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David J. Sanderson [a] & Alec Black [b]
[a] Biomechanics Laboratory, School of Human Kinetics , University of British Columbia
[b] Shriners Gait Lab , Sunny Hill Health Centre for Children , Vancouver, BC, Canada
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The effect of prolonged cycling on pedal forces

DAVID J. SANDERSON[1]* and ALEC BLACK[2]

[1]*Biomechanics Laboratory, School of Human Kinetics, University of British Columbia and* [2]*Shriners Gait Lab, Sunny Hill Health Centre for Children, Vancouver BC, Canada*

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The aim of this study was to determine whether cyclists modify the pattern of force application to become more effective during a prolonged ride to exhaustion. Twelve competitive male cyclists completed a steady-rate exercise ride to exhaustion at 80% of their maximum power output at 90 rev · min^{-1} on a cycle ergometer. Pedal force, pedal and crank angle data were collected from an instrumented bicycle for three pedalling cycles at the end of the first and final minutes of the exercise test with simultaneous video recording of the lower limbs. Kinematic and force data were combined to compute hip, knee and ankle joint moments. There were changes in the pattern of force application, joint kinematics and joint moments of force. Comparison of the first minute and the final minute ride revealed significantly increased peak effective force (340 $\pm$ 65.0 and 377 $\pm$ 74.8 N for the first and final minute, respectively; $F_{1,11} = 7.44$, $P = 0.02$), increased positive (28.4 $\pm$ 4.5 and 30.5 $\pm$ 4.8 N · s for the first and final minute, respectively; $F_{1,11} = 7.80$, $P = 0.02$) and negative angular impulses (–1.5 $\pm$ 1.6 and –2.4 $\pm$ 1.5 N · s for the first and final minute, respectively; $F_{1,11} = 4.50$, $P = 0.06$). Contrary to our initial assumptions, it would appear that riders became less effective during the recovery phase, which increased the demand for forces during the propulsive phase. Training the pattern of force application to improve effectiveness may be a useful strategy to prolong an endurance ride.

Keywords: cycling, duration, force, muscle, training.

Introduction

Pedalling mechanics have been well documented (Redfield and Hull, 1984; Sanderson and Cavanagh, 1985; Patterson and Moreno, 1990). Many studies have investigated changes in the mechanics of pedalling when the system is perturbed, including changes in saddle height (Nordeen-Snyder, 1977), power output (Sanderson, 1990) and cadence (Patterson and Moreno, 1990; Sanderson, 1990; Redfield and Hull, 1994).

Few biomechanical studies have probed the changes that occur as a result of whole-body fatigue. In running, kinematic changes that result from fatigue have been investigated by Bates and Osternig (1977), Elliott and Roberts (1980), Chapman (1981, 1982), Elliott and Ackland (1981) and Siler and Martin (1991). Bates and Osternig (1977) and Chapman (1982) measured kinematic changes that result from whole-body fatigue during running. They found measurable changes in the patterns of motion that the athletes exhibited during separate running protocols leading to fatigue. With the exception of Elliott and Roberts (1980), velocity was measured in each of the above studies but there was no control or maintenance of constant over-ground velocity. Therefore, as the athlete exercised, the kinematics that changed were a result not only of fatigue, but of reductions in over-ground velocity as well. Siler and Martin (1991) removed the confounding variable of decreased over-ground speed by exercising the athletes on a treadmill at constant velocity. By maintaining the velocity of the run, Siler and Martin were able to quantify the kinematic changes in running that accompanied the fatigue process.

Sprague and Mann (1983) described the kinetic changes that accompany the fatigue process in running. They compared joint moments during foot contact with the ground at the beginning of a 400 m race to the same joint moments at the end of the race. Although differences were found, these differences may have been a result of changes in over-ground velocities, which decreased from 9.51 m · s^{-1} at the start to 7.53 m · s^{-1} at the end, and not a reflection of the fatigue process during this exercise, as was the case in many of the kinematic studies.

* Address all correspondence to David Sanderson, Biomechanics Laboratory, School of Human Kinetics, University of British Columbia, 210 War Memorial Gym, 6081 University Boulevard, Vancouver, British Columbia V6T 1Z1, Canada.
e-mail: david.sanderson@ubc.ca

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The present study used cycling to minimize the confounding issues introduced above. The operational definition of fatigue was the voluntary cessation of exercise after a steady-rate period of cycling. The riders were encouraged to ride as long as possible and to maintain the riding cadence within 5 rev · min^{-1} of the target of 90 rev · min^{-1}. Although it was likely that the duration of the rides would vary among individuals, we anticipated that the dependent variables would change in a similar fashion. The aims of this study were to measure and describe the changes in biomechanical measures, specifically changes in joint kinematics, pedal force kinetics and joint moments, during this ride and to examine whether cyclists modify the application of pedal force such that they become more effectively oriented.

Methods

Participants

The study group consisted of 12 male cyclists selected on the basis of competitive standard and training background using the criteria of the current Canadian Cycling Association (CCA) road racing license of category three or higher (1 or 2) or CCA mountain bike racing license of Sportsman, Expert or Elite. We felt that athletes with these credentials would be able to complete the test protocol, would be familiar with riding at or above their anaerobic threshold and would be sufficiently well trained. All participants were fully informed of the experimental details before the testing began. They were told that they were able to withdraw from the test procedure at any time without prejudice and they provided signed informed consent, which conformed with the ethical guidelines of the university. The characteristics of the participants are summarized in Table 1.

Instrumentation

The study used a standard bicycle instrumented with two piezo-electric tri-axial force transducers (Kistler 9251A) in the right pedal (Sanderson *et al.*, 2000). This pedal has a force resolution of 0.01 N over the load range used in this study. A continuous output potentiometer monitored pedal angle and an optical encoder monitored data collection and top dead centre. The bicycle was mounted on a Schwinn Velodyne electronically braked cycle ergometer. The Schwinn Velodyne simulates inertial characteristics of road riding and modulates power outputs based on cadence. The bicycle was set up so that it would match, as closely as possible, the athlete's own bicycle with one exception: the bicycle seat height was set to 100% trochanteric height. This height was found to be optimal in terms of oxygen consumption by Nordeen-Snyder (1977). Sanderson (1990) has shown that asymmetry in force application does not vary significantly with variations in cadence or power output. Consequently, data from one side only were recorded.

Force data were recorded continuously at a rate of 240 Hz. Data collection was triggered by the optical encoder indicating top dead centre. Force data were collected at the end of each minute of the test protocol

Table 1. Characteristics of the participants

Participant	Category	Age (years)	Mass (kg)	Height (m)	Maximal power output (W)	$\dot{V}O_{2max}$ (ml · kg^{-1} · min^{-1})	Duration at 80% (min)	%HR$_{max}$ at end
1	CCA Mtn Expert	26	81.7	1.84	425	67.6	16.5	96
2	CCA Mtn Elite	22	74.7	1.78	475	65.5	8.0	95
3	CCA Mtn Sport	25	75.6	1.87	425	63.3	15.0	96
4	CCA Veteran B	46	88.5	1.81	375	50.3	14.5	106
5	CCA level 1	24	54.4	1.64	325	69.5	8.0	92
6	CCA level 2	23	73.5	1.78	450	68.7	7.5	95
7	CCA level 1	27	88.5	1.89	525	68.1	13.0	94
8	CCA Mtn Expert	23	74.3	1.81	375	67.4	18.0	89
9	CCA level 2	30	87.6	1.87	400	66.1	15.0	90
10	CCA level 3	28	78.5	1.80	350	63.7	22.0	90
11	CCA level 1	32	63.4	1.71	425	79.5	4.0	93
12	CCA level 1	27	60.0	1.72	400	70.7	15.0	95
Mean		27.8	75.1	1.79	413	66.7	13.0	94
s		6.5	11.2	0.08	54.9	6.7	5.2	5

Abbreviations: $\dot{V}O_{2max}$ = maximal oxygen uptake, HR$_{max}$ = maximal heart rate, CCA = Canadian Cycling Association.

and lasted for three revolutions of the crank per collection period. The outputs from the two force transducers were summed before amplification, translated from analog to digital (12-bit) and then stored for subsequent processing.

Protocol

The participants visited the laboratory three times. At the first visit, each cyclist completed a 5 min warm-up at the cadence and power output of his choice and then proceeded to perform a test to maximal oxygen uptake. The test began at a power output setting of zero and a cadence of 90 rev $\cdot$ min^{-1}, giving an effective power output of approximately 60 W. After the first minute, the power output of the cycle ergometer was increased to 100 W. At the end of each subsequent minute, the power output was increased by 25 W until such time as the athlete was unable to maintain the test cadence (90 ± 5 rev $\cdot$ min^{-1}). Metabolic data were collected at 15 s intervals using a Rayfield open-circuit system. This system consists of the Rayfield computational software, a Rayfield gasmeter, which measures ventilation and volume, an Amtek oxygen analyser and a Beckman carbon dioxide analyser. The total system measures oxygen, carbon dioxide, ventilation, respiratory rate, tidal volume and heart rate and plots these data at 15 s intervals. In the initial test, the maximum power output of each cyclist was determined. This determination of the maximal power output was then used to calculate the appropriate power output of the cycle ergometer for the second visit.

At the second visit, the athlete completed a steady-rate ride at 80% of his maximum power output achieved in the initial test. This ride began with the cyclist choosing a power output and cadence of his choice for a 5 min warm-up. At the end of the warm-up, the power output setting on the cycle erogometer was set to correspond to the 80% value of the initial visit. The test continued until the cyclist was no longer able to maintain the test cadence of 90 ± 5 rev $\cdot$ min^{-1}, indicated by cadence outside the boundary for more than 10 s. The inability to continue to ride at the target cadence and power output was defined as fatigue. Video and force data were collected continuously at the end of the first minute and the final minute of the test. Heart rate data were collected continuously using a Sport Tester PE 3000 portable heart rate monitor (Polar Electro Inc., Finland). The final heart rate was computed as the average heart rate over the final minute during which force data were acquired.

There was a possibility that having the participants ride for a particular time might have resulted in changes in the mechanical variables, simply as a consequence of a long ride. To account for this, the cyclists were requested to perform a third ride at a substantially reduced power output, thus limiting the chance of any fatigue. We hypothesized that there would be no change in any kinetic or kinematic variable, which would allow one to consider changes at the 80% condition to be the consequence of the exertion associated with that intensity. At the third visit, each cyclist completed a steady-rate ride at 30% of his maximum power output in the initial test. This ride began with the rider choosing a power output and cadence of his choice for a 5 min warm-up. At the end of the warm-up, the power output setting on the cycle ergometer was set to correspond to 30% of the athlete's maximum power output determined at the initial visit. The test continued for as long as the 80% ride was maintained. Video and force data were collected at the end of each minute as described above.

Data analysis

Kinetic data

Normal and tangential pedal force data were collected for three pedalling cycles during the steady-rate exercise test at the end of each minute. From these data, mean resultant force, the effective components of the resultant force and the overall index of effectiveness for the entire pedal cycle were calculated, in addition to crank torque, total work, positive work, negative work, power, angular impulse, positive angular impulse, negative angular impulse and the positions of where the negative angular impulse began and ended within the pedal cycle. Effective force is that component of the resultant force perpendicular to the crank. The index of effectiveness was determined as the ratio of the linear impulse of the effective component of force to the linear integral of the resultant pedal force (Lafortune and Cavanagh, 1983). These mean kinetic data were averaged for the group to allow comparison of the initial and final minutes of the tests.

Kinematic data

The kinematics of the limb segments were recorded at 60 Hz using a Panasonic video camera (WDV 5100), with the lens axis oriented perpendicular to the sagittal plane of the rider. The pattern of limb motion was defined by highly reflective markers placed over the front and centre of the toe clip, lateral aspect of the fifth metatarsal head, posterior lateral aspect of the calcaneus, lateral malleolus and the greater trochanter of the femur. Three cycles from each minute of the test were digitized using the Peak Performance Technologies (version 5.4) software package and the two-dimensional coordinates for the limb segment markers were calculated as well as segment angles with a resolution of 5 mm in the sagittal plane. These coordinate data were then time-matched to the corresponding force data file for subsequent processing. Time matching was achieved using a square-wave pulse from the bicycle crank, generated at top dead centre, and recorded as one channel of the analog-to-digital converter. This pulse also generated a white square on the video signal using an event synchronization unit. Using an in-house computer routine, these data were adjusted in time using linear interpolation so that the arrays became the same length, from one top dead centre to the next.

Calculations of joint moments were done using conventional inverse dynamics as described previously (Redfield and Hull, 1984; Gregor *et al.*, 1985; Hull and Jorge, 1985; Ericson, 1986). Joint moments for three cycles of data were averaged for each participant. A group average was used in the calculations of the total average ankle, knee, hip and propulsive moments.

Statistical analysis

Data from the 30% and 80% conditions were analysed separately using a one-way analysis of variance (ANOVA) with time (initial test, final test) as a repeated measure. Significance was set at $P<0.05$. Two cyclists exhibited ride duration times that were 1.7 standard deviations from the mean. Because of this heterogeneous response, effect size determinations were computed based on Cohen (1962), Potvin and Schutz (2000) and Park and Schutz (1999). The effect size varied from zero for maximum thigh angle to 1 for minimum thigh angle. For the variables for which the results were significant, the effect size was 0.5 or greater. It appeared that the range of responses did not have an adverse effect on the results.

Results

We included the 30% condition to ensure that there was no contamination of data from the 80% ride that could be attributed to riding itself for a similar duration but at a very low intensity. A one-way ANOVA of the kinematic and kinetic data, with time (initial and final) as the repeated measure, showed no difference in any selected variables for the 30% riding condition. Subsequent analysis focused, therefore, on the 80% condition and a one-way ANOVA, with time as the repeated measure, was used for all variables.

As the aim of the study was to have the riders perform to a level beyond which they felt they could no longer ride, heart rate was used as a verification that they were performing close to or at their maximum. The data in Table 1 indicated that their performances, as reflected by the heart rate in the final minute, were very good, including that of participant 11 who had only a very short ride. On average, their heart rate was 94% of maximum.

Changes in the kinematics and kinetics of riding between the initial and final minute of riding at 80% of maximum power output are summarized in Table 2. At the end of the exercise test, there was a significant increase in the maximum normal force (354 ± 65.0 and 391 ± 74.8 N for the first and final minute, respectively; $F_{1,11}=7.44$, $P=0.02$) but no difference in the maximum or minimum tangential force (85.6 ± 18.3 and 85.3 ± 19.4 N for the first and final minute, respectively; $F_{1,11}=0.003$, $P=0.96$). However, there was a change in the shape of the tangential force profile such that more time was spent pulling back on the pedal (Fig. 1a,b). This was a consequence of an earlier transition from pushing forward to pulling rearward, from 133° to 121° of crank position, and a delayed return to forward pushing, from about 300° to about 330° of crank rotation. This difference, however, was not significant.

Mean effective force, about the crank axis, for the 80% condition is plotted as a function of crank angle for the first and final minutes of the steady-rate exercise test in Fig. 1c. There was a significant increase in peak propulsive torque (57.8 ± 11.0 and 64.1 ± 12.7 N·m for the first and final minute, respectively; $F_{1,11}=7.44$, $P=0.02$) and in peak negative or retarding torque (-5.4 ± 5.1 and -8.7 ± 4.77 N·m for the first and final minute, respectively; $F_{1,11}=8.13$, $P=0.02$). Net angular impulse, computed over the complete pedalling cycle, was not significantly different between the first and the final minute (26.9 ± 4.5 and 28.2 ± 4.1 N · s for the first and final minute, respectively; $F_{1,11}=2.55$, $P=0.139$). Angular impulse revealed that the positive angular impulse was significantly greater ($F_{1,11}=7.80$, $P=0.018$) in the final than in the first minute and that the negative angular impulse was smaller in the first than in the final minute, although not significantly so ($F_{1,11}=4.50$, $P=0.058$). Similarly, the overall index of effectiveness was lower in the final minute (Fig. 1d) but primarily in the second 180° of crank rotation.

The angular kinematics of the thigh, shank and foot segments and the joint-angle conventions are plotted in Fig. 2 and the peak values displayed in Table 2. Thigh segment angle became significantly more vertically oriented at bottom dead centre in the final minute ($F_{1,11}=12.2$, $P=0.005$) and although there was a shift to a more vertically oriented shank, the shift was not significant ($F_{1,11}=0.666$, $P=0.432$). Foot segment angle became more horizontal between 90° and 180° of crank rotation. This change indicated that the ankle became more dorsi-flexed during this phase but the change was not significant (Table 2).

Table 2. *F*-score (degrees of freedom = 1,11) and associated *P*-value for single-factor repeated measures on timing design, for selected variables averaged across individuals and trials (mean ± *s*)

Variable	First minute	Final minute	*F*-score	*P*-value
Maximum Fe (N)	340 ± 65.0	377 ± 74.8	7.441	0.0197
Crank angle at maximum Fe (°)	103 ± 6.9	118 ± 47.5	1.119	0.3128
Minimum Fe (N)	–31.8 ± 30.4	–51.5 ± 28.1	8.124	0.0158
Crank angle at minimum Fe (°)	277 ± 18.4	281 ± 26.6	0.736	0.4092
Index of effectiveness	0.67 ± 0.11	0.63 ± 0.09	2.580	0.1365
Positive work (J)	119 ± 18.9	128 ± 20.1	7.786	0.176
Negative work (J)	–6.2 ± 6.8	–10.0 ± 6.3	4.488	0.0577
Positive angular impulse (N · s)	28.4 ± 4.5	30.5 ± 4.8	7.796	0.0175
Negative angular impulse (N · s)	–1.5 ± 1.6	–2.4 ± 1.5	4.499	0.0575
Segment kinematics				
Maximum thigh segment angle (°)	66.1 ± 3.0	66.2 ± 2.8	0.067	0.801
Maximum shank segment angle (°)	–3.8 ± 3.0	–4.4 ± 3.0	0.666	0.432
Maximum foot segment angle (°)	92.6 ± 3.6	94.1 ± 4.6	1.245	0.228
Minimum thigh segment angle (°)	24.5 ± 1.8	22.2 ± 2.6	12.203	0.005
Minimum shank segment angle (°)	–46.8 ± 3.1	–48.1 ± 4.2	6.505	0.027
Minimum foot segment angle (°)	41.5 ± 4.8	41.9 ± 6.7	0.074	0.791
Joint moments				
Maximum ankle dorsi-flexion (N · m)	2.5 ± 2.9	1.4 ± 3.0	1.179	0.3008
Maximum ankle plantar flexion (N · m)	41.6 ± 8.2	48.7 ± 10.3	14.626	0.0028
Maximum knee extensor (N · m)	31.8 ± 9.8	30.9 ± 4.6	0.105	0.7525
Maximum knee flexor (N · m)	52.8 ± 11.8	59.2 ± 11.5	5.098	0.0453
Maximum hip flexor (N · m)	18.7 ± 3.2	17.5 ± 3.7	1.991	0.1859
Maximum hip extensor (N · m)	80.0 ± 20.7	94.3 ± 17.3	7.634	0.0185

Abbreviation: Fe = effective component of resultant force.

Hip, knee and ankle joint moment of force data are presented in Fig. 3 and the peak values displayed in Table 2. They indicate that there was a general increase in moments about each of the three joints, primarily in the first 180° of pedalling, between the first and final minutes of the ride. The maximum ankle plantar flexor moment was significantly greater between the first and final minutes, as was the maximum hip extensor moment. However, there were no significant differences at the knee joint. In each case, these maximum moments occurred in the propulsive phase of the pedal cycle.

Discussion

We hypothesized that as the athlete fatigued, the force applied to the crank would be more effectively applied to maintain the test cadence. We believed that a more effectively applied force would reduce the resultant force applied to the pedals and, therefore, muscle stress. However, our results indicate that although there was a change in the effective force, it was in the opposite direction to that hypothesized. The peak positive effective force increased significantly rather than decreased. However, the peak negative effective component also changed by becoming significantly more negative. Furthermore, although the net angular impulse, computed over the complete pedalling cycle, was not significantly different between the first and the final minute, there was a significant increase in the positive angular impulse from the first to the final minute. There was also an increase in the negative angular impulse from the first to the final minute. We were unable to identify which came first – increased positive or increased negative impulse. However, it is likely that as the athlete fatigued and, one might argue, style became compromised, the recovery phase degraded more quickly. The increased negative angular impulse would necessitate an increased positive angular impulse for the rider to maintain the overall power output. This would explain the need for increased peak positive effective force to overcome the more negative component. Thus it would appear that as the riders fatigued, they became less effective during the recovery phase and needed to increase the force application during the propulsive phase to account for that. Clearly, this warrants further investigation, perhaps by training participants during

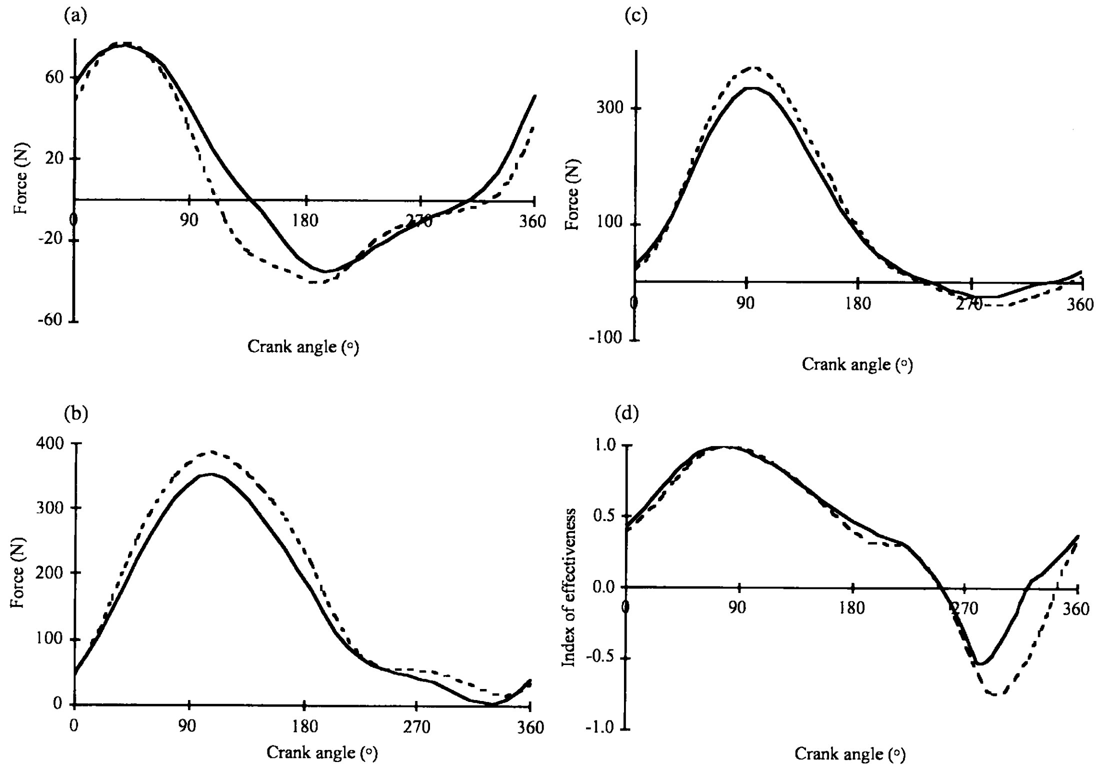

Fig. 1. Mean data for the first (solid line) and final minute (broken line) for (a) tangential pedal force, (b) normal pedal force, (c) effective force and (d) the index of effectiveness.

particular phases of the pedalling cycle, as done by Sanderson and Cavanagh (1989).

The segmental kinematic data showed that the thigh was significantly more vertically oriented at the end of the propulsion stroke (near 180° of crank rotation), there was little change in the shank segment and a more horizontally oriented foot segment. That is, the hip angle became more extended as the foot angle became more dorsi-flexed. Both Black *et al.* (1993) and Kautz *et al.* (1991) have reported a sensitivity in the joint kinematics to increased power output. Their riders were grouped into two separate categories based on the changes in pedal angle as power output was increased. One of the two groups showed no change in the pedal angle as the power output was increased and the other group, called the 'ankling group', showed a more dorsi-flexed pedal style throughout one pedal cycle.

It was suggested here that these kinematic changes were linked to the change in the pattern of force application used by the athlete to maintain the given standard of performance. At maximum dorsi-flexion, there was an anatomical barrier or limitation to motion of the ankle joint. This limitation to further movement could enhance the plantar flexor muscle groups of the lower limb. Instead of combined ankle support and ankle plantar flexion action, the muscles could focus solely on plantar flexion. To maintain a high plantar flexor moment, the ankle dorsi-flexed to its maximum position, at which point there was a mechanical barrier to motion, thus stabilizing the ankle joint and allowing the transmission of force from the larger muscle further up the leg through the lower limb and into the pedal. These results support those of both Black *et al.* (1993) and Amoroso *et al.* (1993), who found that, as an athlete fatigues, the plantar flexors fatigue first, perhaps because they are a relatively small muscle group compared with the quadriceps group. Range of motion data were not reported for these riders; however, normative data from the American Orthopedic Association indicated a maximum ankle dorsi-flexed angle of 18°. The maximum dorsi-flexed ankle angle in the present study was about 16°, indicating that the ankle might have been near its end of range of motion, which indicates some support for the hypothesis. Clearly, this needs further investigation.

Tangential force (Fig. 1b) indicated that there was an earlier shift to pulling back across the pedal in the final minute than in the first minute of exercise. It is reasonable to conclude that this change in force pattern was linked to changes in the joint angle at

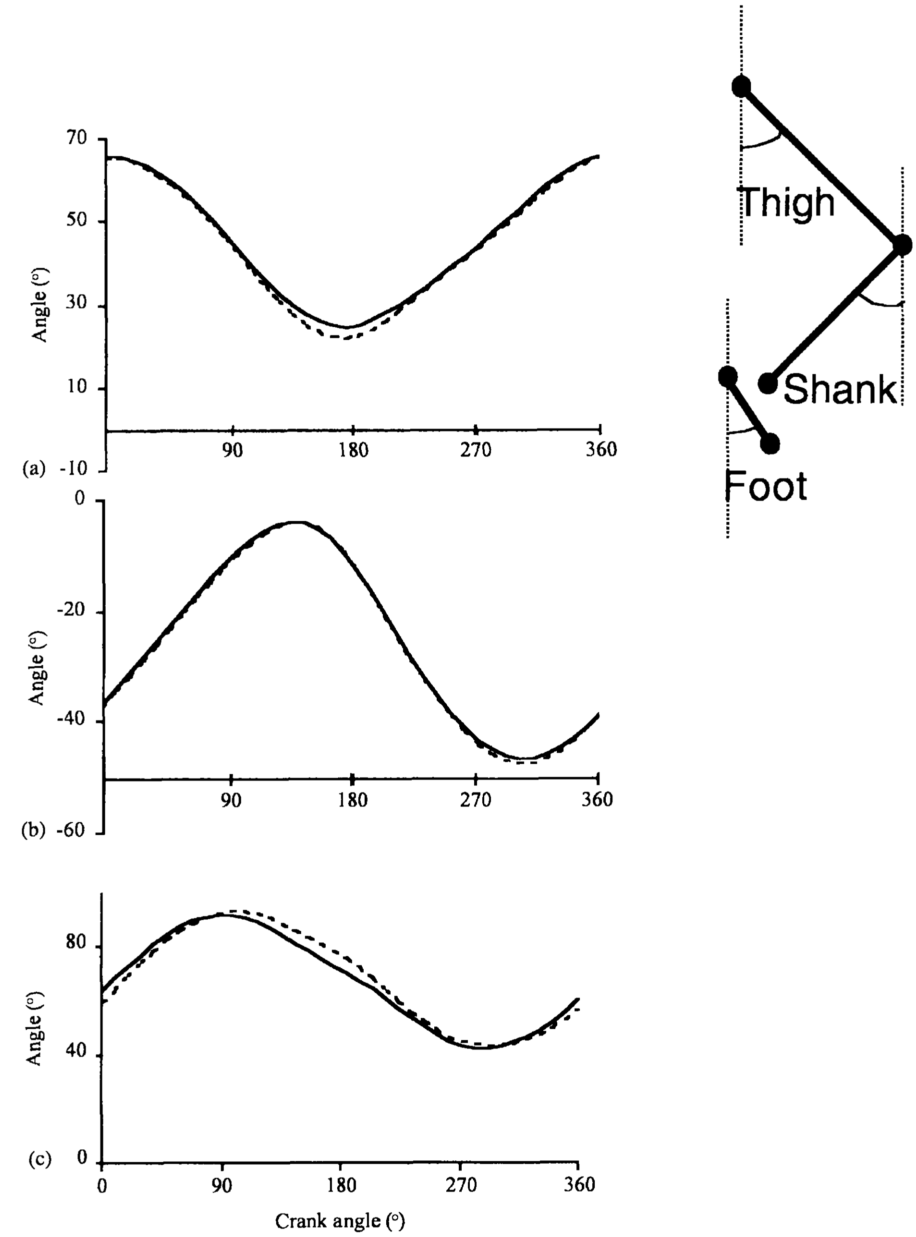

Fig. 2. Mean segment kinematic data for the first (solid line) and final minute (broken line) for the (a) thigh, (b) shank and (c) foot segments.

the ankle joint, because if the ankle was locked at a maximum dorsi-flexion position once the crank passed 90°, the direction of applied force to the pedal would be rearward. This orientation would increase the effectiveness of the applied force and thus contribute to the production of power. There was some support for this when the index of effectiveness was examined (Fig. 1d). However, there was substantial decreased effectiveness during the final 100° of crank rotation that resulted in an overall decrease in effectiveness from 0.67 to 0.63. This appears to have been the consequence of reduced effective force application during the recovery phase.

There was a general increase in the moments of force about each of the three joints that occurred primarily in the first 180° of crank rotation. These changes were probably the result of the increased need for propulsive force during this phase and possibly the result of the larger muscle groups of the thigh taking a more active role in propulsion and the smaller plantar flexor groups of the lower leg taking a smaller role as a result of fatigue. The ankle plantar flexor moment was increased

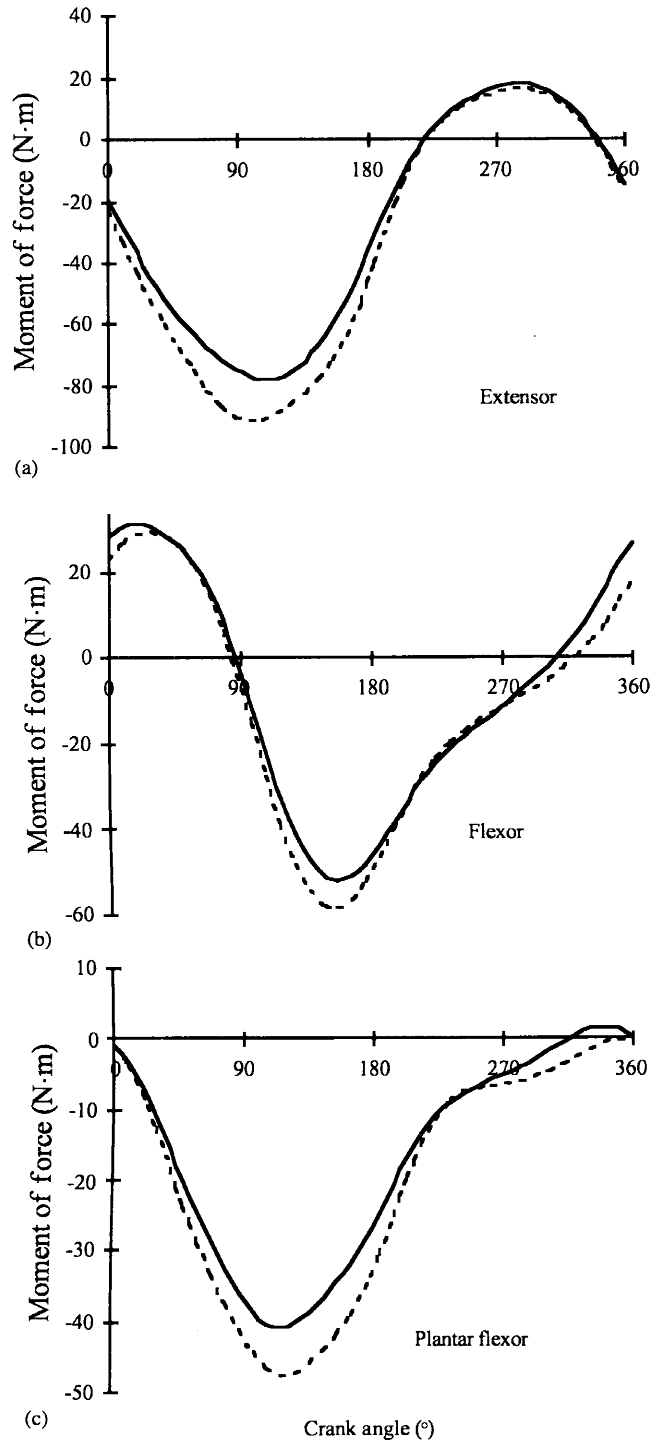

Fig. 3. Mean joint moment data for the first (solid line) and final minute (broken line) for the (a) hip joint, (b) knee joint and (c) ankle joint.

because of the limited range of motion in the ankle joint in the dorsi-flexed position.

Our results suggest that as a cyclist fatigues, the retarding forces generated in the recovery phase result in increased force production during the propulsion phase if power output is to remain constant. This suggests that increased endurance may be achieved if training focuses on maintaining an effective recovery force profile. Sanderson and Cavanagh (1989) showed that the pattern of force production during the recovery phase can be modified with appropriate feedback. Focusing on this segment of the pedalling cycle may well offer improvement in performance. Although recovery forces are small, they can nonetheless contribute in an important way as the exercise progresses. To ensure that the rider can perform to maximum, improving the effectiveness during the recovery phase may be an important factor worthy of more detailed examination.

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