An integrated biomechanical analysis of high speed incline and level treadmill running

STEPHEN C. SWANSON and GRAHAM E. CALDWELL

The Orthopedic Specialty Hospital, Salt Lake City, UT; and Department of Exercise Science, University of Massachusetts, Amherst, MA

ABSTRACT

SWANSON, S. C., and G. E. CALDWELL. An integrated biomechanical analysis of high speed incline and level treadmill running. *Med. Sci. Sports Exerc.*, Vol. 32, No. 6, pp. 1146–1155, 2000. **Purpose:** Recent sprint training regimens have used high-speed incline treadmill running to provide enhanced loading of muscles responsible for increasing forward running speed. The goal of this study was to document the joint kinematics, EMG, and swing-phase kinetics of incline treadmill running at 4.5 m·s^{-1} with a 30% grade, and compare these data to that of level running under similar conditions. **Methods:** Sagittal plane video (200 Hz) and EMG from eight lower extremity muscles were recorded during each of three locomotion conditions: incline running at 4.5 m·s^{-1} and 30% grade (INC), level running at 4.5 m·s^{-1} (LSS), and level running at the same stride frequency as INC (LSSF). A rigid body model was used to estimate net muscle power and work values at the hip, knee, and ankle during swing. Timing and amplitude of EMG signals for each muscle relative to footstrike were compared between conditions. **Results:** Stride frequency and percentage of stride spent in stance were significantly higher during INC (1.78 Hz; 32.8%) than in the LSS (1.39 Hz; 28.8%) condition. Stride frequency played an important role, as most measures were more similar between INC and LSSF. Extensor range of motion of all joints during push-off was higher for INC. During INC, average EMG amplitude of the gastrocnemius, soleus, rectus femoris, vastus lateralis, and gluteus maximus were higher during stance, whereas the hamstrings activity amplitudes were lower. Average power and energy generated during hip flexion and extension in the swing phase were greatest during INC. **Conclusions:** These data suggest that compared with LSSF and LSS, INC provides enhanced muscular loading of key mono- and bi-articular muscles during both swing and stance phases. **Key Words:** COORDINATION, BIARTICULAR MUSCLES, EMG, KINETICS, INCLINE RUNNING, TREADMILL RUNNING, KINEMATICS, SPRINT RUNNING, SPRINT TRAINING

Lower extremity kinematics and kinetics during level running have been well-documented (5,28,37). Calculation of net moments and powers at specific joints has highlighted the relative contribution of individual muscle groups, whereas electromyography (EMG) has permitted the analysis of specific muscle activity within the running cycle (18,19,24,26,30). The integration of kinematic, kinetic, and EMG data has provided a wealth of information on the coordination of the musculoskeletal system during level running (18,25).

In contrast, fewer studies have focused on the biomechanics of incline running, although substantial changes in lower extremity motion have been reported with alterations in grade of 5% or more (11,21,28). Studies on the kinetics of graded running have been limited to the stance phase only (3,17). Although the effect of grade on lower extremity muscle activity has been well documented for walking (35), similar studies on graded running are absent. However, research on rat locomotion revealed substantial increases in gastrocnemius and soleus EMG with increases in treadmill grade (34).

Recent training regimens intended to improve sprinting performance have included incline treadmill running at speeds above 4.5 m·s^{-1} with grades over 30% (8). These training protocols are designed to enhance muscular loading of the hip, knee, and ankle extensors during stance and the hip flexors and extensors during recovery (8). It has been suggested that these muscle groups are primarily responsible for generating forward propulsion during running and sprinting (18,19,24). Studies on explosive lower extremity extension movements such as sprinting suggest that coordination of the mono- and bi-articular muscles crossing these joints is important for optimal performance (18,19,20,32). An integrated biomechanical analysis during high-speed incline sprinting would provide insight into the nature of muscular loading and coordination during these conditions, and would enhance our understanding of the effectiveness of such training programs.

Therefore, the purpose of this study was to document the joint kinematics, EMG, and swing-phase kinetics of incline treadmill running at 4.5 m·s^{-1} with a 30% grade and to compare these data with that of level running at the same

0195-9131/00/3206-1146/0
MEDICINE & SCIENCE IN SPORTS & EXERCISE®

Submitted for publication April 1999.
Accepted for publication September 1999.

speed. It was hypothesized that inclined running would exhibit changes in lower extremity joint kinematics and kinetics and enhanced activity in muscles targeted by incline treadmill running protocols. Further, it was hypothesized that these alterations would be due in part to the higher stride frequency needed for the incline running (28). These hypotheses were tested through comparison of selected kinematic, kinetic, and EMG variables between incline and level running conditions.

METHODS

Subjects and experimental protocol. Twelve healthy, college-aged men (mass: 68 ± 5 kg, height: 1.76 ± 0.04 m; 7 collegiate sprinters and 5 lacrosse athletes) who had successfully completed a 6-wk incline sprint training program gave their informed written consent to serve as subjects in the study. In addition to weight training and low-impact plyometrics, a main component of the training program was incline treadmill running at speeds and grades ranging from 4.5 to 9.0 m·s^{-1} and 0 to 40%, respectively. All testing occurred in the week after the 6-wk training program.

Three locomotion conditions were examined: incline running at 4.5 m·s^{-1} and 30% grade (INC); level running at 4.5 m·s^{-1} (LSS); and level running at the same stride frequency as the incline condition (LSSF). Anthropometric measurements of the lower extremity were taken as described by Winter (39). After a brief warm-up, the subjects performed a series of five standard isometric contractions (SICs) as part of the EMG data collection protocol (see Appendix for details). Upon completion of the SICs, three sets of five consecutive strides of INC running were timed with a stopwatch to determine the nominal stride frequency for the LSSF condition. The subjects then ran at 0% grade for several (≤ 4) sets of five strides while the treadmill speed was adjusted incrementally to elicit the same stride frequency as INC. The LSSF speed was selected when the last two sets at one speed were within ± 5% of the INC stride frequency. After a 5- to 10-min rest period, the subjects completed five, 6-s trials in each locomotion condition, with the condition order randomized between subjects. High-speed video and EMG from eight muscles of the lower extremity were collected for 3–4 complete strides during each trial.

Kinematics and swing-phase kinetics. The sagittal plane motion of six retro-reflective markers defining the foot, leg, thigh, and trunk segments (Fig. 1) was recorded at 200 Hz using a NAC video camera (Motion Analysis Corp., Santa Rosa, CA). The video data were digitized using a Motion Analysis VP-110 microprocessor interfaced to a microcomputer. After proper scaling, the raw coordinate data were smoothed with a dual-pass, fourth-order Butterworth low-pass filter (fc = 12 Hz). One representative stride (right foot contact to next right foot contact) was selected from each of the five trials in each condition for further analysis (5 trials per condition). Stance phase was identified using an LED within the camera view that was triggered by

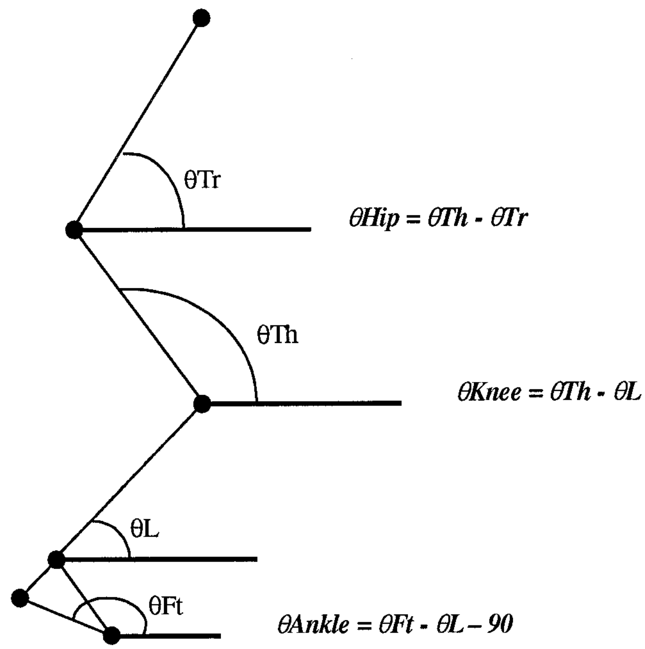

Figure 1—Convention used for calculation of segmental and joint angles from marker coordinate data.

a piezo-resistive footswitch mounted under the bed of the treadmill. Segment angles were calculated with respect to the right horizontal, and joint angles were calculated as the angles between adjacent segments (Fig. 1). The measured anthropometric data and marker coordinates were used to calculate the mass, moment of inertia, and position of the center of mass for each segment (39). Finite difference methods were used to calculate linear and angular velocity and acceleration data for each segment. Selected kinematic parameters were calculated at key events in the gait cycle (footstrike, toe-off, etc.), and ensemble averages for joint kinematics were calculated. Stride frequency was calculated from the footswitch indicator, and stride length was estimated by dividing treadmill speed by stride frequency and adding horizontal displacement of the center of mass of the foot.

Standard link segment modeling and Newtonian equations were used to calculate instantaneous net muscle moments at the hip, knee, and ankle during the swing phase of the gait cycle, with hip flexor, knee extensor, and ankle dorsi-flexor moments defined as positive. Net muscle power for each joint was defined as the product of net muscle moment and joint angular velocity (33). Ensemble averages of joint angular velocity, muscle moment, and muscle power were calculated for each subject, and then across all subjects, in each condition. Selected kinetic variables such as average muscle power and net muscular work were calculated for distinct periods throughout the swing phase.

Electromyography. The electrical activity of eight lower extremity muscles was monitored with preamplified Ag/AgCl surface electrodes (Model EMG-544, Therapeutics Unlimited, Iowa City, IA) interfaced to an amplifier/processor module (CMRR 87 dB at 60 Hz; input impedance > 15 megohms at 100 Hz; 20–4000 Hz band-pass).

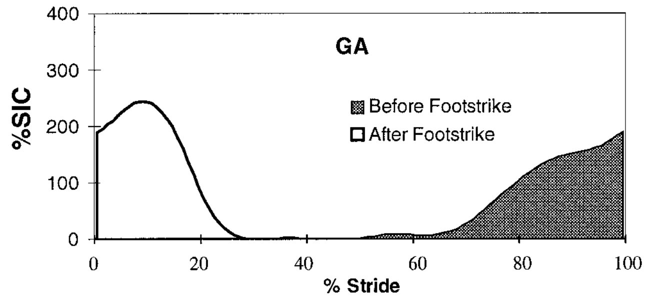

Figure 2—Schematic depicting calculation of pre- and post-footstrike variables.

The eight muscles analyzed were tibialis anterior (TA), gastrocnemius (GA), soleus (SOL), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH), biceps femoris (BF), and gluteus maximus (GM). After standard skin preparation (1), the electrodes were placed distal to the approximate motor point of each muscle (40). A common ground electrode was placed on the dorsal right wrist just lateral to the ulnar styloid process. Amplification of each EMG signal was set to maximize resolution. EMG channels were sampled for 2 s during each trial at 1000 Hz using a 12-bit analog to digital (A/D) converter. The footswitch/LED control signal was also captured by the A/D board to allow temporal synchronization of the EMG and video data.

After removal of any possible baseline offset (bias), the raw EMG data were high-pass filtered at 20 Hz using a dual-pass, fourth-order Butterworth filter to eliminate possible movement artifact. Times of activity onset, offset, and burst duration in each stride were identified, using a criterion threshold of 10 μV above baseline lasting for more than 10 ms. The EMG data were then full-wave rectified and low-pass filtered at 12 Hz using a dual-pass, fourth-order Butterworth filter to create linear envelopes. Muscular activity occurring before and after footstrike was quantified using average EMG amplitude, expressed as a percentage of the mean amplitude of the highest one second portion of EMG activity during each muscle's SIC (Fig. 2). Note that the EMG data were not shifted in time relative to the kinematic and kinetic data, as true electromechanical delay is very brief and invariant between our running conditions (7).

Descriptive statistics were calculated for selected kinematic, kinetic, and EMG dependent variables. A within-subject repeated measures analysis of variance design (RM ANOVA) was used to test the main effect of running condition, using *post hoc* orthogonal contrasts (SPSS, Inc., Chicago, IL). All significant differences reported are at $P < 0.01$ unless otherwise noted.

RESULTS

Kinematics. The subjects exhibited significantly higher stride frequency in the INC condition (1.78 vs 1.39 Hz) than in LSS (Table 1). To attain this stride frequency in the LSSF condition, the treadmill speed was set at 7.61 m·s^{-1}, significantly higher than the INC and LSS speed of 4.47 m·s^{-1}. A greater percentage of the gait cycle during INC was spent in stance as compared with the level conditions. However, the absolute stance time was longest during LSS and shortest during LSSF. The joint kinematic patterns (Fig. 3) were typical for running, with relatively low coefficients of variation (CV) for joint displacements, especially for the hip and knee (Table 1).

Compared with level running, INC elicited several kinematic changes during stance, including greater joint flexion at footstrike for all three joints (Table 2 and Fig. 3). On the incline, the hip was extending rapidly at footstrike, more slowly during the impact phase (from footstrike to peak knee flexion), and rapidly again during push-off (peak knee flexion to toe-off). In contrast, the hip flexed slightly at impact during the level conditions. Thus, in INC hip ROM during impact was extensor (−) and significantly different than the flexor (+) ROM seen on the level. Further, hip extensor ROM during push-off was significantly higher for the INC condition. The knee angle was relatively constant during impact followed by rapid extension throughout push-off in the INC condition, whereas a more typical flexion-extension sequence was seen on the level during stance. Thus, knee flexion ROM during impact was significantly greater during the level conditions, whereas knee extension ROM during push-off was greater during INC. Interestingly, the knee was significantly more flexed at toe-off during LSSF than in either INC or LSS. At the ankle, dorsi-flexion followed by plantar-flexion was found during stance in all conditions. During INC, the ankle was more dorsiflexed at footstrike and exhibited significantly less dorsi-flexion ROM than LSSF or LSS. However, INC showed greater dorsi-flexion angles at the beginning of push-off and significantly greater plantar-flexion ROM during push-off than the level conditions. Also, stance phase ankle angular velocities were considerably higher during LSSF than in either INC or LSS.

In the INC condition, the hip was more flexed during swing, with maximum hip flexion angle and total joint ROM being greater than in either LSSF or LSS. The INC condition demonstrated higher hip flexor velocities in early swing, with late swing hip extensor velocity increasing in INC but decreasing before footstrike in both level conditions. At the knee, maximum flexion angle and ROM during swing were significantly greater for LSSF. The rapid deceleration of knee extension into flexion before footstrike seen in level running was absent during INC, which demonstrated continual knee extension into stance. Finally, although ankle velocity profiles during swing were similar between conditions, the ankle ROM was greatest for INC.

Swing-phase kinetics. The hip moments and powers (Fig. 4) reveal a similar pattern for recovery in all conditions—early eccentric hip flexor activity (Ecc HF), followed by a substantial concentric hip flexor phase (Conc HF) from 10–50% swing, a brief eccentric hip extensor period (50–65%), and concluding with a major concentric hip extensor phase (Conc HE). For the knee, the initial 50% of recovery was comprised of eccentric knee extension, followed by prominent eccentric knee flexor (Ecc KF)

TABLE 1. Summary of general kinematic variables by condition.

Dependent	Incline Mean	Incline SD	LSSF Mean	LSSF SD	LSS Mean	LSS SD
Treadmill speed ($m \cdot s^{-1}$)	4.47	0.02	7.61	0.23[a]	4.47	0.02[b]
Stride length (m)	1.26	0.10	2.12	0.12[a]	1.61	0.07[a,b]
Stride frequency (Hz)	1.78	0.14	1.80	0.11	1.39	0.07[a,b]
Stance duration (% stride)	32.8	1.0	25.6	1.7[a]	28.8	1.5[a,b]
Stance duration (actual time in s)	0.184	0.013	0.142	0.009[a]	0.207	0.012[a,b]
Coefficient of variation						
Hip	9.0		13.1		14.3	
Knee	8.9		7.3		6.5	
Ankle	13.7		15.6		19.6	

[a] Indicates means significantly different from the incline condition.
[b] Indicates means significantly different from the LSSF condition.

activity starting at ~70% of swing. Ankle kinetic variables were of small magnitude and have been omitted for clarity.

Table 3 presents muscle power and work data for the four major power production phases (Ecc HF, Conc HF, Conc HE, and Ecc KF). The most salient feature is that all power and work variables during LSS were significantly reduced compared with the INC and LSSF conditions, including positive work (energy generated) during concentric phases

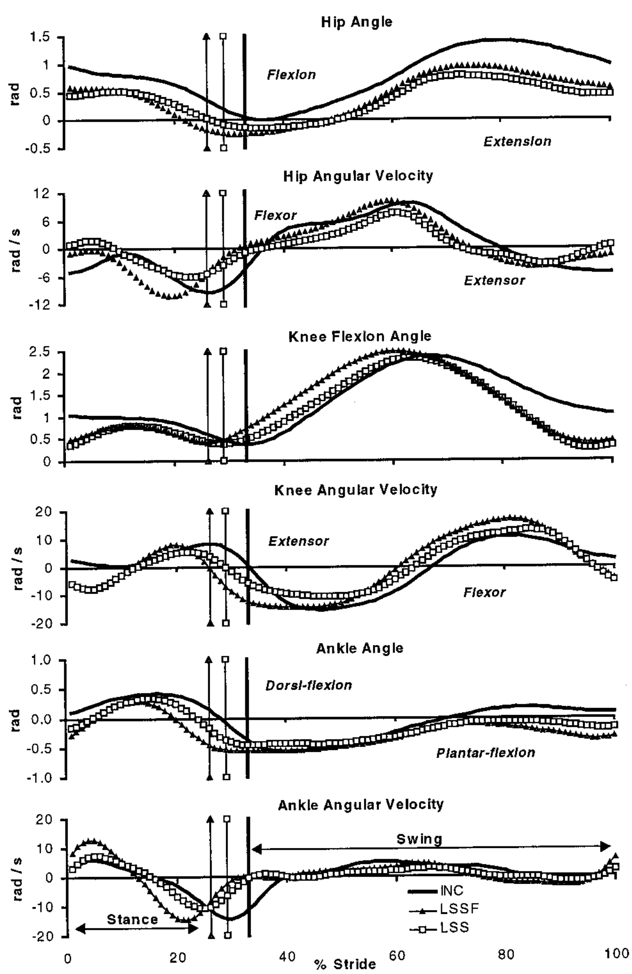

Figure 3—Ensemble angular position and velocity data for the three lower extremity joints graphed as a function of standardized stride cycle (0–100% from right footstrike to next right footstrike). Running condition is indicated in the bottom panel, along with depiction of stance and swing phases. Vertical bars in each panel show the end of stance phase in each condition.

and negative work (energy absorbed) during eccentric phases. The INC condition demonstrated greater power and work values compared with LSSF in Conc HF, due primarily to much greater hip flexor velocities (Fig. 4). In the earlier Ecc HF phase, the amount of energy absorbed was significantly greater in LSSF compared with INC. However, no significant differences in *average* muscle power were found between INC and LSSF during Ecc HF, due to the substantially shorter time of this phase in INC. A shorter absolute duration was also responsible for higher Conc HE muscle power during INC compared with LSSF, despite similar positive work values. Peak Conc HE power was actually higher during LSSF, but due to a steady rise in hip extensor velocity the INC power levels stayed high for an extended time (Fig. 4). Therefore, average Conc HE power in late swing was much higher for INC than LSSF despite lower hip extensor moments. Finally, for the late Ecc KF phase both negative work and average power were greater in LSSF than INC.

EMG. The linear envelope EMG patterns were similar in each locomotion condition, with all muscles active before and during stance (Fig. 5), although TA activity occurred primarily during swing. The most striking EMG differences were large increases in magnitude during the stance phase of INC for GA, SOL, RF, VL, and GM. In contrast, the biarticular MH and BF exhibited lower stance amplitudes for INC compared to LSSF. Another salient difference was an early swing RF activity burst associated with hip and knee flexion in INC and LSSF that was greatly reduced in LSS.

The hip extensors BF, MH, and GM became active around 60% of stride in each condition, with GM onset time significantly earlier in LSSF than in either INC or LSS (Fig. 6). GM activity continued until late in stance, with the offset time again significantly earlier for LSSF. In LSS MH was activated earlier ($P < 0.05$) and remained active longer. The offset time of BF was earliest in LSSF and latest in LSS. The biarticular GA was the next muscle to become active, and it too remained active throughout stance, with both onset and offset significantly earlier in the LSSF condition. The VL, SOL, and RF were the last muscles to be activated before footstrike, with earlier onsets and offsets in LSSF for all three muscles. VL activity after footstrike ended later during INC compared to both level conditions. TA activity was biphasic, during early swing to assist with toe clearance and

TABLE 2. Summary of selected joint kinematic variables.

Variable	Joint	Incline		LSSF		LSS	
		Mean	SD	Mean	SD	Mean	SD
Angle at footstrike (°)	Hip	54.3	7.4	32.5[a]	6.2	25.4[a]	6.4
	Knee	59.7	6.7	26.7[a]	6.3	21.0[a]	5.9
	Ankle	6.7	5.8	−16.6[a]	4.4	−9.1[a]	10.6
Angle at toe-off (°)	Hip	2.2	5.4	−10.5[a]	7.8	−4.6[a]	6.5
	Knee	21.5	5.4	28.3[a]	8.8	22.0[b]	6.9
	Ankle	−21.7	5.7	−25.2	5.8	−21.0	8.6
Range of motion during impact (°)	Hip	−11.2	2.2	5.2[a]	2.7	4.4[a]	2.8
	Knee	−4.5	1.3	19.6[a]	3.6	25.3[a,b]	2.6
	Ankle	18.3	2.3	32.6[a]	5.3	28.4[a]	4.7
Range of motion during push-off (°)	Hip	−41.9	3.8	−36.4[a]	3.6	−29.9[a,b]	4.3
	Knee	−34.0	3.3	−22.4[a]	4.9	−23.5[a]	3.9
	Ankle	−46.5	3.1	−41.4[a]	3.0	−40.4[a]	5.5
Max angle during swing phase (°)	Hip	80.6	7.1	55.2[a]	9.9	46.2[a,b]	9.4
	Knee	136.7	4.4	143.3[a]	3.7	134.1	7.2
	Ankle	24.0	6.6	17.0[a]	5.7	20.2[a]	8.9
Range of motion swing phase (°)	Hip	81.0	5.1	70.4[a]	6.7	54.9[a,b]	10.4
	Knee	116.3	8.7	123.7[a]	8.5	120.3	9.0
	Ankle	44.6	5.2	31.6[a]	4.8	32.6[a]	6.3

[a] Indicates means significantly different from the incline condition.
[b] Indicates means significantly different from the LSSF condition.

later in preparation for footstrike. The earlier onset time during LSS was probably due to increased swing time in this condition.

The magnitude of muscle activity before footstrike for GM, MH, RF, and SOL was similar for LSSF and INC but significantly reduced in LSS (Fig. 7). BF and VL amplitudes during this period were greater in INC than in either of the level conditions. Swing phase TA activity was also significantly higher during INC compared with either LSSF or LSS. In contrast, GA was the only muscle that showed more pre-footstrike activity during LSSF than in INC. After footstrike, activity in the monoarticular extensors GM, VL, SOL, and the biarticular RF and GA were highest during INC and lowest in LSS. Average post-footstrike activity in MH was significantly greater in LSSF than in either INC or LSS, whereas BF activity during this period was greatest in LSSF and lowest in LSS.

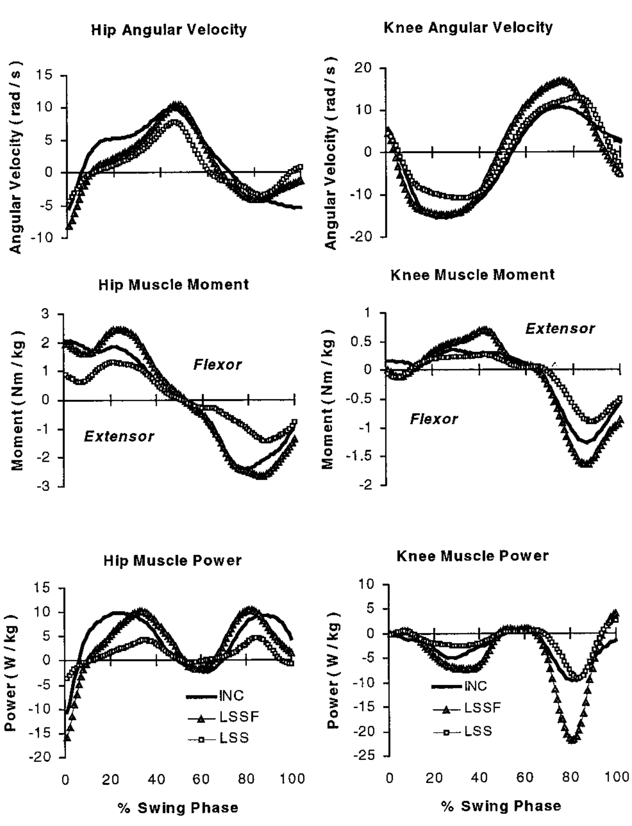

Figure 4—Ensemble joint angular velocity, net moment and power for the hip and knee graphed as a function of standardized swing phase (0–100% from right toe-off to right footstrike). Running condition is indicated in the bottom panels.

DISCUSSION

Our data illustrate that high-speed incline running elicited distinct changes in segmental and muscular coordination compared with level running at the same speed. These effects are due in part to the higher stride frequency in the incline condition, and the LSSF condition provided valuable insight to the influence of stride frequency alone. The results supported our hypotheses that the incline condition would generate kinetic and kinematic differences accompanied by increases in the activity of specific muscles. Further, similarities between the incline and LSSF conditions illustrated the importance of the increased stride frequency. In the discussion we will focus on the integration of EMG, kinematic, and kinetic data in the different running conditions.

Level running. The level running joint kinematic and EMG patterns were consistent with other studies (26,28). For example, hip and knee angles at footstrike were similar to those reported by Milliron and Cavanagh (28) for running at 4.5 m·s^{-1}, whereas ankle angles at footstrike were typical of those exhibited by mid-foot strikers or during sprinting (18). Kinematic patterns throughout stance mirrored previous studies of level running: relatively isometric hip motion, knee flexion, and ankle dorsi-flexion during the impact phase, followed by hip and knee extension and ankle plantar-flexion during push-off (18,28).

As in previous studies, peak EMG amplitudes occurred during the first half of stance, coinciding with the highest ground reaction forces (19,26). EMG timing and patterns were also similar to other studies, with the hamstrings ac-

TABLE 3. Net muscular work and power during selected phases of recovery.

Variable	Phase	Incline		LSSF		LSS	
		Mean	SD	Mean	SD	Mean	SD
Average net muscle power (W·kg^{-1})	Ecc HF	−9.00	1.41	−9.92	1.96	−2.33[a,b]	0.92
	Conc HF	7.02	0.79	6.09[a]	0.76	2.47[a,b]	0.70
	Conc HE	7.91	1.18	6.76[a]	2.05	2.29[a,b]	1.17
	Ecc KF	−5.79	0.97	−12.20[a]	2.08	−5.01[a,b]	1.21
Net muscular work (J·kg^{-1})	Ecc HF	−0.31	0.05	−0.55[a]	0.07	−0.16[a,b]	0.06
	Conc HF	1.19	0.09	1.08[a]	0.13	0.52[a,b]	0.14
	Conc HE	0.92	0.15	0.92	0.29	0.42[a,b]	0.20
	Ecc KF	−0.76	0.08	−1.44[a]	0.20	−0.69[a,b]	0.14

[a] Indicates means significantly different from the incline condition.
[b] Indicates means significantly different from the LSSF condition.

tivated first before footstrike, followed by the GM and GA, and finally the SOL, VL, and RF (18,19,24,26). The monoarticular extensors VL, SOL, and GM and biarticular RF ceased firing first, followed by the biarticular GA, MH, and BF. The hamstrings BF and MH displayed a bimodal pattern, with a second burst of activity during push-off. The second burst of RF activity during early swing has also been reported (19,24,26) and is suggested to control the amount of knee flexion during the swing phase (31). Further, the general increase in EMG amplitudes seen in the higher speed LSSF condition is typically associated with increases in running velocity (24,25).

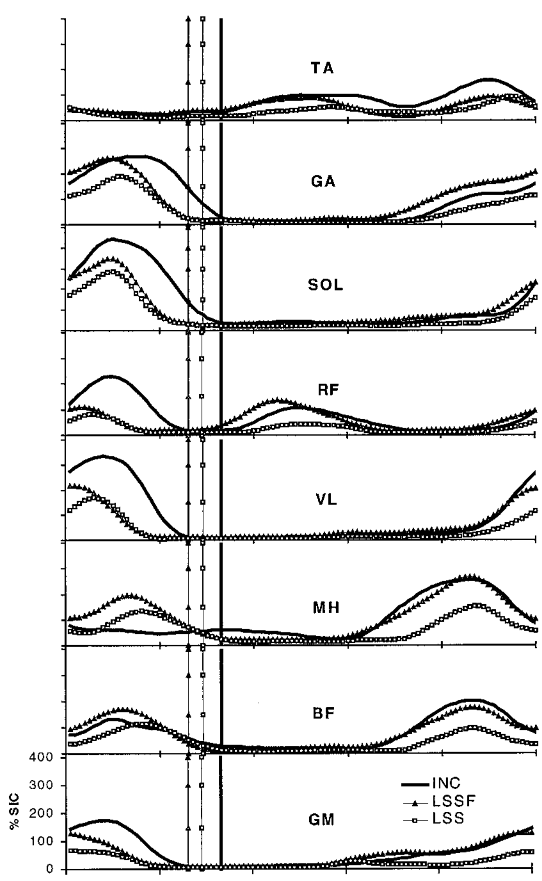

Figure 5—Ensemble linear envelope EMG data graphed as a function of standardized stride cycle (0–100% from right footstrike to next right footstrike). EMG amplitudes have been standardized using the SIC contractions, with calibrated scale shown in the bottom panel. Running condition is also indicated in the bottom panel. Vertical bars in each panel show the end of stance phase in each condition.

The swing-phase kinetic patterns indicated energy generation at the hip during both flexion and extension and energy absorption at the knee during late swing, all of which have been reported for level running at similar speeds (4,6,37). In particular, hip and knee power values during LSS (4.5 m·s^{-1}) and LSSF (7.6 m·s^{-1}) were nearly identical to those reported by Caldwell and Forrester (4) and Chapman and Caldwell (6) for running at ~5.0 m·s^{-1} and 7.6 m·s^{-1}, respectively.

Joint kinetic patterns during stance in overground level running are well documented, with dominant extensor/plantar-flexor moments at the knee and ankle accompanied by extensor hip moments during the impact phase and flexor hip moments during the later push-off (9,18,19,37). The

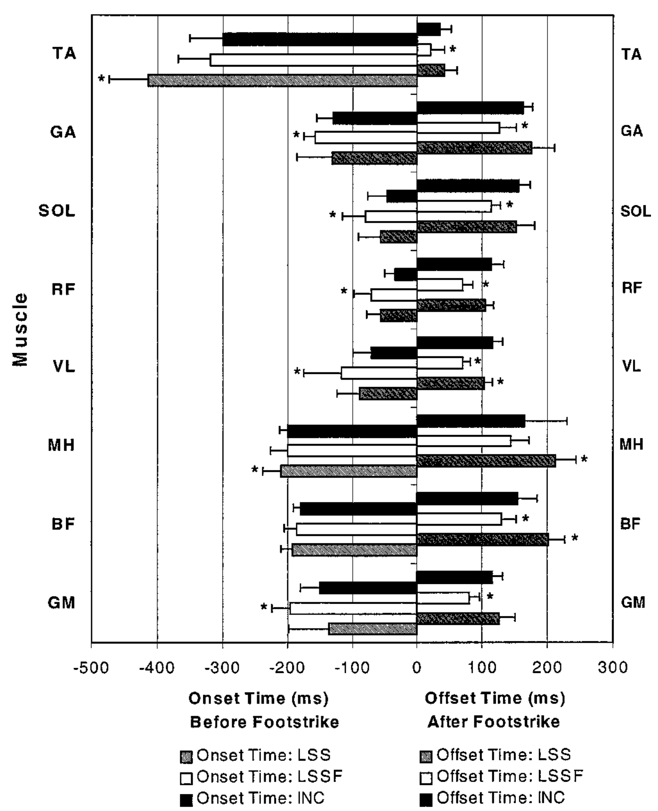

Figure 6—Mean timing data for each muscle activity in relation to footstrike. The onset times are shown on the left side (negative values, before footstrike), and the offset times are shown on the right (positive values, after footstrike). Running conditions are indicated by differently shaded bars as defined at bottom of graph. Whiskers indicate one standard deviation above the mean. Asterisks denote values significantly different from INC condition.

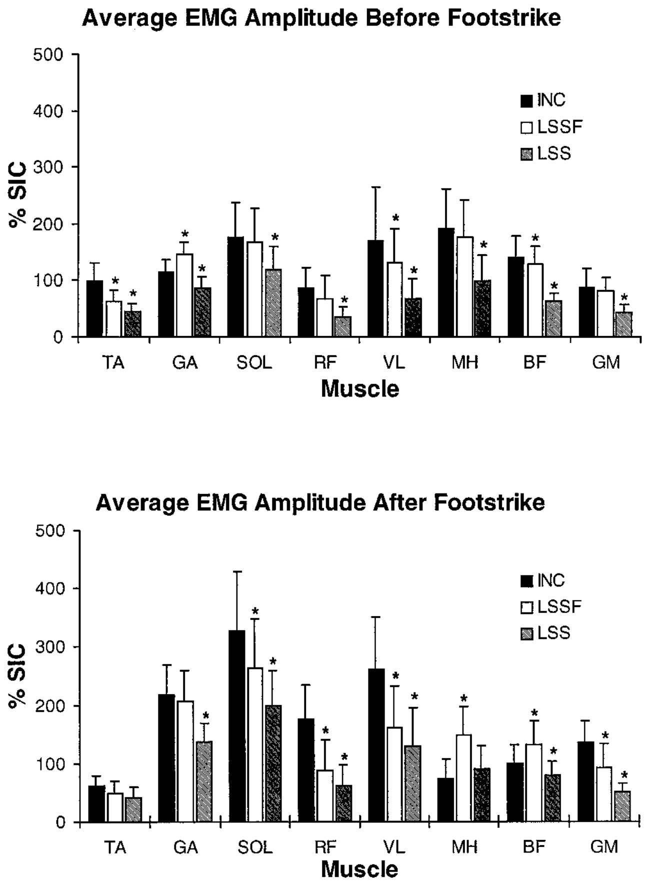

Figure 7—Average EMG amplitudes for each muscle before (top panel) and after (bottom panel) footstrike. Running conditions are indicated by differently shaded bars as defined in each panel. Whiskers indicate one standard deviation above the mean. Asterisks denote values significantly different from INC condition.

joint kinematics that accompany these net moments are very similar to those found in the present study, and it is reasonable to assume that our subjects produced similar joint moment patterns during stance. Nigg et al. (29) suggested that studies performed on solidly built treadmills (as in the present study) would likely agree with overground studies. A recent paper by Kram et al. (22) supports this contention. Using a treadmill of solid construction mounted on a force platform, they demonstrated that both the vertical and anterior/posterior ground reaction forces were nearly identical to that of overground running.

Incline running. The INC condition resulted in higher stride frequency and a greater proportion of the stride cycle spent in stance compared with level running. However, the actual time spent in stance for INC was less than for LSS. Although stride length was reduced during INC, the horizontal distance from the metatarsal marker to the hip marker at footstrike for the INC condition was similar to both LSSF and LSS. Iverson and McMahon (17) noted a similar phenomenon and proposed a "hanging triangle" hypothesis, postulating that the angle of the leg with respect to the vertical at footstrike does not change with treadmill inclination (17). This suggests that runners tend to place their leg in relatively the same position before footstrike, regardless of surface grade. Winter (38) has also suggested that foot placement is one of the key "subtasks" of gait. Our data support this view in that a condition as extreme as INC caused little change in the total horizontal stance distance relative to the hip. To maintain similar foot placement during INC, flexion angles of the hip, knee, and ankle were increased dramatically.

With increasing grade, both the reaction force normal to the treadmill and upward vertical acceleration decrease in magnitude (17,23). These findings provide insight to the changes observed during INC. To maintain a constant center of mass position above the treadmill, the average resultant treadmill reaction force during stance must roughly equal body weight to counter the effects of gravity. If the reaction force normal to the treadmill bed decreases, the force component parallel to the treadmill must concomitantly increase. Also, an increase in percentage stance duration would allow relatively more time to produce force against the treadmill, allowing an increase in total impulse ($F \times t$). This predicted increase in force parallel to the treadmill bed is supported by the joint kinematics and EMG data in the present study. During INC, the absence of knee flexion during the impact phase suggests a decrease in normal treadmill reaction force and shock attenuation requirements (9). Winter (37) has suggested that a primary function of the knee during early stance is to absorb energy. The lack of knee flexion during INC could indicate lower impact forces and thus reduced need for energy absorption.

For all three joints, extensor ROM and angular velocities during push-off were significantly greater for INC than in LSS. The monoarticular extensors GM, VL, and SOL and biarticular RF and GA were all significantly more active during INC, while the activity of biarticular MH and BF during push-off were slightly lower or similar. Combined, these muscle activities and joint kinematics suggest an increase in energy generated during push-off for the INC condition. Although we have no stance phase kinetic data, the joint kinematics and corresponding EMG amplitudes provide evidence that greater net torque was likely produced during push-off.

Another feature of the incline condition was the significantly shorter swing phase duration related to the higher stride frequency. As a result, muscular loading at the hip during swing was significantly higher in INC, with greater hip ROM and flexor/extensor velocities. Kinetic analysis revealed that much more energy was absorbed and subsequently generated by the hip flexors during the first half of swing for INC. During late swing the hip extensors generated more energy and higher moments, accompanied by more activity in the GM, BF, MH, and RF muscles. Evidence of enhanced muscular loading at the knee and ankle was also seen, as VL, SOL, and GA had higher activation levels in late swing before footstrike. This increased preactivation during INC may tune the extensors for the stretch/shorten cycle of early stance, and enhance their ability to produce force in the concentric phase (25). The differences in joint angles at footstrike and the increased joint ranges of motion during push-off for INC support this scenario.

Incline running—effect of higher stride frequency. Examination of the LSSF condition reveals that the increase in stride frequency accounts for some of the differences between INC and LSS. As in other studies, the higher running speed in LSSF resulted in increases in the amplitude of most kinematic, kinetic and EMG variables with only slight changes in their patterns (6,24–26,28). Increasing stride frequency elevates the inertial contribution of each segment during swing, resulting in large increases in kinetic variables between LSS and LSSF. Such kinetic increases are reflected in greater muscular activity of the GA, RF, MH, BF, and GM during swing for INC and LSSF compared with LSS. INC and LSSF also displayed similar amounts of preactivation of the VL, SOL, and RF before footstrike. However, several kinematic and kinetic alterations at the hip and knee during swing and large differences in EMG amplitude and joint kinematics during stance were found between INC and LSSF, suggesting changes in muscular loading unique to the incline condition.

The swing phase was significantly shorter during the INC condition compared with LSSF, whereas the hip joint ROM increased by over 10°. As a result, more energy was generated and average power was higher in Conc HF during the INC condition. Although we have no EMG data for monoarticular hip flexors, higher hip flexor velocities coupled with similar hip moments during early swing suggest greater hip flexor activity. The biarticular RF became active after the initiation of Conc HF and peaked after the occurrence of maximum Conc HF power, suggesting that its contribution to hip flexion was aided by the monoarticular hip flexors. Our data support the view that the primary function of the RF is to control the amount of knee flexion during swing (31) and suggest that the monoarticular hip flexors are loaded at higher levels during INC than in LSSF.

In late swing, GM, MH, BF, and GA all contributed to energy generation at the hip and/or absorption at the knee. The GM became active during the brief Ecc HE phase and increased its activity as more energy was generated during Conc HE. GM activity before footstrike was similar for LSSF and INC despite significantly greater average hip power during Conc HE of INC. The timing of biarticular MH and BF activity suggests that they function to assist in both eccentric and concentric hip extension while also controlling the amount of knee extension. However, the similarities in late swing EMG activity for the MH and BF between INC and LSSF were accompanied by much greater energy absorption at the knee in LSSF. The biarticular GA, activated earlier and to a greater extent during LSSF than in INC, was likely responsible for this energy absorption.

The data suggest that the muscular coordination patterns exhibited during late swing may be influenced by position-specific muscular capabilities. Hoy et al. (13) report that the hamstrings contribution to the net hip extensor moment is greatest between 60 and 90° (1.0–1.4 rad) of hip flexion, whereas the GM contribution is greatest at hip flexion angles less than 40° (0 to ~0.7 rad). Similarly, the hamstrings greatest contribution to knee flexion supposedly occurs between 20 and 50° (0.4–0.9 rad) knee flexion (36). These authors also state that hamstring length is much more sensitive to changes in hip flexion than changes in knee flexion. Hip and knee kinematics during late swing suggest that the MH and BF contribute more to energy generation at the hip during INC, while serving more to absorb energy at the knee during LSSF. Further, increased hip angles during INC allowed the hamstrings to produce greater hip extensor moments than in LSSF (13). In similar fashion, the GA is much better able to generate knee flexor moments at knee angles of less than 40° flexion ($< \sim 0.7$ rad) and provides its highest contribution to the knee flexor moment near full extension (13). Thus, facilitation of the GA before footstrike during LSSF may be to absorb additional energy at the knee, and subsequently decelerate the leg in preparation footstrike.

The largest differences between the INC and LSSF conditions occurred during stance. Hip, knee, and ankle extensor ROM as well as the EMG amplitudes of the monoarticular GM, VL, and SOL muscles during stance were all greater in the INC condition, suggesting greater energy generation. Although we have no stance kinetic data, it is likely that a extensor/flexor moment pattern at the hip and dominant knee extensor and ankle plantar-flexor moments also took place during INC, as the velocity patterns were similar to the level conditions. In addition, the EMG amplitudes of the antagonists MH and BF were significantly lower during INC than in LSSF, suggesting that the net extensor knee moment may have been increased further still. Caution should accompany this interpretation as inhibition of the hamstrings may reduce the transfer of energy from knee to hip during powerful leg extensions, theoretically reducing the hip extensor power (10,16,20). However, Jacobs et al. (20) have estimated that the hamstrings contribution to hip extension via energy transfer from knee to hip is only 11% during sprint running. Thus, it seems that inhibition of the hamstrings may be desirable trade-off in order to provide less antagonistic activity to the crucial knee extensor moment at the knee during push-off.

The phenomenon of monoarticular energy generation and biarticular energy transfer has been illustrated in several movements (14–16,18–20,32). Our data indicate that the INC condition elicited the general proximal to distal muscle activity sequence associated with explosive leg extensions in sprinting and vertical jumping (2,18,20). It has been suggested that the biarticular RF transfers energy generated by the monoarticular hip extensors to the knee joint to assist in its concomitant extension (18–20,32). Similarly, the biarticular GA is purported to transfer energy generated by knee extensors to the ankle to assist in plantar-flexion. This transfer mechanism can increase the effective energy at the knee and ankle by about 30% during a sprint push-off (20). The large increases in EMG of the monoarticular GM, VL, and SOL and the biarticular RF and GA combined with the greater extensor ROM of all three joints suggest that such a transfer mechanism was facilitated during stance in the INC condition. During late swing a similar mechanism involving MH and BF may transfer energy from the rapidly decelerating knee joint to the hip to help with hip extension.

Sprint training implications. Many athletic trainers and therapists agree that the most effective training programs are those that provide increased muscular loading in a sport-specific manner, which for sprinting would include high-velocity movements (8). Several aspects of incline running suggest it may be a viable activity for enhancing sprint speed. The INC condition elicited significantly higher levels of EMG activity during both the stance and swing phases of the gait cycle as compared to level running at either the same speed or stride frequency. In addition, all lower extremity joint angular velocities were significantly higher during the push-off phase of incline running. However, the general patterns of EMG activity and joint kinematics were similar between incline and level running. Thus, incline running seems to satisfy the requirements of being both a sport-specific and high velocity training activity for improving sprint speed.

Previous studies have suggested that the extensor muscles are primarily responsible for generating propulsive force during the push-off phase (20,32). The present data reveal that the EMG amplitudes of all monoarticular extensors were significantly higher during the INC condition. Coupled with an increase in the extensor range of motion during push-off, it is reasonable to assume that more energy was generated by these monoarticular extensors during INC. Further, EMG amplitudes of the biarticular RF and GA and differences in lower extremity joint kinematics during the stance phase suggest that facilitation of energy transfer from hip to knee and from knee to ankle may take place during incline running.

Swing-phase kinetic analysis revealed that significantly more energy was generated during hip flexion and hip extension in the incline condition. Average hip flexor velocities were also significantly higher during incline running. Mann et al. (24) suggested that the main muscle groups that increase the speed of gait were the hip flexors and the knee extensors. Thus, the enhanced muscular loading of the hip flexors that takes place during incline running would be conducive to enhancing running speed.

This study was supported in part by Acceleration Products, Inc., Fargo, ND. The expert technical assistance of Saunders Whittlesey is gratefully acknowledged.

Address for correspondence: Stephen C. Swanson, M.S., The Orthopedic Specialty Hospital 5848 South 300 East, Murray, UT 84107; E-mail: tosswans@ihc.com.

REFERENCES

1. Basmajian, J. V., and C. J. Deluca. *Muscles Alive, Their Functions Revealed by Electromyography*. Baltimore: Williams & Wilkins, 1985, pp. 19–64.
2. Bobbert, M. F., P. A. Huijing, and G. J. Van Ingen Schenau. An estimation of power output and work done by the human triceps surae muscle-tendon complex in jumping. *J. Biomech.* 18:899–906, 1986.
3. Buzcek, F. L., and P. R. Cavanagh. Stance phase knee and ankle kinematics and kinetics during level and downhill running. *Med. Sci. Sports Exerc.* 22:669–677, 1990.
4. Caldwell, G. E., and L. W. Forrester. Estimates of mechanical work and energy transfers: demonstration of a rigid body power model of the recovery leg in gait. *Med. Sci. Sports Exerc.* 24: 1396–1492, 1992.
5. Chapman, A. E., and G. E. Caldwell. Factors determining changes in lower limb energy during swing in treadmill running. *J. Biomech.* 16:69–77, 1983.
6. Chapman, A. E., and G. E. Caldwell. Kinetic limitations of maximal sprinting speed. *J. Biomech.* 16:78–83, 1983.
7. Corcos, D. M., G. L. Gottlieb, M. L. Latash, G. L. Almeida, and G. C. Agarwal. Electromechanical delay: an experimental artifact. *J. Electromyogr. Kinesiol.* 2:59–68. 1992.
8. Delecluse, C., H. Van Coppenolle, E. Willems, M. Van Leemputte, R. Diels, and M. Goris. Influence of high-resistance and high-velocity training on sprint performance. *Med. Sci. Sports Exerc.* 27:1203–1209, 1995.
9. Derrick, T. R., J. Hamill, and G. E. Caldwell. Energy absorption of impact during running at various stride lengths. *Med. Sci. Sports Exerc.* 30:128–135, 1998.
10. Doorenbosch, C. A., T. G. Welter, and G. J. Van Ingen Schenau. Intermuscular co-ordination during fast contact control leg tasks in man. *Brain Res.* 751:239–246, 1997.
11. Hamill, C. L., T. E. Clarke, E. C. Frederick, L. J. Goodyear, and E. T. Howley. Effects of grade running kinematics and impact force. *Med. Sci. Sports Exerc.* 16:185, 1984.
12. Hintermeister, R. A., D. O'Connor, C. J. Dillman, C. L. Suplizio, G. W. Lange, and J. R. Steadman. Muscle activity in slalom and giant slalom skiing. *Med. Sci. Sports Exerc.* 27:315–322, 1995.
13. Hoy, M. G., F. E. Zajac, and M. E. Gordon. A musculoskeletal model of the human lower extremity: the effect of muscle, tendon, and moment arm on the moment-angle relationship of musculotendon actuators at the hip, knee, and ankle. *J. Biomech.* 23:157–169, 1990.
14. Ingen Schenau, G. J. Van, M. F. Bobbert, and A. J. Van Soest. The unique action of bi-articular muscles in leg extensions. In *Multiple Muscle Systems: Biomechanics and Movement Organization,* J. Winters and S. L.-Y. Woo, (Eds.). Berlin: Springer, 1990, pp. 591–607.
15. Ingen Schenau, G. J. Van, P. J. M Boots, G. De Groot, R. J. Snackers, and W. W. L. M. Van Woensel. The constrained control of force and position in multi-joint movements. *Neuroscience* 46:197–207, 1992.
16. Ingen Schenau, G. J. Van, W. M. M. Dorssers, T. G. Welter, A. Beelen, G. De Groot, and R. Jacobs. The control of monoarticular muscles in multi-joint leg extensions in man. *J. Physiol.* 484:247–254, 1995.
17. Iverson, J. R., and T. A. McMahon. Running on an incline. *J. Biomech. Eng.* 114:435–441, 1992.
18. Jacobs, R., and G. J. Van Ingen Schenau. Intermuscular coordination in a sprint push-off. *J. Biomech.* 25:953–965, 1992.
19. Jacobs, R., M. F. Bobbert, and G. J. Van Ingen Schenau. Function of mono- and biarticular muscles in running. *Med. Sci. Sports Exerc.* 25:1163–1173, 1993.
20. Jacobs, R., M. F. Bobbert, and G. J. Van Ingen Schenau. Mechanical output from individual muscles during explosive leg extensions: the role of bi-articular muscles. *J. Biomech.* 29:513–523, 1996.
21. Klein, R. M., J. A. Potteiger, C. J. Zebas. Metabolic and biomechanical variables of two incline conditions during distance running. *Med. Sci. Sports Exerc.* 29:1625–1630, 1997.
22. Kram, R., T. M. Griffin, J. M. Donelan, and Y. H. Chang. Force treadmill for measuring vertical and horizontal ground reaction forces. *J. Appl. Physiol.* 85:764–769, 1998.
23. Lafortune, M., and E. Hennig. Effects of velocity and uphill slope on tibial shock during running. Proceedings of the Fifth Biennial Conference and Human Locomotion Symposium of the Canadian Society for Biomechanics. C. E. Cotton, et al. (Eds.). University of Ottawa, Canada, 1988, pp. 94–95.
24. Mann, R. A., G. T. Moran, and S. E. Dougherty. Comparative electromyography of the lower extremity in jogging, running, and sprinting. *Am. J. Sports Med.* 14:501–510, 1986.

25. Mero, A., and P. V. Komi. Force, EMG, and elasticity-velocity relationships at submaximal, maximal and supramaximal running speeds in sprinters. *Eur. J. Appl. Physiol.* 55:553–561, 1986.
26. Mero, A., and P. V. Komi. Electromyographic activity in sprinting at speeds ranging from sub-maximal to supra-maximal. *Med. Sci. Sports Exerc.* 19:266–274, 1987.
27. Mero, A., P. V. Komi, and R. J. Gregor. Biomechanics of sprint running. *Sports Med.* 13:376–392, 1992.
28. Milliron, M. J., and P. R. Cavanagh. Sagittal plane kinematics of the lower extremity during distance running. In: *Biomechanics of Distance Running,* P. R. Cavanagh (Ed.). Champaign, IL: Human Kinetics, Inc., 1990, pp. 65–100.
29. Nigg, B. M., R. W. Deboer, and V. Fisher. A kinematic comparison of overground and treadmill running. *Med. Sci. Sports Exerc.* 27:98–105, 1995.
30. Nummela, A., H. Rusko, and A. Mero. EMG activities and ground reaction forces during fatigued and nonfatigued sprinting. *Med. Sci. Sports Exerc.* 6:605–609, 1994.
31. Piazza, S. J., and S. L. Delp. The influence of muscles on knee flexion during the swing phase of gait. *J. Biomech.* 29:723–733, 1996.
32. Prilutsky, B. I., and V. M Zatsiorsky. Tendon action of two-joint muscles: transfer of mechanical energy between joints during jumping, landing, and running. *J. Biomech.* 27:25–34, 1994.
33. Robertson, D. G. E., and D. A. Winter. Mechanical energy generation, absorption, and transfer amongst segments during walking. *J. Biomech.* 13:845–854, 1980.
34. Roy, R. R., D. L. Hutchinson, D. J. Pierotti, J. A. Hodgson, and V. R. Edgerton. EMG patterns of rat ankle extensors and flexors during treadmill locomotion and swimming. *J. Appl. Physiol.* 70:2522–2529, 1991.
35. Tokuhiro, A., H. Nagashima, and H. Takechi. Electromyographic kinesiology of lower extremity muscles during slope walking. *Arch. Phys. Med. Rehabil.* 66:610–613, 1985.
36. Visser, J. J., J. E. Hoogkamer, M. F. Bobbert, and P. A. Huijing. Length and moment arm of human leg muscles as a function of knee and hip-joint angles. *Eur. J. Appl. Physiol.* 61:453–460. 1990.
37. Winter, D. A. Moments of force and mechanical power in jogging. *J. Biomech.* 16:91–97, 1983.
38. Winter, D. A. Biomechanics of normal and pathological gait. *J. Mot. Behav.* 21:337–355, 1989.
39. Winter, D. A. *Biomechanics and Motor Control of Human Movement.* New York: John Wiley & Sons, Inc., 1990, pp. 37–61.
40. Zipp, P. Recommendations for the standardization of lead positions in surface electromyography. *Eur. J. Appl. Physiol.* 50:41–54, 1982.

APPENDIX

The subjects performed five standard isometric contractions (SICs) to provide a relative reference of EMG activity in each muscle during the test conditions (12,18). During these SICs, joint angles were measured using a manual goniometer, while external resistance was provided by immovable objects. Two seconds of EMG data were collected during each SIC, performed for approximately 3 s in the exact order as follows:

SIC 1. For TA: maximal dorsi-flexion against external resistance with ankle joint at 90°.

SIC 2. For GA and SOL: maximal plantar flexion standing on one leg.

SIC 3. For RF and VL: maximal right leg extension while standing with the right knee at 0° flexion.

SIC 4. For GM: maximal right hip extension while standing on the left leg. The subjects were instructed to extend their right leg behind them as far as possible.

SIC 5. For BF and MH: maximal knee flexion against external resistance with the knee joint at 35° flexion.