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Trunk muscles activation during pole walking vs. walking performed at different speeds and grades

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ABSTRACT

Given their functional role and importance, the activity of several trunk muscles was assessed (via surface electromyography—EMG) during Walking (W) and Pole Walking (PW) in 21 healthy adults. EMG data was collected from the external oblique (EO), the erector spinae longissimus (ES), the multifidus (MU), and the rectus abdominis (RA) while performing W and PW on a motorized treadmill at different speeds (60, 80, and 100% of the highest speed at which the participants still walked naturally; PTS_{60}, PTS_{80} and PTS_{100}, respectively) and grades (0 and 7%; $GRADE_0$ and $GRADE_7$, respectively). Stride length, EMG area under the curve ($_{AUC}$), muscles activity duration ($_{ACT}$), and percentage of coactivation (CO-ACT) of ES, MU and RA, were calculated from the averaged stride for each of the tested combinations.

Compared to W, PW significantly increased the stride length, EO_{AUC}, RA_{AUC} and the activation time of all the investigated muscles, to different extents depending on treadmill speeds and grades. In addition, MU_{AUC} was higher in PW than in W at $GRADE_0$ only (all speeds, $p < 0.01$), while ES_{AUC} during W and PW was similar at all the speeds and grades. These changes resulted in longer CO-ACT in PW than W, at $GRADE_0$-PTS_{100} ($p < 0.01$) and $GRADE_7$ (all speeds, $p < 0.01$). In conclusion, when compared to W, PW requires a greater engagement of the abdominal muscles and, in turn, a higher control of the trunk muscles. These two factors taken together may suggest an elevated spinal stability while walking with poles.

1. Introduction

Pole walking (PW) is a walking-based physical activity that implies the use of a pair of poles in opposition to the lower limb locomotion [1]. This activity has proved to be effective in maintenance/improvement of the cardiovascular system function [1] and, when compared to walking (W), to increase both heart rate and oxygen consumption to a higher extent [2]. Nevertheless, the muscular responses to PW, which could help understand the potential benefits and/or drawbacks of this exercise mode, have only been partially investigated. When W and PW have been compared, the analysis of the muscle activity revealed that the upper limb muscles are generally more active during uphill PW, while uphill W appears to activate more the lower limb muscles [3]. However, only one study focused on the differences between W and PW at the trunk level, and found the same activation amplitude of the erector spinae longissimus between uphill W and PW while carrying a backpack [4].

The trunk muscles are fundamental for the balance of the whole body, and it is thought that the neuromuscular system acts through their coactivation to provide adequate spinal stability in different conditions [5]. They also assist the movement of the arms and legs during locomotion and other physical activities [6], and modulate their activity and function according to a specific task (e.g. changing the W speed) [7]. For instance, while W the erector spinae muscles preserve the body balance perturbed by arm swing [8] and anticipate and support the pelvis movements [9]. Conversely, the external oblique muscles switch their activation pattern from tonic to phasic in a speed-dependent way, reflecting both their stabilizing and mobilizing role during W [7].

Given the multiple functions and overall importance of this muscle groups, it is pivotal to examine the role of trunk muscles during PW and how this compares to W. Elucidating the effect of speed and grade on the activity of trunk muscles will provide additional insights into their neuromuscular response to these two modes of human locomotion. Accordingly, this study aimed to concurrently measure and describe the activity of several trunk

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muscles in a healthy population whilst performing both PW and W at different speeds and grades.

2. Methods

2.1. Participants

Ten healthy males (age: 28.5 ± 5.6 y, mass: 78.3 ± 9.5 kg, height: 1.77 ± 0.06 m) and eleven females (age: 33.0 ± 10.1 y, mass: 66.2 ± 7.5 kg, height: 1.68 ± 0.09 m) were recruited. The study was approved by the Ethical Research Committee of the Sports, Health and Exercise Science Department of the University of Hull (UK). The participants signed a written informed consent before their inclusion in the study and had to fill in a pre-exercise medical questionnaire. All participants were free from chronic low-back pain and were asked to rest the day before each testing session.

2.2. Experimental procedure

Participants attended a minimum of two testing sessions, at least 24 h apart. During the first visit, the preferred transition speeds (PTS), i.e. the highest speed at which the participants still walk naturally, (at both 0% ($GRADE_0$) and 7% ($GRADE_7$) grades were determined ($GRADE_0$ mean ± SD: 1.96 ± 0.16 m/s; $GRADE_7$ mean ± SD: 1.84 ± 0.14 m/s) on a motorized treadmill (Pulsar—h/p/cosmos Sports & Medical, Nussdorf-Traunstein, Germany). The PTS was identified using a modified version of the Hreljac's protocol [10]: each stage duration was set at 20 s and the speed increment/decrement during each trial was set at 0.2 km/h. In the same testing session, the participants were familiarized to PW, but additional familiarization sessions were planned if required to meet the following criteria: walk fluently while looking forward; keeping the poles inclined backwards with the elbows slightly flexed; extending the arms behind the body at the end of the pushing phase. Because different PW techniques exist [1], these criteria were chosen as they are those mainly met by nordic walkers [11], thus allowing a similar PW technique across the participants.

During the last visit, surface EMG data was collected, on the dominant side (defined by asking the participant which foot they would use to kick a ball [12]), from the external oblique (EO), the erector spinae longissimus (ES), the multifidus (MU) and the rectus abdominis (RA). After the equipment setup, the baseline EMG activity was collected with the participants standing still for 30 s. Then, a 5-min warm-up (PW-$GRADE_0$-60% PTS) was performed prior to four randomized tests combining either W or PW with $GRADE_0$ and $GRADE_7$. Each test required three 1-min bouts of exercise at 60, 80 and 100% of the PTS (PTS_{60}, PTS_{80} and PTS_{100}, respectively). Pilot tests revealed that the PTS_{60} and PTS_{80} trials were generally well tolerated by the participants (6–14 range of the Borg's 6–20 rate of perceived exertion scale; RPE [13]). Conversely, the PTS_{100} trials were more challenging (13–17 RPE range). Therefore, to reduce the effect of fatigue, 1-min rest was allowed after the PTS_{60} and PTS_{80} trials, whereas, at the end of the PTS_{100} trials, the participants sat until the heart rate dropped to the resting value (measured for 5 min while sitting before the warm-up). The recovery was assumed to be completed when the heart rate was steadily within the resting value ± 5 bpm for at least 1 min.

2.3. Equipment setup

A heart rate monitor (RS800CX—Polar Electro Oy, Kempele, Finland) was worn by the participants during all the testing sessions.

The study was conducted using a pair of trekking telescopic aluminium poles (Forclaz 500—Quechua, Passy, France) with adjustable wrist straps and hard rubber covers at the distal ends, specifically made to allow to incline the poles backwards. The pole length was adjusted to each participant's body size [14].

After the skin was shaved, slightly abraded and cleaned with an alcohol swab, EMG electrodes (BlueSensor N—Ambu, Copenaghen, Denmark) were placed and secured parallel to the muscles fibres (with 2 cm inter-electrode distance) as follows: EO, 3 cm anterior to the mid-point of the line between the lateral pelvic crest and the lateral lower ribcage margin [15]; ES, 2 cm apart of the spinal process of L1 [16]; MU, about 2 cm apart from the back midline at L5 level [16]; RA, 86% of a line parallel to the linea alba (approximately 2 cm apart) starting from the xiphoid process and ending at level of the superior anterior iliac spine [17]. A 24 G tri-axial accelerometer was placed and secured on the dominant tibia mid-way of the line between its medial condyle and the lateral malleolus.

2.4. Data collection and processing

All data was collected synchronously (sampling rate: 1500 Hz; input impedance: >100 MΩ; CMRR: >100 dB; baseline noise: <1 μV RMS; base gain: 200; final gain: 500) and stored on a computer using a 16 bit resolution wireless system (Desktop DTS—Noraxon USA Inc., Scottsdale, Arizona, USA).

Raw EMG data was processed firstly applying a 2nd order, phase-corrected, band-pass Butterworth filter with bandwidth cut-off of 10–500 Hz. Secondly, the heart beats artefacts were removed by a 2nd order, phase-corrected, high-pass Butterworth filter with cut-off of 30 Hz [18]. Thirdly, the Teager–Kaiser energy operator was applied in order to improve the muscles activation onset detection during the subsequent analysis [19] (see below). Finally, the signal was full-wave rectified and the linear envelope was obtained through a 2nd order, phase-corrected, low-pass Butterworth filter with cut-off frequency of 10 Hz [20].

The static accelerometer tilt was corrected as described by Kavanagh [21], then the anterior-posterior accelerations of the tibia were used for strides detection [21] (see the appendix in supplementary material for the MATLAB code).

For each trial, the central 30 consecutive strides were selected and time-normalized to 101 points prior to the calculation of their point-by-point average. For each participant, the 12 average strides obtained (resulting from the combination of two locomotion types, two treadmill grades, and three speeds) were normalized to the peak of the average stride representing the PW condition at $GRADE_0$ and PTS_{100}.

The average stride length was calculated as the product of the treadmill speed with the average stride duration. The area under the curve of the EMG signal of the normalized average stride was computed, using the trapezoid method, for each muscle (EO_{AUC}, ES_{AUC}, MU_{AUC}, RA_{AUC}) as measure of their EMG amplitude. The time at which each muscle was active during each stride (EO_{ACT}, ES_{ACT}, MU_{ACT}, RA_{ACT}) has been calculated as the percentage of the stride duration at which the normalized signal was higher than the baseline mean value plus 7 standard deviations [19]. Finally, the coactivation time (CO-ACT) of flexors and extensors muscles of the spine was calculated as the percentage of the average stride duration at which at least one of the trunk extensors (ES, MU) and RA were active at the same time [20].

2.5. Statistical analysis

Generalized estimating equations were used to test the effects of locomotion type, treadmill grade and speed on the calculated parameters, as this approach does not require distributional assumptions of the data [22]. For each dependent variable, robust sandwich standard errors were calculated and the model's distribution family, link function and working correlation matrix