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Effects of Step Rate Manipulation on Joint Mechanics during Running

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Abstract

Purpose—The objective of this study was to characterize the biomechanical effects of step rate modification during running on the hip, knee and ankle joints, so as to evaluate a potential strategy to reduce lower extremity loading and risk for injury.

Methods—Three-dimensional kinematics and kinetics were recorded from 45 healthy recreational runners during treadmill running at constant speed under various step rate conditions (preferred, ± 5% and ± 10%). We tested our primary hypothesis that a reduction in energy absorption by the lower extremity joints during the loading response would occur, primarily at the knee, when step rate was increased.

Results—Less mechanical energy was absorbed at the knee ($p<0.01$) during the +5% and +10% step rate conditions, while the hip ($p<0.01$) absorbed less energy during the +10% condition only. All joints displayed substantially ($p<0.01$) more energy absorption when preferred step rate was reduced by 10. Step length ($p<0.01$), center of mass vertical excursion ($p<0.01$), breaking impulse ($p<0.01$) and peak knee flexion angle ($p<0.01$) were observed to decrease with increasing step rate. When step rate was increased 10% above preferred, peak hip adduction angle ($p<0.01$), as well as peak hip adduction ($p<0.01$) and internal rotation ($p<0.01$) moments, were found to decrease.

Conclusion—We conclude that subtle increases in step rate can substantially reduce the loading to the hip and knee joints during running and may prove beneficial in the prevention and treatment of common running-related injuries.

Keywords

energy absorption; knee; stride length; injury prevention; rehabilitation

Introduction

It is expected that approximately 56% of recreational runners and as high as 90% of runners training for a marathon will sustain a running-related injury each year (33). Approximately 50% of all running-related injuries occur at the knee with nearly half of those involving the patellofemoral joint (32). While several injury risk factors have been suggested (33,35), the

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inability of the lower extremity joints to adequately control the loads applied during initial stance is often identified (16,27) and the focus of injury prevention strategies (17,29).

In the interests of reducing loads to the lower extremity joints during the loading response (LR) of running, several popular strategies have been proposed including minimalist footwear and alterations in running form (7,13,31). A common outcome from these different strategies is an increased step rate. By increasing one's preferred step rate by 10% or greater (with a proportional decrease in step length assuming a constant speed), reduced impact load on the body is achieved due, in part, to less vertical center of mass (COM) velocity at landing (11,18). Subsequently, less energy absorption (negative work) is required by the lower extremity joints with the greatest effect observed at the knee (11). Thus, adopting a step rate greater than one's preferred may prove beneficial in reducing the risk of developing a running-related injury or facilitating recovery from an existing injury (4,10,14).

While a 10–20% increase in step rate substantially reduces joint loading, such a large deviation from one's self-selected step rate may prove challenging to adopt and compromise performance. For example, greater oxygen consumption is required when step rate is increased by more than 10% of preferred, while increases less than or equal to 10% of preferred reveal minimal change in metabolic cost (2,18). However, it is unknown whether the reduction in mechanical energy absorption by the joints occurs when subtle changes (≤10%) are applied. Reductions in tibial accelerations have been observed with only a 5% increase in step rate suggesting small alterations may result in measurable differences in joint loading (4).

The objective of this study was to characterize lower extremity joint biomechanics during running at constant speed under various step rate conditions (preferred, ± 5% and ± 10%). We tested our primary hypothesis that a reduction in energy absorption by the lower extremity joints would occur, primarily at the knee, when step rate was increased. We also compared the joint kinematics and ground reaction forces between running conditions to better understand the biomechanical adaptations to step rate manipulation.

Methods

Subjects

Forty-five healthy adult volunteers (age, 32.7 ± 15.5 yrs; height, 176.3 ± 10.3 cm; mass, 69.5 ± 13.1 kg) familiar with treadmill running agreed to participate in this study. All subjects ran a minimum of 24.1 km/wk (15 miles/wk; average volume, 29.8 ± 15.5 km/wk) and had been running for at least 3 months prior to study enrollment. Subjects were excluded if they experienced a leg injury in the prior 3 months; had undergone hip, knee, or ankle joint surgery; or currently had pain in their back or lower extremities while running. Based on a 20% change (11) between conditions in our primary outcome variable (knee joint energy absorption) with standard deviation equal to the estimated change, a sample size of 38 subjects was required to achieve a minimum power of 80% (α=0.05). The testing protocol was approved by the Health Sciences Institutional Review Board at the University of Wisconsin-Madison and subjects provided written informed consent in accordance with institutional policies.

Experimental protocol

Before data collection, each subject's preferred speed (2.9 ± 0.5 m/s) and step rate (172.6 ± 8.8 steps/min) were determined while running on treadmill for 5 min. Subjects were instructed to adjust the speed as needed over this period until identifying a speed that was representative of a typical moderate intensity run. Step rate was visually determined over a 30 s period by counting the number of right foot-strikes and multiplying by four. The process was repeated to ensure accuracy with the average value used. Subjects were then asked to run at their preferred speed under five step rate conditions: preferred, ±5% and ±10% of preferred. The order of step rate conditions was randomized for each subject, with 15 s of data recorded for each condition. Subjects ran with a digital audio metronome to facilitate the appropriate step rate. Data collection did not begin until the subjects were able to maintain the prescribed step rate for a minimum of 1-min determined by visual inspection. Upon completion of each condition, ratings of perceived exertion (RPE) were self-determined using the 15-point Borg Scale (3).

Data acquisition

Whole body kinematics were recorded (200 Hz) during all running conditions using an 8-camera passive marker system (Motion Analysis Corporation, Santa Rosa, CA, USA), which tracked 40 reflective markers placed on each subject, with 21 located on anatomical landmarks. An upright calibration trial was performed to establish joint centers, body segment coordinate systems, segment lengths and the local positions of tracking markers. A voluntary hip circumduction movement was also performed, with the corresponding kinematic data used to estimate the functional hip joint center in the pelvis reference frame (28). Kinematic data were low-pass filtered using a bidirectional, 4^{th} order Butterworth filter with a cutoff frequency of 12 Hz. Three dimensional ground reaction forces and moments were simultaneously recorded at 2000 Hz using an instrumented treadmill (Bertec Corporation, Columbus, OH). These ground reactions were then low-pass filtered using a bidirectional, 6^{th} order Butterworth filter with a cutoff frequency of 100 Hz. Foot contact and toe-off times were identified when the vertical ground reaction force exceeded or fell below 50 N, respectively, and were used to determine the stance and swing portions of the gait cycle. Five successive strides of the right limb for each subject were analyzed during each step rate condition.

Musculoskeletal model

The body was modeled as a 14-segment, 31 degree of freedom (DOF) articulated linkage. Anthropometric properties of body segments were scaled to each individual using the subject's height, mass, and segment lengths (8). The functional hip joint centers were used to scale the medio-lateral width of the pelvis. The hip joint was modeled as a ball and socket with three DOF. The knee joint was represented as a one DOF joint, in which the tibiofemoral translations and non-sagittal rotations were constrained functions of the knee flexion-extension angle (34). The ankle-subtalar complex was represented by two revolute joints aligned with anatomical axes (9). The lumbar spine was represented as a ball and socket joint at approximately the 3^{rd} lumbar vertebra (1). For each stride, joint angles were computed at each time step using a global optimization routine to minimize the weighted sum of squared differences between the measured and model marker positions (24). To compute COM, each model segment position was multiplied by the respective mass; these were then summed and divided by the total mass of the body. In addition, a segment-by-segment inverse dynamics analysis was used to calculate joint moments from the ground reaction forces and kinematic data. The joint powers were computed as the product of the moment and angular velocity for each joint, with mechanical work determined by integrating (function trapz, Matlab, MathWorks, Inc., Natick, MA) the respective negative (energy absorbed) and positive (energy generated) portions of each joint power curve.

Outcome Measures

All outcome measures were determined for each stride and averaged within each condition. Spatiotemporal gait descriptors were calculated including step length, stance duration, vertical excursion of the COM, foot inclination angle at initial contact (with respect to the horizontal), and the horizontal distance between the COM and heel at initial contact. The ground reaction forces were characterized using the peak vertical ground reaction force and the braking impulse, the integral of the anterioposterior ground reaction force from initial contact until midstance. The occurrence of a distinct impact transient was determined from the vertical ground reaction force on a per stride basis. For each condition, subjects were classified into three categories based how many of the 5 trials displayed an impact transient: rare, 0–1 trials; occasional, 2–3 trials; and frequency, 4–5 trials.

To address our primary hypothesis, the mechanical energy absorbed and generated at the hip, knee and ankle in the sagittal plane were determined. Energy absorbed was specific to the LR, defined from initial contact to peak knee flexion angle during stance (19), while energy generated was calculated throughout stance. In addition, the following discrete joint angles were identified during the LR: peak hip flexion, adduction and internal rotation; peak knee flexion; and knee flexion at initial contact. Similarly, the following joint moments were determined during the LR: hip extension moment at initial contact; peak hip abduction and hip internal rotation moments; and peak knee extension moment. All kinetic variables were normalized to subject body mass.

Statistics

All continuous variables were compared across conditions using a one-way ANOVA with repeated measures (STATISTICA 6.0, StatSoft, Inc, Tulsa, OK, USA), with significant main effects evaluated using Tukey's HSD. The distribution of the impact transient occurrence was compared between conditions using chi-square analysis. The criterion α level was set to 0.05. Because we were primarily interested in determining the effect of deviating from one's preferred step rate on running biomechanics, only those pair-wise comparisons involving the preferred step rate condition are reported.

Results

Step length ($p<0.01$), COM vertical excursion ($p<0.01$), horizontal distance from the COM and heel at initial contact ($p<0.01$), and braking impulse ($p<0.01$) were inversely related to step rate and displayed significant changes from preferred at both ±5% and ±10% conditions (Table 1). As step rate increased, step length was shorter with less COM vertical excursion; the heel was placed horizontally closer to the COM at initial contact with a reduction in the braking impulse (Figure 1). Foot inclination angle at initial contact ($p<0.01$), peak vertical GRF ($p<0.01$), and step duration ($p<0.01$) only differed if step rate was changed from preferred by 10% (Table 1). RPE increased ($p<0.01$) only when step rate was 10% greater than preferred. As step rate increased, the impact transient occurrence was found to decrease ($\chi^2 = 33.8$, $p< 0.001$) (Table 1).

The mechanical energy absorbed at the knee during LR was inversely related to step rate, with significant changes ($p<0.01$) from preferred during all conditions (Table 2 and Figure 2). That is, ~20% and ~34% less energy was absorbed at the knee when preferred step rate was increased 5% and 10%, respectively (Figure 3). A decrease in the preferred step rate produced a similar increase in the energy absorbed at the knee. Regarding the hip and ankle, a 10% decrease in preferred step rate produced a significant increase ($p<0.01$) in energy absorption, while a 10% increase resulted in less energy absorption at the hip only ($p<0.01$) (Table 1 and Figure 2). The knee had the greatest percent contribution to energy absorption during the LR (Figure 3) and showed the largest absolute change with step rate (Table 2).

Mechanical energy generation at the knee ($p<0.01$) and ankle ($p<0.01$) across stance was observed to decrease with an increase in step rate, with most conditions being significantly different from preferred (Table 1 and Figure 3). The energy generated by the hip was similar across conditions, with the exception of the −10% condition when an average increase of 40% over preferred was observed.

Kinematic analysis revealed a more flexed knee at initial contact ($p<0.01$) when step rate was increased 10%, with less peak knee flexion during stance ($p<0.01$) across conditions (Table 3). Similarly, as preferred step rate increased, the hip achieved less peak flexion ($p<0.01$) and adduction ($p<0.01$) during the LR, with a reduction in the peak abduction ($p<0.01$) and internal rotation ($p<0.01$) moments at the +10% condition (Table 3).

Discussion

The objective of this study was to characterize the influence of step rate on lower extremity biomechanics during running at a constant speed, with an emphasis on the change in mechanical energy absorbed at the hip, knee and ankle. In partial support of our hypothesis, we observed a substantial reduction in energy absorption at the knee and hip when step rate was increased above preferred. Our findings demonstrate that subtle changes in step rate can reduce the energy absorption required of the lower extremity joints, which may prove beneficial in the prevention and treatment of running injuries.

The decreased energy absorption observed at the knee and hip as step rate increased is likely due primarily to the corresponding change in step length and lower extremity posture at initial contact (10,22). Indeed, when step length and step rate were manipulated independent of each other, energy absorption was observed only if step length decreased (26). Because subjects in our study ran at their preferred speed for all conditions, an increase in step rate necessitated a proportional decrease in step length. As such, the heel was located more underneath the COM at initial contact with an accompanying decrease in the braking impulse. Similarly, peak knee flexion during stance and the COM vertical excursion were observed to decrease as step rate increased, suggestive of greater lower extremity stiffness (15). Of note, many of the biomechanical changes we found when step rate increased are similar to those observed when running barefoot or with minimalist footwear (12,20,23,31).

Our findings regarding energy absorption were comparable to those of Derrick et al. (11), despite calculating the negative work over different portions of the stance phase. Specifically, Derrick et al (11) was interested in the impact phase, defined as the initial ~20% of stance, while we considered the entire LR, representing the initial ~42% of stance. As a result, our absolute energy absorption values are greater; however, the relative change between conditions and joints is comparable.

While systematic kinematic and kinetic alterations were observed across the step rate conditions, the knee joint appeared to be most sensitive to changes in step rate. In particular, only the knee displayed significant changes in energy absorption between all step rate conditions, with a 20% decrease observed when preferred step rate was only increased by 5%. When combined with the significant reduction (18%) in energy generation at the knee during the same step rate condition, it is clear that a substantial decrease in mechanical work performed at the knee occurs with as little as a 5% increase in step rate.

Despite the clear reduction in the magnitude of knee joint loading when step rate is increased, the corresponding increase in the number of steps required for a given distance (i.e. loading cycles) may offset any potential benefit to injury reduction. That is, the cumulative loading incurred by the lower extremity may be the same for a given running distance. However, running with shorter stride lengths has been suggested to reduce the risk of a tibial stress fracture, despite the greater number of loading cycles (14). Thus, it appears that the benefits of reducing the magnitude of loading outweigh the detriments of increased loading cycles. Whether this same injury-reducing benefit is realized for other common running-related injuries (e.g., anterior knee pain, iliotibial band syndrome) has yet to be determined.

The reduced energy absorption at the hip and knee when running with an increased step rate may prove useful as an adjunct to current rehabilitation strategies for running injuries involving these joints and associated tissues. That is, injured runners could be instructed using a metronome to increase their step rate while maintaining the same speed. The associated reduction in loading may enable injured individuals to continue running without aggravating symptoms, while receiving care for their injuries. Similarly, utilizing an increased step rate may prove beneficial following injury recovery as part of a progressive return to running. Recent work has demonstrated that runners can be taught to modify their gait to reduce impact loading and that this modification can be maintained at a 1-month follow-up (6). The effectiveness of such strategies in reducing symptoms, facilitating injury recovery, and promoting a return to full running performance, however, remains unknown.

Excessive hip motion during running, specifically adduction and internal rotation, has been associated with anterior knee pain and iliotibial band syndrome (16,27,30). Our findings indicate that a 5–10% increase in step rate can significantly reduce peak hip adduction during the LR. Interestingly, an associated reduction in the hip abduction and internal rotation moments was not realized until step rate was increased by 10%. Regardless, it appears that running with a step rate greater than preferred reduces the biomechanical demands incurred by the hip in the frontal and transverse planes of motion, and therefore may be useful in the clinical management of running injuries involving the hip. However, it is uncertain whether injured individuals display the same biomechanical changes to step rate manipulation, or if existing symptoms or impairments would interfere.

Because preferred step rate and length are closely aligned with minimizing metabolic energy cost (2), modifying an individual's step rate may have a metabolic consequence. For example, subjects in the current study reported a greater RPE when step rate increased 10% above preferred. However, given the novelty of the modified step rate conditions to the subjects, we believe that this increase in perceived effort may be reflective of increased attentional focus (5), rather than an actual metabolic response (21). Indeed, increasing one's step rate to 10% above preferred has demonstrated no significant increase in oxygen consumption or heart rate (18). Further, the reduction in peak knee flexion observed at the higher step rate conditions in the current study has been associated with an improved economy (2,25).

Certain limitations with the present study should be considered when interpreting its findings. Despite subjects receiving adequate time to achieve the prescribed step rate, it is uncertain whether the observed biomechanical changes persist beyond the short-term. As one becomes more experienced running at a faster step rate, further biomechanical changes may occur. The step rate was determined through visual inspection and may have been prone to measurement error; however, post-hoc analysis of the force plate data confirmed the accuracy of the step rate assessments. Further, our step rate modification protocol was conducted on a treadmill, potentially limiting its generalizability to overground running. However, our findings are consistent with those performed overground (11,14). Finally, due to a limited number of markers during our experimental capture, non-sagittal kinematics and kinetics at the knee and ankle were not calculated. Given the clinical relevance of these additional degrees of freedom, their inclusion in future studies is warranted.

In conclusion, our findings indicate that a substantial reduction in energy absorption occurs at the hip and knee when step rate is increased to 10% above preferred with a constant running speed, while a 5% increase appears to reduce the total work performed by the knee.

Thus, the reduction in joint loading via step rate manipulation may have distinct benefits in the treatment and prevention of common running-related injuries involving the knee and hip.

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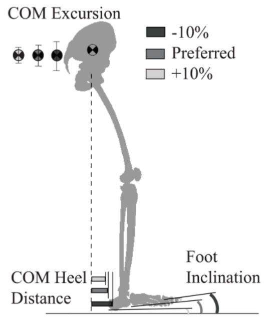

Figure 1.
Center of mass (COM) vertical excursion, horizontal distance from COM to heel at initial contact and foot inclination at initial contact decreased as step rate increased.

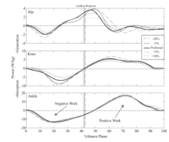

Figure 2.
At the hip and knee joints, energy absorption (negative work) and generation (positive work) were observed to decrease with increasing step rate. The ankle joint displayed a reduction in energy generation with step rate, while energy absorption remained relatively consistent. Negative work was determined during loading response (defined as foot contact to peak knee flexion angle) and positive work was determined throughout stance phase. Data are from a representative subject.

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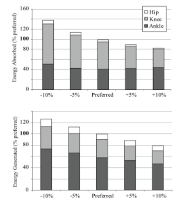

Figure 3.

While the knee joint showed the greatest absolute change in mechanical energy absorption with step rate, the hip joint showed the greatest percent change. Despite the overall reduction in mechanical energy absorption across joints at the higher step rate conditions, the ankle joint was responsible for a greater proportion. The mechanical energy generated by each joint during stance phase remained proportional across the step rate conditions. All data are reported as a percentage of the preferred condition.

Table 1

Mean (SD) step and ground reaction force measures during each step rate condition. Running speed was self-selected by each subject and maintained across conditions. GC, gait cycle; IC, initial contact; COM, center of mass; GRF, ground reaction force.

	Step Rate Condition				
Measure	**−10%**	**−5%**	**Preferred**	**+5%**	**+10%**
Step Length (cm)	111.8(18.2)*	105.9(17.4)*	100.8(16.5)	96.0(15.9)*	91.9(15.2)*
Stance Duration (%GC)	34.2(3.8)*	34.9(3.7)	35.4(3.4)	35.7(3.7)	36.2(3.3)*
IC COM - Heel Distance (cm)	11.4(4.2)*	10.2(3.7)*	9.2(4.0)	7.8(3.8)*	7.0(3.9)*
COM Vertical Excursion (cm)	10.7(1.5)*	9.6(1.4)*	8.7(1.3)	8.0(1.3)*	7.3(1.1)*
IC Foot Inclination (°)	7.9(10.0)*	6.6(8.6)	5.5(7.6)	3.3(8.1)	1.2(8.6)*
Braking Impulse (N* s/kg)	382.4(111.1)*	337.2(90.8)*	306.0(87.5)	274.1(78.4)*	256.7(77.4)*
Peak Vertical GRF (N/kg)	24.2(2.7)*	24.0(2.5)	23.6(2.3)	23.4(2.5)	23.0(2.5)*
Rating of Perceived Exertion	11.2(1.4)	11.1(1.2)	11.2(1.2)	11.4(1.4)	11.9(1.6)*
Impact Transient Occurrence (%)†					
Rare (0–1 strides)	34[a]	31	22	42 [a]	56 [a]
Occasional (2–3 strides)	16	27	31	21	19
Frequent (4–5 strides)	50	42	47	37	26

* significantly different from preferred, $p<0.05$

† percentage of subjects classified into the three categories defined by the number of trials in which a impact transient was evident, i.e. a subject was classified as “rare” for that condition if an impact transient occurred in 1 or less of the 5 strides.

[a] distribution significantly different from preferred ($\chi^2 = 33.8$, $p<0.001$)

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Table 2

Mean (SD) mechanical energy (J/kg) absorbed and generated in the sagittal plane during each step rate condition. Negative work was determined during the loading response of stance, while positive work was determined across all of stance. Running speed was self-selected by each subject and maintained across conditions.

	Step Rate Condition				
Measure	**−10%**	**−5%**	**Preferred**	**+5%**	**+10%**
Hip					
Energy Absorbed	1.2(0.8)*	0.9(0.7)	0.7(0.8)	0.5(0.8)	0.3(0.4)*
Energy Generated	4.6(2.9)*	3.8(2.7)	3.3(2.3)	3.2(2.3)	3.0(2.3)
Knee					
Energy Absorbed	13.5(4.9)*	11.1(4.0)*	9.2(3.6)	7.4(3.4)*	6.1(3.2)*
Energy Generated	13.8(5.1)*	12.3(5.2)	11.3(4.3)	9.3(3.7)*	8.4(3.3)*
Ankle					
Energy Absorbed	8.4(4.9)*	7.0(4.0)	6.7(4.0)	6.9(4.1)	7.2(3.8)
Energy Generated	24.5(6.5)*	22.0(6.2)*	19.3(5.5)	17.6(4.7)*	15.5(4.7)*

* significantly different from preferred, $p<0.05$

Table 3

Mean (SD) joint angle and moment measures during each step rate condition. All values were determined during the loading response, defined as initial contact (IC) to peak knee flexion angle during stance. Running speed was self-selected by each subject and maintained across conditions.

	Step Rate Condition				
Measure	**−10%**	**−5%**	**Preferred**	**+5%**	**+10%**
Hip					
Peak Flexion Angle (°)	30.7 (5.7)*	27.9(5.8)	26.7(5.5)	25.3(5.5)	23.6(6.0)*
Peak Adduction Angle(°)	11.3 (3.6)*	10.8(3.3)	10.4(3.3)	9.5(3.1)*	8.7(3.1)*
Peak Internal Rotation Angle(°)	1.3 (4.9)*	0.8(4.6)	0.4(4.3)	0.3(4.3)	0.4(4.4)
IC Extension Moment (Nm/kg)	0.2 (0.5)	0.3(0.6)	0.3(0.5)	0.4(0.5)	0.4(0.5)
Peak Abduction Moment (Nm/kg)	1.9 (0.5)	1.8(0.4)	1.8(0.4)	1.8(0.4)	1.7(0.4)*
Peak Internal Rotation Moment(Nm/kg)	0.7 (0.2)*	0.6(0.2)	0.6(0.2)	0.6(0.2)	0.5(0.2)*
Knee					
IC Flexion Angle (°)	16.9 (4.2)	17.0(4.1)	17.8(4.0)	18.7(3.9)	19.6(4.2)*
Peak Flexion Angle(°)	50.6 (4.8)*	48.0(4.7)*	46.3(4.5)	44.1(4.7)*	42.8(4.4)*
Peak Extension Moment(Nm/kg)	2.7 (0.6)*	2.7(0.6)	2.5(0.6)	2.4(0.6)	2.2(0.4)*

* significantly different from preferred, $p<0.05$