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Adaptation to walking with an exoskeleton that assists ankle extension

S. Galle *, P. Malcolm [1], W. Derave [2], D. De Clercq [3]

Department of Movement and Sport Sciences, Ghent University, Watersportlaan 2, B-9000 Ghent, Belgium

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ABSTRACT

The goal of this study was to investigate adaptation to walking with bilateral ankle-foot exoskeletons with kinematic control that assisted ankle extension during push-off. We hypothesized that subjects would show a neuromotor and metabolic adaptation during a 24 min walking trial with a powered exoskeleton. Nine female subjects walked on a treadmill at $1.36 \pm 0.04\ \mathrm{ms}^{-1}$ during 24 min with a powered exoskeleton and 4 min with an unpowered exoskeleton. Subjects showed a metabolic adaptation after 18.5 ± 5.0 min, followed by an adapted period. Metabolic cost, electromyography and kinematics were compared between the unpowered condition, the beginning of the adaptation and the adapted period. In the beginning of the adaptation (4 min), a reduction in metabolic cost of 9% was found compared to the unpowered condition. This reduction was accompanied by reduced muscular activity in the plantarflexor muscles, as the powered exoskeleton delivered part of the necessary ankle extension moment. During the adaptation this metabolic reduction further increased to 16%, notwithstanding a constant exoskeleton assistance. This increased reduction is the result of a neuromotor adaptation in which subjects adapt to walking with the exoskeleton, thereby reducing muscular activity in all leg muscles. Because of the fast adaptation and the significant reductions in metabolic cost we want to highlight the potential of an ankle-foot exoskeleton with kinematic control that assists ankle extension during push-off.

1. Introduction

Because of the frequent use of walking in daily life and the walking recommendations to increase quality of life [1], a robotic exoskeleton that assists walking has the potential to become an assistive device in disabled people or a performance-increasing tool when walking endurance is challenged. Although the technical aspect of exoskeletons and powered prostheses is developing rapidly, little attention is paid to how people interact with exoskeletons [2–4] whilst this is an unavoidable step in the development of an exoskeleton for daily use. In this context, researchers attribute great potential to pneumatically powered ankle-foot exoskeletons [2,3]. These exoskeletons consist of a shell around the lower leg and can be powered by compliant pneumatic actuators, operating as artificial muscles. Similar to human skeletal muscles these actuators can contract, thereby assisting ankle extension when origin and insertion are located at the rear side of the exoskeleton.

The metabolic cost of walking with a powered exoskeleton can be lower than that of walking with an unpowered exoskeleton. Sawicki and Ferris [5–7] found reductions up to 13% for a bilateral exoskeleton with a control algorithm based on *m.* soleus electromyography (EMG) after an adaptation period up to 90 min. A drawback of these EMG-based control algorithms is that they need a relatively long adaptation period. This could be explained by the continuous interaction between the control algorithm (based on EMG), and the neuromotor adaptation to walking with the exoskeleton that will alter these EMG signals. Indeed, Norris et al. [8] reported an increased metabolic efficiency of 15% for powered versus unpowered walking after only 7 min of adaptation, using a more simple kinematic control method that assisted ankle extension. Malcolm et al. [9] showed that the kinematic control method can be done based on timing of stance phase with metabolic reductions up to 17%. One study compared a kinematic control method with an EMG control method [4] and favored EMG but their kinematic control method was activated too early during stance according to the findings of Malcolm et al. [9].

There seems a discrepancy between the two control methods and although most attention is faced toward EMG control the theoretical modeling of Kuo [10], which is supported by findings of Malcolm et al. [9], highlights the potential of an exoskeleton with kinematic control and optimal actuation timing. However, adaptation to an exoskeleton with kinematic control is not yet investigated in detail. Therefore, we aimed to investigate the

* Corresponding author. Tel.: +32 9 264 94 37; fax: +32 9 264 86 64.
E-mail addresses: samuel.galle@ugent.be (S. Galle), philippe.malcolm@ugent.be (P. Malcolm), wim.derave@ugent.be (W. Derave), dirk.declercq@ugent.be (D. De Clercq).
[1] Tel.: + 32 9 264 86 59; fax: +32 9 264 86 64.
[2] Tel.: + 32 9 264 63 26; fax: +32 9 264 86 64.
[3] Tel.: + 32 9 264 63 22; fax: +32 9 264 86 64.

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adaptation to simple bilateral ankle-foot exoskeletons with ankle extension assistance during a 24 min walking trial [11] in terms of the metabolic adaptation, the neuromotor adaptation, and the interaction between both. We hypothesized that subjects will show a neuromotor adaptation in their kinematics and muscle recruitment, and as such have accommodated metabolically within this adaptation period.

2. Methods

2.1. Subjects

Nine healthy female subjects (age 20.8 ± 0.7 years; body mass 61.4 ± 4.9 kg; stature 167.8 ± 5.0 cm; mean ± s.d.) participated in the experiment. They had no previous experience with exoskeleton walking. All participants signed an informed consent, approved by the ethical committee of the Ghent University hospital.

2.2. Exoskeletons

The bilateral exoskeletons were constructed similar to other ankle-foot exoskeletons [4–7,11–13]. They had a weight of 0.76 kg each and fitted in running shoes where footswitches were build in (IP67, Herga Electric, Suffolk, UK). Based on footswitch signals, a computer program (Labview, National Instruments, Austin, TX, USA) triggered onset and offset of the pneumatic muscles [14] actuation by means of a feedforward algorithm. Actuation timing of the pneumatic muscles was set at 43% and deactivation timing at 63% of stride [9].

2.3. Protocol

Each subject completed a powered and unpowered condition in randomized order. Treadmill was set at a constant dimensionless speed of 0.47 (e.g. 1.36 ms^{-1} for a leg length of 0.90 m) [15]. In the powered condition, subjects walked for 24 min as this was the adaptation period reported by Gordon and Ferris [11] for a unilateral exoskeleton with EMG control. In the unpowered condition subjects walked for 4 min as pilot testing showed that this was sufficient to metabolically adapt to unpowered walking.

2.4. Data collection

Lower limb kinematics were analyzed by 20 reflective markers on the right lower limb, recorded with 11 infrared camera's (200 Hz; Pro Reflex, Qualisys AB, Gothenburg, Sweden) and Qualisys software. EMG was measured (200 Hz; Zerowire, Noraxon, Scottsdale, AZ, USA) with surface electrodes that were placed in accordance with SENIAM guidelines [16] on the *m.* soleus (bilateral), medial *m.* gastrocnemius (bilateral), *m.* tibialis anterior (bilateral), *m.* vastus lateralis (right leg) and *m.* rectus femoris (right leg). Because of difficulties with electrode placement beneath the exoskeleton, EMG measurements were successful in only 7 subjects. Spatiotemporal parameters were calculated based on recordings of a high speed camera (200 Hz; Bassler, Ahrensburg, Germany) with MaxTraq software (Innovasion Systems, Columbiaville, MI, USA). Tensile force of the right pneumatic muscle was measured with a loadcel (200 Hz; A.L. Design, Buffalo, NY, USA). Kinematics, EMG, high-speed video and loadcell data were synchronized and collected during 10 s every minute. O_2 consumption and CO_2 production were recorded during the entire experiment (Oxycon Pro, Jaeger GMBH, Höchberg, Germany).

2.5. Data processing

Metabolic data were calculated with the formula of Brockway [17]. Resting metabolic cost was subtracted from metabolic data in the walking conditions and normalized by bodyweight to calculate net metabolic cost. During the powered condition the lowest 2-min-average was considered the minimum. The average of the individual quadratic regressions (Fig. 1) of the metabolic cost every 30 s during the powered condition illustrates the adaptation (the time to reach the minimum), followed by the adapted period once the minimum was reached.

Reflective marker data were filtered with a 4th order Butterworth low-pass filter (cutoff frequency 6 Hz) and sagittal lower limb kinematics were calculated in Visual 3D (C-Motion, MD, USA). Ankle and knee angles for the right leg were time-normalized from right heel contact to right heel contact (Fig. 2). Moment arm of the pneumatic muscle was calculated as the distance between the ankle joint and the pneumatic muscle of the right leg, which varied over the gait cycle. Torque of the right pneumatic muscle was calculated by multiplying pneumatic muscle force with moment arm. We calculated mechanical power of the exoskeleton by multiplying torque with ankle joint angular velocity and divided this by body mass. Exoskeleton mechanical power was also time-normalized from right heel contact to right heel contact (Fig. 2).

EMG data were rectified and moving root mean square (RMS) was taken to create linear envelopes which were averaged and time-normalized from right heel contact to right heel contact (Fig. 3). The mean RMS per stride was a measure for the activity of the muscles per distance as the absolute speed in the powered and unpowered condition was similar between subjects (1.36 ± 0.04 ms^{-1}). To express changes in EMG amplitude, average values of the linear envelop per stride were normalized to the average of the unpowered condition (Fig. 4).

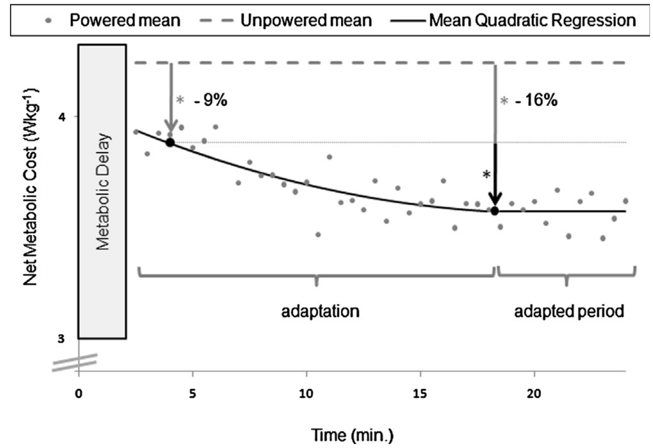

Fig. 1. Net metabolic cost. Gray dots represent the mean net metabolic cost (30 s values) during the 24 min powered condition. Black solid line represents the mean quadratic regression over these 30 s values during the powered condition until the minimum is reached after 18.5 ± 5 min. The time to reach this minimum is called the adaptation, the time period after this minimum is called the adapted period. Gray dashed line represents the mean net metabolic cost of min 2–4 in the unpowered condition. Black dotted line represents the mean net metabolic cost of min 2–4 in the 24 min powered condition. Percentages represent reductions in net metabolic cost in the beginning of the adaptation (min 2–4) and in the adapted period (min 16.5–18.5) compared to the unpowered condition (min 2–4). Gray box represents the metabolic delay until steady state is achieved. * = sign. different from the beginning of the adaptation (ANOVA, $p \leq 0.05$); * = sign. different from the unpowered condition (ANOVA, $p \leq 0.05$).

Repeated measures ANOVA was done with SPSS Statistics 20 (IBM, Armonk, NY, USA) with Tukey Honestly Significant Difference post hoc tests ($p \leq 0.05$) to compare metabolic data (2-min-averages), spatiotemporal parameters (stride time and stance time), kinematics (mechanical power, ankle and knee angle every 10% of stride and instant of peak values) and EMG (instant of peak activation and stride averages) between three conditions: the unpowered condition (metabolic cost of min 2–4 in combination with EMG and kinematic data of the fourth min of the unpowered condition), the beginning of the adaptation (metabolic cost of min 2–4 in combination with EMG and kinematic data of the fourth min of the powered condition) and the adapted period (metabolic cost of min 16.5–18.5 in combination with EMG and kinematic data of the 24th min of the powered condition).

3. Results

Subjects reached the minimum for metabolic cost in the 24 min powered condition after an average adaptation period of 18.5 ± 5.0 min. No subject showed the lowest metabolic cost in the last 2 min. After the adaptation a steady state in metabolic cost was achieved during the adapted period (Fig. 1), as net metabolic cost after the average adaptation time (3.56 ± 0.70Wkg^{-1}; min 16.5–18.5) was not significantly different from the end of the powered condition (3.55 ± 0.68Wkg^{-1}; min 22–24; $p = 0.99$). Therefore, net metabolic cost after the average adaptation time was taken as a reference for the entire adapted period (Fig. 1).

Net metabolic cost in the unpowered condition (4.25 ± 0.59 Wkg^{-1}; min 2–4) was significantly higher than at the beginning of the adaptation (3.88 ± 0.73 Wkg^{-1}; min 2–4; $p \leq 0.05$) or than in the adapted period (3.56 ± 0.70 Wkg^{-1}; min 16.5–18.5; $p \leq 0.05$). During the adaptation the net metabolic cost significantly decreased from the beginning of the adaptation (3.88 ± 0.73 Wkg^{-1}; min 2–4) to the adapted period (3.56 ± 0.70 Wkg^{-1}; min 16.5–18.5; $p \leq 0.05$).

Stride time did not differ between the unpowered condition (1.06 ± 0.05 s), the beginning of the adaptation (1.06 ± 0.05 s) and the adapted period (1.08 ± 0.05 s). Differences in stance time were