Influence of Trunk Posture on Lower Extremity Energetics during Running

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ABSTRACT

TENG, H., and C. M. POWERS. Influence of Trunk Posture on Lower Extremity Energetics during Running. *Med. Sci. Sports Exerc.*, Vol. 47, No. 3, pp. 625–630, 2015. **Purpose**: This study aimed to examine the influence of sagittal plane trunk posture on lower extremity energetics during running. **Methods**: Forty asymptomatic recreational runners (20 males and 20 females) ran overground at a speed of 3.4 $m{\cdot}s^{-1}$. Sagittal plane trunk kinematics and lower extremity kinematics and energetics during the stance phase of running were computed. Subjects were dichotomized into high flexion (HF) and low flexion (LF) groups on the basis of the mean trunk flexion angle. **Results**: The mean (±SD) trunk flexion angles of the HF and LF groups were 10.8° ± 2.2° and 3.6° ± 2.8°, respectively. When compared with the LF group, the HF group demonstrated significantly higher hip extensor energy generation (0.12 ± 0.06 vs 0.05 ± 0.04 $J{\cdot}kg^{-1}$, $P < 0.001$) and lower knee extensor energy absorption (0.60 ± 0.14 vs 0.74 ± 0.09 $J{\cdot}kg^{-1}$, $P = 0.001$) and generation (0.30 ± 0.05 vs 0.34 ± 0.06 $J{\cdot}kg^{-1}$, $P = 0.02$). There was no significant group difference for the ankle plantarflexor energy absorption or generation ($P > 0.05$). **Conclusions**: Sagittal plane trunk flexion has a significant influence on hip and knee energetics during running. Increasing forward trunk lean during running may be used as a strategy to reduce knee loading without increasing the biomechanical demand at the ankle plantarflexors. **Key Words:** ENERGETICS, KINEMATICS, RUNNING, TRUNK POSTURE

Running is a popular form of exercise, with approximately 36 million people engaging in this activity in the United States alone (30,36). Despite the positive health effects associated with running (21,25), there remains a high incidence of lower extremity injuries in this population (19%–79%) (20,35,37,38). Among the lower extremity joints, the knee accounts for 50% of all lower extremity running injuries (35,37). In addition, half of the injuries occurring at the knee are related to the patellofemoral joint (35,37).

Many running injuries have been attributed to a rear foot strike running pattern (9). For example, it has been reported that runners who strike the ground with the heel exhibit higher impact forces and loading rates (7,14,15,24,31). As such, transitioning to a midfoot or forefoot strike pattern has been proposed to reduce the risk of running injuries (4,6,28,39). Although this modification in foot strike pattern has been shown to reduce knee extensor moments, power, and energy absorption (1,4,6,16,39), several studies have reported that a midfoot or forefoot strike pattern leads to increases in ankle flexor moments, power, and energy absorption (1,4,6,16,28,29,39). This shift in mechanical demand from the knee joint to the ankle joint has been proposed to contribute to higher risk of foot and ankle injuries in runners (4,9,15,39). Therefore, transition to a forefoot or midfoot strike pattern would require a significant accommodation period and may not be practical for all runners.

Recent studies have suggested that altering trunk posture can affect the moment distribution among the lower extremity joints during weight bearing activities. Given that the trunk segment (defined as the mass from the greater trochanters to glenohumeral joints) constitutes approximately 50% of the body mass (10), small changes in trunk orientation can have significant influence on the mechanical demands of the lower extremity. For example, studies have reported that a forward trunk lean is associated with lower knee extensor moments and higher hip extensor moments during walking, stair ascent, and hop landing (2,23,27). As such, it is logical to hypothesize that sagittal plane trunk posture may affect the biomechanical demands of the lower extremity joints during running.

The purpose of the current study was to compare lower extremity energetics between individuals who run with a forward lean trunk posture and individuals who run with a more upright trunk posture. On the basis of existing literature, we hypothesized that persons who exhibit a greater degree of forward lean trunk would demonstrate lower energy absorption and generation of the knee extensors and higher energy absorption and generation of the hip extensors when compared with runners who demonstrate relatively less trunk flexion.

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METHODS

Participants. Twenty male runners (age, 27.1 ± 7.0 yr; height, 1.74 ± 6.9 m; weight, 71.1 ± 8.2 kg; running distance per week, 24.1 ± 20.0 km) and 20 female runners (age, 26.2 ± 5.8 yr; height, 1.65 ± 7.4 m; weight, 60.6 ± 6.6 kg; running distance per week, 22.7 ± 10.9 km) participated in this study. An *a priori* power analysis using pilot data obtained from both males and females indicated that 20 subjects per group would provide adequate statistical power (80%) to detect differences for each variable of interest (using an alpha level of 0.05).

Participants were natural rear foot strikers, which was verified using sagittal plane images from the high-speed video (sampling rate, 125 Hz). Subjects were excluded from participation if they reported any of the following: 1) current lower extremity or low back pain, 2) previous history of lower extremity or low back surgery, and 3) lower extremity or low back pathology that caused pain or discomfort during running 6 months before participation. Only symptom-free subjects were recruited in this study to avoid the potential confounding effects of pain on trunk and lower extremity biomechanics.

Instrumentation. Three-dimensional trunk and lower extremity kinematics were collected using an 11-camera motion capture system (Qualisys, Gothenburg, Sweden) at a sampling rate of 250 Hz. Ground reaction force data were obtained at a rate of 1500 Hz using a single force plate (AMTI, Newton, MA). Marker and ground reaction force data were collected and synchronized using a motion capturing software (Qualisys Track Manager version 2.8).

Procedures. Data were collected at the Jacquelin Perry Musculoskeletal Biomechanics Research Laboratory at the University of Southern California. Before participation, the objectives, procedures, and potential risks were explained to each subject. An informed consent as approved by the Health Science Institutional Reviewed Board of the University of Southern California was then obtained.

Subjects were evaluated wearing shorts, tank tops, and their personal running shoes. Data were obtained from each subject's dominant leg. Leg dominance was determined by asking the subject which leg they preferred to use when kicking a ball.

Before the running assessment, 21 anatomical markers (reflective, 14-mm spheres) were placed on the following bony landmarks: end of second toes, first and fifth metatarsal heads, medial and lateral malleoli, medial and lateral epicondyles of the femurs, greater trochanters, iliac crests, L5–S1 junction, and acromioclavicular joints. In addition, tracking marker clusters mounted on semirigid plastic plates were placed on the lateral surfaces of the subject's thighs, shanks, and heel counter of the shoes. A standing calibration trial was first obtained to define the segmental coordinate systems and joint axes. After the calibration trial, anatomical markers were removed, except for those at iliac crests, L5–S1 junction, and acromioclavicular joints. The tracking markers remained on the subject throughout the data collection session.

Consistent with previous running studies (12,32,40), participants were instructed to run at a controlled speed of $3.4\ m{\cdot}s^{-1}$ along a 14-m runway. Practice trials were permitted to allow subjects to become familiar with the running speed. Five successful running trials were obtained. A successful trial was defined when the foot of the dominant leg fell within borders of the force plate from initial contact to toe-off, and the running speed was within ±5% of the target speed.

Data analysis. Kinematic, kinetic, and energetic data were processed and analyzed using Visual3D (C-Motion, Germantown, MD) and MATLAB software (MathWorks, Inc., Natick, MA). Marker trajectories data were low-pass–filtered at 12 Hz using a fourth-order Butterworth filter (3,13,22). The trunk segment was defined by markers placed on bilateral iliac crests and acromioclavicular joints (26). The pelvis and trunk segments were modeled as cylinders, and the lower extremity segments were modeled as frusta of cones (18). The local coordinate systems of the trunk, pelvis, thigh, shank, and foot segments were derived from the standing calibration trial (42).

Joint kinematics were calculated using Cardan angles with a rotation sequence of flexion/extension, abduction/adduction, and internal/external rotation (8,17,42). The trunk angle was calculated as the motion of the trunk segment relative to the global coordinate system (global vertical axis). Lower extremity kinematics were calculated as the motion of the distal segment relative to the proximal reference. Net joint moments of force were computed using inverse dynamics equations and were normalized to each subject's body mass.

Net moment power was calculated as the scalar product of angular velocity and net joint moment. The positive and negative values of the joint power were used to identify phases of energy generation and absorption, respectively (41). Energy absorption and generation of the extensor muscle groups (i.e., ankle plantarflexors, knee extensors, and hip extensors) were calculated by integrating the respective power–time curves during the stance phase of running (34). For instance, energy absorbed by hip extensors was computed as the integral of negative power, respective to the time when the hip extensor moment was positive.

Kinematic variables of interest included the mean trunk flexion angle, which was calculated as the average sagittal plane trunk posture during the stance of running. Additional kinematic variables of interest included peak hip flexion, knee flexion, and ankle dorsiflexion angles during the stance phase of running. Energetic variables of interest consisted of the energy absorption and generation of the hip extensors, knee extensors, and ankle plantarflexors during the stance phase of running. The stance phase was defined when the vertical ground reaction force exceeded 30 N. All variables were calculated for each stride, and the average values of five strides were used for statistical analysis.

Statistical analysis. Subjects were dichotomized into two groups on the basis of their mean trunk flexion during the stance phase of running. Individuals who ran with a trunk flexion angle greater than the median of the entire group were assigned to the high flexion (HF) group (n = 20), whereas individuals whose trunk flexion angles were lower than the median were allocated to the low flexion (LF) group (n = 20). A one-way between-group multivariate ANOVA (MANOVA) was used to compare group differences with respect to the kinematic and energetic variables of interest. Nine dependent variables were included in the analysis: peak hip flexion angle, peak knee flexion angle, peak ankle dorsiflexion angle, energy absorption of the hip extensors, knee extensors and ankle plantarflexors, and energy generation of the hip extensors, knee extensors, and ankle plantarflexors. If the MANOVA indicated a significant main effect, *post hoc* Bonferroni testing was used. Statistical analyses were performed using PASW statistical software (Chicago, IL) with a statistical significance level of 0.05.

RESULTS

Thirteen male and seven female runners were assigned to the HF group, whereas seven male and 13 female runners were allocated to the LF group. There was no significant sex effect with respect to group allocation (chi-square test, P = 0.13). The mean trunk flexion angles were 10.8° ± 2.2° for the HF group and 3.6° ± 2.8° for the LF group (Fig. 1 and Table 1).

The overall MANOVA test assessing for group differences was significant (P < 0.001). *Post hoc* analysis revealed that the HF group exhibited significantly lower energy absorption and generation of the knee extensors (P = 0.001 and P = 0.02, respectively) and higher energy generation of the hip extensors (P < 0.001) when compared with the LF group (Fig. 2 and Table 1). No significant group difference was observed for hip extensor energy absorption or ankle plantarflexor energy absorption and generation (Fig. 2 and Table 1). In addition, there were no significant group differences with respect to peak hip flexion, knee flexion, and ankle dorsiflexion angles (Fig. 3 and Table 1).

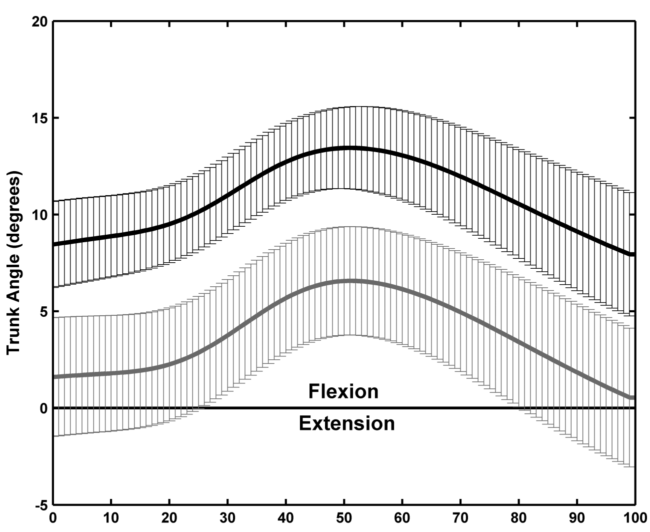

FIGURE 1—Sagittal plane trunk kinematics during the stance phase of running for the HF (*black line*) and LF (*gray line*) groups. *Error bars* represented ±1 SD.

TABLE 1. Trunk and lower extremity kinematics and lower extremity energetics (work done per kilogram body weight) during the stance phase of running for the HF and LF groups.

	Trunk Groups		
	HF	LF	P Value
Kinematics			
Mean trunk flexion angle (°)	10.8 ± 2.2	3.6 ± 2.8	<0.001
Peak hip flexion angle (°)	34.9 ± 5.7	34.9 ± 7.0	0.97
Peak knee flexion angle (°)	43.5 ± 3.7	44.9 ± 3.8	0.23
Peak ankle dorsiflexion angle (°)	23.1 ± 2.5	23.7 ± 3.0	0.46
Energy absorption			
Hip extensor (J·kg^{-1})	0.07 ± 0.04	0.06 ± 0.04	0.45
Knee extensor (J·kg^{-1})	0.60 ± 0.14	0.74 ± 0.09	0.001
Ankle plantarflexor (J·kg^{-1})	0.41 ± 0.08	0.37 ± 0.05	0.06
Energy generation			
Hip extensor (J·kg^{-1})	0.12 ± 0.06	0.05 ± 0.04	<0.001
Knee extensor (J·kg^{-1})	0.30 ± 0.05	0.34 ± 0.06	0.02
Ankle plantarflexor (J·kg^{-1})	0.62 ± 0.13	0.61 ± 0.12	0.66

Values are presented as mean ± SD.

DISCUSSION

The objective of this study was to examine the influence of sagittal plane trunk posture on lower extremity energetics during running. Our findings revealed that a relatively small difference in the mean trunk flexion angle (7.2°) resulted in significant differences in hip and knee energetics during the stance phase of running. Specifically, individuals who ran with a more flexed trunk posture exhibited lower energy absorption and generation of the knee extensors and higher energy generation of the hip extensors. Conversely, individuals who ran with a more upright trunk posture exhibited higher knee extensor energy absorption and generation and lower hip extensor energy generation.

The findings of the current study are consistent with previous studies that have evaluated the influence of trunk posture on lower extremity biomechanics (2,23,27). Leteneur et al. (23) reported that individuals who walked with a forward trunk lean exhibited lower peak knee extensor moments and higher hip extensor moments compared with persons who walked with a more extended trunk posture. Moreover, studies have reported that individuals with various knee pathologies (i.e., severe knee osteoarthritis and anterior cruciate ligament deficiency) adopt a forward lean trunk posture to reduce peak knee extensor moments during stair ascent and hop landing (2,27).

The HF and LF groups did not differ with respect to lower extremity joint kinematics. This finding is in disagreement with those of Saha et al. (31) who reported that adopting a forward trunk lean resulted in greater hip, knee, and ankle flexion angles during walking. However, it is important to note that Saha et al. (31) used a much greater degree of trunk flexion in their study (25°–50°) compared with the 7.2° group difference observed in the current investigation.

The lack of differences in joint angles in the current study suggests that the observed difference in hip and knee energetics was due to differences in the locations of center of

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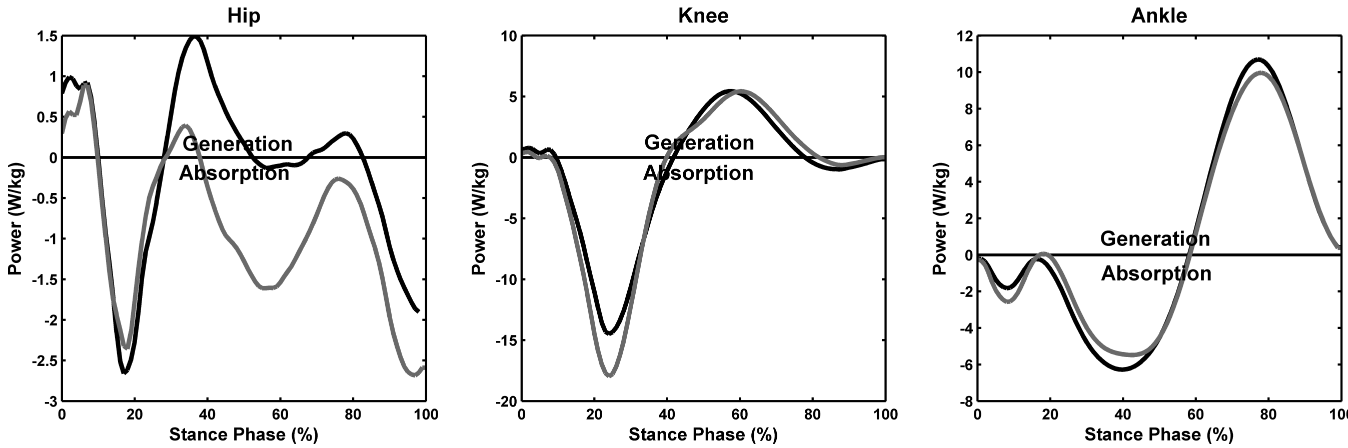

FIGURE 2—Comparison of sagittal plane hip, knee, and ankle powers during the stance phase of running between the HF (*black line*) and LF (*gray line*) groups.

mass and center of pressure as opposed to lower extremity kinematics. For instance, a more forward lean trunk posture would bring the body center of mass and center of pressure more anteriorly. In turn, this would result in lower knee extensor moment and higher hip extensor moment (2,23,27,33). Given that joint power is a product of joint moment and angular velocity, an increase or decrease in the moments at the hip and knee would influence the respective joint energy absorption and generation values (assuming no change in angular velocity).

Modification of the foot strike pattern has been advocated as a means to reduce the risk of lower extremity injuries during running. For example, converting a rear foot strike pattern to a midfoot or forefoot strike pattern has been shown to result in a significant reduction in the knee extensor moments, power, and energy absorption (1,16,39). Furthermore, barefoot running, which typically results in a midfoot or forefoot strike pattern (24), has been promoted to reduce knee joint loading (4,5,39). However, studies have reported that adopting a midfoot or forefoot strike pattern also leads to significant increases in ankle plantarflexor moments, power, and energy absorption (1,4,6,16,28,29,39). Bonacci et al. (4) reported that converting runners from shod to barefoot running resulted in a 23.7% reduction in knee energy absorption and a 23.8% increase in ankle energy generation. Moreover, Williams et al. (39) reported that a forefoot strike pattern led to a decrease in knee negative power and a corresponding increase in ankle negative power when compared with a rear foot strike pattern.

Taken together, current literature on running suggests that converting to a midfoot or forefoot strike pattern reduces knee loading during running by shifting the mechanical demand to the ankle joint. This is potentially problematic because the ankle joint normally absorbs and generates 40%–50% of the total energy in the lower extremity during running (11,19). Shifting the mechanical demand to the calf musculature may lead to increased potential for foot and ankle injuries, which has been observed in persons who are habitually midfoot or forefoot strikers.

The results of the current study suggest that modifying sagittal plane trunk posture can be used as an alternative strategy to reduce knee loading during running. For example, a 7.2° greater trunk flexion angle corresponded to 23.3% lower energy absorption of the knee extensors and 13.3% lower energy generation of the knee extensors. Importantly,

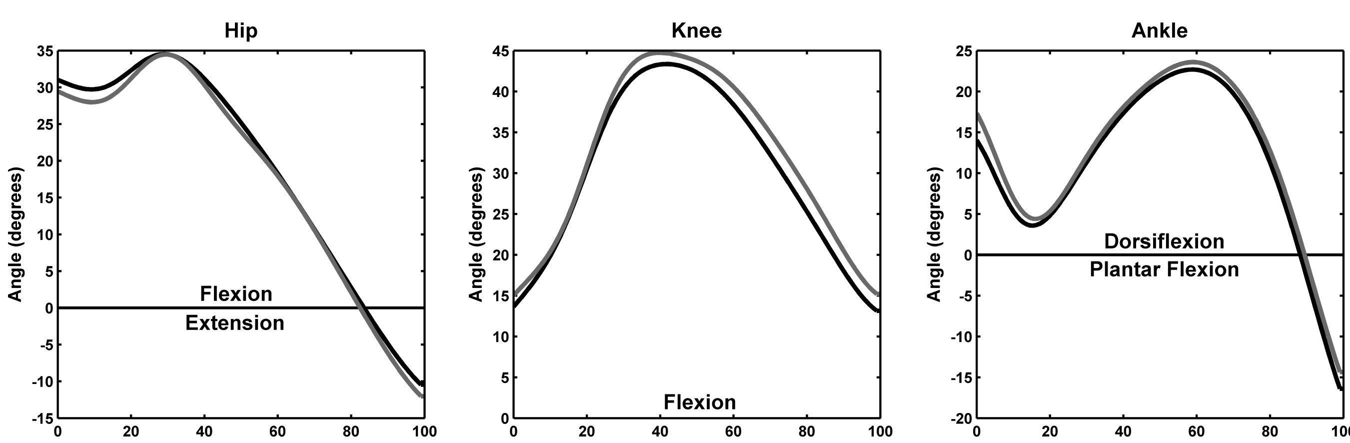

FIGURE 3—Comparison of sagittal plane hip, knee, and ankle kinematics during the stance phase of running between the HF (*black line*) and LF (*gray line*) groups.

this change in knee energetics occurred without a change in ankle energetics. The observed decrease in energy absorption at the knee is comparable with studies that have evaluated changes in knee loading associated with a forefoot/midfoot strike pattern and barefoot running (4,6).

A potential downside of using a forward lean trunk to reduce knee extensor demand is that there was a 140% increase in hip extensor energy generation. This may contribute to gluteus maximus or hamstrings strain. However, the hip joint absorbs and generates less than 10% of the total energy in the lower extremity during running (11,19) and is the least common injury site among all lower extremity joints (35,37). Therefore, it could be argued that shifting the mechanical demand to the least loaded joint as opposed to increasing the mechanical demand in the musculature that already bears a large mechanical demand (i.e., ankle plantarflexors) may be a more preferable strategy to reduce knee running injuries.

There are several limitations that need to be considered when interpreting the findings of the current study. First, subjects were asked to run at a controlled speed ($3.4\ m \cdot s^{-1}$) to eliminate the potential influence of speed on trunk posture and lower extremity biomechanics. Although all subjects were capable of running at the controlled speed, participants may not have run with their natural running posture owing to this constraint. Second, inverse dynamics methods were used to compute lower extremity joint kinetics. As such, cocontraction of the flexors and extensors was not taken into account and could have resulted in an underestimation of the muscle moment and, therefore, joint power and energy absorption and generation. Third, net joint moment and power curves were used to calculate energy absorption. Although negative work is performed primarily by the eccentric contraction of muscles, it is important to note that passive structures can also contribute to energy absorption. Finally, no direct causal relationships between trunk posture and lower extremity injury can be inferred from this study. Although a more extended trunk posture was associated with a higher demand on the knee extensors, all the subjects were free of symptoms at the time of participation. Further study is needed to examine whether incorporation of a forward trunk lean in persons with knee pain reduces symptoms.

CONCLUSIONS

Sagittal plane trunk flexion has a significant influence on hip and knee energetics during running. More specifically, persons who run with a relatively extended trunk posture exhibit higher energy absorption and generation of the knee extensors and lower energy generation of the hip extensors. In contrast, persons who run with a more flexed trunk posture exhibit higher energy generation of the hip extensors and lower energy absorption and generation of the knee extensors.

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The authors have no conflict of interest to disclose for this research.

The results of the present study do not constitute endorsement by the American College of Sports Medicine.

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