THE EFFECTS OF RUNNING VELOCITY AND LOWER EXTREMITY LOADING ON BIARTICULAR LEG MUSCLES DURING TREADMILL RUNNING

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The purpose of this study was to investigate the changes in muscle activity levels in treadmill running while using wearable resistance as a function of running speed. Nine recreational runners participated in this study; they were requested to perform the running without and with wearable resistance at four discrete speeds ranging from 2.5 to 7.0 m/s on a treadmill. The mass of wearable resistance was set at one third of each participant's shank and foot mass. Repeated-measures two-way ANOVA analysis was then used to explore the effect of loaded and speed. The data showed that shank loading enlarged the activation amplitude of the biceps femoris (BF) for the concentric action of hip extension following the stretch of knee extension at a high running speed. The loaded condition facilitated the rectus femoris (RF) to be stretched.

KEYWORDS: ELECTROMYOGRAPHY, KINEMATICS, HIP, KNEE, GAIT.

INTRODUCTION: Running is a form of repetitive movement that involves the cycle of the stance-swing phase of the lower limb. The muscles of the lower extremities function repetitively to produce specific features of the electromyography signals for joints movements. Albertus-Kajee et al. (2011) found the activations of the rectus femoris (RF) and biceps femoris (BF) increased as running speed increased. The RF and BF are two-joint muscles that cross the hip and knee joints and play the roles of hip flexor/knee extensor and hip extensor/knee flexor, respectively. The BF is one of the hamstrings complex that performs the eccentric action to decelerate the knee extension in the late swing phase and the concentric action on hip extension in the early stance phase (Howard, Conway, & Harrison, 2018; Yu et al., 2008). The RF is active from the late swing phase (pre-activation) to the early stance phase for the knee extension. However, the RF is the important hip flexor that carries the thigh forward in the early swing phase (Mero & Komi, 1987).
Resistance training is a common strategy to increase load for the enhancement of strength and/or endurance, and wearable resistance training is a conformable sport-specific movement. Athletes could properly wear added load on the selected body segment(s) to perform the training task of their sport, such as attached weight on segment(s) for running, boxing, batting, or kicking. Couture et al. (2020) observed that attaching a loaded of 5% body mass (1.5-2.25 kg per leg) to the shanks did not affect flight time, ground contact time, stride length and center of mass displacement during 3.9 m/s treadmill running. However, the limb inertia changes with the additional mass on the ankles, leading to influencing the acceleration and deceleration of muscle actions, especially in conditions of limb movement at high speeds. There is a need for more studies to explore muscle activations during running gait cycle phases in response to changes in limb inertia. The manipulation of the added mass on distal limbs provides insight into understanding muscle action and function, offering professional guidance for developing a wide variety of training strategies. The aim of this study was to examine whether added mass on the distal shanks exhibits differences in the activation patterns of the BF and RF during treadmill running at four steady-state speeds.

METHODS: Nine recreationally active men, who were physical education students at the university, participated in the study after providing IRB-approved informed consent. Participants had a mean ± 1 SD age of 20.5 ± 1.5 years, height of 171.9 ± 6.6 cm, and body mass of 69.7 ± 13.6 kg. None of the participants had experience running with wearable loaded prior to laboratory activities. All were free of any prior injuries that would impair their running movements. Participants wore their own running shoes and athletic shorts for testing. Kinematic and EMG data were obtained from the right low extremity of each participant. Seven

Noraxon inertial measurement units (IMU) sensors (200 Hz) and two Noraxon surface EMG sensors (2000 Hz) were attached to the pelvic, thighs, shanks, feet, BF, and RF using elastic bandages. Pairs of bipolar Ag/AgCl surface electrodes (Blue Sensor, Ambu Inc., Denmark) were mounted over the BF and the RF for the right limb. The target running speeds were 2.5 m/s (jogging), 4.0 m/s (slow-pace running), 5.5 m/s (medium-pace running), and 7.0 m/s (fast-pace running) on a treadmill with safety arch and harness (pulsar 3p, h/p/cosmos sports & medical gmbh, Germany). The commercial wearable resistance suit (Twdimas Sport CO., Taiwan) was used for loaded running. The mass of wearable resistance was adjusted and set at one third of the individual mass of the shank and foot, according to the anthropometric data of Dempster (1955). The average mass of the wearable resistance was 1.4 ± 0.3 kg per leg.
Participants started with a warm-up of unloaded and loaded running at the target speeds on the treadmill to familiarize themselves with the test tasks. Thereafter, the participants completed eight trials of treadmill running, four at discrete speeds under randomised unloaded and loaded conditions, interspersed with five minutes of rest. Kinematic and EMG data were acquired for ten stride cycles at the steady-state target speed for each running trial condition. Hip and knee joint angles were smoothed using a fourth order Butterworth filter with a cutoff frequency of 6 Hz. EMG data were filtered using a root mean square with a window 80 ms following a band-pass Butterworth filter (10-450 Hz) to produce a linear envelope. The events of foot-strike and toe-off were determined from the accelerometer data of IMUs on feet. Data were created by averaging six running stride cycles after normalizing to 100% of the running stride cycle for each trial condition. All parameters of interest were compared using 2 (load) × 4 (speed) repeated-measures two-way ANOVA tests with Tukey post hoc comparisons. Statistical analyses were performed using IBM SPSS (version 20.0) with a significance alpha level set at .05 for all analyses. Effect size (reported using partial eta squared "η^2_p") was described as small (< .01), medium (.06), and large (.14). The observed statistical power (SP) of the test was calculated in this study.

RESULTS: Regarding the stance phase variables, the peak angular velocities of the hip extension ($p < .001$, $\eta^2_p = .947$, SP = 1.00) and the knee extension ($p = .046$, $\eta^2_p = .279$, SP =.65) were significantly increased with the running speed increment, and significantly decreased with the loaded condition as compared to the unloaded condition (hip: $p < .001$, $\eta^2_p = .838$, SP = 1.00; knee: $p = .001$, $\eta^2_p = .688$, SP =.97). The main effects of the speed and loaded were non-significant in the peak angular velocity of the knee flexion ($p > .05$) during the stance phase. The peak amplitude of the RF (shown in Table 1) displayed a significant main effect for the running speed ($p = .001$, $\eta^2_p = .196$, SP = .97) but not for the loaded condition ($p = .669$), while there was non-significant interaction between the running speed and the loaded condition ($p = .316$). The peak amplitude of the BF presented a significant interaction between the running speed and the loaded condition ($p = .004$, $\eta^2_p = .423$, SP = .92). The highest peak amplitude of the BF during the stance phase was observed at a running speed of 7.0 m/s ($p < .05$).
When the running speed changed from 2.5 to 7.0 m/s, the peak angular velocities of the hip flexion, the hip extension, the knee flexion, and the knee extension significantly increased during the swing phase (ps < .001). For running speeds at 4.0 and 7.0 m/s, the peak angular velocities of the hip flexion for the loaded condition decreased significantly ($p < .05$). The peak angular velocity of the knee extension during the swing phase displayed the largest under the loaded condition at a running speed of 7.0 m/s ($p < .05$). The peak angular velocity of the hip extension during the swing phase was significantly lower with the loaded condition as compared to the unloaded condition ($p = .006$, $\eta^2_p = .638$, SP =.91). The peak amplitude of the RF during the swing phase (shown in Table 1) displayed a significant main effect for the running speed ($p < .001$, $\eta^2_p = .754$, SP = 1.00) and the loaded condition ($p = .018$, $\eta^2_p = .524$, SP =.74), while there was non-significant interaction between the running speed and the loaded condition ($p = .843$). The peak amplitude of the BF during the swing phase displayed a significant interaction between the running speed and the loaded condition ($p = .022$, $\eta^2_p = .324$, SP =.75). The peak amplitude of the BF was higher at running speeds of 5.5 and 7.0 m/s compared to the lower running speed during the swing phase (ps < .05).

Table 1: Means and standard deviations of the peak rectified EMG amplitudes (unit: μV) for loaded conditions as a function of running speed conditions.

	2.5 m/s		4.0 m/s		5.5 m/s		7.0 m/s	
Muscles	M	SD	M	SD	M	SD	M	SD
				Stance phase				
Rectus femoris†								
Unloaded	149.6	38.5	210.6	73.5	228.0	74.6	269.7	118.7
Loaded	134.7	36.0	203.0	75.6	265.6	97.2	232.9	85.2
Biceps femoris†								
Unloaded	229.3	83.5	337.2	164.6	408.7	105.9	520.6	206.0
Loaded	212.9	91.8	290.1	99.3	472.0	144.5	663.6*	187.8
				Swing phase				
Rectus femoris†								
Unloaded	54.0	81.5	168.4	72.1	265.6	123.1	362.9	170.7
Loaded	72.2*	49.9	186.7*	80.9	305.6*	134.1	407.4*	163.3
Biceps femoris†								
Unloaded	247.5	93.9	349.9	159.6	488.6	184.6	677.5	128.9
Loaded	269.6	128.9	356.8	127.4	570.2*	180.0	750.5*	205.0

*Denotes significantly different to unloaded condition ($p < .05$).
†Denotes significantly different among running speed ($p < .05$).

DISCUSSION: There was no hip flexion during the stance phase when the running speed exceeded 4.0 m/s. The BF plays a role as the hip extensor and increases the peak amplitude with the running speed increments during the stance phase. This finding is consistent with previous studies by Schache et al. (2012) and Higashihara et al. (2010). The peak angular velocities of hip extension at a running speed of 5.5 and 7.0 m/s were significantly higher with the loaded condition than the unloaded condition during the stance phase. The peak amplitude of BF at a running speed of 7.0 m/s was larger with the loaded condition than the unloaded condition. This indicates that the shank loaded wearable resistance used in this study stimulates the action of BF for hip extension under the conditions of the high-speed running. The angular velocity of hip extension increased with the running speed increments during the late swing phase, but it was lower under the loaded condition than the unloaded condition. This suggests that participants needed to slow down the thigh downward until the knee extension for landing during the late swing phase. It also stated that the action of the RF was lengthened for the duration (Figure 1). However, the shank loaded wearable resistance caused an increment in the peak velocity of the knee extension at high speed running during the swing phase. The BF acted as a knee flexor and played the role of the eccentric action at high speed running during the end of the swing phase. The shank loaded wearable resistance enlarged the activation amplitude of the BF for the eccentric action of the knee extension prior to foot landing and the concentric action of the hip extension while landing at running speed of 5.5 to 7.0 m/s. This was the phenomenon of the mechanical energy transfer through the BF.
During the stance phase, the amplitude of the RF increased with a running speed but was not affected by the shank loaded wearable resistance. The RF flexed the hip joint after toe-off (Mero & Komi, 1987). More knee flexion occurred and carried the swing lower extremity forward under the loaded condition. This showed that the RF was stretched to diminish the disadvantage of the active insufficiency. The loaded condition provided the RF with a favorable position for the activation of the hip flexor during the initial swing phase.

CONCLUSION: The application of load on the distal shanks during high-speed running resulted in an increased stretching load on the knee flexor at the end of the swing phase. This led to a higher concentric action of the hip extensor during the initial stance phase. The loaded condition also increased the trend of knee flexion to stretch the BF for the activation of hip flexion. This finding provides empirical evidence to better understand the function of various running speeds for loaded wearable resistance training.

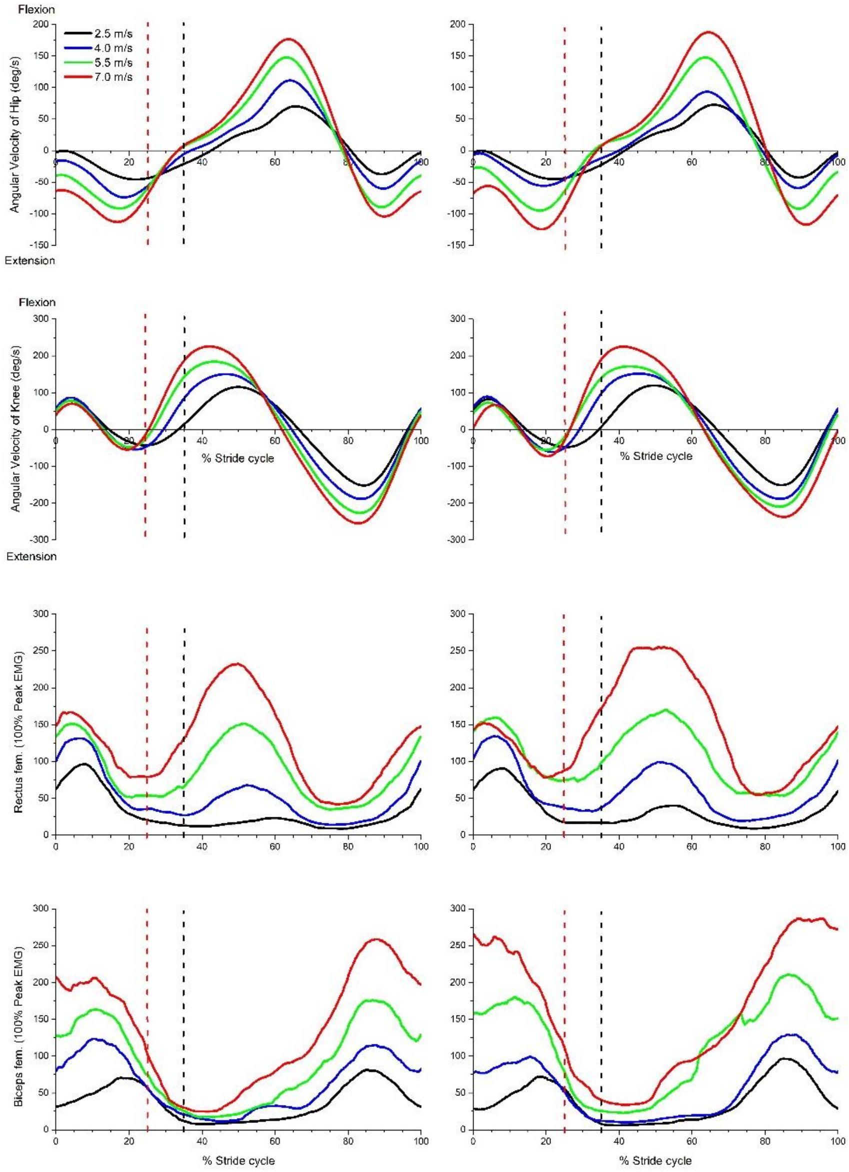

Figure 1: The averaged profiles of hip flexion angular velocity, knee flexion angular velocity, rectified rectus femoris, and rectified biceps femoris during stride cycle at four different speeds under the conditions of unloaded (left) or loaded (right). The dot lines (black: 2.5 m/s; red: 4.0, 5.5, and 7.0 m/s) mean the instant of toe-off.

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