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Effects of general fatigue induced by incremental maximal exercise test on gait stability and variability of healthy young subjects

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ABSTRACT

The purpose of this study was to determine whether general fatigue induced by incremental maximal exercise test (IMET) affects gait stability and variability in healthy subjects. Twenty-two young healthy male subjects walked in a treadmill at preferred walking speed for 4 min prior (PreT) the test, which was followed by three series of 4 min of walking with 4 min of rest among them. Gait variability was assessed using walk ratio (WR), calculated as step length normalized by step frequency, root mean square (RMS_{ratio}) of trunk acceleration, standard deviation of medial-lateral trunk acceleration between strides (VAR_{ML}), coefficient of variation of step frequency (SFCV), length (SLCV) and width (SWCV). Gait stability was assessed using margin of stability (MoS) and local dynamic stability (λ_s). VAR_{ML}, SFCV, SLCV and SWCV increased after the test indicating an increase in gait variability. MoS decreased and λ_s increased after the test, indicating a decrease in gait stability. All variables showed a trend to return to PreT values, but the 20-min post-test interval appears not to be enough for a complete recovery. The results showed that general fatigue induced by IMET alters negatively the gait, and an interval of at least 20 min should be considered for injury prevention in tasks with similar demands.

1. Introduction

The effects of general fatigue on postural control is relatively well established (Paillard, 2012). Exercises such as running, cycling, cycle ergometer, maximal exercise test have been shown to affect postural control (Hoffman et al., 1992; Nardone et al., 1997, 1998; Derave et al., 2002; Nagy et al., 2004; Vuillerme and Hintzy, 2007). Such type of exercises involving the whole body induces general fatigue that deteriorates the sensory proprioceptive and exteroceptive information (Lepers et al., 1997), and decreases muscle force (Nardone et al., 1998). They mobilize a large number of body muscles inducing physiological alterations with important mechanical impacts on the musculoskeletal system that, in turn, decrease the effectiveness of the postural control system, with a recovery interval of approximately 15–20 min (Nardone et al., 1998; Fox et al., 2008).

Energy expenditure increases when performing general muscular exercise, which, in turn, increases liquid movements and cardiac and respiratory muscular contractions, with increasing cardiac and breathing rhythm. These alterations accentuate the amplitude of body sway in upright stance. After an exhaustive physiological exercise such as maximal oxygen uptake test that generates a large post-exercise oxygen deficit, postural control is degraded (Mello et al., 2010). Postural control is also negatively affected after a treadmill walking exercise when its intensity is superior to the lactate accumulation threshold (Nardone et al., 1997).

The metabolites released by the fatiguing muscle exert an inhibitory reflex effect on homonymous α-motoneurons from activation of fatigue-sensitive group III and IV muscles afferents disturbing postural control (Garland and Kaufman, 1995). Dehydration induced by general muscular exercise alters the vestibular function that, in turn, disturbs balance (Derave et al., 1998). In addition, sensory information is altered by general muscular exercise. Prolonged stimulation of visual input during exercises as running or walking influences postural control (Derave et al., 2002). Repetitive vertical and lateral movements of the head during running and walking alter otholitic sensitivity decreasing the contribution of vestibular information for the motor control systems (Derave et al., 2002). Proprioception information is also altered

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during running and walking. Walking involves eccentric actions of the triceps surae (Nardone et al., 1997) and eccentric muscle actions produces more muscle damage and soreness than concentric action (Vissing et al., 2008) that, in turn, deteriorates proprioception and disturbs the senses of force and limb position (Paschalis et al., 2007).

However, there is a lack in literature of studies assessing the effects of general fatigue on gait. The studies of fatigue effects on gait stability have focused on local fatigue of specific muscles. The unilateral hip abductor muscle fatigue increased stride time variability and step-to-step asymmetry in frontal plane of healthy older adults, although gait stability in terms of local dynamic stability was not affected (Arvin et al., 2015). In a study on balance control in perturbed and unperturbed gait, healthy older adults were able to cope with substantial unilateral leg muscle fatigue during gait and demonstrated faster balance recovery in terms of Lyapunov exponents in the fatigued than in the unfatigued condition (Toebes et al., 2014).

The effects of general fatigue observed in postural control studies can influence gait stability and variability after a general muscular exercise because balance and posture control are essential mechanisms in locomotion (Dimitrijevic and Larsson, 1981). The selection of strategies of movement control and of a certain gait pattern may be related to improving gait stability and reducing fall risk rather than minimizing energy cost (Hak et al., 2013), and this can be the case after a strenuous exercise.

Different methods have been proposed to describe gait characteristics. Besides spatiotemporal gait features, it is important to assess gait variability and stability, which are influenced by the ability to optimally control gait from one stride to the next (Hausdorff et al., 2001). The root mean square of trunk acceleration is often used as a measure of gait variability (Sekine et al., 2013), as well as the standard deviation of medial-lateral (ML) trunk acceleration between strides (Dingwell et al., 2001). A method to asses gait stability is the estimation of local dynamic stability (LDS), which is derived from nonlinear dynamic system theory (maximal Lyapunov exponent) (Bruijn et al., 2013). This method assumes that motor control ensures a dynamically stable gait if the divergence remains low between trajectories in a reconstructed state space that reflects gait dynamics.

The purpose of this study was to evaluate the effects of fatigue induced by general muscular exercise on gait variability and stability using spatiotemporal and nonlinear descriptors, and to identify the variables that are sensitive to these effects. We investigate gait stability before and after an incremental maximal exercise test (IMET) and hypothesized that such exercise degrades gait stability and alters gait variability with a recovery interval similar to that observed in postural control studies.

2. Methods

2.1. Subjects

A total of 22 healthy male subjects participated in this study (22.9 ± 3.7 years old, 76.7 ± 10.4 kg, 1.77 ± 0.06 m). All subjects were instructed to avoid any kind of exercise during 24 h prior to the experiment. After written informed consent, the subjects were submitted to protocols previously approved by the Local Research Ethics Committee.

2.2. Equipment

Reflective markers were attached to the lateral malleoli, the heels, the head of second and fifth metatarsals and T1 of all subjects. A rigid squared cluster of four reflective markers spaced (8 cm apart) was attached using a Velcro belt in the midpoint between left and right posterior superior iliac spines. The reflective markers were used for movement registration with a 3D motion capture system using ten infrared cameras operating at 100 Hz (Vicon Nexus, Oxford Metrics, Oxford, UK).

2.3. Protocol

Subjects walked and ran on a level treadmill. First, the preferred walking speed (PWS) was evaluated following previous reported protocol (Kang and Dingwell, 2008). The subjects performed a pre-trial for 6 min of treadmill walking at PWS. The first 2 min were used to become familiar with treadmill walking. The final 4 min were used to collect kinematic data (PreT). Next, the subjects were submitted to an adapted IMET on the treadmill with 0% slope (de Lira et al., 2013). The adapted IMET consisted of the following phases: the subjects began with a 6 min walking at PWS and then the speed was adjusted to 6 km/h, followed by progressive increases of the treadmill speed at a rate of 1 km/h every minute until subject exhaustion.

A rest of 30 s was allowed to minimize dizziness (Nardone et al., 1997), followed by a post-trial (PosT-0) for 4 min of treadmill walking at PWS, 4 min of rest, a second post-trial for 4 min at PWS (PosT-1), 4 min of rest, and a third post-trial for 4 min at PWS (PosT-2). This protocol (Fig. 1) was adopted to asses a possible recovery effect.

Subjects wore a heart-rate meter during the entire experimental session. The subjective exertion level was assessed during the exercise through the Borg 0–20 scale of perceived exertion (Borg, 1982). We considered the exercise suprathreshold when the heart rate exceeded 60% of the maximum heart rate for that subject, estimated on the basis of the following calculation: 220 – age (years) – 0.6 (Wasserman, 1984).

2.4. Data analysis

Before data analysis, except for the calculation of LDS and approximate entropy (ApEn), kinematic data were low-pass filtered with a fourth order, zero-lag Butterworth filter with a cut-off frequency of 6 Hz. All parameters were calculated for 150 strides. First, the initial and final 15 s of each trial were discarded (Hak et al., 2013). Next, all steps were detected as the zero-cross of the heel markers velocity (Zeni et al., 2008). Then, the intermediate 150 strides were selected, discarding the initial and final strides exceeding 150. A customized Matlab code was used for data analysis.

Step frequency (SF) was determined as the inverse of the average duration of the steps. The average step length (SL) was computed from the average step frequency and the average treadmill speed (SL = PWS/SF) (Terrier and Reynard, 2015). Walk ratio (WR) was calculated as SL normalized by SF (WR = SL/SF) and represents what would be SL if SF was equal to 1 step/s (Terrier and Reynard, 2015). Step width (SW) was determined as the ML distance between two subsequent heel-strikes.

Gait variability was assessed using two descriptors: normalized root mean square of trunk acceleration (RMS_{ratio}) (Sekine et al., 2013) and standard deviation of ML trunk acceleration between strides (VAR_{ML}) (Toebes et al., 2015). To compute RMS_{ratio} of trunk acceleration we first calculate the vector norm of the 3D trunk acceleration (Ta) for each sample n (Eq. (1)),

$$Ta_n = \sqrt{ax_n^2 + ay_n^2 + az_n^2} \tag{1}$$

where (ax, ay, az) is 3D trunk acceleration. Next, the RMS of Ta is calculated as follows:

$$Ta_{RMS} = \sqrt{\frac{1}{N}\sum_{n=1}^{N}(L_n)^2} \tag{2}$$