Muscular and metabolic costs of uphill backpacking: are hiking poles beneficial?

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ABSTRACT

KNIGHT, C. A. and G. E. CALDWELL. Muscular and metabolic costs of uphill backpacking: are hiking poles beneficial? *Med. Sci. Sports Exerc.*, Vol. 32, No. 12, 2000, pp. 2093–2101. **Purpose:** The purpose of the present study was to compare pole and no-pole conditions during uphill backpacking, which was simulated on an inclined treadmill with a moderately heavy (22.4 kg, 30% body mass) backpack. **Methods:** Physiological measurements of oxygen consumption, heart rate, and RPE were taken during 1 h of backpacking in each condition, along with joint kinematic and electromyographic comparisons from data collected during a third test session. **Results:** The results showed that although imposing no metabolic consequence, pole use elicited a longer stride length (1.27 vs 1.19 m), kinematics that were more similar to those of unloaded walking, and reduced activity in several lower extremity muscles. Although pole use evoked a greater heart rate (113.5 vs 107 bpm), subjects were backpacking more comfortably as indicated by their ratings of perceived exertion (10.8 vs 11.6). The increased cardiovascular demand was likely to support the greater muscular activity in the upper extremity, as was observed in triceps brachii. **Conclusion:** By redistributing some of the backpack effort, pole use alleviated some stress from the lower extremities and allowed a partial reversal of typical load-bearing strategies. **Key Words:** LOAD-CARRIAGE, HIKING, TREKKING, ELECTROMYOGRAPHY, KINEMATICS, OXYGEN CONSUMPTION

Prolonged backpacking typically requires the carriage of a substantial load of equipment and provisions. Physical manifestations of this load carriage include kinematic and inertial consequences, energetic effects, and local muscle fatigue (8). For example, Han et al. (10) showed that increasing the load in a backpack produced greater trunk flexion, greater knee flexion at heelstrike, and a more extended knee during stance. The physical stresses of load carriage have been studied extensively in attempts to develop strategies to minimize them and improve the comfort and safety of backpacking (7,9,14,19). One strategy that has received little attention is the use of hiking poles, which are essentially modified ski poles that can be adjusted in length. Historically, the use of hiking poles has been more prevalent in Europe, although their use is expanding with the increased popularity of outdoor recreation and endurance competitions such as the Eco-Challenge.

According to proponents, hiking poles alleviate the discomfort of sore knees, swollen feet, and a hyperflexed climbing posture while improving balance. If these anecdotal claims are true, then pole use may help to reduce the incidence of overuse injuries of the musculoskeletal system over the course of months and years of frequent backpacking (6,14). However, there is currently little research supporting any of these claims. In one of the few studies of which we are aware, subjects walked (without a backpack) over mixed terrain using ski poles that were instrumented with spring gauges (18). During a given stride the poles accommodated 10, 13, and 7 kg of load on uphill, downhill, and level terrain, respectively. When extrapolated to a longer time frame, this represents 28,800, 33,600, and 13,500 $kg \cdot h^{-1}$ in these three conditions. The authors suggested that this accommodation of load justifies pole use during recovery from lower extremity injuries and to prevent overuse injuries in hikers. For the backpacker, hiking poles may have an even greater contribution to load accommodation.

Therefore, the purpose of this study was to investigate the use of hiking poles and their potential to reduce the stresses imposed by a heavy backpack load. Metabolic, kinematic and electromyographical (EMG) data were collected as subjects backpacked on an inclined treadmill both with and without hiking poles. It was hypothesized that the use of poles would result in altered kinematics as described by angular displacement and velocity of the trunk and limb segments, and that pole use would result in a shift in

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muscular activity from the lower to the upper extremities. Further, it was hypothesized that the oxygen cost of backpacking with hiking poles would not differ from that of backpacking without poles, despite the use of additional upper body musculature.

METHODS

Ten volunteers (5 male and 5 female) were recruited as subjects, each of whom were regular backpackers who took more than two trips per year carrying a full size pack for at least 4 h. Each subject provided their written informed consent and answered a Physical Activity Readiness Questionnaire (PAR-Q). Volunteers were not selected as subjects if they identified potential risks associated with their participation. Mean ($\pm$ SD) age, height, and body mass were 30 ± 12 yr, 172 ± 7 cm, and 75 ± 10 kg, respectively. Each subject was required to visit the laboratory on three occasions for testing.

On two different days (separated by 48–72 h), each subject backpacked for 60 min, in either a poles (condition P) or no-poles (condition NP), presented in a counterbalanced order. For each subject, the poles were adjusted to an appropriate length in which the elbow was at 90° while the pole was held in a vertical position and in contact with the ground. Subjects walked on a treadmill inclined at 5°, carrying an internal frame backpack loaded to approximately 30% body mass (21–30 kg). Previous research has determined this load to be appropriate for prolonged backpacking (24,28). The backpack was loaded such that the approximate center of mass location was similar to that of actual backpacking. Backpack adjustments were optimized to improve fit and comfort for each subject and maintained throughout testing. The treadmill velocity was established *a priori* such that the subject's heart rate was between 55 and 65% of their age-predicted maximum in the NP condition. During each of the two 60-min trials, ratings of perceived exertion (RPE, Borg Scale 6–20 [2]), were collected at 10-min intervals. Heart rate was measured at 2-min intervals with a Polar Vantage heart rate monitor (Polar Electro Inc., Port Washington, NY), whereas a Teem-100 portable gas analysis system (Aerosport Inc., Ann Arbor, MI) was used to measure $\dot{V}O_2$ ($mL \cdot kg^{-1} \cdot min^{-1}$) at 1-min intervals. To reduce the effects of thermal overload associated with prolonged exercise, subjects ingested $\approx$250 mL H_2O 15 min before and again at 30 min into testing. Subjects were tested at the same time of day, and were asked to eat similar meals before each metabolic test session.

The same testing conditions and assigned presentation order were used during the third laboratory visit, when kinematic and EMG data were collected. The metabolic sessions had given each subject at least 1 h of experience in each condition before kinematic and EMG data acquisition. Sagittal view kinematics were videotaped with a 200-Hz NAC camera as the subject backpacked on the treadmill for approximately 15 min. Reflective markers were placed over the humeral head, greater trochanter, lateral aspect of the tibial tuberosity, lateral malleolus, lateral aspect of the calcaneus, and the head of the fifth metatarsal. The positions of these markers in each frame were digitized from five strides in each condition, using a VP110 video processor (Motion Analysis Corp., Santa Rosa, CA). Kinematic data were smoothed using a fourth-order low-pass Butterworth filter, with cut-off frequencies (10-12 Hz) determined using a residual analysis technique (11). Ankle, knee, hip, and trunk angles were calculated, and angular velocities were determined using finite difference methods. A mechanical heel switch illuminated an LED within the view of the video camera to define individual stride cycles. Discrete variables from the angular position and velocity records were calculated, including peak angles, peak velocities and their timing, peak knee flexion during early stance, and joint ranges of motion.

EMG data were recorded using Ag/AgCl preamplified bipolar surface electrodes (model 544 Therapeutics Unlimited, Iowa City, IA, frequency response 20–4000Hz,) from gastrocnemius (GA), soleus (SO), rectus femoris (RF), vastus lateralis (VL), semimembranosis (SM), left and right erector spinae (lumbar portion; ESL, ESR), and triceps brachii (TB). To reduce skin/electrode impedance, the areas of electrode placement were shaved, abraded with sandpaper, and washed with rubbing alcohol, with electrolyte gel used to improve conductance. The preamplified EMG was coupled to an active two-pole high-pass filter (-3 db cut-off of 20 Hz, common-mode rejection 87 db at 60 Hz) to improve the signal to noise ratio and minimize cable artifact. The EMG signals were amplified (gain accuracy $\pm$ 10%, gain linearity $> \pm$ 0.5% full-scale output) at selected gains from 1K to 20K to optimize resolution. A 12-bit analog-to-digital conversion board (Computer Boards Inc., Mansfield, MA) was used to sample (1 kHz) the eight EMG channels and the heel switch channel that lit the LED. Five individual strides of EMG data were identified for further analysis.

For each EMG channel, the DC-offset was removed and the data scaled to the voltage at the skin/electrode interface. Any possible movement artifact was removed by a zero-lag high-pass 20-Hz Butterworth filter. The signal was full-wave rectified, and average amplitude and integral of EMG throughout each stride were calculated. For display purposes, the rectified signals were low pass filtered (Butterworth, 20 Hz cut-off) and standardized in time to 100% stride cycle. Ensemble averages for each condition were calculated using these standardized linear envelope EMG signals.

A two-factor repeated measures analysis of variance was used for statistical analysis of RPE, HR, and $\dot{V}O_2$ results. Condition and time were tested as main effects in addition to the condition by time interaction. For kinematic and EMG data, intraclass correlation coefficients were calculated. Because the coefficients were typically greater than 0.95, the five strides from each subject in each condition were averaged. A paired *t*-test was applied to the resulting means. To evaluate mean differences that were typically small, effect size (ES) was calculated as the mean difference between conditions divided by the standard deviation of the no-pole condition. The effect sizes and *P*-values were evaluated

TABLE 1. Individual and mean (SD) subject characteristics and testing parameters.

Subject	Gender	Age (yr)	Height (cm)	Mass (kg)	Load (kg)	Load (% mass)	Velocity ($m \cdot min^{-1}$)	Grade (°)
1	M	31	168	70	21	30	57	5
2**	F	20	170	70	21	30	57	5
3	F	56	163	72	21	29	49	2
4*	F	19	170	70	21	30	54	4
5	M	22	175	91	21	23	60	4
6	F	21	175	60	21	35	55	5
7	M	31	185	87	21	24	53	5
8*	M	26	178	85	26	31	71	5
9	F	46	163	63	21	33	57	5
10**	M	27	175	75	30	39	96	5
Mean		29.9	172.2	74.5	22.4	30.4	60.8	4.5
SD		(12.1)	(7.0)	(10.3)	(3.1)	(4.7)	(13.5)	(1.0)

* Only kinematic and EMG results.
** Only $\dot{V}O_2$, HR, and RPE results.

together as the basis for discussing differences between conditions.

RESULTS

Individual subject characteristics are given in Table 1. Due to technical difficulties, metabolic results ($\dot{V}O_2$, HR, and RPE) were not available for subjects 4 and 8, while kinematic and EMG results do not include subjects 2 and 10. To promote successful completion of all three experimental sessions, three subjects were tested at inclines lower than the prescribed 5°. In these subjects, treadmill velocity was adjusted to achieve the same 55–65% heart rate intensity. The average backpack load was 22.4 ± 3.1 kg (30.4 ± 4.7% body mass), and the average testing velocity was 60.8 ± 13.5 $m \cdot min^{-1}$ (3.7 ± 0.8 $km \cdot hr^{-1}$).

Metabolic Responses

Figure 1 depicts oxygen consumption results expressed relative to body mass. Over the entire 60 min, the average $\dot{V}O_2$ for P and NP were 18.9 ± 6.6 and 18.2 ± 6.8 $mL \cdot min^{-1} \cdot kg^{-1}$, respectively. To assess changes in metabolic cost over time, 10-min periods from the beginning (after steady-state was established) and the end of testing were compared. The average metabolic cost from min 11 to 20 was 18.8 ± 6.7 for P and 17.6 ± 6.5 $mL \cdot min^{-1} \cdot kg^{-1}$ for NP. During the later period (51–60 min), the metabolic cost was 19.1 ± 6.9 for P and 18.7 ± 7.5 $mL \cdot min^{-1} \cdot kg^{-1}$ for NP. The metabolic cost did not change with pole use (ES = 0.10, P = 0.126) or over time (ES = 0.11, P = 0.230), and there was no interaction between condition and time (F=1.32, P = 0.288).

During the 60-min test, mean HR (Fig. 1) was higher (ES = 0.52, $P \leq 0.010$) with poles (113.5 ± 12.8 bpm) than without (107 ± 12.6 bpm). In NP, the corresponding exercise intensity level was 56 ± 5.7% of the average age-predicted maximal HR. The mean HR for P was consistent from early (113.5 ± 13.7 bpm) to late test (114.1 ± 13.2 bpm), whereas HR increased (ES = 0.38, $P \leq 0.090$) from 105.9 ± 11.4 to 110.2 ± 14.2 in NP. There was no condition by time interaction (F=3.02, P = 0.126, Fig. 1).

The perceived exertion (RPE, Fig, 2) throughout the one h test was less with poles (10.8 ± 0.36) than without poles (11.6 ± 0.43; ES = 1.9, $P \leq 0.003$) and, in both conditions combined, it increased from the beginning (10.6 ± 0.44) to the end of testing (11.6 ± 0.71, ES = 2.3, $P \leq 0.002$). There was no interaction between condition and time (P = 0.670).

Kinematics

At the same treadmill velocity, subjects displayed longer stride length (1.27 m) and lower stride frequency (0.76 Hz)

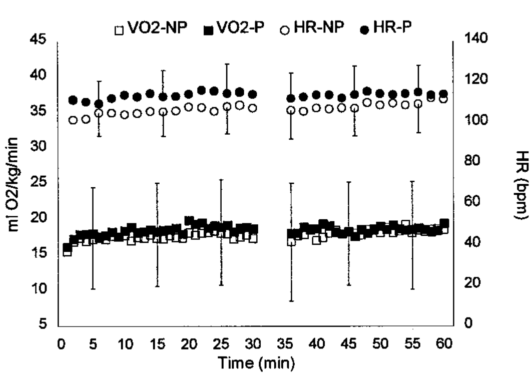

FIGURE 1—Mean (±SD) metabolic cost ($\dot{V}O_2$, squares) was similar while backpacking with (*solid markers*) and without (*open markers*) hiking poles despite a greater heart rate (*circles*) with poles. Measurements were not acquired between 30 and 35 min while subjects were allowed to ingest water. For clarity, only a few SD bars are shown.

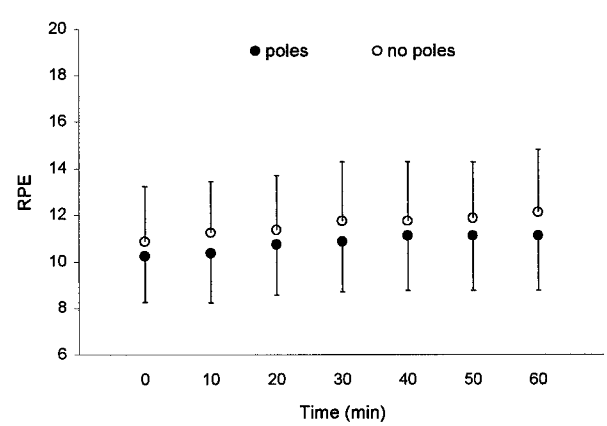

FIGURE 2—Mean (±SD) ratings of perceived exertion were consistently less while backpacking with hiking poles (*solid circles*) than in the no-pole condition (*open circles*; Borg Scale 6–20, ref. 2).

TABLE 2. Mean (SD) kinematic variables with and without hiking poles.

Kinematic Variable	No Poles	Poles	Δ	ES	*P*
Stride length (m)	1.19 (0.15)	1.27 (0.14)	0.08	0.53	0.0194
Stride frequency (Hz)	0.80 (0.08)	0.76 (0.09)	−0.05	0.63	0.0179
Ankle ROM (°)	35.2 (5.7)	34.9 (7.5)	−0.3	0.05	0.7630
Knee ROM early stance (°)	9.1 (5.0)	11.8 (5.9)	2.8	0.56	0.0642
Hip ROM (°)	49.2 (7.8)	50.7 (7.0)	1.6	0.21	0.2054
Trunk ROM (°)	8.7 (2.4)	8.2 (1.8)	−0.6	0.25	0.2688

Δ, mean difference (P-NP); ES, effect size.

in the pole condition compared with the no pole condition (1.19 m and 0.80 Hz). Therefore, the average stride length was 6.7% longer (ES = 0.51, $P \leq 0.020$) and stride frequency was 6.3% less (ES = 0.65, $P \leq 0.018$) when backpacking with poles (Table 2). Figures 3 and 4 display the ensemble angular displacement and velocity curves from each condition, normalized to percent stride (right heel contact to the next right heel contact).

In early stance phase, the ankle plantarflexed to a neutral angle followed by approximately 15° of dorsiflexion as the leg rotated forward over the foot while in stance. Near the end of stance (≈60%), the ankle plantarflexed strongly into swing and then dorsiflexed for toe clearance as the foot came forward. One salient feature is seen in early stance (0–20%), when the knee flexed to an early peak as load was accepted. For the pole condition, the knee range of motion during this initial loading phase (Table 2) was 2.76° greater than without poles (ES = 0.59, $P \leq 0.065$). The knee approached full extension during stance, followed by a large flexion/extension phase during swing before the next heelstrike. Beginning near its most flexed position, the hip extended throughout stance and then returned toward peak flexion before the next heelstrike. Finally, the trunk segment, which maintained a flexed posture throughout the stride, extended slightly after heelstrike, flexed throughout the majority of stance, and extended to a more erect posture after toe-off.

Although the kinematic patterns were similar, Tables 3 and 4 illustrate differences between conditions. For the pole condition, the knee was 3.4° less flexed at heelstrike (ES = 1.10, $P = 0.132$), minimum knee flexion (near toe off) was 1.09° greater (ES = 0.22, $P \leq 0.037$), and maximum knee flexion (during swing) was 1.79° less (ES = 0.30, $P \leq 0.046$). Also in the pole condition, joint velocities were lower in many, but not all, joints (Table 4). For example, maximum knee flexion velocity just after toe-off was lower by $29.1°\cdot s^{-1}$ (ES = 0.76, $P = 0.017$), maximum hip extension velocity in early stance was $22.9°\cdot s^{-1}$ lower (ES = 0.60, $P \leq 0.045$), and maximum hip flexion velocity in swing phase decreased by $12.23°\cdot s^{-1}$ (ES = 0.43, $P = 0.123$). Pole use also affected trunk movement, with maximum trunk flexion velocity (during stance) lower by $8.2°\cdot s^{-1}$ (ES = 0.58, $P = 0.056$) and the maximum trunk extension velocity (near toe-off) decreased by $6.0°\cdot s^{-1}$ (ES = 0.87, $P = 0.042$). The only joint to show increased velocity in the P condition was the ankle, which had higher peak dorsiflexion velocity during swing by $7.43°\cdot s^{-1}$ (ES = 0.32, $P = 0.011$).

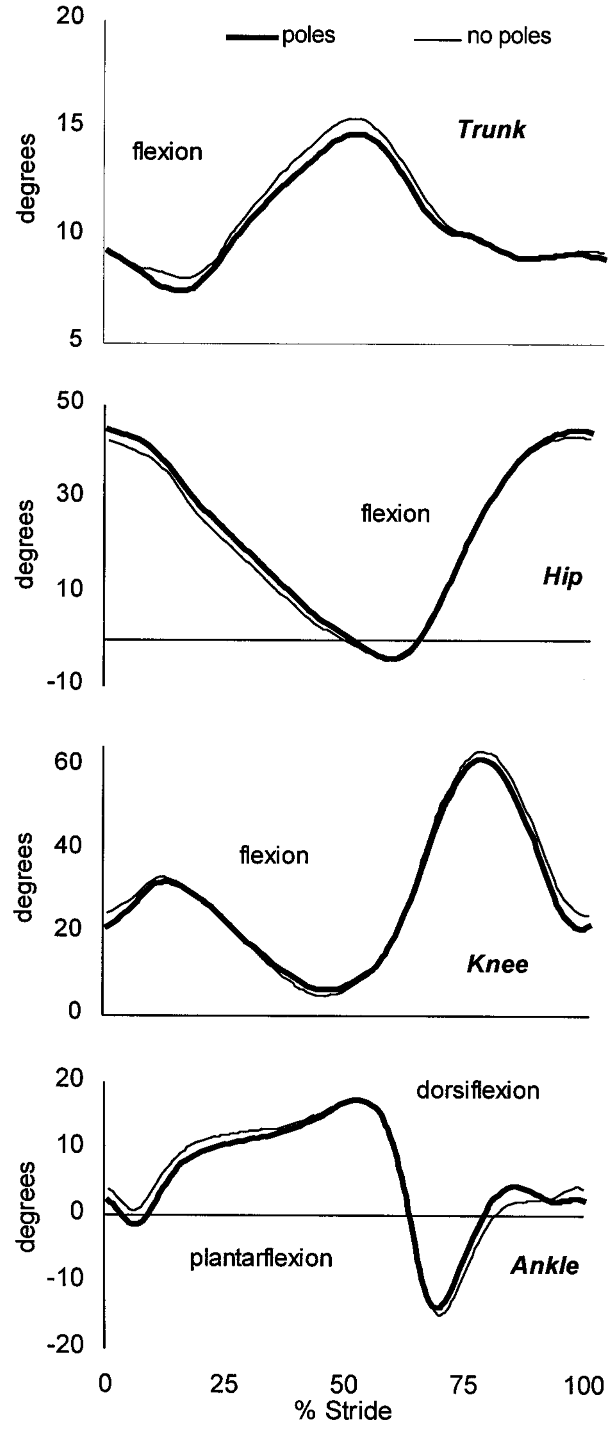

FIGURE 3—Angular displacement of the ankle, knee, hip, and trunk expressed as a function of % stride. Mean ensemble averages are shown for pole (*thick line*) and no-pole (*thin line*) conditions.

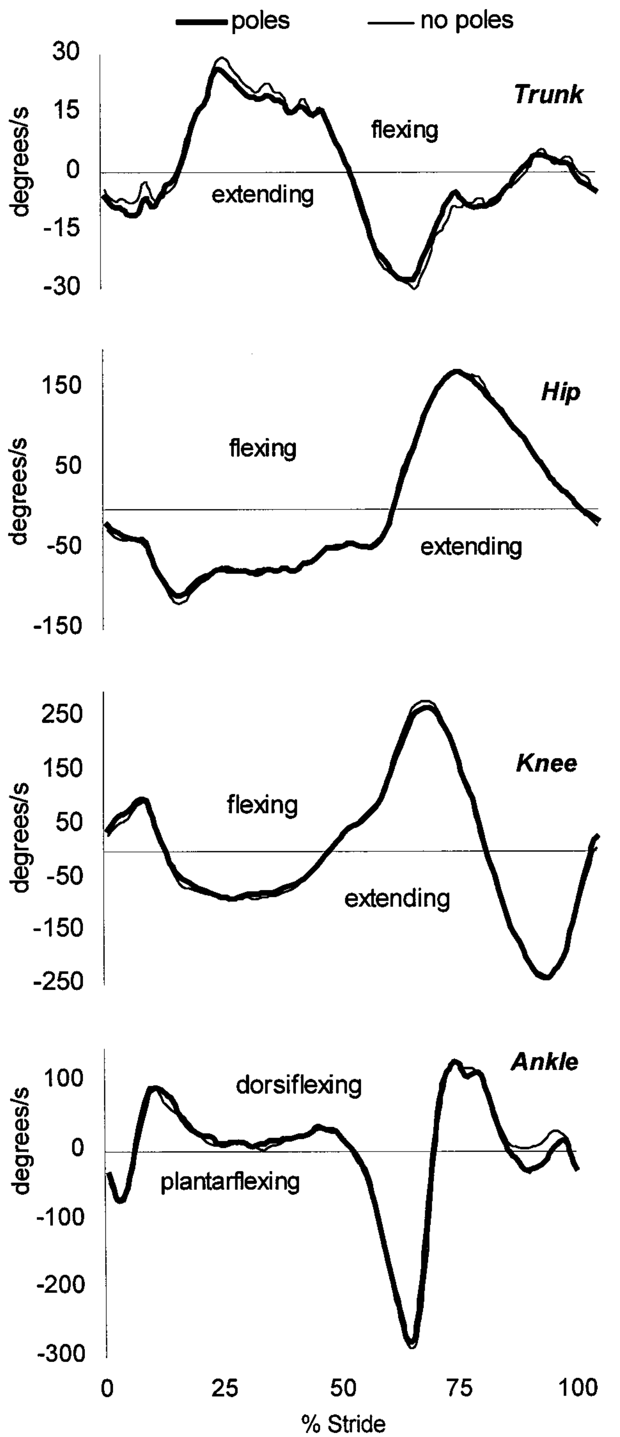

FIGURE 4—Angular velocity of the ankle, knee, hip, and trunk expressed as a function of % stride. Mean ensemble averages are shown for pole (*thick line*) and no-pole (*thin line*) conditions.

Electromyography

Figure 5 displays the ensemble EMG linear envelopes from each condition normalized to percent stride, whereas selected EMG variables are shown in Tables 5 and 6. For the thigh muscles, RF and VL were active primarily in early stance, whereas BF was active in early stance and before heelstrike. In the more distal muscles, SO was active early and throughout stance, whereas GA was active before toe-off. The back muscles (ESL and ESR) were active before and after right heelstrike, and near right toe-off, which was coincidental with left heelstrike. TB showed very little activity without poles, but in the pole condition high activity was seen beginning near right toe-off, which corresponded to the load bearing phase of the right pole during stance phase of the left lower extremity.

Several muscles displayed changes in their myoelectric activity with pole use. As expected, the activity of TB increased in both the integral (iEMG, ES = 8.02, $P \leq 0.006$, 295% higher) and average amplitude (ES = 7.56, $P \leq 0.008$, 268% higher) measures in the pole condition. Concurrently, several lower extremity muscles showed lower average EMG values in the pole condition. For the hamstrings muscle BF, the activity decrease was seen in both average amplitude (ES = 0.51, $P \leq 0.013$) and iEMG (ES = 0.24, $P \leq 0.046$) despite the increase in P condition stride time. In contrast, the knee extensor VL average amplitude decreased (ES = 0.34, $P \leq 0.056$), but its iEMG remained the same (ES = 0.19, $P \leq 0.246$). Notice that both quadriceps muscles (RF and VL, Fig. 5) had lower peak amplitude in the early part of stance during weight acceptance with the poles. However, RF showed no EMG differences between conditions because of higher activity following this phase and throughout stance with pole use. Also in condition P, GA had less activity as indicated by both measures (iEMG ES = 0.20, $P = 0.052$; Avg. Amp ES = 0.32, $P \leq 0.024$). Its counterpart SO displayed lower average amplitude (ES = 0.35, $P \leq 0.091$) with no change in iEMG (ES = 0.16, $P \leq 0.267$), likely due to prolonged activity with the increased stride length.

DISCUSSION

Due to the lack of information in the literature, in this study a multidisciplinary approach was used to examine several facets of hiking pole use while backpacking. We found that while using poles the subjects increased their stride length and reduced their stride frequency. Lower extremity joints and the trunk segment saw lower peak velocities, whereas the subjects displayed less knee flexion at heelstrike and greater knee range of motion during early stance weight acceptance. These kinematic changes were accompanied by evidence of less muscular activity in the lower extremity BF, VL, GA, SO, and RF muscles, with a large increase in activity for the arm extensor TB. Metabolic response as measured by $\dot{V}O_2$ did not change with pole use, although average HR increased. Finally, the subjects perceived the pole condition to be less taxing than without poles, as demonstrated by lower RPE values. These changes with pole use are consistent with acceptance of our stated hypotheses.

Although field testing would be a more ecologically valid measure of pole use effectiveness, we felt that strict experimental control was critical in this initial study. Therefore, heavy uphill backpacking conditions were simulated on a treadmill. In this simulation there was limited friction between the pole tips and the treadmill belt, and fewer gait

TABLE 3. Mean (SD) angular position at various stride landmarks and peak angular displacement with and without hiking poles.

All Angles in °	No Poles	Poles	Δ	ES	P
Maximum ankle plantarflexion	−16.6 (7.5)	−16.4 (9.4)	0.2	0.03	0.8800
Maximum ankle dorsiflexion	18.6 (3.7)	18.4 (4.3)	−0.1	0.03	0.8509
Knee angle at heelstrike	25.0 (3.1)	21.6 (4.4)	−3.4	1.1	0.1320
Peak knee flexion early stance	34.0 (5.1)	33.4 (7.2)	−0.7	0.14	0.6639
Minimum knee flexion	4.4 (4.9)	5.5 (5.2)	1.1	0.22	0.0365
Maximum knee flexion	65.0 (5.8)	63.2 (5.5)	−1.8	0.31	0.0458
Maximum hip extension	−4.6 (9.6)	−4.7 (8.8)	−0.1	0.01	0.8173
Maximum hip flexion	44.6 (12.2)	46.0 (12.5)	1.5	0.12	0.2051
Minimum trunk flexion	7.0 (8.4)	6.9 (8.1)	−0.1	0.01	0.7219
Maximum trunk flexion	15.8 (8.0)	15.0 (7.9)	−0.7	0.09	0.1906

Δ, mean difference (P-NP); ES, effect size.

perturbations than one might expect during outdoor backpacking. Therefore, the present results may be a conservative representation of those that would be observed in the field. The subtle differences reported may be different than with varying terrain where the pole-ground interaction may allow more effective pole use compared with the treadmill (21).

Our metabolic and cardiovascular results compare well with the results of Kirk and Schneider (13), specifically in their internal frame backpack condition (carrying 33% body mass at 3% grade at a 86 $m \cdot min^{-1}$ pace). Higher variability in the present $\dot{V}O_2$ results can be explained by the heterogeneity of our sample compared to their all-female sample (13). Their average heart rate was greater than the present NP data (134 ± 30 vs 107.0 ± 12.6 bpm), possibly due to a gender difference or differences such as velocity and/or grade. Legg and Mahanty (15) reported more similar heart rate values (105 ± 13 bpm) and a similar average RPE value (11.9 ± 0.8 vs 11.6 ± 0.43 in NP) with male subjects carrying 35% body mass load at 0% grade and 4.5 $km \cdot h^{-1}$.

The kinematic and EMG data also agree with other literature. The kinematics share similar characteristics with normal treadmill walking (17), with the most obvious differences being increased knee and hip flexion throughout the stride. The load and incline likely caused the more flexed knee, whereas the hip angles can be attributed to the more flexed trunk. Our peak trunk flexion values around mid stride were similar to those of Carlson et al. (5), but we observed relatively little flexion near the ends of the stride, possibly due to attenuated trunk movement under the heavy load. The ensemble burst patterns of the EMG from erector spinae show a similar bimodal pattern as documented by Bobet and Norman (1) and Carlson et al. (5). The patterns of SO, BF, VL, and RF compare roughly to the linear envelopes presented by Winter (27) in a description of normal walking, and GA activity corresponds to the description by Tokuhiro et al. (26) in their examination of slope walking.

As hypothesized, there was no metabolic consequence for the use of hiking poles while backpacking. The increase in metabolic cost of active upper extremity muscles (e.g., TB) was likely offset by a reduced metabolic cost in the lower extremities. Although there was a greater cardiovascular demand (+6% HR) with poles, the subjects perceived their level of exertion to be lower (−7%, RPE). This could be related to the upward HR drift that was seen in NP but not in P. Based on the similar oxygen consumption ($\dot{V}O_2$) in the two conditions, the higher HR in P was not a response to increased overall workload, but rather a result of increased delivery of blood to the upper extremities (23). The higher heart rate in P could perhaps result from increased peripheral resistance associated with upper body exercise or it could be due to the pressor response while gripping the poles.

The kinematic and electromyographic data displayed changes with pole use indicative of an underlying adaptive strategy. Previous studies have illustrated that subjects adjust their kinematics in response to a heavy backpack load. These adjustments include a shortened stride length (16), greater knee flexion at heelstrike, and a straighter knee at mid stance (10). One possible reason for these alterations is to attenuate the potentially higher impact on the lower extremity just after heelstrike caused by heavy backpack loads. This normal backpacking strategy was partially reversed in the P condition, as the poles likely supported some of the load during early stance that would otherwise contribute to the impact. The increase in TB activity with the pole in contact with the ground is indicative of this contribution. The poles allowed the subjects to adopt more normal walking kinematics, including increased stride length and a straighter knee at heelstrike. This strategy also allowed the subjects to decrease the activity level of a number of lower extremity muscles. Although these kinematic and myoelectric differences were slight, it should be noted that the testing duration for these parameters was brief (15 min/

TABLE 4. Mean (SD) angular velocities of the ankle, knee, hip, and trunk with and without hiking poles.

All velocities in $°\cdot S^{-1}$	No Poles	Poles	Δ	ES	P
Peak ankle plantarflexion	188 (79)	212 (92)	24	0.30	0.6779
Peak ankle dorsiflexion	338 (86)	346 (118)	8	0.09	0.0110
Peak knee flexion	308 (42)	278 (60)	−30	0.71	0.0170
Peak knee extension	256 (48)	258 (46)	2	0.04	0.8190
Peak hip flexion	201 (31)	189 (27)	−12	0.39	0.1230
Peak hip extension	149 (40)	126 (19)	−23	0.58	0.0448
Peak trunk flexion	45 (17)	36 (7)	−9	0.53	0.0560
Peak trunk extension	43 (9)	37 (9)	−6	0.67	0.0413

Δ, mean difference (P-NP); ES, effect size.

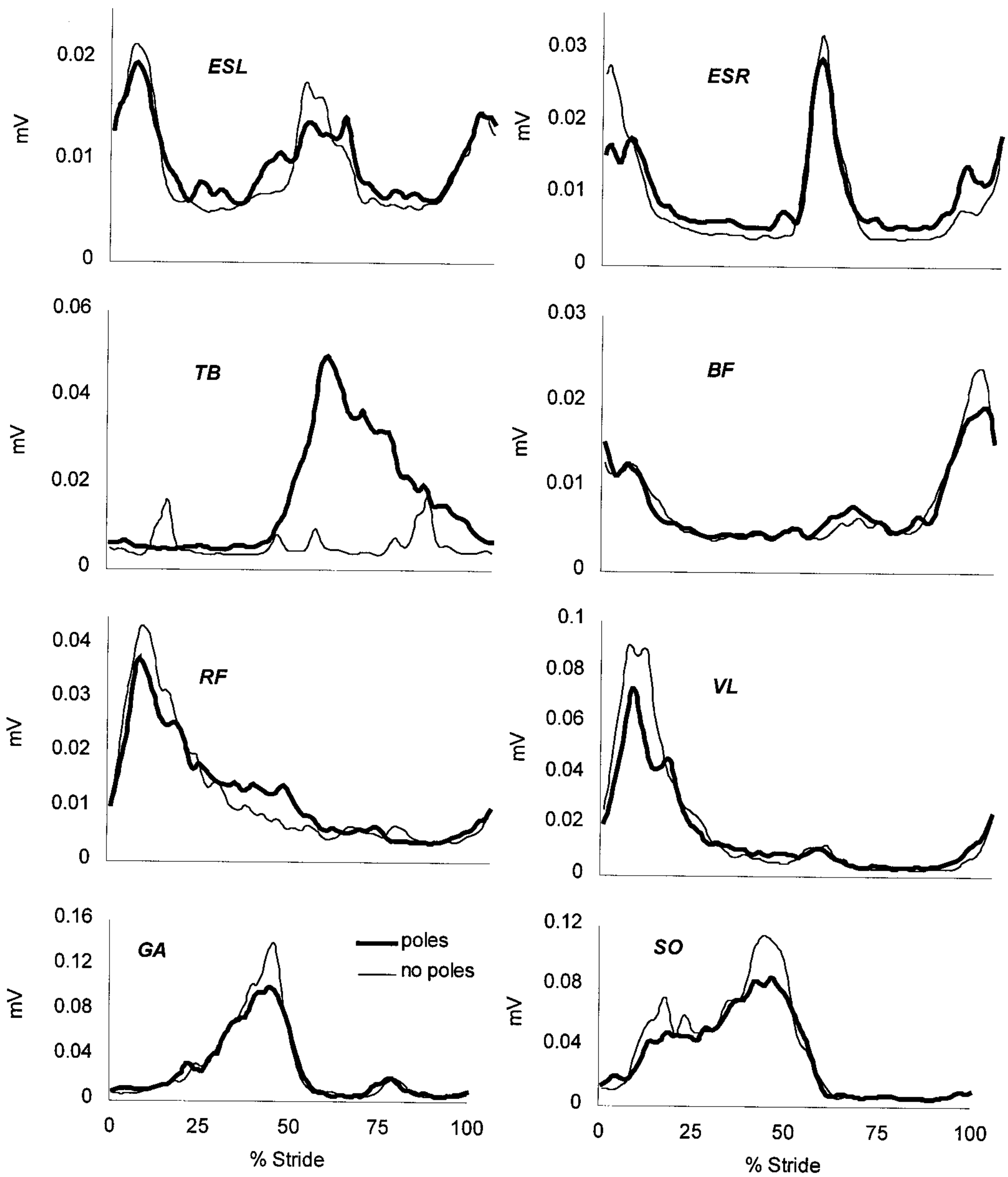

FIGURE 5—Full-wave rectified EMG for test muscles as a function of % stride. Mean ensemble averages are shown for pole (*thick line*) and no-pole (*thin line*) conditions.

condition). It may be the case that prolonged test duration would reveal even greater differences between conditions as fatigue becomes more pronounced.

Another possible strategy in normal backpacking is to reduce muscular strain by stiffening the knee joint and limiting its range of motion during weight acceptance. With poles, the knee was straighter at heelstrike but flexed to the same early peak, thus increasing its range of motion during weight acceptance. At the same time, the corresponding peaks in EMG activity of RF and VL were reduced. Again,

TABLE 5. Mean (SD) EMG integral values from selected muscles with and without hiking poles.

Muscle	No Poles (mV·ms)	Poles (mV·ms)	%Δ	ES	*P*
GA	31.3 (16.4)	28.1 (13.4)	−10.2	0.20	0.0520
SO	36.7 (11.7)	38.4 (11.9)	−4.6	0.15	0.2669
RF	12.0 (5.0)	12.1 (5.2)	0.8	0.02	0.7770
VL	17.4 (6.1)	16.2 (5.8)	−6.9	0.20	0.2459
BF	8.1 (1.8)	7.7 (2.0)	−4.9	0.22	0.0456
ESL	9.8 (5.2)	9.9 (5.1)	1.0	0.02	0.8369
ESR	9.2 (3.5)	10.1 (5.3)	9.8	0.26	0.2228
TB	5.0 (1.8)	19.7 (11.8)	294	8.02	0.0058

%Δ, % difference (P-NP)/NP; ES, effect size.

TABLE 6. Mean (SD) EMG average amplitude values from selected muscles with and without hiking poles.

Muscle	No Poles (mV)	Poles (mV)	%Δ	ES	*P*
GA	0.0249 (0.0121)	0.0210 (0.0094)	−15.6	0.32	0.0243
SO	0.0297 (0.0096)	0.0263 (0.0089)	−11.3	0.35	0.0907
RF	0.0096 (0.0042)	0.0092 (0.0040)	−4.6	0.10	0.4264
VL	0.0139 (0.0050)	0.0122 (0.0045)	−12.3	0.34	0.0551
BF	0.0065 (0.0015)	0.0058 (0.0014)	−11.5	0.51	0.0129
ESL	0.0078 (0.0042)	0.0076 (0.0043)	−3.6	0.07	0.4626
ESR	0.0074 (0.0032)	0.0078 (0.0043)	5.2	0.12	0.5503
TB	0.0040 (0.0014)	0.0147 (0.0092)	267.5	7.56	0.0075

%Δ, % difference (P-NP)/NP; ES = effect size.

because some load was borne by the poles, the normal backpacking strategy to limit the range of motion of the knee was not invoked, and less demand was placed on the knee extensors in this weight acceptance phase.

In our title, we posed the question *Are hiking poles beneficial?* Metabolically, the hiking poles could be considered beneficial if they lowered metabolic requirements, or contributed to the task of backpacking with no metabolic consequence, or with an associated metabolic cost that is redeemed somehow (e.g., less risk of injury). In the current circumstances, with a heavy load on an inclined treadmill, several kinematic and muscular benefits were achieved without additional metabolic cost. However, this may not be the case when poles are used in other conditions. In a pole versus no-pole level gait comparison with no backpack load at self-selected pace, Porcari et al. (20) reported increased metabolic cost, cardiovascular demand, and perceived exertion while using poles. However, under their conditions, the poles may have simply become part of the load carriage task, especially because arm movements may have been exaggerated to raise exercise intensity.

Overuse injuries in backpacking are likely related to the abnormal loads on the lower extremity. Crouse and Josephs (6) reported that musculoskeletal complaints and traumatic injury were frequent and resulted in an average loss of 4.7 d of hiking among Appalachian trail hikers. Hiking poles can perhaps reduce this incidence of injury by shifting some stresses from the lower extremity to the upper extremity. Other results have suggested that the vertical component of the ground reaction force on a walking pole accounts for 26% of total body weight (4). A recent study examined the loads on the knee during steep downhill walking and similarly demonstrated that a part of the load can be borne by hiking poles (22). Although this shift in load may have consequences for the upper extremities as well, it may be that the greater repertoire of available locomotion patterns with poles can be alternated to allow recovery of fatigued muscles. Even if the use of poles becomes unfavorable, the cost of carrying this additional mass (≈0.6 kg) is minimal, estimated at 20.1 $mLO_2 \cdot min^{-1}$ with an elevation of HR by only 0.66 bpm (3).

Another possible benefit of pole use is prevention of traumatic injuries caused by falls on uneven terrain. Jacobson et al. (12) reported significantly longer balance times with a 15-kg backpack on a stability platform using two poles compared with either one or no poles. The authors extrapolated their results to a reduced possibility of falling while standing on loose alpine terrain. Their findings, although encouraging, are limited in their applications to the dynamic stability required while backpacking over similar terrain. The requirement for stability increases with load mass as the backpacker's COM location is raised further above their base of support. In theory, the poles will provide a more stable situation during swing phase because the normal base of support over one foot is improved with the addition of the contralateral pole.

Also linked to variable terrain are the inertial consequences of the movement of the trunk and backpack throughout a gait cycle. In a mechanical analysis of rifle movement during winter biathlon, Frederick (8) found it optimal to minimize the movement of the rifle throughout a stride cycle. Although the movement of the backpack throughout a stride is far less than that of a biathlon rifle, the mass of a heavy backpack is much greater. In the present study, peak trunk flexion and extension velocities were reduced with poles (18% and 14%, respectively). If this trend were augmented in the field where poles can stabilize the trunk and backpack against perturbations, an energy analysis may show even further cost reduction from hiking pole use.

In this study, under the conditions of simulated uphill backpacking, the effect of hiking pole use included the benefits of improved backpacking kinematics, a redistribution of muscular demand, with no additional metabolic cost and improved comfort. Aside from the limitations imposed by the treadmill, other features of hiking pole use could be addressed in future studies. For example, the description of trunk movement could be improved with three-dimensional analysis to add axial rotational information. Further, some conclusions were based on a presumed accommodation of load via the poles. A description of the forces applied to the poles throughout the stride would be useful in quantifying the degree of load accommodation, as others have shown for steep down hill walking (22) and for cross-country skiing (25). Also, the addition of a no-pole, no-load condition would be an asset for interpretations in the backpacking context. On a theoretical level, the strategies for knee motion in load carriage and the changes that occur with pole use could serve as a practical model to study joint stiffness as it pertains to muscular response, energetic demands, impacts, and injury.

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