Gait & Posture 40 (2014) 616–621

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Can external lateral stabilization reduce the energy cost of walking in persons with a lower limb amputation?

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ARTICLE INFO

Article history:
Received 26 March 2014
Received in revised form 24 June 2014
Accepted 14 July 2014

Keywords:
Gait
Energy cost
Gait economy
Lower limb amputation
External lateral stabilization
Balance control

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 difference in 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.

1. 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 9and 33% for transtibial, and 66 and 100% for transfemoral amputees [1–3]. 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 [4]. 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 [5]. However, correlations between these adaptations and the elevated energy cost of walking are moderate at best [4,6,7], suggesting a role for other factors, possibly not directly related to forward progression. One such factor could be the impaired balance control in LLA [8,9]. While the energy demand of the motor responses associated with balance control is relatively low in healthy

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http://dx.doi.org/10.1016/j.gaitpost.2014.07.013
0966-6362/

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 [1].

Especially in the frontal plane, the most unstable direction during walking, active feedback control appears necessary to ensure stability [10,11]. Primary strategies for medio-lateral balance control are a stepping strategy, a lateral ankle strategy, and a hip strategy [12]. 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 [13].

These impairments can be dealt with by taking wider steps to ensure a sufficient margin of stability [14]. Indeed, an increase in step width has been observed in LLA compared to controls, with larger increases for transfemoral amputees [12,15–17]. 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 [18,19]. 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 [20,21] due to increased mechanical work to redirect the center of mass from side-to-side [20–22], or increased muscle activity to ensure adequate foot placement [23].

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. [24] 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% [24–27]. 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.

2. Method

2.1. 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 min 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 1. 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.

2.2. Study protocol

Subjects completed two 5-min 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 min 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 min in a seated position after 10 min of rest. Thereafter, both experimental conditions were practiced at a comfortable speed for 3 min to familiarize subjects with the experimental conditions. Subsequently, the subjects' preferred walking speed (PWS) was determined without stabilization following a previously described protocol [28]. This PWS was used in both experimental trials.

2.3. 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 (Fig. 1). 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 [26], 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 (~18.5 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 [25].

2.4. 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

Table 1
Descriptive characteristics of study population.

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

* Significantly different from CO.
† Significantly different from TT.
a Cause of amputation: traumatic, vascular or other (e.g. bacterial, cancer) in number of subjects.