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Lower extremity joint kinetics and energetics during backward running

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ABSTRACT

DEVITA, P. and J. STRIBLING. Lower extremity joint kinetics and energetics during backward running. *Med. Sci. Sports Exerc.*, Vol. 23, No. 5, pp. 602–610, 1991. The purpose of this study was to measure lower extremity joint moments of force and joint muscle powers used to perform backward running. Ten trials of high speed (100 Hz) sagittal plane film records and ground reaction force data (1000 Hz) describing backward running were obtained from each of five male runners. Fifteen trials of forward running data were obtained from one of these subjects. Inverse dynamics were performed on these data to obtain the joint moments and powers, which were normalized to body mass to make between-subject comparisons. Backward running hip moment and power patterns were similar in magnitude and opposite in direction to forward running curves and produced more positive work in stance. Functional roles of knee and ankle muscles were interchanged between backward and forward running. Knee extensors were the primary source of propulsion in backward running owing to greater moment and power output (peak moment = 3.60 $N \cdot m \cdot kg^{-1}$; peak power = 12.40 $W \cdot kg^{-1}$) compared with the ankle (peak moment = 1.92 $N \cdot m \cdot kg^{-1}$; peak power = 7.05 $W \cdot kg^{-1}$). The ankle plantarflexors were the primary shock absorbers, producing the greatest negative power (peak = −6.77 $W \cdot kg^{-1}$) during early stance. Forward running had greater ankle moment and power output for propulsion and greater knee negative power for impact attenuation. The large knee moment in backward running supported previous findings indicating that backward running training leads to increased knee extensor torque capabilities.

LOCOMOTION, MOMENT OF FORCE, MUSCLE POWER, HIP, KNEE, ANKLE, FORCE PLATFORM, CINEMATOGRAPHY, WORK

The investigation of backward locomotion has recently received attention from biomechanics researchers. Several studies on backward walking (BW) have sought to provide insight into the neural control mechanisms used in gait. Vilensky et al. (15) used kinematic variables and Thorstensson (13) used both kinematic and electromyographic (EMG) data to investigate the hypothesis that BW is produced by a temporal reversal of the forward walking (FW) muscle activation pattern. These investigations reported contradictory conclusions, however, owing to the limitations of using kinematic variables to assess neural function and underlying movement causes. Winter et al. (18), using joint kinetic and energetic variables along with EMG data, provided stronger evidence that BW was accomplished using a reversed FW muscle excitation pattern.

The second major focus of backward locomotion studies has been the training and rehabilitative effects of this gait pattern. These studies have only reported comparative descriptions of forward and backward gait patterns and the overall effect of backward running (BR) on the function of selected muscles. Two studies comparing forward gait with BW and BR (1,2) reported that, owing to observed differences in kinematic variables, backward locomotion may be a viable training and rehabilitative modality. BW kinematic and EMG descriptors were reported (7) for use by physical therapists in developing rehabilitation procedures. BR training regimens have been reported to improve knee joint stability (9) and to increase peak concentric isokinetic torque of the quadriceps muscles (14).

Although BR is used for rehabilitation of lower extremity injuries and to increase muscle strength (6), the underlying kinetics and energetics producing BR have not been investigated. The purpose of this study was to measure lower extremity joint moments of force and joint muscle power patterns used to perform BR. Assessments of these variables will identify the loads on various lower extremity muscle groups and provide quantitative data for use in developing both rehabilitation and training protocols. BR results were also compared with a set of corresponding FR data from one of the subjects.

METHODS

Subjects. Five healthy, male runners (mean mass: 71 kg; mean age: 25 yr) volunteered as subjects for the study. One subject normally used BR as part of his

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regular training regimen, and another had previously performed BR on a treadmill and over a force platform. The remaining subjects were naive to extended BR. All subjects signed an informed consent form prior to participation in accordance with university and American College of Sports Medicine policy.

Instrumentation. A force platform interfaced to a computer was located in the center of a running lane in a large gymnasium and between two photocells which were used to monitor running speed (3.0 ± 0.1 $m \cdot s^{-1}$) over a 4 m interval. The running lane provided for 20 m of BR for each trial. A black line was painted along the center of the lane to assist the subjects through the testing area. The vertical and anteroposterior ground reaction forces (GRF) along with the moment around the force platform mediolateral axis were sampled at 1000 Hz.

A 16 mm LoCam camera, operating at 100 Hz and located 14 m from the force platform, was used to make sagittal plane film records of each trial. The camera field of view covered approximately 3 m on the approach side and 1 m past the platform. This arrangement was used to film the swing phase and subsequent force platform stance phase of the leg closer to the camera (right leg).

Experimental protocol. The experimental protocol included two test sessions per subject. The subjects wore their own running shoes and shorts during both sessions. The first session was a 30 min practice session in which each subject performed BR around the gym and through the experimental environment. The practice session familiarized the subject with the general movement pattern, running at the selected velocity and contacting the force platform with the correct foot. This session was conducted the day before the experimental test session.

Prior to data collection in the experimental session, black circular marks (2 cm in diameter) were applied over the fifth metatarsal head, back edge of the shoe at mid-upper height, lateral malleolus, lateral femoral condyle, greater trochanter, and shoulder to aid in the digitizing process. Also, circumference measures of the upper thigh, knee ankle, and metatarsal heads were taken for later use in the mathematical human body model (5). The subject then practiced BR around the gym and through the experimental area for 10 min. He then practiced at the selected speed, and, when he was consistent at this pace and at contacting the platform in a normal stride, the data were collected.

Each subject performed 25 successful trials, from which ten were randomly selected for analysis. The performance of 25 trials and the random selection of ten were done to improve the accuracy of the data and to ensure that representative trials were obtained. All trials were visually monitored to verify that the measured stride was similar to the other strides in the trial.

Data reduction. The six body points and the front edge of the force platform were digitized during the right limb swing and subsequent force platform stance phases. The film records were digitized starting at eight frames before the swing phase and continuing until four frames after the force platform stance phase. The extra frames were included to improve the accuracy of the data near the performance boundaries (19). The force platform point was used as a reference point and to subsequently locate the center of pressure in the kinematic reference frame.

The average swing and stance phase relative durations were obtained from the film data. The film records were then smoothed with an interactive cubic spline routine and interpolated to produce 300 frames of data with the respective swing and stance phase proportions, therefore normalizing for both swing and stance phase durations.

The GRF data were then scaled and interpolated so that each stance phase body position had a corresponding applied GRF. The accuracy of the interpolation routine was tested to verify that error introduced by the interpolation was minimal. Overall, the peak and average force values of the interpolated curves were within 1% of the original GRF data.

The GRF and force platform mediolateral axis moment data were then used to calculate the center of pressure, which was expressed as a point along the horizontal axis of the kinematic reference frame. The center of pressure was used to locate the point of application of the GRF on the runner.

The Hanavan body model (5), along with the average segmental mass proportions in Winter (16), was used to estimate the magnitude and location of lower extremity segmental mass centers and their moments of inertia. An inverse dynamics analysis, combining the anthropometric, film, and GRF data, was used to calculate the lower extremity joint reaction forces and joint moments of force throughout the stride. A convention of positive moments as extensor or plantarflexor torques was selected.

The use of inverse dynamics to calculate lower extremity joint moments tends to increase the random noise component of the derived values as the analysis moves further away from the point of application of the externally applied GRF. Therefore, the moment curves were digitally filtered with a second-order, low pass digital filter using average cutoff frequencies of 40, 15, and 10 Hz for the ankle, knee, and hip, respectively. The raw and smooth moment curves were visually examined to verify that the smooth curves retained the general characteristics of the raw curves but not the high frequency oscillations. The reflection extrapolation procedure described by Smith (12) was incorporated into the filter to increase the accuracy of the ends of the curves.

Relative joint position and velocity of the hip, knee, and ankle were calculated from the kinematic data with a similar convention of positive values as extended positions or extensor velocities. Zero degrees for the three joints corresponded to an erect, standing position with the trunk, thigh, and leg in a straight line and the foot at a right angle to the leg. The joint muscle powers were obtained as a product of the joint moments and joint angular velocities. Positive power phases indicated concentric muscle contractions and positive work performed by the torque-generating muscle group, while negative phases indicated eccentric contractions and negative work. The areas under selected portions of the curves were calculated and represent the work done in those phases.

The three moment curves were summed to produce a support moment curve (17), and the three power curves were summed to produce a total power curve. The support moment and total power curves represent the net torque and power outputs of the entire extremity over the stride period. The moment and power data along with the work values were normalized to subjects' body mass and were expressed in units of $N \cdot m \cdot kg^{-1}$, $W \cdot kg^{-1}$, and $J \cdot kg^{-1}$, respectively.

Reference data. The subject experienced with treadmill and force platform BR was also tested in forward running at a 3.0 ± 0.1 $m \cdot s^{-1}$ pace. The experimental procedures and analysis for this test were identical to those for BR, except for the running direction and the fact that 15 trials were analyzed. The derived joint position, moment, and power results were used for a general comparison with the BR results.

All statements describing FR and BR movements or torques are relative to the body's position. For example, forward hip rotation is hip flexion in both FR and BR but will move the limb toward and away from the direction of progression in FR and BR, respectively.

Reliability of results. The reliability of the moment and power calculations was evaluated by reanalyzing five BR trials and comparing these results with those of the original analysis on a trial-by-trial basis. The five trials were redigitized and smoothed by a research assistant who did not participate in the original data reduction. Visual comparison of the corresponding curves indicated that the swing phase results were highly reliable since the matched curves were virtually identical during this time.

Stance phase moment and power analyses were less reliable. Maximum stance phase moments and powers were derived for each curve, and the absolute differences between the corresponding trial pairs were then calculated. The mean absolute differences in maximum stance phase moments were 0.40, 0.20, 0.57, and 0.31 $N \cdot m \cdot kg^{-1}$ for support, hip, knee, and ankle moments, respectively. These values represent approximately 15% of the magnitude of the maximum stance phase moments for each curve. For BR, therefore, reliable estimates of the maximum stance phase moments can be considered as the observed values plus or minus 15%. Also, in comparisons of maximum moments between conditions, minimum differences of the magnitudes noted above must be observed before the conditions can reliably be considered as different.

The mean absolute differences in maximum stance phase powers were 1.92, 1.05, 0.74, and 2.55 $W \cdot kg^{-1}$ for total, hip, knee, and ankle powers, respectively. These values represent approximately 21% of the magnitude of the maximum stance phase powers for each curve. Reliable maximum power estimates during BR stance can, therefore, be considered as the observed values plus or minus 21%, and, when comparing maximum powers between BR and FR, minimum differences of the magnitudes noted above must be observed before the conditions can reliably be considered as different.

RESULTS

Kinematics. BR swing and stance phase relative durations were 63 and 37% of the total stride. Similar FR swing (65%) and stance (35%) relative durations were observed. For the purpose of graphic presentation, BR and FR curves were adjusted to 64 and 36% swing and stance phase relative durations, respectively.

The mean BR stride length and frequency, along with their standard deviations, were 2.10 ± 0.21 m and 1.43 ± 0.20 Hz, respectively. The corresponding FR mean values showed this gait to have a longer stride length (2.30 m) and a lower stride frequency (1.34 Hz). Figure 1 presents equal interval stick-figure representations of BR and FR trial and is shown to provide a visual reference of the movement patterns. The mean joint position curves for two subjects during BR and for the FR data are presented in Figure 2. The lower BR curves are from the subject who regularly performed BR, and the other BR data are from a naive subject. The curves are representative of the other subjects' results and show that the practice session and practice trials before data collection were effective in providing time for the subjects to learn the task. The standard deviation bandwidths around both subjects' BR curves were reasonably small and were comparable to the FR results. The mean intrasubject standard deviation bandwidths over the entire stride were approximately 2.5 and 1.9 degrees for BR and FR, respectively.

The mean BR joint position curves for all subjects along with the FR position data are shown in Figure 3. The between-subject variability was larger than the within-subject variability over trials but was still not excessively large, indicating a reasonably similar movement pattern across subjects. The BR joint range of

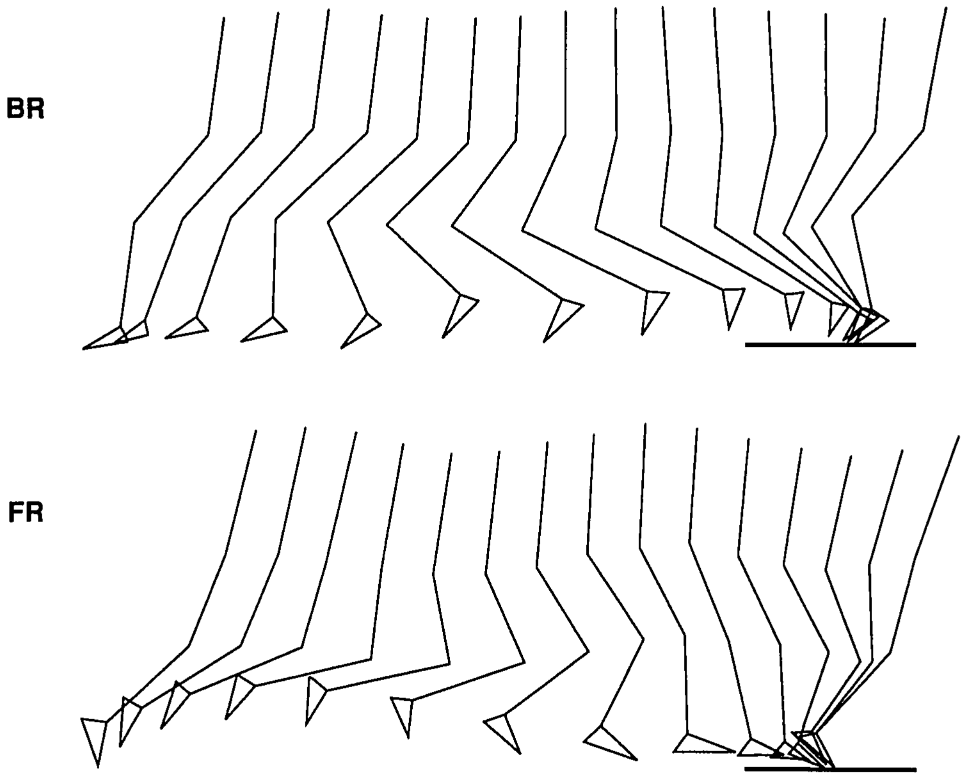

Figure 1—Stick-figure representations of BR (*top*) and FR (*bottom*). Both sequences display running from left to right. Swing and stance phases are shown, and *horizontal lines* represent the force platform.

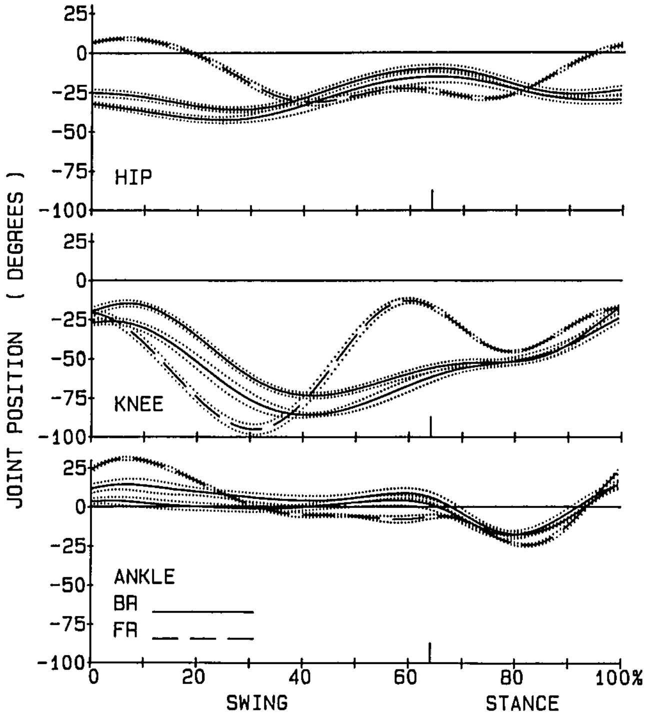

Figure 2—Individual subject BR and FR mean joint position curves. *Dotted lines* represent mean ±1 SD. Swing and stance phases are 0–64% and 64–100%. Positive values are extended or plantarflexed positions. The lower BR curves are from the subject experienced in BR. The narrow standard deviation bandwidths show that the subjects had sufficient BR practice and performed the task nearly as consistently as FR was performed.

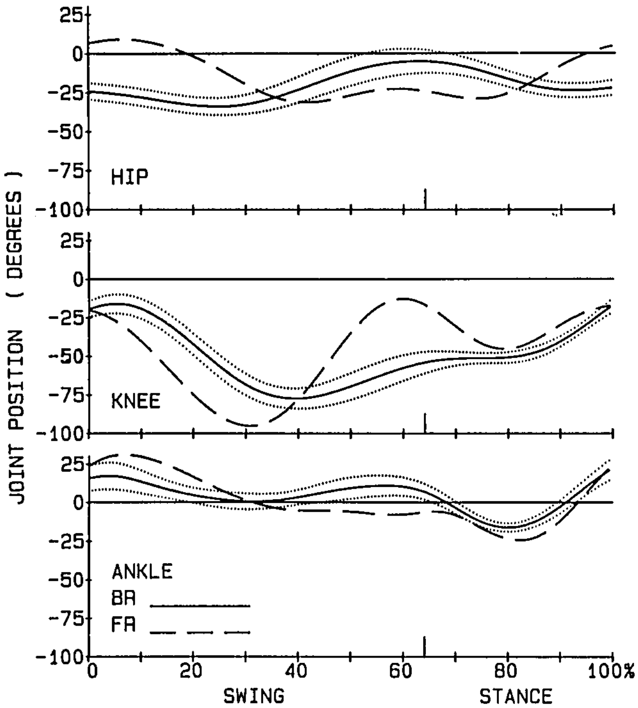

Figure 3—Mean BR joint position curves for all subjects and mean FR joint position curves for one subject. *Dotted lines* represent mean ±1 SD. Swing and stance phases are 0–64% and 64–100%. Positive values are extended or plantarflexed positions. Kinematic differences between BR and FR were observed at all joints. BR had only flexed hip positions, whereas FR had extended and flexed positions. Less knee flexion in late swing was observed in BR compared with FR, and the knee remained stationary in early BR stance. As stance approached, the ankle dorsiflexed in FR but remained plantarflexed in BR.

motion (ROM) was less than that observed in FR, which is in agreement with a previous comparison (2). Mean ROM values for the hip, knee, and ankle were 27, 60, and 43 degrees in BR and 40, 83, and 55 degrees in FR.

The joint position curves (Fig. 3) showed several kinematic differences between BR and FR. The hip was continually flexed in four of the five subjects in BR, with only one subject having a small amount of hip extension in late swing. In contrast, FR had extended hip positions in late stance through the first third of swing. Knee joint motion was different between the two gait patterns during late swing and early stance. As the BR stance phase approached, the knee remained more flexed, and, during early BR stance, no knee flexion occurred. Ankle joint kinematics were similar between BR and FR during the first half of swing, but then FR had a dorsiflexed position into stance whereas BR had a second plantarflexion phase until ground contact. The movement at the ankle joint was similar in stance for both gait patterns and consisted of dorsiflexion and then plantarflexion motions.

Joint moments and muscle powers. Figures 4 and 5 show the mean moment and power curves for BR and FR. The hip and knee curves generally had opposite polarities but similar magnitudes between the running conditions during the swing phase. In BR, the hip had eccentrically acting extensor and then flexor moments during swing as indicated by the two negative hip power phases. The initial extensor moment stopped the forward limb rotation, and the flexor moment stopped the limb extension, terminating the recovery action. These moments produced $-0.44\ J \cdot kg^{-1}$ of work. A smaller amount of positive work, $0.24\ J \cdot kg^{-1}$, was performed by the hip flexors at 8–32% of the stride and then by the hip extensors at 43–58% of the stride to rotate the extremity forward and then backward in FR swing.

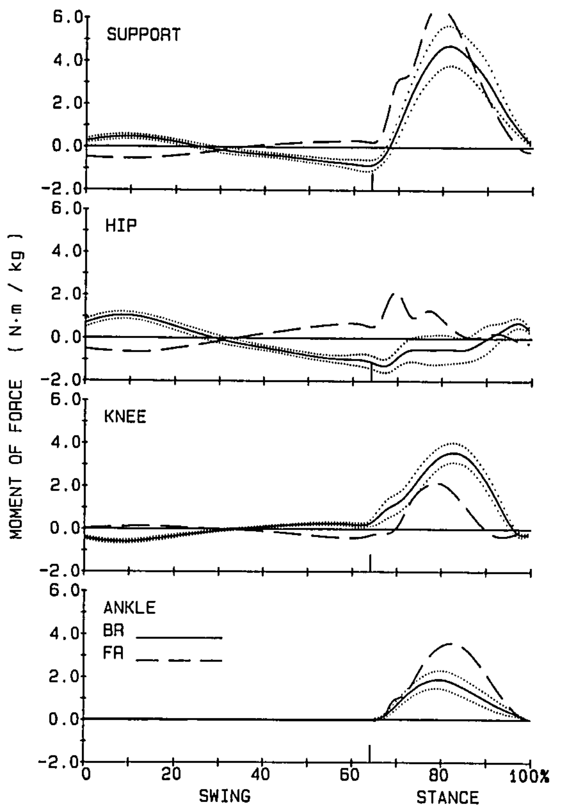

Figure 4—Mean BR joint moment curves for all subjects and mean FR moment curves for one subject. *Dotted lines* represent mean ±1 SD. Swing and stance phases are 0–64% and 64–100%. Positive values are extensor or plantarflexor moments. Support moment is the sum of the individual joint moment curves and shows the overall torque output of the extremity. Moments of similar magnitude and opposite polarity were observed between BR and FR in the support, hip, and knee curves during swing and in the hip curves during stance. FR had greater support and ankle stance phase moments, and BR had a greater knee stance phase moment.

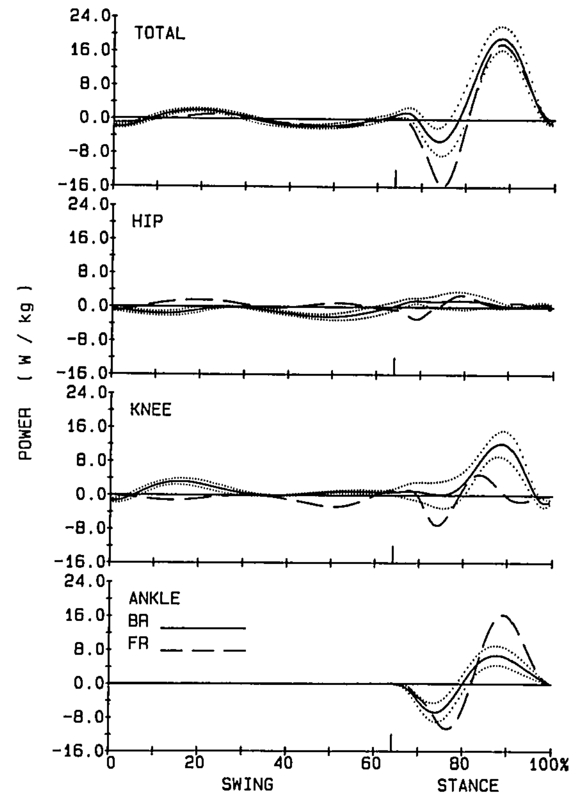

Figure 5—Mean joint power curves for all subjects and mean FR power curves for one subject. *Dotted lines* represent mean ±1 SD. Swing and stance phases are 0–64% and 64–100%. Total power is the sum of the individual joint powers. Positive and negative values indicate energy generation (concentric muscle action) and energy absorption (eccentric muscle action). *Areas under the curves* indicate the work done by the dominant torque-producing muscle group. Swing phase powers were similar in magnitude and opposite in polarity at the hip and knee but were similar in both features in total power. During stance, BR had less negative total power due to absence of eccentric knee action and less eccentric ankle action compared with FR. BR had greater knee power and less ankle power in late stance compared with FR.

The hip flexor moment continued into the stance phase in BR, concentrically acting at a low power level and producing $0.22\ J \cdot kg^{-1}$ of work to propel the runner backward. In contrast, the FR stance phase hip extensor moment worked eccentrically to stop hip flexion, as indicated by the negative power at 64–73% of the stride. This moment then worked concentrically in midstance (positive power phase) to propel the runner. The net work contribution of this moment was only 0.08 $J \cdot kg^{-1}$.

In both running gaits, the hip was flexed at ground contact, and the vertical GRF produced an upward knee joint reaction force by the leg acting on the thigh. This force produced a flexor moment at the hip which assisted the hip muscle moment to flex the thigh in BR. In FR, however, the joint reaction force flexor moment counteracted the hip extensor moment, producing hip flexion in early stance. Also, the hip was in a more flexed position at ground contact in FR, which increased the joint reaction force moment arm and, therefore, also increased the counterproductive torque.

In BR, the knee had a small, eccentrically acting flexor moment as indicated by the negative knee power at 0–6% of the stride. The flexor moment functioned to stop knee extension in early swing. The flexor moment then worked concentrically until midswing, shortening the limb for backward rotation by flexing the knee. This action carried the foot in the progression direction. At midswing, the net knee moment changed to extensor dominance, which worked concentrically to lower the foot and provide tension across the knee joint in preparation for ground contact. The concentric action is indicated by the positive knee power at 42–64% of the stride. The total work produced by the knee moment during BR swing was 0.41 $J \cdot kg^{-1}$.

In FR, the knee extensor moment performed a small amount of negative work from 0 to 28% of the stride to control knee flexion in early swing. The knee flexors then worked eccentrically until 60% of the stride to limit forward rotation of the leg prior to ground contact. The total negative work done by these muscle groups was −0.43 $J \cdot kg^{-1}$.

The dominant stance phase moment in BR was observed at the knee, which showed a net extensor pattern for 90% of stance and reached a relatively large peak value of 3.60 $N \cdot m \cdot kg^{-1}$. The combination of a net extensor moment entering stance and its large magnitude prevented knee flexion and energy absorption by the quadriceps muscle group after ground contact. The BR knee power curve showed negligible power in early stance followed by a large power output producing 0.95 $J \cdot kg^{-1}$ of work to propel the runner upward and backward.

In contrast, the FR knee moment was flexor dominant at ground contact, reached a smaller peak extensor moment of 2.19 $N \cdot m \cdot kg^{-1}$, and changed to flexor dominance at 72% of stance. This torque pattern produced two energy absorption phases at 70–79% and 90–100% of the stride and two energy generation phases at 64–70% and 79–90% of stride. The net work output of the stance phase knee moment was −0.06 $J \cdot kg^{-1}$.

Ankle moment and power stance phase patterns were temporally similar between BR and FR but were smaller in magnitude in BR. Peak ankle plantarflexor torques were 1.92 and 3.60 $N \cdot m \cdot kg^{-1}$ for BR and FR, respectively. The plantarflexor dominant torque produced energy absorption (negative ankle power) and generation (positive ankle power) phases in both gaits during stance. In BR, the ankle plantarflexors produced −0.36 and 0.51 $J \cdot kg^{-1}$ of work in the two phases, for a net output of 0.16 $J \cdot kg^{-1}$. The corresponding FR values were −0.65 and 1.16 $J \cdot kg^{-1}$, which produced a net result of 0.51 $J \cdot kg^{-1}$ and indicated that the ankle plantarflexors were more important in propelling the body center of mass in FR than in BR.

The combined output of all three moment curves was summarized in the support moment curves. The swing phase support moments were dominated by the hip torques in BR and FR and had the same pattern as the hip moments. The support moment curves were also opposite in polarity during swing; however, the total joint muscle power curves had similar polarities throughout the swing phase. This result was due to the identical functional demands on the limb during swing. At the beginning of both BR and FR swing, energy must be absorbed from the limb to stop its rotation away from the progression direction and then transferred to the limb for recovery. This energy absorption action can be seen in the total power curves from 0 to 8% of the stride, and the energy transfer or generation phase followed at 8–32% of the stride. These functions were accomplished by extensor and flexor dominant support moments in BR and FR, respectively. The recovery movement was then stopped with a second energy absorption phase at 32–60% of the stride. Finally, a small amount of positive work was performed near the end of swing to begin rotating the limb away from the progression direction for support and propulsion in stance. These actions were produced by flexor and extensor dominant support moments in BR and FR, respectively.

The stance phase support moment curves had a net extensor moment in both BR and FR; however, greater support was provided by the extremity in FR. This result was due to a FR hip extensor moment compared with a BR flexor moment and a larger FR ankle plantarflexor moment. The BR hip flexor moment dominated the support moment in early stance and resulted in a brief active lowering of the body center of mass.

Due to the positive hip power, absence of negative knee power, and low negative ankle power in the first half of stance, the BR total power curve showed only a small negative power output, producing only −0.24 $J \cdot kg^{-1}$ of work. The corresponding FR value was −0.88 $J \cdot kg^{-1}$. The total power peak value and work done during the stance propelling phase were similar in BR and FR. For example, a net amount of 1.49 and 1.40 $J \cdot kg^{-1}$ of work was done by the lower extremity muscles in BR and FR during the stance propelling phase.

The sequence of peak stance phase moments was different in BR and FR. FR had peak hip, knee, and ankle extensor moments at 14, 39, and 53% of the stance phase, which were the same order and similar magnitudes as previously shown in slow FR jogging (17). BR peak moments occurred at 8, 53, and 42% of the stance phase for the hip, knee, and ankle joints, showing a reversed contribution order for the knee and ankle moments.

DISCUSSION

Although the present FR data were based on a single subject, their overall agreement with previously reported FR data (4,17) validates their use for a general comparison of backward and forward running. The FR moment curves were similar to those of 11 normal subjects during slow jogging (17), except for a smaller hip flexor moment and the knee flexor moment in late stance. The present stance phase moments were also similar to results from six average runners performing at a faster pace (4.29 $m \cdot s^{-1}$) (4). The power curves were similar to those of Winter (17), showing nearly identical phasic relationships.

Reliability estimates of maximum moment and power values were presented as a range around the observed sample values. These data provide a quantitative assessment of the accuracy of both the results and the comparisons between the running conditions. The observed maximum moment values in BR and FR differed by approximately 1.64 $N \cdot m \cdot kg^{-1}$ in the support, knee, and ankle moments and by 3.50 $N \cdot m \cdot kg^{-1}$ in the hip moment. These differences were 2.5–17.5 times larger than the magnitude of the reliability range values listed above. The most notable differences between BR and FR muscle powers were in the maximum knee and ankle powers, along with the minimum total power during stance. The observed differences between the running conditions ranged from 6.67 to 9.67 $W \cdot kg^{-1}$ for these measures and were 3.4–9.0 times larger than the magnitude of the reliability values. Based on these results, the observed differences between backward and forward running moments and powers appear to be true biomechanical differences and not measurement error.

In a comparison of BW and FW, Winter et al. (18) noted that, when viewing FW and reverse BW film records, observers could not correctly identify the gait pattern. Additionally, these researchers reported high correlation coefficients ($r = 0.95$) between FW and temporally reversed BW joint position curves at both the hip and the knee. Present curve correlations for temporally reversed BR and FR joint position data from the subject who performed both gaits were $r = 0.92$, 0.77, and 0.60 for the hip, knee, and ankle, respectively. These results indicated that there was less similarity between backward and forward running movement patterns than between backward and forward walking.

When viewing reversed BR and FR file records, observers correctly identified the gait patterns because of magnitude differences in the joint positions. The continually flexed hip position gave the appearance of downhill FR when viewing reversed BR films, as did the greater amount of knee flexion in midstance (3). The higher curve correlations at the hip and knee identified similar temporal features between gait kinematics but were insensitive to magnitude differences.

The overall temporal similarity between BR and FR was also shown in the nearly identical relative swing and stance phase durations for the two gait patterns. Additionally, the presently observed BR swing and stance relative durations were similar to previously reported FR values at a similar running speed (17). Threlkeld et al. (14) also reported similar phasic relationships between BR and FR, although the magnitude of their results differed from the present values. Their stance phase was 15% shorter, probably due to a faster (19%) running speed.

The presently observed shorter stride length and higher stride frequency in BR compared with FR were in agreement with previously reported BR results (14). Those investigators also reported a 30% lower GRF vertical impulse in BR. The present BR mean GRF vertical impulse (3.63 $N \cdot s \cdot kg^{-1}$) was 14% lower than the FR mean GRF vertical impulse (4.21 $N \cdot s \cdot kg^{-1}$) for the subject who performed both movements. Also, the BR mean GRF vertical impulse across all subjects (3.25 $N \cdot s \cdot kg^{-1}$) was 12% lower than the corresponding FR value (3.71 $N \cdot s \cdot kg^{-1}$) reported previously (10) for the same running speed. The reduced BR vertical impulse directly affected the stride length and frequency by reducing the vertical velocity at the end of stance, which, therefore, produced shorter flight times.

The lower GRF vertical impulse in BR was due to the combination of a net hip flexor moment and a lower ankle plantarflexor moment compared with FR. This combination of moments, along with the linear momentum of the runner entering stance, resulted in a more horizontal trajectory for the runner after stance. The reduced GRF vertical impulse and resultant vertical velocity after stance may partially explain the reduced negative total power observed in stance. With a lower vertical takeoff velocity, the impact velocity on the subsequent stance phase will also be reduced, and, therefore, the negative work required to stop the fall of the body center of mass will be less. As noted earlier, the large BR knee extensor moment was also partially responsible for the reduced negative work in stance.

In FR, the knee had a low flexor moment at ground contact to ensure the occurrence of knee flexion and energy absorption by the quadriceps. This moment was partially a result of the tension developed in the biarticular hamstring muscle group (11) to provide a hip extensor moment in early stance. In BR, however, two kinematic actions required the knee to exhibit a low extensor moment entering stance and a high extensor moment in midstance. First, hip flexion prior to and during stance was needed to propel the runner backward, and this action was produced by a hip flexor moment. The hip moment was probably partially produced by the biarticular rectus femoris, which also effected a low knee extensor moment. Second, the knee was flexed 50 degrees at ground contact and in early stance, giving the appearance of a more seated type of gait. In this body position, the external GRF produced a larger flexing moment at the knee. Therefore, a relatively high knee extensor moment was required to prevent the runner from collapsing.

The need for the knee extensor mechanism to support the body center of mass is made evident by comparing the joint moment curves with the support moment curve. The flexor hip moment reduced the total trunk support provided by the extremity, and the low ankle moment provided less support in BR compared with FR. Ratios between the mean stance moments for each joint and the mean support moment indicated that the knee provided 74% of the support function whereas the ankle and hip accounted for 40 and −14%, respectively. In contrast, FR results indicated that the ankle was primarily responsible for support (58%) while the hip and knee provided equal but less support (21%) through their extensor moments.

The inverse relationship between knee and ankle torque patterns in BR and FR led to an exchange in functional roles for the extensor muscles at these joints. In FR, the knee negative power burst produced −0.30 $J \cdot kg^{-1}$ of work to absorb the impact shock and stop the fall of the body center of mass. In BR, only the ankle muscles absorbed the impact shock by doing −0.36 $J \cdot kg^{-1}$ of work. The primary power phase for propulsion in FR came from the ankle plantarflexors, which did 1.16 $J \cdot kg^{-1}$ of work, and the secondary contribution came from the knee extensors, which did 0.29 $J \cdot kg^{-1}$ of work. In BR, the knee extensors provided the primary propulsive work of 0.95 $J \cdot kg^{-1}$, while the ankle plantarflexors contributed only 0.51 $J \cdot kg^{-1}$. Overall, in comparison with FR, BR had increased torque and power demands on the knee extensors and reduced demands on the ankle plantarflexors.

Previous BR studies have reported increased knee extensor torque production (14) and increased knee power and benefits to knee ligamentous instability problems (9) after extended BR training. Also, elite track athletes have used BR to strengthen the quadriceps muscles (6). The observed knee moment patterns support these findings and practices by identifying the high knee extensor moment necessary for BR. Surprisingly, the mean BR peak knee extensor moment (256 $N \cdot m$) was greater than the peak knee extensor moment observed in the squat weight lifting exercise (242 $N \cdot m$) performed by skilled lifters at 90% of their 1 repetition maximum (8).

Winter et al. (18) reported that BW was accomplished by a simple temporal reversal of FW, resulting in interchanged energy generation and absorption patterns of the lower extremity muscle groups throughout the stride. Bates et al. (2) hypothesized this process for BR. The present results support this hypothesis during the swing phase. The joint moment curves identified both flexor and extensor dominant phases during swing at the hip and knee joints for both gait patterns. The hip and knee power curves, however, showed interchanged functional demands between BR and FR, indicating that the net hip torques worked only eccentrically in BR and only concentrically in FR while the knee had only concentric and eccentric actions in BR and FR, respectively.

The hypothesis of interchanged energy generation and absorption profiles was not supported during the stance phase, especially at the knee and hip. BR had mostly a net extensor knee torque, which contracted isometrically and then concentrically, while FR had concentric flexor, eccentric extensor, concentric extensor, and eccentric flexor work phases. Hip joint muscle function was nearly entirely concentric during BR stance but had both eccentric and concentric phases during FR stance.

The conclusion of a simple muscle function reversal in walking (18) was supported by opposite power curves throughout the stride and also by similar moment and power magnitudes at each joint. This conclusion was further supported by EMG results from the two gaits. BR and FR had similar moment and power magnitudes during swing but not during stance at the knee and ankle. Generally, the functional roles of these two joints were interchanged, with the knee generating more torque and power in BR and the ankle doing so in FR. The differences in the moment and power curves between BR and FR, which produced kinematic patterns that were not mirror images, suggested that BR and FR were not controlled by a reversed muscle activation pattern, as were BW and FW (18). Further investigations into this question utilizing EMG analyses are necessary, however before a definitive conclusion can be derived.

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