THE EFFECT OF POSTURAL ALTERATIONS ON METABOLIC COST IN RUNNING ON INCLINE SLOPES

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ABSTRACT

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In spite of an intuitive relationship between running technique and performance, the influence of specific techniques on metabolic cost remains unclear. While recent studies suggest that postural lean affects metabolic cost during level ground running, little information exists regarding the influence of postural lean on the metabolic cost of running on an incline slope. Purpose: This study sought to investigate the effect of postural lean on metabolic cost, kinematics, and muscle activation of running across a range of incline slopes. Methods: Sixteen healthy adult runners (21±2 years, 8M/8F) of different competitive standards participated in twelve running trials on a motorized treadmill at a speed of 6.0 mph (2.68 ms^{-1}) with varying incline slopes (0%, 4% and 8%) and postural leans (preferred, 0°, 4.5° and 9°). Metabolic power, kinematics, and muscle activation data were recorded for all trials. Results: Increasing postural lean angle across a range of incline slopes had no effect on metabolic cost ($p=0.329$). Leaning forward increased peak flexion of the hip, knee, ankle dorsiflexion and pelvic angle during the stance phase ($p<0.05$). In addition, when running with a 4.5° lean and a 9° lean medial gastrocnemius muscle activation decreased by 4% ($p=0.003$) and 9% ($p=0.019$)

respectively. Conclusion: These findings suggest that kinematics and muscle activation may adjust to maintain the same metabolic cost when running with various postural leans.

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TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

INTRODUCTION

With a record high participation in long distance running events in 2014, running as a form of exercise and for competition has continued to increase in popularity (Lamppa, 2015). One common outcome seen in runners is lower extremity overuse injuries. Although the exact cause of overuse injuries has not been determined, various researchers have found an estimated 27% to 70% of runners experience overuse injuries during any one year period (Hreljac, Marshall, & Hume, 2000). Due to increasing participation and injury rates, research on the biomechanical factors that may improve performance and influence an individual's risk of injury is needed.

With an increase in running participation, clinicians, researchers, and coaches continue to analyze running in order to evaluate and rehabilitate injuries, and enhance performance. When assessing running, the movement is often broken down into gait cycles. The gait cycle begins when one foot makes contact with the ground and ends when the same foot contacts the ground again (Novacheck, 1998). Practitioners are better able to prescribe injury treatments and performance enhancements with an understanding of the biomechanical events in the gait cycle during running (Dugan & Bhat, 2005). The gait cycle can be divided into two phases including the stance phase and the swing phase (Dugan & Bhat, 2005; Novacheck, 1998)

Due to the storage and return of elastic potential energy during the stance phase, running has been modeled as a spring mass system (Alexander, Ker, Bennet, Bibby, & Kester, 1987; Arellano & Kram, 2014). Elastic properties of muscles, tendons, and

ligaments create a spring like behavior while running (Farley, Glasheen, & McMahon, 1993). During the absorption phase of stance, the leg compresses and allows the storage of mechanical energy (Arellano & Kram, 2014). This mechanical energy is then released as the body's center of mass is accelerated upward and forward due to knee and hip extension as well as ankle plantar flexion (Arellano & Kram, 2014; Dugan & Bhat, 2005; Novacheck, 1998).

The elastic energy stored and returned during the stance and swing phases of the gait cycle is believed to influence an individual's optimal stride frequency (Snyder & Farley, 2011). Stride frequency is the number of strides taken in a given period of time, usually measured in Hertz (strides per seconds) (Dugan & Bhat, 2005). Stride length is the horizontal distance from foot strike of one foot to ipsilateral foot strike. Trained runners typically develop a preferred stride frequency and stride length that minimizes metabolic cost (Cavanagh & Kram, 1989; Högberg, 1952). Speed is a product of stride length and stride frequency. As speed increases, a runner will increase stride frequency and/or stride length (Cavanagh & Kram, 1989).

Similar to stride frequency, ground contact time is a strong determinant of running energetics and leg stiffness, explaining 90% of leg stiffness variance (Morin, Samozino, Zameziati, & Belli, 2007). Time of contact (T_c) is the duration of the stance phase relative to the duration of the entire gait cycle. Increased stride frequency results in decreased ground contact time (Cavagna, Thys, & Zamboni, 1976; Snyder & Farley, 2011). The inverse of ground contact time, also known as the rate of force application, is

considered a key determinant of the metabolic cost while running, whereby as T_c increases and rate of force and metabolic cost decrease (Morin et al., 2007).

The cost of generating force to support body weight is inversely proportional to the contact time of the foot with the ground (Kram & Taylor, 1990; Roberts, Kram, Weyand, & Taylor, 1998). Metabolic cost during running is largely determined by the cost of generating force. The cost of generating force hypothesis suggests muscles use energy to generate tension, allowing tendons to store and return elastic energy while running (Hoogkamer, Taboga, & Kram, 2014). Faster, less economical muscle fibers are required to generate force more rapidly when there is shorter ground contact time (Roberts et al., 1998).

The net rate of metabolic energy consumption is related to both the cost of generating force, and the cost of performing mechanical work (Kram, 2000). When compared to walking, muscles are more active for a greater proportion of the gait cycle while running (Thordarson, 1997). Three primary tasks are performed by the muscles, all of which consume metabolic energy. While running, muscles are required to swing the leg forward, support body weight against gravity, and to lift, accelerate, and decelerate the runner's center of mass (Kram, 2000). More economical running has been associated with less dependence on the hip, and ankle and greater reliance on the knee for energy generation (Heise, Smith, & Martin, 2011). Prior research suggest that the large extensor muscles at the hip require substantial metabolic energy when active therefore, metabolic energy may be conserved when hip extensor forces are reduced (Roberts & Belliveau, 2005).

While much is known about the biomechanics and energetics of running on flat ground, few studies have investigated incline running. Runners experience a diverse range of terrains and slopes resulting in changes of kinematics, kinetics, and energetics. There is an interaction between stride length, stride frequency, ground reaction force and metabolic cost, not only when running on flat surfaces, but also when running uphill (Padulo, Powell, Milia, & Ardigò, 2013).

Compared to level running, muscles work differently when running on an incline. Greater net mechanical work is needed to increase the vertical height and thus potential energy of the runner (Roberts & Belliveau, 2005). In order to generate force and perform enough mechanical work to lift the body weight of a runner uphill, muscle fascicles shorten whereas muscles likely remain more isometric in level running (Kram, 2000). Increased net work output is seen at the hip with increasing incline, while the knee and ankle function remains similar at all slopes (Roberts & Belliveau, 2005; Telhan et al., 2010). As slope increases, moments of force increase until 75% of positive net work is performed by the hip (Roberts & Belliveau, 2005).

Although changes in net power are experienced primarily at the hip, alterations in joint kinematics are observed throughout the lower limb. Increasing the duration of the stance phase leads to changes in extensor range of motion at the knee, hip and ankle during push-off while running uphill (Swanson & Caldwell, 2000). Compared to level running, knee flexion angles remain relatively constant during impact, followed by greater range of motion during push-off (Swanson & Caldwell, 2000). The hips experience greater joint flexion at foot strike followed by extension at toe off (Swanson

& Caldwell, 2000). At the ankle, greater dorsi-flexion is observed at foot strike followed by greater plantar-flexion during push-off (Swanson & Caldwell, 2000).

Incline slope running results in metabolic adaptations as well as mechanical adaptations. The rate of metabolic energy consumption by a runner increases when running uphill (Kram, 2000) as the cost of lifting the body against gravity increases while the cost of perpendicular bouncing remains constant (Hoogkamer et al., 2014). The observed increase in metabolic cost can be predominantly explained by greater stride frequency and an increase in vertical mechanical work (Padulo et al., 2013). Changes in stride frequency parallel changes in internal work (Minetti, Ardigo, & Saibene, 1994). As stride frequency increases, the internal work necessary to swing the runner’s limbs relative to the center of mass increases (Snyder & Farley, 2011). Moreover, greater net external mechanical work is required to increase potential energy of the body and raise the body’s center of mass during incline running (Vernillo et al., 2016). When produced over a shorter period of time, the cost of muscular force increases, resulting in greater metabolic cost (Kram & Taylor, 1990).

Whether running on level ground or on inclined slopes, runner and coaches want to improve performance and decrease metabolic cost. Research has found that mechanical demands of the lower extremity joints are influenced by changes in trunk posture (Teng & Powers, 2015). Individuals running with a more flexed trunk posture experience higher energy generation of hip extensors and lower energy generation and absorption of the knee extensors (Teng & Powers, 2015). The opposite is observed when individuals run with a more upright trunk posture with higher energy generation and absorption of the

knee extensors and lower energy generation of the hip extensor (Teng & Powers, 2015). Leaning forward in the sagittal plane has been associated with decreased knee joint stress while upright posture can lead to greater stress of the patellofemoral joint (Teng & Powers, 2015). A forward lean of 5.9° has been found to be the most economical posture while running (Williams & Cavanagh, 1987).

With a lack of research on running posture on incline slopes, runners and coaches have not reached an agreement on the "proper technique" for running uphill. Athletes and coaches often seek ways to improve performance and decrease risk of injury. Some individuals claim an upright posture is most efficient and runners should avoid any forward lean at the hips. Others believe leaning into the hill is the best posture to maintain. There is further dispute as to whether the lean should be from the hips or from the ankle.

Although researchers have examined postural changes while running, there is limited research on how these adjustments influence running performance and injury prevention. While the kinematics of running at different slopes has been researched, few studies have investigated postural lean or discussed the influence of postural alterations on metabolic energy requirements while running on incline slopes. Therefore, the purpose of this study was to investigate the effects of postural lean on metabolic cost of running across a range of incline slopes.

METHODS

Participants included sixteen healthy adult runners (21± 2 years, 8M/8F) who self-reported running a minimum of 40 minutes, three or more times per week during the six months prior to the study, with a 5 kilometer time ≤ 22 minutes. All participants met inclusion criteria and were given informed consent. Subjects participated in one experimental session. Each participant performed a standing trial and twelve running trials. Subjects ran on a motorized treadmill (Trackmaster TMX425C, Full Vision Inc., Newton, KS) at varying incline slopes (0%, 4% and 8%) and postural leans (preferred, 0°, 4.5° and 9°) at a speed of 6.0 mph (2.68 ms^{-1}). Each trial lasted five minutes with a minimum of 5 minutes of rest between trials.

To help the subject achieve the prescribed postural lean during the running trials, real-time video feedback (60 Hz) of the participant's running posture in the sagittal plane throughout each trial was provided. Specifically, on the monitor 1 meter in front of the subject, the prescribed lean angle along with the real-time video of their postural lean was projected. For each trial, subjects were asked to align their body from the ankle to their ear (their postural lean angle) with the prescribed angle shown on the video monitor.

Throughout each trial, rates of oxygen consumption ($\dot{V}O_2$) and carbon dioxide ($\dot{V}CO_2$) production were recorded in order to determine metabolic rate. The average $\dot{V}O_2$ and $\dot{V}CO_2$ for the last two minutes of each trial were calculated. Stride kinematics including postural lean and lower limb muscle activation for a 30 second duration during the last two minutes of each trial was measured.

We measured $\dot{V}O_2$ and $\dot{V}CO_2$ using an open-circuit expired gas analysis (ParvoMedics, Inc., Sandy, UT) during the standing and running trials to determine the participant's energy expenditure. A sterilized face mask worn by the subject was used to collect the air that had been expelled during each trial. Average metabolic rate per kilogram body mass (W kg^{-1}) was calculated using the average $\dot{V}O_2$ (ml O_2 min^{-1}) and $\dot{V}CO_2$ (ml CO_2 min^{-1}) for a two minute time period between minutes 3 and 5 when metabolic steady-state had been achieved. We then subtracted the standing metabolic rate from the running metabolic rate to calculate net metabolic power (J kg^{-1} s^{-1}).

We determined stride kinematics and postural lean angle using a nine camera motion capture (200 fields/s, Vicon, Centennial, CO, USA) for twenty steps during the last two minutes of each trial. The marker set used for this study was based on the calibrated anatomical systems technique (CAST) using a six degree of freedom (DOF) model (Cappozzo, 1995). Passive reflective markers (14 mm) were placed over the anterior/posterior lateral side of the head, 7th cervical vertebrae, right scapular inferior angle, sterno-clavicular notch, xyphoid process, 10th thoracic vertebrae, acromion processes, lateral shoulder joint, lateral elbow joint, medial/lateral wrist joint, second metacarpal heads (distal), posterior–superior iliac spines, anterior–superior iliac spines, lateral femoral epicondyles, lateral malleoli, calcanei, distal and proximal second metatarsal heads, and distal 5th metatarsal heads. Rigid marker clusters were placed on the upper arms, thighs, and shanks to aid in 3D tracking. Anatomical calibration markers were also placed on the medial elbow, knee and medial malleoli. Calibration markers

were removed after collecting static and dynamic calibration trials and prior to the running trials.

Motion capture marker data was processed in Vicon Nexus in order to identify anatomical and tracking markers. Kinematic parameters were then quantified using Visual 3-D (C-Motion, Germantown, MD) after marker data was filtered using a fourth order low-pass zero-lag Butterworth filter with a cut-off frequency of 10 Hz. Joint kinematic data was normalized to 100% of the gait cycle and then averaged across the 20 strides for the hip, knee and ankle. The joint kinematic variables calculated include 1) angle at foot strike, 2) angle at toe off, 3) range of motion from foot strike to toe off during stance and 4) peak angle during stance. Postural lean angle was calculated as the angle between the body at mid-stance, measured from the calcaneal marker to 7^{th} cervical marker, relative to the vertical axis.

To determine lower limb muscle activation, electromyographic data (EMG) from seven muscles of the right leg was collected. After preparing the shaved skin with an impedance lowering abrasive gel (NuPrep, Aurora, CO), we placed differential electrodes with wireless pre-amplifiers (Trigno, Delsys Incorporated, Boston, MA) over the following muscles of each subjects' right leg: gluteus maximus (GM), biceps femoris (BF), rectus femoris (RF), vastus medialis (VM), medial gastrocnemius (MG), soleus (SOL), and tibialis anterior (TA). We verified electrode positions and signal quality by visually inspecting the EMG signals, sampled at 1000Hz, while subjects contracted each muscle. We recorded all data during the same 30 second period in the last two minutes of

each running trial as the kinematic data was collected. Prior to data analysis the Delsys hardware band-pass filtered the EMG signals (20–300 Hz).

Using a custom Visual 3D analysis program (Germantown, MD), we full wave rectified the raw EMG signals and calculated the normalized root mean square (40 ms window) EMG amplitude (EMG_{RMS}). EMG_{RMS} amplitude of each muscle was normalized relative to the peak EMG_{RMS} amplitude during the 0% slope, preferred lean running trial. EMG_{RMS} was also time normalized to the gait cycle from foot strike (0%) to subsequent ipsilateral foot trike (100%). We calculated the average EMG_{RMS} muscle activation magnitude from for the loading phase (foot strike to mid stance, propulsion phase (mid stance to toe off), and across the entire gait cycle.

An analysis of variance (ANOVA) for repeated measures was used to evaluate the effect of running postural lean and incline slope on metabolic energy consumption with a $p<0.05$ criterion. Additional planned comparisons were used to evaluate differences between the different levels of postural lean. In order to better understand in the effects of postural lean on metabolic cost, two secondary ANOVAs were be performed to determine the main effect and interaction effect of postural lean and incline slope on muscle activation and joint kinematics.

RESULTS

Metabolic Cost

Increasing postural lean angle across a range of incline slopes had no effect on metabolic cost ($p=0.329$, Figure 1). Moreover, metabolic cost increased as slope increased ($p<0.001$). When increasing slope from 0% incline to 8% incline, metabolic cost increased by 56% ($p<0.001$, Table 1). There was no interaction between postural lean angle and incline slope ($p=0.072$).

Kinematics

Despite a lack of metabolic changes, postural lean alterations resulted in kinematic changes across a variety of incline slopes. Across the range of leans from upright running to the largest lean of 9^0, peak hip flexion increased by approximately 9% ($p<0.001$, Figure 2). Specifically, peak hip flexion was 5% greater when running with a 4.5^0 lean than when utilizing a 0^0 lean ($p=0.003$). When compared to the 4.5^0 lean, the 9^0 lean resulted in 3% greater peak hip flexion ($p=0.039$). Peak hip flexion decreased by approximately 7% when utilizing a prescribed 0^0 lean compared to a preferred lean angle ($p<0.001$). Running posture also resulted in differences of peak knee flexion during stance across the range of leans and incline slopes ($p=0.005$ and $p=0.001$ respectively, Figure 3). Post-hoc analysis revealed that peak knee flexion was 3% greater for the 4.5^0

lean than the 0^0 lean ($p=0.012$). Additionally, peak knee flexion decreased by 1.5% when running with a 9^0 postural lean compared to running with a 4.5^0 lean ($p=0.018$). When compared to a preferred lean angle, peak knee flexion was 1.5% greater for the 4.5^0 lean ($p=0.038$). As slope increased from 0% to 8%, peak knee flexion increased by 5% ($p=0.002$). Peak dorsiflexion during stance was influenced by lean angle and incline slope ($p=0.036$ and $p<0.001$ respectively). Subjects used approximately 2% greater peak dorsiflexion when running with a 4.5^0 lean compared to running with a 0^0 lean ($p=0.009$, Figure 4). Furthermore, when compared to 0% incline, peak dorsiflexion increased by an average of 9% on 4% incline ($p<0.001$) and an average of 21% on 8% incline ($p<0.001$). Lean angle and treadmill slope also had an effect on pelvic tilt angle during stance ($p=0.004$, and $p=0.001$ respectively). Average pelvic tilt angle increases 11% from the 0^0 lean to the 9^0 lean ($p=0.044$). When compared to a preferred lean angle, differences in pelvic tilt were observed. An average 11% increase in pelvic tilt was observed when running with a 4.5^0 lean, in addition to a 13% increase in pelvic tilt when running with a 9^0 lean ($p=0.018$ and $p<0.001$ respectively). Increase in slope incline from 0% to 4% also resulted in a 16% increase in pelvic tilt ($p=0.018$). An additional increase of 11% in pelvic tilt was observed when slope was raised from 4% to 8% ($p=0.006$). Differences in stride length and ground contact time due to lean angle or incline slope were not observed ($p>0.05$, Table 2).

Muscle Activation

Postural lean alterations influenced muscle activation patterns minimally. There was no main effect of either postural lean angle or incline slope on upper leg (VM, RF, BF, GM) average muscle activation during stance phase ($p>0.05$, Table 3). Additionally, there was no interaction between lean angle and slope for upper leg muscles. In contrast, postural lean influenced MG activation in the stance phase ($p=0.002$, Figure 5). Compared to 0^0 lean, MG activation decreased by 4% when running with a 4.5^0 lean ($p=0.003$) and by 9% when utilizing a 9^0 lean ($p=0.019$). Specifically, in the early portion of the stance phase (i.e. 10-20% of the gait cycle), MG activation was an average of 12% less when running with a 9^0 lean than when running with a 0^0 lean ($p=0.004$). For the entire stance phase, an interaction between prescribed lean angle and slope for MG activation was observed ($p=0.007$). Moreover, SOL average activation during stance was influenced by incline slope ($p=0.005$). As slope increased from 0% to 8%, SOL activation increased by 21% ($p=0.010$, Table 3). Finally, there was no change in TA activation due to prescribed lean angle or incline slope during the stance phase of running ($p>0.05$).

DISCUSSION

The purpose of this study was to evaluate the effect of postural lean on metabolic cost, muscle activation, and kinematics of running across a range of incline slopes. Our findings revealed that metabolic cost of running was not influenced by posture when running on incline slopes. In contrast, kinematic changes were observed when running with a greater forward lean across a range of incline slopes. Postural lean alterations and incline slope had only a minimal effect on leg muscle activation. Specifically, upper leg muscle activation during the stance phase of running did not change significantly as a result of changing lean angle or incline slope. However, medial gastrocnemius and soleus activation did change with both postural lean and incline slope. These findings provide further understanding of how postural lean and incline slope influence the interrelations between metabolic cost, kinematics, and muscle activation during running that may be insightful to runners, clinician and researchers alike.

Changes in metabolic cost due to postural lean were not observed in the present study. When comparing postural leans, our results found no changes in ground contact time, stride frequency or stride length (Table 2). These results support the cost of generating force hypothesis. Prior studies have shown that a primary contributor of the total metabolic cost of running is the cost of generating force, which is inversely proportional to the contact time of the foot with the ground (Kram & Taylor, 1990; Roberts et al., 1998). The lack of change in ground contact time, stride frequency and stride length due to postural lean parallels the lack of change in metabolic cost. Although

no differences in metabolic cost due to postural lean was observed, metabolic differences due to incline slope were consistent with results of previous literature (Kram, 2000; Padulo et al., 2013).

In contract to our results, previous research found that running with an upright posture or a more moderate forward lean decreases metabolic cost (Carson, 2016). However, participants in the previous study ran at a quicker speed of 8.0 mph (3.58 ms^{-1}) compared to participants of our study who ran at a speed of 6.0 mph (2.68 ms^{-1}). These results suggest a possible interaction between speed, postural lean, and metabolic cost. Postural lean may not influence metabolic cost at slower speeds but may have an effect when running with a quicker pace. Further studies should investigate the influence of speed on postural lean and metabolic cost.

Despite a lack of metabolic changes, our results showed an increase in peak flexion of the hip, knee, and ankle joints due to postural lean. According to the spring mass model of running, mechanical energy is stored during the stance phase of running. Greater flexion at the hip, knee and ankle, allows the leg to compress more, storing greater amounts of mechanical energy. The release of this energy accelerates the body's center of mass upward and forward. The increase of released energy due to kinematic changes may allow the runner to utilize various postural leans without negatively affecting metabolic cost. In other words, runners may alter joint kinematics when running with a forward lean in part to prevent an increase in metabolic cost.

Metabolic cost has also been related to the cost of performing mechanical work. Previous studies have shown that large hip extensor muscles require substantial metabolic

energy when active (Roberts & Belliveau, 2005). Our results show no increase in muscle activation of the upper leg due to postural lean or incline. These findings indicate that metabolic cost may be conserved due to the lack of changes in upper leg muscle activation.

Although no changes in upper leg muscle activation occurred, MG activation decreased with greater postural lean angles. The increase in ankle dorsiflexion due to postural lean decreases the reliance on ankle plantar flexors, reducing the muscle activation of the MG. Furthermore, as the ankle flexes, the Achilles tendon stretches. Previous research has found that as the ankle dorsiflexes, elastic strain energy is stored in the Achilles tendon (Alexander, 1991) while minimizing the muscle fascicle shortening of the plantar flexors (Fletcher & MacIntosh, 2014). Reduction of the muscle energy cost due to minimized muscle fascicle shortening, and elastic energy return from the Achilles tendon may contribute to the lack of changes in metabolic cost due to postural lean observed in our study. In contrast to the effect of postural lean on MG activation, running on steeper incline slopes increased SOL activation. These findings are consistent with prior research that showed SOL activation increases by as much as 14% with a 10% increase in incline slope (Sloniger, Cureton, Prior, & Evans, 1997). This increase in SOL activation has been associated with an increase in concentric muscle activation required for running uphill (Minetti, 1994).

A limitation of our study may be related to the running experience of the participants tested. Based on fitness requirements, experienced runners were selected for participation in this study. In response to physiological changes, trained runners have

been shown to self-optimize running biomechanics (Moore, 2016). Thus, the results of our study may not be generalized to novice runners. Furthermore, the variability in our subject's ability to accurately match the prescribed postural lean angle using live video feedback may have contributed to the large variability in metabolic cost and muscle activation and thus contributed to a lack of statistically significant findings.

To further investigate the relationship between postural lean, incline slope, and metabolic cost, future studies should use faster speeds or more drastic incline slopes. Using faster speeds may allow researchers to explore an interaction between postural lean and slope. Although our study found no effect of postural lean on metabolic cost, lean may have a greater influence when running at a faster speed. Previous research found that metabolic cost decreased when running with an upright posture at a speed of 8.0 mph (3.58 ms^{-1}) (Carson, 2016). Metabolic cost may be more dependent on postural lean at faster speeds. Additionally, as slope increases, kinematics and muscle activation may experience greater changes compared to level running. Running with various postural lean angles across a greater range of incline slope may result in greater differences in metabolic cost due to postural lean.

CONCLUSIONS

In conclusion, our results suggest that postural lean does not affect metabolic cost when running on incline slopes. However, changes in leg kinematics and lower leg muscle activation were observed. Our findings suggest that kinematics and muscle activation may adjust to maintain the same metabolic cost when running with various postural leans. Based on the results of this research, future studies may investigate the effects of postural lean on metabolic cost with the use of more drastic changes in incline slope or faster speeds. Additionally, kinetic data may provide a further understanding of the factors affecting metabolic cost while running with various postural leans.

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FIGURES

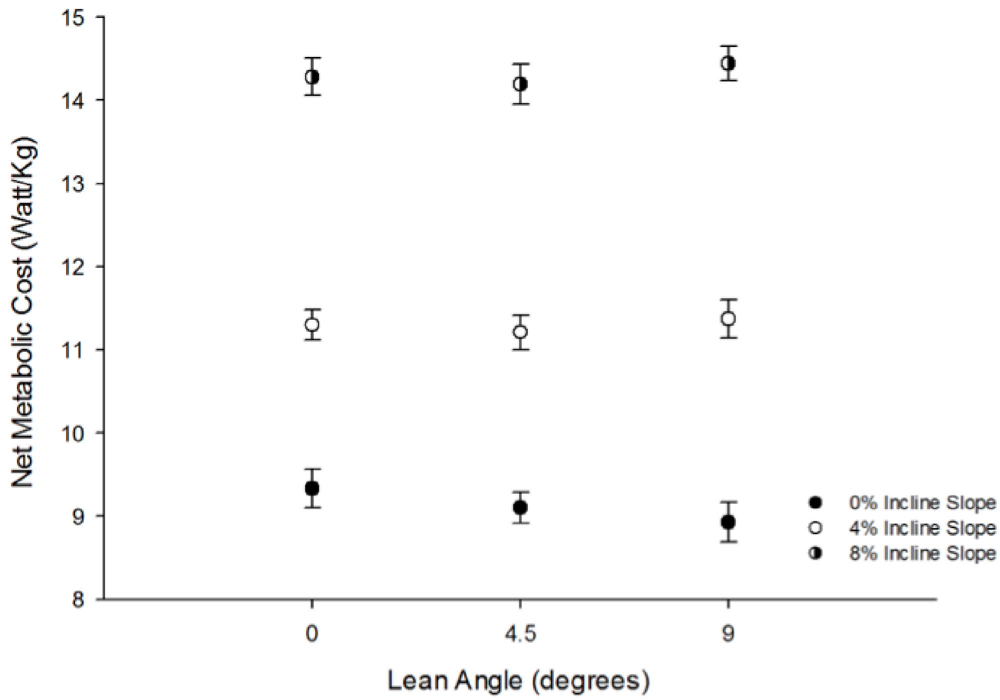

Figure 1: Net metabolic power ($W \cdot kg^{-1}$) plotted as a function of postural lean angle for runners on various incline slopes. There was no significant difference in metabolic power between postural lean angles ($p=0.329$).

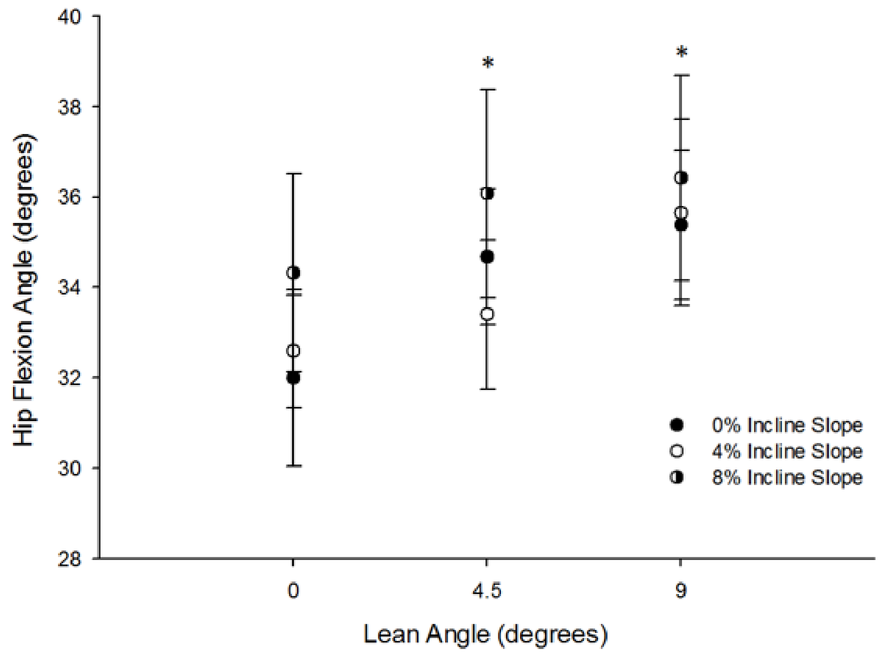

Figure 2: Mean (SE) stance phase peak hip angle (°) plotted as a function of postural lean angle for runners on various incline slopes. Asterisks (*) indicate significant differences from 0° lean angle ($p<0.05$).

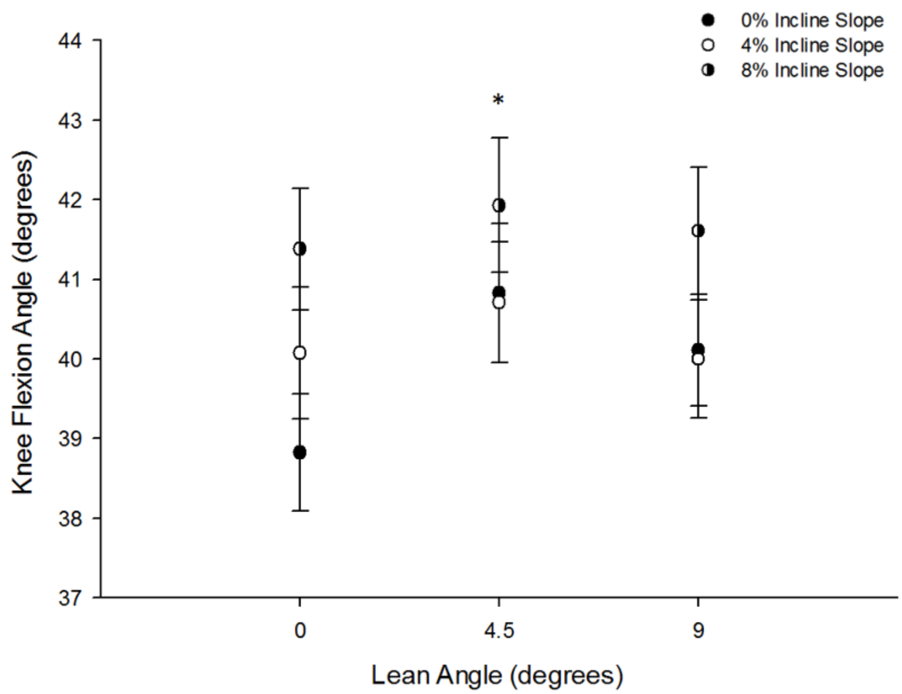

Figure 3: Mean (SE) stance phase peak knee flexion angle (°) plotted as a function of postural lean angle for runners on various incline slopes. Asterisk (*) indicates significant differences from 0° lean angle (p<0.05).

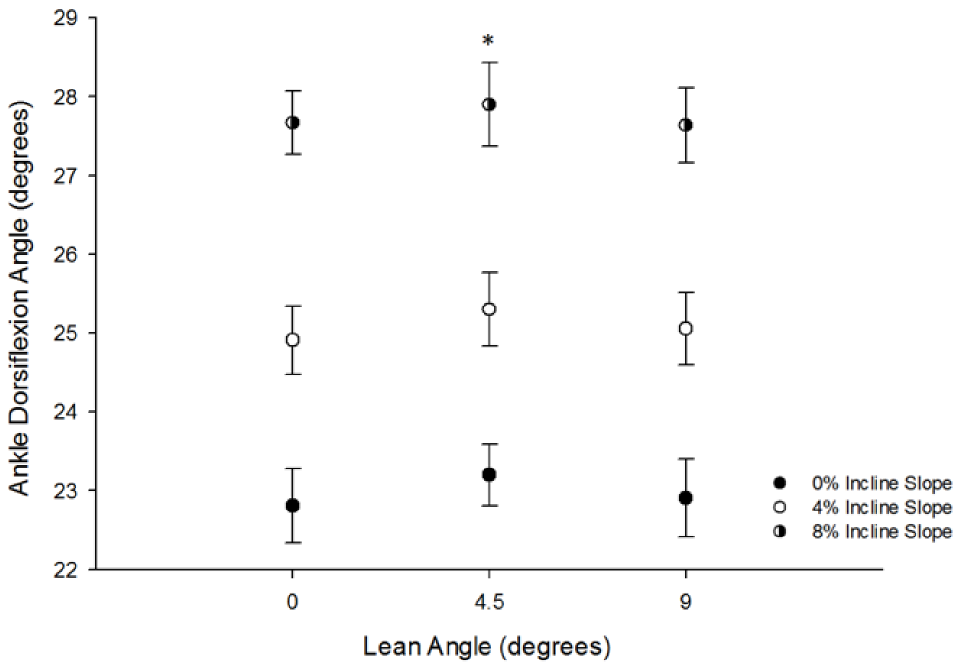

Figure 4: Mean (SE) stance phase ankle dorsiflexion angle (°) plotted as a function of postural lean angle for runners on various incline slopes. Asterisk (*) indicates significant differences from 0° lean angle (p<0.05).

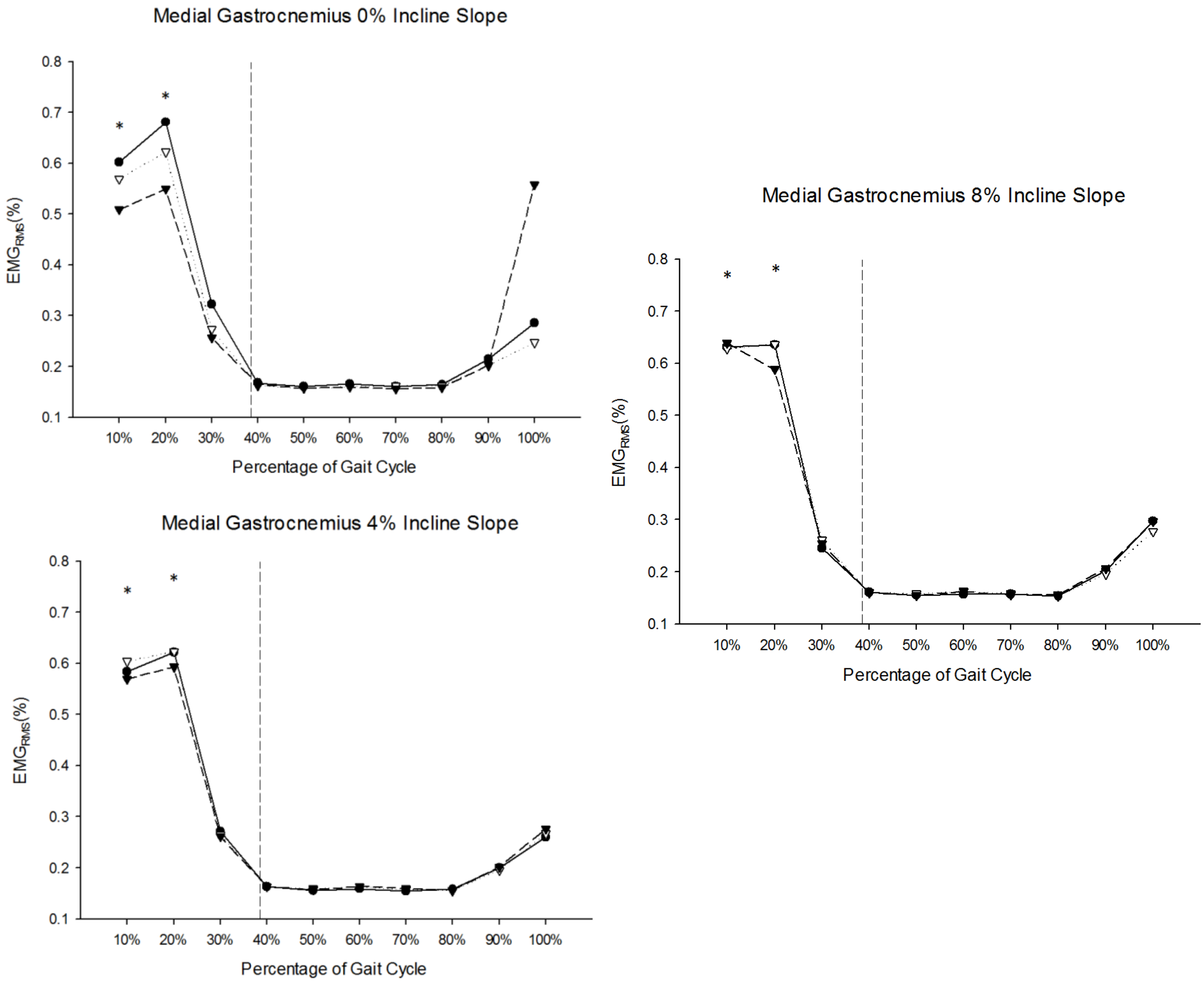

Figure 5: Averaged normalized EMG_{RMS} signals of the medial gastrocnemius for the 0° lean (●), 4.5° lean (▽), and the 9° lean (▼) conditions over a complete stride (0-100% of gait cycle) across a range of incline slopes (0%, 4%, 8%). Asterisks (*) indicate significant differences from 0° postural lean angle ($p<0.05$). Vertical dashed line indicates the end of the stance phase and the beginning of the swing phase.

TABLES

Table 1. Net metabolic power (Watts/kg) for conditions averaged from the final two minutes of each trial ± standard error of the mean (N=16). Cross indicates significant differences from 0% slope ($p<0.05$).

	Incline Slope	**Prescribed lean angle**	**Mean ± SEM**
Metabolic Power (Watts/kg)	0%	0º	9.33 ± 0.23
		4.5º	9.10 ± 0.19
		9º	8.93 ± 0.24
		Preferred	9.23 ± 0.19
	4% †	0º	11.30 ± 0.18
		4.5º	11.21 ± 0.21
		9º	11.37 ± 0.23
		Preferred	11.14 ± 0.17
	8% †	0º	14.28 ± 0.23
		4.5º	14.19 ± 0.24
		9º	14.44 ± 0.21
		Preferred	14.14 ± 0.22

Table 2. Mean values for kinematic variables ± standard error of the mean (N=16). Asterisk indicates significant differences from prescribed 0° lean ($p<0.05$). Cross indicates significant differences from 0% slope ($p<0.05$).

	Prescribed lean angle	**Incline Slope (%)** 0	4	8
Contact time (sec)	0^0	0.26 ± 0.01	0.26 ± 0.00	0.26 ± 0.00
	4.5^0	0.26 ± 0.01	0.26 ± 0.01	0.26 ± 0.01
	9^0	0.26 ± 0.01	0.26 ± 0.01	0.26 ± 0.01
	Preferred	0.26 ± 0.01	0.27 ± 0.00	0.26 ± 0.00
Stride length (m)	0^0	1.92 ± 0.02	1.91 ± 0.02	1.88 ± 0.02
	4.5^0	1.93 ± 0.02	1.93 ± 0.02	1.88 ± 0.02
	9^0	1.94 ± 0.02	1.91 ± 0.02	1.88 ± 0.02
	Preferred	1.93 ± 0.02	1.92 ± 0.02	1.90 ± 0.02
Peak hip flexion * (deg)	0^0	32.00 ± 1.96	32.58 ± 1.26	34.31 ± 2.20
	4.5^0	34.67 ± 1.49	33.40 ± 1.65	36.07 ± 2.30
	9^0	35.58 ± 1.64	35.65 ± 2.06	36.42 ± 2.27
	Preferred	34.44 ± 1.95	34.87 ± 1.90	36.30 ± 2.02
Peak knee flexion *† (deg)	0^0	38.82 ± 0.74	40.07 ± 0.83	41.38 ± 0.76
	4.5^0	40.83 ± 0.88	40.71 ± 0.76	41.93 ± 0.85
	9^0	40.11 ± 0.70	40.00 ± 0.75	41.61 ± 0.80
	Preferred	39.36 ± 0.70	40.89 ± 0.65	41.44 ± 0.75
Peak dorsiflexion *† (deg)	0^0	22.80 ± 0.47	24.91 ± 0.43	27.67 ± 0.40
	4.5^0	23.20 ± 0.39	25.30 ± 0.47	27.90 ± 0.53
	9^0	22.91 ± 0.50	25.05 ± 0.46	27.64 ± 0.48
	Preferred	23.00 ± 0.47	25.37 ± 0.43	27.72 ± 0.46
Pelvic angle (deg) *†	0^0	2.01 ± 0.19	2.21 ± 0.19	2.48 ± 0.20
	4.5^0	1.80 ± 0.20	2.08 ± 0.21	2.30 ± 0.24
	9^0	1.67 ± 0.19	1.95 ± 0.23	2.37 ± 0.23
	Preferred	2.00 ± 0.19	2.43 ± 0.23	2.48 ± 0.25

Table 3. Mean normalized EMG_{RMS} values during stance phase ± standard error of the mean (N=16). All values are expressed as percent (%) of peak activation during the 0° postural lean trial performed on a level (0% slope) treadmill angle. Asterisk indicates significant differences from prescribed 0° lean ($p<0.05$). Cross indicates significant differences from 0% slope ($p<0.05$).

Muscle	Prescribed lean angle	Incline Slope (%) 0	4	8
Tibialis Anterior (TA)	0^0	0.17 ± 0.02	0.16 ± 0.02	0.15 ± 0.02
	4.5^0	0.16 ± 0.02	0.16 ± 0.02	0.15 ± 0.02
	9^0	0.16 ± 0.02	0.15 ± 0.02	0.16 ± 0.02
	Preferred	0.15 ± 0.01	0.15 ± 0.02	0.16 ± 0.02
Soleus † (SOL)	0^0	0.30 ± 0.02	0.44 ± 0.03	0.49 ± 0.04
	4.5^0	0.37 ± 0.02	0.44 ± 0.04	0.44 ± 0.03
	9^0	0.36 ± 0.03	0.48 ± 0.06	0.45 ± 0.03
	Preferred	0.40 ± 0.02	0.43 ± 0.03	0.46 ± 0.03
Medial Gastroc.* (MG)	0^0	0.47 ± 0.02	0.43 ± 0.03	0.44 ± 0.04
	4.5^0	0.43 ± 0.02	0.43 ± 0.03	0.43 ± 0.04
	9^0	0.39 ± 0.03	0.41 ± 0.03	0.42 ± 0.04
	Preferred	0.47 ± 0.01	0.43 ± 0.03	0.43 ± 0.04
Vastus Medialis (VM)	0^0	0.32 ± 0.02	0.37 ± 0.04	0.40 ± 0.09
	4.5^0	0.28 ± 0.02	0.41 ± 0.08	0.30 ± 0.03
	9^0	0.27 ± 0.02	0.32 ± 0.05	0.36 ± 0.05
	Preferred	0.29 ± 0.02	0.29 ± 0.02	0.34 ± 0.04
Rectus Femoris (RF)	0^0	0.38 ± 0.03	0.41 ± 0.06	0.47 ± 0.08
	4.5^0	0.37 ± 0.03	0.40 ± 0.06	0.45 ± 0.09
	9^0	0.36 ± 0.04	0.39 ± 0.06	0.42 ± 0.08
	Preferred	0.35 ± 0.02	0.36 ± 0.04	0.40 ± 0.06
Biceps Femoris (BF)	0^0	0.25 ± 0.02	0.22 ± 0.03	0.24 ± 0.03
	4.5^0	0.27 ± 0.02	0.27 ± 0.03	0.25 ± 0.03
	9^0	0.27 ± 0.02	0.27 ± 0.03	0.28 ± 0.03
	Preferred	0.28 ± 0.02	0.24 ± 0.03	0.27 ± 0.03
Gluteus Maximus (GM)	0^0	0.33 ± 0.04	0.35 ± 0.05	0.30 ± 0.04
	4.5^0	0.31 ± 0.04	0.27 ± 0.03	0.31 ± 0.05
	9^0	0.31 ± 0.03	0.28 ± 0.03	0.31 ± 0.05
	Preferred	0.28 ± 0.02	0.32 ± 0.03	0.31 ± 0.04