effect size.

In addition, we analyzed the relation between walking ability and the effect of support using Pearson correlation between self-selected unsupported walking speed (on either the treadmill or overground) and the reduction in energy cost with support, from which we calculated the coefficient of determination r^2.

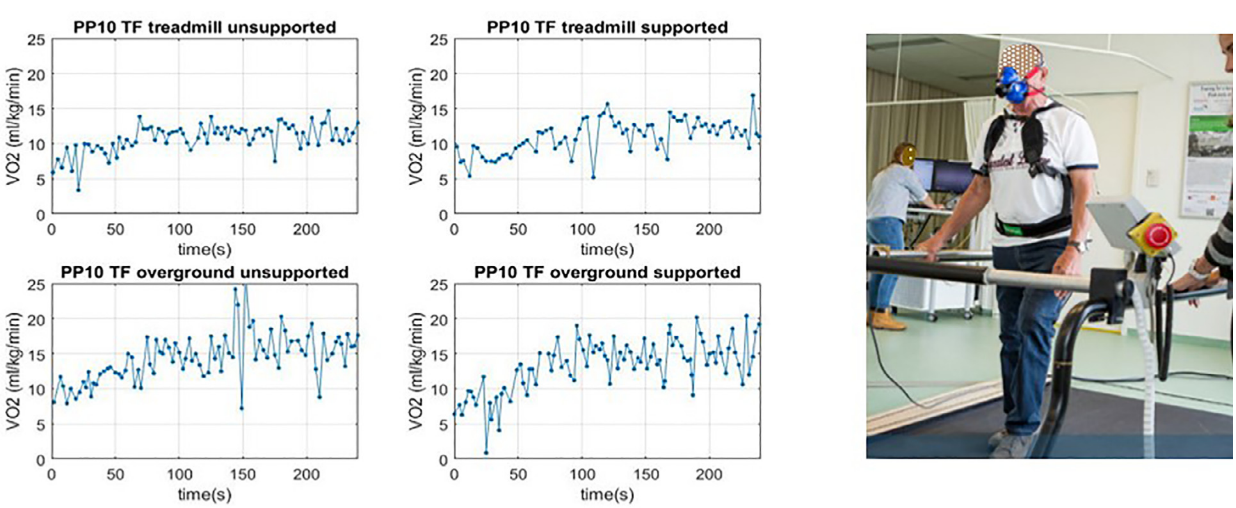

Fig 1 Left, typical example of breath-by-breath $\dot{V}o_2$ recordings for the treadmill and overground unsupported and supported condition of a participant with a nonvascular transfemoral amputation. Note that steady state $\dot{V}o_2$ was reached after 2 min. Right, experimental setting; a left-sided transtibial amputee walks on the treadmill with handrail support on the right while breath-by-breath $\dot{V}o_2$ data is collected.

Results

A total of 26 people with a lower limb amputation participated in this study: 7 with a transfemoral, 3 with a knee disarticulation (grouped together in TF+KD group), and 16 with a transtibial amputation (TT group). Nine participants underwent amputation because of vascular deficiency. Seventeen underwent amputation because of nonvascular causes: 12 because of trauma, 3 because of cancer, 1 because of bacterial infection, and 1 because of surgical complications.

There were no significant differences between the TT and TF+KD subgroups regarding demographic characteristics, balance confidence, or walking speed (table 1). However, significant differences existed between subgroups stratified by cause of amputation: people with a vascular cause of amputation had significantly shorter time since amputation, significantly higher body mass index and body mass, and significantly slower preferred walking speed overground than those with an amputation due to nonvascular causes.

Inspection of the energy cost data showed 1 outlier, with vascular cause of amputation at TF level in the energy cost data on the treadmill (>3 SD from the group mean), which affected the normal distribution of the data (Shapiro-Wilk test: P=.066, P=.027, P=.133, P=.273 for treadmill without and with support and overground without and with support, respectively). However, statistical conclusions results remained stable with a reanalysis without the data from this participant. Because this outlier was a result of biological variation and not measurement error, it was decided to include this participant in the analyses.

Main effect of support on energy cost

There was a significant main effect of support on energy cost for treadmill ($EC_{unsupported}$=4.68±2.96 J/kg/m; $EC_{supported}$=4.42±2.69 J/kg/m; F=11.783; P=.002; η_p^2=0.349) but not overground walking ($EC_{unsupported}$=3.40±1.09 J/kg/m; $EC_{supported}$=3.48±0.93 J/kg/m; F=0.002; P=.962; η_p^2=0.000). On the treadmill, handrail support on average resulted in a 6% lower energy cost of walking.

Influence of the etiology of amputation

There was a significant interaction effect of support times cause for treadmill walking with a moderate effect size (F=9738; P=.005; η_p^2=0.307) (fig 2). Post hoc analysis showed that balance support resulted in a significant 11% reduction in energy cost of walking for vascular amputees (t=2.60; df=11; P=.025) on average, but no significant difference was found for those with an amputation due to nonvascular causes (t=0.584; df=15; P=.568).

There was also a significant interaction effect of support times cause for overground walking with a small effect size (F=4.299; P=.050; η_p^2=0.163) (see fig 2). Post hoc analysis, however, did not show a significant difference for vascular amputees (t=−0.181; df=11; P=.860) but did show a significant increase of 5% on average when using support in those with an amputation due to nonvascular causes (t=−2.55; df=15; P=.022).

Influence of the level of amputation

No significant interaction effect of support times level of amputation was found for treadmill (F= .117; P=.736; η_p^2=0.005) or overground walking (F=1.295; P=.267; η_p^2=0.056), indicating that the level of amputation did not significantly influence the effect of support on the energy cost of walking (see fig 2).

Relation with self-selected walking speed

The reduction in energy cost with support was significantly correlated with self-selected walking speed for both treadmill (r^2=0.202, P=.021) as well as overground (r^2=0.234, P=.012) walking (fig 3).

Effects on step parameters and handrail support forces

Assessment of handrail support forces (see supplemental appendix S2, available online only at http://www.archives-pmr.org/) demonstrated that participants did not exert substantial forces in

Table 1 Characteristics of study population (N=26)

Participant Characteristics	Total Group (N=26)	Nonvascular (n=17)	Vascular (n=9)	*P* Value Cause	TT (n=16)	TF/KD (n=10)	*P* Value Level
Level of amputation (TT/TF+KD)*	16/10	10/7	6/3	.696	.	.	.
Etiology (nonvascular/vascular)*	17/9	.	.	.	10/6	7/3	.696
Sex (M/F)*	23/3	14/3	9/0	.400	13/3	10/0	.245
Age (y), mean ± SD	58.7±13.7	57.0±15.7	61.8±8,6	.408	61.3±15.4	54.4±9.7	.217
BMI, mean ± SD	27.0±4.3	25.7±3.7	29.6±4.4	.027†	26.3±4.0	28.2±4.8	.284
Body mass (kg), mean ± SD	88.7±18.5	81.9±13.4	101.6±20.6	.007†	85.9±15.5	93.1±22.7	.345
Time since amputation (y), median (range)*	6.8 (0.3-58)	15.0 (2-58)	1.5 (0.3-10)	.008†	5.3 (0.3-58)	8.5 (1-43)	.780
ABC score (%), mean ± SD	74.1±23.3	76.8±21.2	69.0±27.4	.433	75.4±23.7	72.1±23.8	.733
Preferred walking speed treadmill (m/s), mean ± SD	0.74±0.30	0.81±0.26	0.60±0.33	.085	0.79±0.32	0.65±0.25	.245
Preferred walking speed overground unsupported (m/s), mean ± SD	1.14±0.26	1.25±0.16	0.94±0.29	.002†	1.18±0.24	1.08±0.28	.323

Abbreviations: ABC, Activities-specific Balance Confidence; BMI, body mass index (calculated as weight in kilograms divided by height in meters squared); F, female; M, male.
* χ^2 test.
† P<.05.

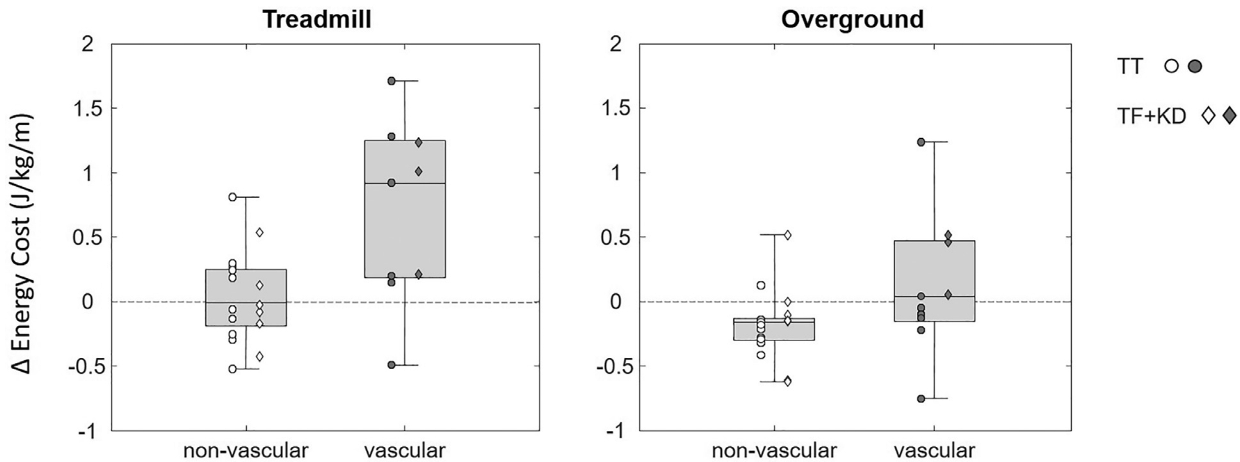

Fig 2 Difference in energy cost of walking with and without support on a treadmill (left) and overground (right). Positive values indicate a reduction in energy cost with support. Box plots represent the effect of cause of amputation (median, interquartile range, total range). Individual data, stratified for level of amputation, visualize individual responses (TT=circles, TF+KD=diamonds).

anterior-posterior direction for propulsion (peak force <1% body weight) or vertical direction for body weight support (peak force <7% body weight). Small forces were exerted in mediolateral direction (peak force <2% body weight), mainly during stance on the prosthetic leg.

Assessment of step parameters during treadmill walking (see supplemental appendix S1, available online only at http://www.archives-pmr.org/) demonstrated an increase in stride time and length, a decrease in step width, and improved step time and length symmetry with handrail support. Variability for stride time and stride length and step width decreased when walking with support.

Discussion

In this study we investigated the effect of handrail and cane support on the energy cost of walking in people with a lower limb amputation and how this depends on level and cause of amputation. We found that handrail support significantly reduced the energy cost of treadmill walking, specifically in people with an amputation due to vascular problems. No significant interaction effect was found for amputation level. During overground walking no main effect of support was found; however, an interaction effect revealed that energy cost even slightly increased when using a cane in people with nonvascular cause of amputation, and a small nonsignificant decrease was observed in the vascular group.

The predominant occurrence of the effect of support on energy cost in people after vascular amputation is in line with our hypothesis: handrail hold on the treadmill reduced energy cost of walking in people after vascular amputation by11% on average. This reduction is larger than reported for state-of-the-art-prosthetic innovations[26,27] and regarded as clinically relevant.[28] This effect might be related to balance control problems, which have been demonstrated in people after vascular amputation compared with those with nonvascular cause of amputation[14,29,30] and able-bodied peers.[31] Peripheral vascular deficiency, especially when related to diabetes mellitus, results in impaired somatosensory perception.[32] Moreover, people in this group often show reduced physical fitness[4] and might lack the muscle strength to execute

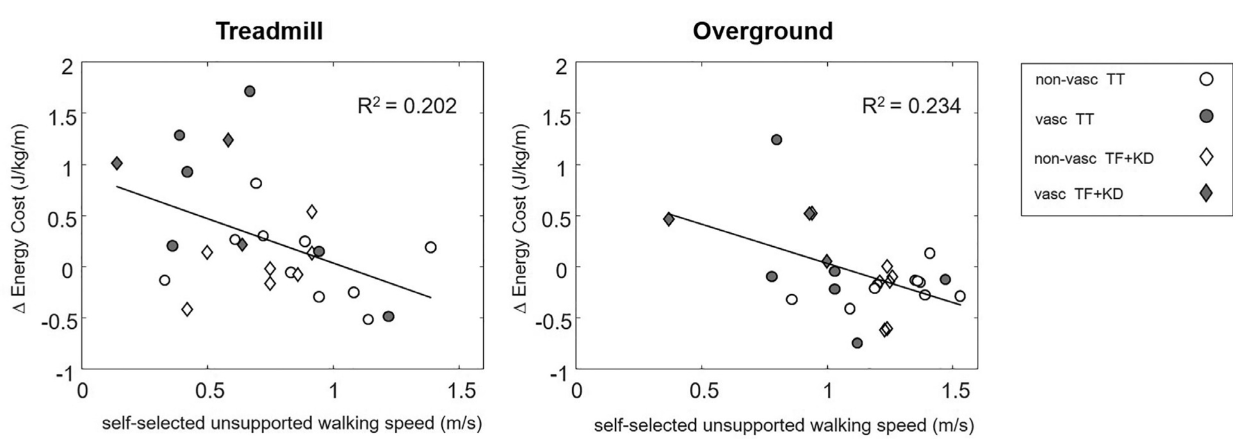

Fig 3 Relation between self-selected unsupported walking speed and the reduction in energy cost with support for treadmill (left) and overground (right) walking. Data are stratified by level (TT=circles, TF+KD=diamonds) and cause (nonvascular=white, vascular=gray) of amputation. Positive difference in energy costs indicates a reduction in energy cost when walking with support. The effect of support is significantly correlated with walking speed on both treadmill (R^2=0.202; P=.021) and overground (R^2=0.234; P=.012).

appropriate corrective muscle actions to control balance in an energy efficient manner. Handrail hold might facilitate balance control, allowing them to adopt a more economic gait pattern.

Remarkably, we did not find an effect of level of amputation in our study. It is generally believed that an amputation at or above the level of the knee, requiring an artificial prosthetic knee joint, imposes an extra challenge on balance control compared with below knee amputations.[15] Possibly, this lack of effect could be attributed to selection bias. In a convience sample, like the one used in this study, patients with greater ability are more likely to volunteer. Especially for the TF-KD subgroup, people with above average walking ability are likely to be included. This is evidenced by the lack of a difference in self-selected walking speed between the TT and TF+KD subgroups (see table 1). This can also be observed in the scatter graph in figure 2, which demonstrates that the participants with an above knee prosthesis were scattered within the similar range of walking speeds as participants with a below knee prosthesis. The significant correlation between self-selected walking speed and the effect of support suggests that the individual's level of walking ability is an important factor that determines the potential benefit of handrail support in addition to level or cause of amputation.

The positive effect of support on energy cost was only found when walking on the treadmill. This suggests that the treadmill condition would impose a larger demand on balance capacity than overground walking. This is in line with previous findings showing that in able-bodied participants gait stability and underlying gait parameters differed between overground and treadmill walking.[33,34] This might be exacerbated by the lower walking speed selected on the treadmill because it has been shown that gait stability might decrease at slower speeds.[35] Alternatively, it could be argued that the mode of support, that is, handrail vs cane, might have made a difference between treadmill and overground condition. People used a fixed handrail in the treadmill condition as opposed to a freely moveable cane in the overground condition. The cane does require picking up and placing down between subsequent steps. The potential additional effort for carrying the cane seems supported by the fact that nonvascular amputees, on average, even had a higher energy cost when walking with support in the overground condition. The additional energy requirement for carrying the cane might offset its potential effect on balance support. On the other hand, holding on to the fixed handrail obstructs natural arm swing, which might increase energy cost[36,37] and also result in underestimation of the potential effect on balance support. In addition, it should be considered that the handrail might also allow a higher degree of weight bearing and/or exertion of propulsive forces during support and that walking with a unilateral cane only allows support during the prosthetic stance phase, whereas handrail holding allows continuous step by step support. However, recordings from the force sensors in the handrail do not show substantial amounts of weight support or fore-aft I believe. The recordings do show that also with the handrail, support is predominantly taken during the prosthetic stance phase (see supplemental appendix S2, available online only at http://www.archives-pmr.org/). Hence, it can be argued that the energy cost of walking seems to depend on walking environment and will be largest in more challenging environments, such as on a treadmill.

Study limitations

Some limitations of this study should be considered. We only analyzed independent effects of cause and level of amputation and walking speed. Statistical power of the study did not allow us to analyze their interactions. Furthermore, the assumption of normality was violated for 1 of the experimental conditions (treadmill with support). Nevertheless, we chose to apply ANOVA to test the results because this test has been shown to be quite robust against this violation,[38] and potential removal of an outlier responsible for this violation did not affect the conclusion. Sample size was calculated from a previous study on patients with stroke and only on the main effect of support. This might affect statistical power of this study, especially for subgroup analysis.

Walking environment affected the results. We included trials on a treadmill as well as overground walking and found effects of support in the potentially more challenging treadmill condition but not in the overground condition. It is not self-evident which condition is most reflective of normal day walking. Although in daily life people do generally not walk on a treadmill, how often people walk in a straight corridor without any obstacles or distractions can also be questioned. The effect of support should be investigated in more realistic daily environments in the future.

Finally, it cannot be ascertained that the observed changes in energy cost can indeed be attributed to improved balance control only because using a handrail or cane could affect other features of gait, such as weight bearing or propulsion. However, as demonstrated in supplemental appendix S2 (available online only at http://www.archives-pmr.org/), analysis of handrail forces did not reveal substantial amount of weight bearing or propulsive forces. In addition, preliminary inspection of step parameters from the instrumented treadmill (see supplemental appendix S2, available online only at http://www.archives-pmr.org/) reveals that changes in step length and width can be observed that are in line with increased balance control.[35] Hence, facilitation of balance control seems an important underlying factor that explains reduced energy cost with handrail hold on the treadmill, but this should be investigated in more detail in the future.

Conclusions

In conclusion, this study reveals that providing support can reduce the increased energy cost of walking in people after vascular amputation while walking in the challenging environment of the treadmill. Therefore, we propose that support aids should be considered when aiming to improve walking economy, especially for people with a vascular amputation and limited walking ability.

Suppliers

a. Oxycon Mobile; CareFusion.
b. C-Mill; ForceLink, Culemborg.
c. Force sensors; AMTI.

Keywords

Gait; Rehabilitation; Oxygen consumption; Amputation; Lower extremity; Artificial limbs

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Appendix I. Spatiotemporal gait parameters

In order to assess the effect of support on balance control, we analyzed spatiotemporal step parameters during walking on the treadmill. Step pattern adaptation provide an important strategy for dynamic balance control as it influences margins of stability and therefore robustness to perturbations[1,2]

Data analysis

Step parameters were calculated for treadmill trials only. During treadmill trials, participants walked on an instrumented treadmill with an embedded force plate (C-Mill, ForceLink, Culemborg, the Netherlands; size: 1m x 1.5m; sampling rate 100Hz). From this force plate, center of pressure positions in anteroposterior and mediolateral direction (COP_{AP} and COP_{ML} respectively) were recorded. The characteristic butterfly pattern of the COP_{AP} against COP_{ML} plots served to identify initial contact and toe-off through peak detection following a previously described method[3]. Instances of initial contact were used to determine mean and variability of step time, step length and step width and step length and step time symmetry. Step time was defined as the time between two consecutive initial contacts, and stride time was calculated by taking the sum of the corresponding left and right step times. Stride length and step length were derived by multiplying stride or step time by belt speed and correcting for the difference in COP_{AP} position between the two respective initial contacts. Step width was defined as the absolute difference in COP_{ML} position between two consecutive initial contacts. Temporal and spatial step symmetry were defined as $(2 \cdot NP/(NP + P)) * 100$, where *NP* and *P* are the step time / length of the non-prosthetic and prosthetic leg, respectively. A value of 100% indicates perfect symmetry, while a value > 100% indicates a higher value for the non-prosthetic leg, and a value < 100% indicates a higher value for the prosthetic leg. We were unable to accurately identify foot contacts on the treadmill for one participant, therefore this participant was left out of all analyses involving gait parameters.

Statistical analysis

To evaluate the effect of handrail hold on the treadmill on the step parameters, a repeated measures Multivariate Analysis of Variance (rMANOVA) was used. Significant effects of the rMANOVA were followed up by univariate equivalents. Partial Eta squared η^2_p, was used as estimate of effect size.

Results

Handrail support had a significant effect on gait parameters ($F(17)=39.00$, $p=.000$ $\eta^2_p=.948$). Univariate testing showed an increase in stride time and length, a decrease in step width and improved step time and length symmetry with handrail support (Figure 1,

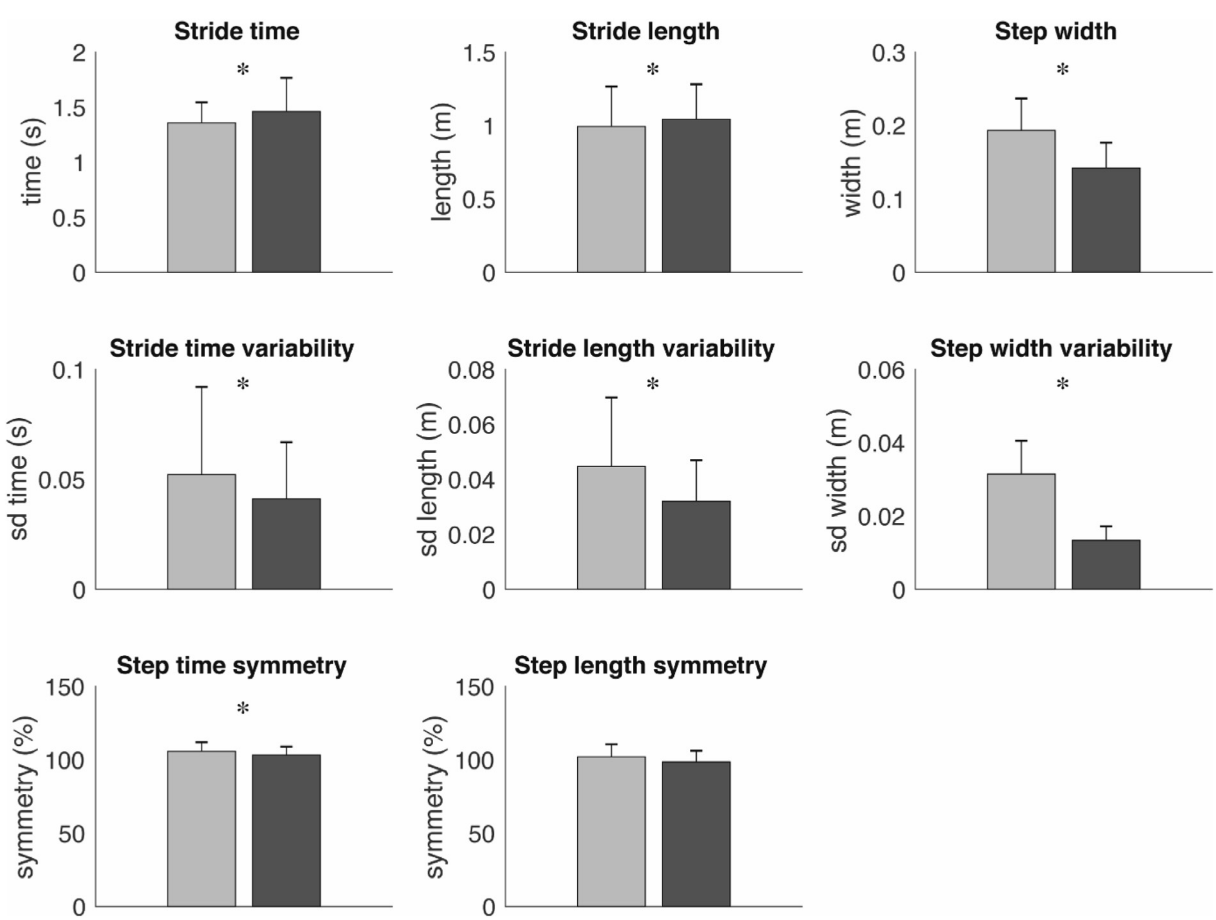

Figure AI.1 Effect of handrail hold on gait parameters during treadmill walking (N=25). White bars represent the average value for unsupported walking, grey bars for supported walking. One-sided error bars represent the standard deviation. * indicates significant difference $p<.05$.

Table AI.1 Effect of handrail hold on gait parameters during treadmill walking (N=25)

Participant Characteristics	Total Group (N=26)	Nonvascular (n=17)	Vascular (n=9)	*P* Value Cause	TT (n=16)	TF/KD (n=10)	*P* Value Level
Level of amputation (TT/TF+KD)*	16/10	10/7	6/3	.696	.	.	.
Etiology (nonvascular/vascular)*	17/9	.	.	.	10/6	7/3	.696
Sex (M/F)*	23/3	14/3	9/0	.400	13/3	10/0	.245
Age (y), mean ± SD	58.7±13.7	57.0±15.7	61.8±8,6	.408	61.3±15.4	54.4±9.7	.217
BMI, mean ± SD	27.0±4.3	25.7±3.7	29.6±4.4	.027†	26.3±4.0	28.2±4.8	.284
Body mass (kg), mean ± SD	88.7±18.5	81.9±13.4	101.6±20.6	.007†	85.9±15.5	93.1±22.7	.345
Time since amputation (y), median (range)*	6.8 (0.3-58)	15.0 (2-58)	1.5 (0.3-10)	.008†	5.3 (0.3-58)	8.5 (1-43)	.780
ABC score (%), mean ± SD	74.1±23.3	76.8±21.2	69.0±27.4	.433	75.4±23.7	72.1±23.8	.733
Preferred walking speed treadmill (m/s), mean ± SD	0.74±0.30	0.81±0.26	0.60±0.33	.085	0.79±0.32	0.65±0.25	.245
Preferred walking speed overground unsupported (m/s), mean ± SD	1.14±0.26	1.25±0.16	0.94±0.29	.002†	1.18±0.24	1.08±0.28	.323

Abbreviations: ABC, Activities-specific Balance Confidence; BMI, body mass index (calculated as weight in kilograms divided by height in meters squared); F, female; M, male.
* χ^2 test.
† $P<.05$.

Table 1). Variability for stride time and stride length and step width decreased when walking with support.

Conclusion

Handrail hold resulted in changes in step pattern that are indicative of an enhanced state of balance. Overall, participants walked with longer and less wide steps, at a lower cadence with reduced variability and increased symmetry. We can therefore conclude that providing handrail support had a general effect on balance control.

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Appendix II. Handrail support forces

In order to analyze to what extent the handrail was used for balance support or other purposes, such as excessive weight bearing[1] or generating propulsion[2], forces exerted on the instrumented handrail of the treadmill were recorded at 100 Hz, using two 6-DoF force sensors incorporated in the handrail. The handrail was positioned only on the non-prosthetic side of the participant during treadmill walking. Force signals were low pass filtered with a 4th order Savitzky-Golay filter with frame size 41. Subsequently, they were cut between subsequent initial contacts of the non-prosthetic leg, time normalized to 0-100% of the stride cycle and averaged over all strides during the final 2 minutes of the supported trial.

Figure AII.1 shows the group average handrail forces during a complete stride during the support condition on the treadmill. The force recordings demonstrate that participants did not exert substantial forces in anterior-posterior direction (peak force <1% body weight). A small amount of support (peak force < 7% body

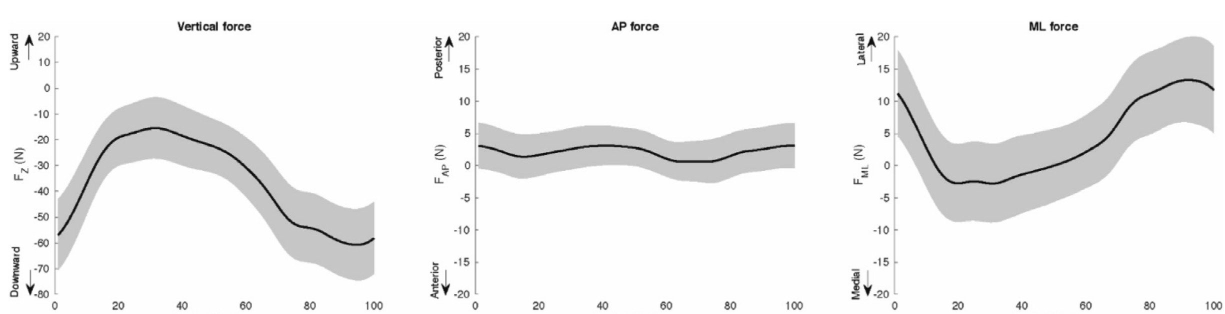

Figure AII.1 Forces exerted on the handrail during the stride cycle in the support condition. Stride cycle starts and ends with initial contact of the non-prosthetic leg. Black line represents the group mean of the ensemble averaged force. Shaded area represents the corresponding standard deviation. Note Forces are expressed as force exerted on the handrail: negative F_z force indicates body weight support, positive FAP indicates propulsion, positive F_{ML} indicates pushing towards the handrail.

weight) was taken in vertical direction, mainly during the stance phase on the prosthetic leg. Small forces were exerted in mediolateral direction (peak force <2% body weight), again mainly during stance on the prosthetic leg.

From these data we conclude that handrail force were mainly exerted to enhance balance control. Potential contribution of body weight support to the reduction in energy cost seems marginal as the vertical impulse is too low to elicit meaningful effects[1] and anterior-posterior forces were negligible.

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