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Original research

Relationship between metabolic cost and muscular coactivation across running speeds

Isabel S. Moore [a,b,*], Andrew M. Jones [b], Sharon J. Dixon [b]

[a] Sports Injury Research Group, Cardiff School of Sport, Cardiff Metropolitan University, UK
[b] Bioenergetics and Human Performance Research Group, Sport and Health Sciences, University of Exeter, UK

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ABSTRACT

Objectives: Muscular coactivation can help stabilise a joint, but contrasting results in previous gait studies highlight that it is not clear whether this is metabolically beneficial. The aim was to assess the relationship between the metabolic cost of running and muscular coactivation across different running speeds, in addition to assessing the reliability and precision of lower limb muscular coactivation.
Design: Eleven female recreational runners visited the laboratory on two separate occasions. On both occasions subjects ran at three speeds (9.1, 11 and 12 km h^{-1}) for six minutes each.
Methods: Oxygen consumption and electromyographic data were simultaneously recorded during the final two minutes of each speed. Temporal coactivations of lower limb muscles during the stance phase were calculated. Five muscles were assessed: rectus femoris, vastus lateralis, biceps femoris, tibialis anterior and gastrocnemius lateralis.
Results: Nonparametric correlations revealed at least one significant, positive association between lower limb muscular coactivation and the metabolic cost of running for each speed. The length of tibialis anterior activation and muscular coactivation of the biceps femoris-tibialis anterior and gastrocnemius lateralis-tibialis anterior decreased with speed.
Conclusions: These results show that longer coactivations of the proximal (rectus femoris-biceps femoris and vastus lateralis-biceps femoris) and leg extensor (rectus femoris-gastrocnemius lateralis) muscles were related to a greater metabolic cost of running, which could be detrimental to performance. The decrease in coactivation in the flexor and distal muscles at faster speeds occurs due to the shorter duration of tibialis anterior activation as speed increases, yet stability may be maintained.

1. Introduction

Muscular coactivation, or cocontraction, concerns the simultaneous contraction of a pair of muscles. It has been argued that such muscular coordination can help stabilise a joint during locomotion.[1] Stability produced in this way can contribute to increased stiffness in the lower limb during dynamic movements.[2]

Biomechanically, coactivation has been proposed as a metabolically efficient muscular coordination during running.[3] It is suggested that coactivation can make a runner's storage of elastic energy more efficient. For example, Heise et al.[3] reported lower oxygen consumption to be related to greater coactivation between the rectus femoris and gastrocnemius, during the stance phase of running, for female runners when performing at self-selected speeds. Both these muscles are biarticular, meaning they cross two joints, Heise et al.[3] found that this coactivation across multiple joints had a stronger relationship with metabolic cost of running (Cr) than did activation of a single muscle. They concluded by suggesting that this activation strategy may decrease Cr. However, this suggestion is only partially supported by their earlier findings examining coactivation, which demonstrated similar relationships that did not attain significance.[4]

On the other hand, physiologically it has been argued that coactivation is an inefficient process that actually increases the metabolic cost of dynamic movement.[5–7] For example, studies utilising standardised speeds of locomotion have reported that greater coactivation in the lower limbs contributes to a higher metabolic cost for elderly individuals whilst walking compared to younger individuals.[5] Furthermore, Frost and colleagues[6,7] reported greater coactivation to be associated with higher metabolic rates for both walking and running in children. Whilst they investigated a variety of speeds, the significant differences in coactivation between age groups occurred at the fastest walking and running speeds.[7] Additionally they found coactivation to be an important predictor of the metabolic rate of both walking and running.[6] Interestingly

* Corresponding author at: Cardiff School of Sport, Cardiff Metropolitan University, Cyncoed, Cardiff CF23 6XD, Wales, UK.
E-mail address: imoore@cardiffmet.ac.uk (I.S. Moore).

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they suggested that the younger children, who employed greater coactivation than the older children, did so for stability purposes despite this making them less metabolically efficient.

There is discrepancy not just between walking[5,8] and running[3] studies in adults, but also between running investigations in children[6,7] and adults[3] regarding whether muscular coactivation is a metabolically beneficial or a detrimental strategy with the potential to either enhance or impair running performance. Furthermore whilst walking at increasing speed results in greater coactivation in adults,[8] the effect of running at greater speeds on coactivation in adults is not yet known.

The aim of this study was, therefore, to determine coactivation across different, standardised running speeds and assess their relationship with Cr. It was hypothesised that coactivation and Cr would be positively related to one another, such that greater coactivation is associated with a greater Cr (i.e. higher oxygen cost), and that greater coactivation would occur at faster speeds. Additionally, the reliability of each coactivation was analysed by quantifying the inter-day variability.

2. Methods

Eleven female recreational runners (age: 21.8 ± 2.9 years; mass: 60.4 ± 6.6 kg; height: 164.8 ± 4.2 cm) took part in the study. All had a minimum of two years running experience. Before participation all participants provided informed consent and declared themselves to be free from injury. Testing took place during two laboratory visits four days apart. The same protocol was used during both testing sessions. Participants wore their own running shoes throughout testing to remove possible gait alterations that may occur whilst adjusting to different running shoes. Ethical approval was obtained from the Sport and Health Sciences Ethics Committee, University of Exeter.

A familiarisation run on the treadmill was performed before the Cr and EMG data were collected. This was performed for a minimum of 6 min to enable a natural running style to be achieved. Additionally, it served as the participants' warm-up during the first visit. A similar run was performed during the second visit as a warm-up. The measurements were made while participants ran on a level treadmill at three test speeds in the following order: 9.1 km h^{-1} (speed 1), 11 km h^{-1} (speed 2) and 12 km h^{-1} (speed 3). Participants were instructed to run at each speed for six minutes, with 10 min rest periods between consecutive running bouts. EMG data were collected for 20 s towards the end of the 5th minute of each speed. Twenty consecutive strides during this 20 s period were used in the analysis. The oxygen consumption data were recorded during the final two minutes of running at each speed.

Surface EMG (Trigno Wireless EMG, Delsys, Boston, MA, USA; parallel bar configuration, contact material 99.9% Ag, interelectrode spacing 10 mm, electrode size 37 mm × 26 mm × 15 mm) was used to analyse the activation and activity of six lower limb muscles: rectus femoris (RF); vastus lateralis (VL); biceps femoris (BF); gastrocnemius lateralis (GL); and tibialis anterior (TA). The electrodes were placed longitudinally with respect to the muscle fibre direction following standardised criteria recommended by SENIAM. The skin surface area was prepared using an abrasive gel and then wiped clean with an alcohol swab. The electrodes were affixed to the lower limb and permanent marker pen used to outline their placement. This outline was kept on the participant's leg until the next testing session so the electrodes could be positioned in the same location on the second visit. Tight shorts and self-adhesive elastic bandage covered the electrodes to minimise their movement.

The raw EMG signal was amplified and band-pass filtered (20–500 Hz) within the Delsys hardware and recorded at a sampling rate of 4000 Hz and a gain of 1000 times. A personal computer was used for off-line analysis and the storage of data. First the data underwent full-wave rectification and then a linear envelope of the EMG signal was created using the Root Mean Square (RMS). The RMS of the EMG (EMG_{RMS}) was calculated using a 50 ms sliding window.

The duration of coactivation was calculated in a similar manner to the previous work of Heise et al.[3] and has also been used in walking studies.[5,9] Specifically, it was temporally quantified as the common duration during stance of muscle on-time between pairs of muscles. This was then recorded as a percentage of stance. The common duration time for each step was divided by the ground contact time for that respective stance period. In total five muscle pairs were considered; three flexor–extensor (RFBF, VLBF and GLTA), one extensor–extensor (RFGL) and one flexor–flexor (BFTA) pairings. These latter two pairing groups were chosen to examine the relationship of muscular coordination and Cr. Peak EMG_{RMS} during the stance phase of 20 consecutive steps for each of the muscles was identified and the mean peak of each muscle calculated. Data were then normalised to the peak EMG_{RMS} of each particular speed and cut-off thresholds applied to the normalised data to identify the onset and offset of muscular activation. These were determined through a customised MatLab (Math Works Inc., Cambridge, MA, USA) script that identified thresholds for specific muscles. Activation had to exceed and be sustained above a certain amplitude threshold for at least 50 ms. Threshold determination followed a similar procedure to that previously used by Steele and Brown,[10] whereby thresholds from 3% to 25% of peak muscle activity were computed and compared to manually derived thresholds. Following this, the following thresholds were chosen: RF 7%, VL 7%, BF 20%, TA 12%, and GL 7% of the mean peak EMG_{RMS} (Fig. 1).

Stance was determined using the triaxial accelerometer integrated in the surface electrode. One electrode was affixed to the right heel of the participant's running shoe specifically for stance detection. This removed any effect of skin movement that may have resulted from using the electrode positioned on the TA. The vertical component was used to identify touch-down and toe-off. This approach was validated separately by simultaneously collecting force and accelerometer data whilst running over a force plate and wearing the surface electrodes. This identified the period of accelerometer data that related to stance phase.

To measure oxygen consumption, participants were fitted with respiratory apparatus. Breath-by-breath respiratory gas exchange and ventilation were measured with an automated gas analysis system with the mean values displayed every 10 s. (Cortex Metalyser II, Biophysik, Leipzig, Germany). Additionally, heart rate was measured via a wireless chest strap telemetry system (Polar Electro T31, Kempele, Finland) and recorded on the 4th, 5th and 6th minute of each run. Mean oxygen consumption during the final 2 min of running was calculated to represent Cr and mean heart rate was determined from the three measurements over the final 2 min.

Results from day 1 were compared to day 2 to analyse day-to-day reliability across each speed separately, using intraclass correlation coefficients (ICC) (2, k)[11] and the standard error of measurement (SEM). Reliability was classed as strong (ICC > 0.80) or moderate (ICC = 0.60–0.80).[12] Precision was expressed using the SEM value, both in absolute and relative (100 × (SEM of variable/mean of variable)) terms. Non-parametric tests were used in further analysis of data from day 2. To determine the differences in coactivation (relative and absolute terms), individual muscle on-times (relative and absolute terms) and stance time over the three speeds a Friedman's ANOVA was used. Post hoc analysis was conducted using Wilcoxon sign-rank tests. Separate Spearman's rank correlations were performed on the Cr for each speed to assess the relationship between coactivations and Cr. Significance level was set at $p \leq 0.05$. Data analysis was conducted using PASW statistics version 18 (SPSS Inc., Chicago, IL).