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CRANK INERTIAL LOAD HAS LITTLE EFFECT ON STEADY-STATE PEDALING COORDINATION

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Abstract—Inertial load can affect the control of a dynamic system whenever parts of the system are accelerated or decelerated. During steady-state pedaling, because within-cycle variations in crank angular acceleration still exist, the amount of crank inertia present (which varies widely with road-riding gear ratio) may affect the within-cycle coordination of muscles. However, the effect of inertial load on steady-state pedaling coordination is almost always assumed to be negligible, since the net mechanical energy per cycle developed by muscles only depends on the constant cadence and workload. This study tests the hypothesis that under steady-state conditions, the net joint torques produced by muscles at the hip, knee, and ankle are unaffected by crank inertial load. To perform the investigation, we constructed a pedaling apparatus which could emulate the low inertial load of a standard ergometer or the high inertial load of a road bicycle in high gear. Crank angle and bilateral pedal force and angle data were collected from ten subjects instructed to pedal steadily (i.e. constant speed across cycles) and smoothly (i.e. constant speed within a cycle) against both inertias at a constant workload. Virtually no statistically significant changes were found in the net hip and knee muscle joint torques calculated from an inverse dynamics analysis. Though the net ankle muscle joint torque, as well as the one- and two-legged crank torque, showed statistically significant increases at the higher inertia, the changes were small. In contrast, large statistically significant reductions were found in crank kinematic variability both within a cycle and between cycles (i.e. cadence), primarily because a larger inertial load means a slower crank dynamic response. Nonetheless, the reduction in cadence variability was somewhat attenuated by a large statistically significant increase in one-legged crank torque variability. We suggest, therefore, that muscle coordination during steady-state pedaling is largely unaffected, though less well regulated, when crank inertial load is increased. Published by Elsevier Science Ltd.

Keywords: Pedaling; Muscle coordination; Crank inertial load.

INTRODUCTION

An understanding of how coordination of leg muscles is affected by environmental interactions, or equivalently, by the effective load encountered by each leg, is needed to elucidate neuromotor strategies (Zajac, 1993). For both a stationary ergometer and an actual road bicycle, the effective load appearing at the crank consists primarily of an inertial and a frictional component (e.g. Fregly, 1993; Gregor *et al.*, 1985; Kautz *et al.*, 1991; Löllgen *et al.*, 1975; Patterson *et al.*, 1983). The frictional load affects muscle coordination (see below) because friction influences the workload (i.e. for a given cadence, the higher the frictional load, the higher the workload, which determines how much *net* mechanical energy the muscles must produce per crank cycle). While the inertial load does not affect workload, it may still affect muscle coordination whenever the crank is accelerated (or decelerated), such as during initiation of pedaling. For steady-state pedaling, when cadence and workload are approximately constant, muscle coordination is almost always assumed to be unaffected by the crank inertial load. Nonetheless, because within-cycle variations in crank angular acceleration exist even under steady-state conditions, crank inertial load may affect muscle coordination in this situation as well (Gregor *et al.*, 1991).

Muscular output and the tangential crank force (i.e. the component of the pedal reaction force which accelerates the crank) have been found to increase with frictional load during stationary ergometer pedaling (crank inertial loads typically $< 6.5\ \mathrm{kg\,m^2}$) (Davis and Hull, 1981; Ericson, 1986; Ericson *et al.*, 1986b; Houtz and Fischer, 1959; Hull and Jorge, 1985; Jorge and Hull, 1986; Kautz *et al.*, 1991). These increases correspond to increases in the peak net muscle joint torques at the hip, knee, and ankle during the downstroke (Ericson, 1986; Ericson *et al.*, 1986b). Furthermore, such changes in net muscle joint torque have been found, in general, to have a positive correlation with changes in electromyographic (EMG) activity (Duchateau *et al.*, 1986; Ericson, 1986; Houtz and Fischer, 1959). For example, increased gluteus maximus EMG correlates well with increased net hip extensor joint torque, increased vastus medialis activity with increased net knee extensor joint torque, and increased soleus activity with increased net ankle plantarflexor joint torque. Similar results have been observed for constant cadence pedaling on roller cylinders (crank inertial loads typically between

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4 and 11 kg m^2), where the frictional load (and inertial load) is altered by changing the bicycle gear ratio (Davis and Hull, 1981; Hull and Jorge, 1985; Jorge and Hull, 1986).

In contrast, pedaling data, though scant, suggest that muscular output may hardly be affected by inertial load. Neither oxygen consumption nor the ratio of tangential crank force to total crank force developed at the pedal spindle has been found to change with inertial load during stationary ergometer pedaling (Patterson *et al.*, 1983; crank inertial loads of 1.4 and 22.7 kg m^2). Furthermore, the perceived exertion of subjects has been found to be essentially unaffected by using a higher mass flywheel (Löllgen *et al.*, 1975; crank inertial loads of 3.5 and 7.7 kg m^2 assuming a standard Monark ergometer). Nevertheless, at lower cadences (e.g. near 40 rpm), minor changes in total crank force have been observed with an increase in ergometer inertial load (Patterson *et al.*, 1983). Although these studies did not show directly how inertial load affects net muscle joint torques and EMG activity, they suggest that only small changes, if any, would occur. On the other hand, they utilized rather low inertial loads (c.f. > 100 kg m^2 for high gear ratios during bicycling; Fregly, 1993).

When pedaling at a constant cadence and against a constant workload, it is clear that crank inertial load does not affect the average mechanical energy dissipated in each cycle. For simplicity of control, the nervous system might, therefore, use identical muscle coordination within the crank cycle to pedal steadily at any given cadence and workload, regardless of the crank inertial load. If this occurred, then since the net joint torques produced by the muscles dominate the contributions to the one-legged crank torque (Fregly and Zajac, 1996), the summed crank torque produced by both legs should be largely unaffected by inertial load. As a result, the peak-to-peak variation in crank angular acceleration within the cycle should be lower with a higher inertia. Also, because a higher inertia implies that the crank kinematics will be less sensitive to slight variations in the crank torque about its mean (i.e. nominal) trajectory, the variation in cadence should be lower as well. For pedaling at a constant cadence and against a constant workload, we therefore hypothesized that the net joint torques produced by muscles at the hip, knee, and ankle would be unaffected by crank inertial load, and furthermore, that crank kinematics would vary less at a higher inertia. Clearly, if the net joint torques produced by muscles are found to be unaffected by inertial load, little impetus will exist for studying coordination of individual muscles when pedaling against different inertias, and coordination principles which apply to low inertia pedaling (e.g. a laboratory ergometer) could then be inferred to apply to high inertia pedaling as well (e.g. a road bicycle outdoors).

METHODS

A Monark ergometer was modified such that subjects could pedal against one of two effective crank inertial loads (6.5 or 130 kg m^2). The low inertial load was similar to that of a Monark 868 ergometer in its standard 52 × 14 gear ratio; the high inertial load to that which a 50th percentile U.S. male riding a 12-speed road bicycle in a 52 × 17 gear ratio would encounter (Fregly, 1993). One-stage and three-stage gearing were used to achieve the two inertial loads (Fig. 1). The effective crank rotational stiffness in both inertial load configurations was adjusted to be 3250 N m rad^{-1}, about that of a standard Monark ergometer or road bicycle in high gear (Fregly, 1993), by using bicycle chains possessing different compliances. Crank rotational stiffness was determined by taking the ratio of the incremental increase in crank torque applied via one pedal to the corresponding incremental increase in crank rotation (see below for measurement details) with the ergometer flywheel locked in place, thereby accounting for all torsional, bending, and other compliance in the drive system. By using a seat tube angle of 73°, a crank arm length of 0.170 m, and a standard road bicycle seat and handlebars, the apparatus emulated the geometry and rider position of a 12-speed racing bicycle.

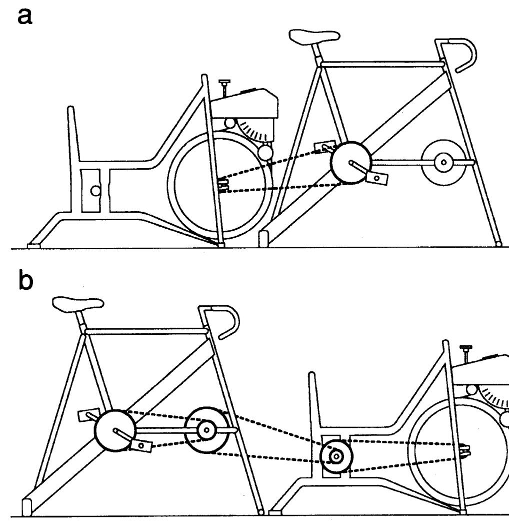

Fig. 1. Configuration of the ergometer to achieve a low or high effective crank inertial load. (a) Low inertia configuration with one-stage gearing (overall gear ratio of 3.7:1) to achieve an effective crank inertial load of 6.5 kg m^2, similar to pedaling on a standard Monark 868 ergometer. (b) High inertia configuration with three-stage gearing (overall gear ratio of 16.4:1) to achieve an effective crank inertial load of 130 kg m^2, similar to a 50th percentile U.S. male pedaling on a 12-speed road bicycle in a 52 × 17 gear ratio. Because a higher gear ratio means increased workload sensitivity to errors in the flywheel belt friction setting, a rotary potentiometer was geared to the front Monark pendulum to improve the precision of friction measurements with the high inertia. The same effective crank rotational stiffness was achieved in both configurations by using bicycle chains possessing different compliances (see text).

Ten male recreational cyclists (age 27.5 ± 1.8 yr, height 1.80 ± 0.03 m, and weight 738 ± 67 N), all in good physical condition with no history of knee or other lower limb injuries, gave informed consent prior to participation. The protocol for the experiments was approved according to the relevant laws and regulations of Stanford University. The seat height was initially set to 100% of trochanteric length (Nordeen-Snyder, 1977) and then adjusted slightly so that the minimum knee flexion angle near bottom dead center was about 35°. This slight adjustment was performed to improve the accuracy of a subsequent inverse dynamics analysis (see below). Prior to data collection, each subject warmed up and became familiar with the apparatus by pedaling for approximately 5 min against both inertial loads but at a lower cadence and workrate (60 rpm and 130 W). During data collection, subjects wore cleated cycling shoes and toe clips and were instructed to pedal steadily (i.e. constant speed across cycles) and smoothly (i.e. constant speed within a cycle) at a 75 rpm cadence using a 225 W workrate, with the actual cadence being displayed by a cycle computer on the handlebars (Hull and Jorge, 1985). This cadence/workrate combination was selected to emulate road riding in a 52×17 gear ratio, where frictional load is proportional to the square of the cadence, primarily due to wind resistance. A higher cadence was not used since the corresponding road-riding workrate, which is proportional to the cube of the cadence, would have been difficult for our recreational cyclists to maintain. Subjects pedaled for about 60 s against each of the two inertial loads (presented in random order), resting for about 4 min between the two trials. When asked before and after each trial if they felt fatigued, the subjects responded negatively.

Each of seven analog data channels was sampled at 1000 Hz. The normal and fore-aft shear forces applied by the subject to the surface of the right and left pedals were measured with dynamometers (Newmiller *et al.*, 1988). Pedal orientation with respect to each crank arm and crank arm orientation with respect to the seat tube were measured with digital optical encoders. Because digital signals were converted to analog voltages, it was necessary to replace a few data points in each encoder's transition zone with linearly interpolated values (Bolourchi and Hull, 1985; Kautz *et al.*, 1991; Newmiller *et al.*, 1988).

For each high and low inertia trial, the ten consecutive cycles of data closest to 75 rpm were ensemble averaged at 5° crank angle increments. Seven important mean experimental trajectories were then computed for statistical analysis, each referenced to the crank angle (θ_c): (i)–(ii) the right and left pedal force (F_t) tangential to the crank arm ('tangential crank force'), (iii)–(iv) the right and left pedal force (F_r) collinear with the crank arm ('radial crank force'), (v)–(vi) the right and left pedal angles (θ_p) referenced to the horizontal, and (vii) the crank angle residual (θ_r), defined as the variation in crank angle from a linear function of time over the cycle (calculated from the relationship $\theta_r = \theta_c - (360/t_f)t$, where θ_c is zero at the start of the cycle, t is the time from the start of the cycle, and t_f is the final time of the cycle). In addition, the mean cadence (*Cadence*) was computed from the duration of the individual cycles.

These seven mean experimental trajectories were used to calculate the (two-legged) crank torque (T_c) as well as the net muscle joint torques developed by each leg. Crank torque (T_c) was calculated by summing the right and left tangential crank forces and multiplying by the crank arm length. The net ankle (T_a), knee (T_k), and hip (T_h) muscle joint torques of each leg were computed from the pedal force and crank and pedal angle data using a sagittal plane linkage model and inverse dynamics (e.g. Hull and Jorge, 1985; Redfield and Hull, 1986). The hip was assumed to remain stationary (Neptune and Hull, 1995), and the joints were assumed to be frictionless and revolute. The ankle axis was assumed to coincide with the lateral malleolus, the knee axis with the lateral epicondyle of the femur, and the hip axis with the superior aspect of the greater trochanter. The distance from the hip center to crank axis-of-rotation, originally estimated from measurement of greater-trochanter to crank-axis distance, was adjusted slightly in the model to assure that the model kinematics produced 35° of knee flexion near bottom dead center, consistent with the experimental design (see above). Limb segment masses, mass centers, and moments of inertia were estimated from measured subject weight and limb segment lengths (Dempster, 1955). Crank and pedal angular velocities and accelerations were estimated via Fourier analysis (Hull and Gonzalez, 1990; Redfield and Hull, 1986), where four harmonics were used to fit the biphasic crank angle residual (Fig. 2(b)) and three to fit the monophasic pedal angle (Fig. 3(d)).

To test for statistically significant differences between low and high inertia pedaling, we characterized each mean trajectory by four variables (e.g. Ruby and Hull, 1993): (i) maximum value (max); (ii) minimum value (min); (iii) range (rng); and (iv) average value (ave). Crank angle residual, however, was characterized by only its range (θ_r rng) and its maximum absolute value ($|\theta_r|$ max), since it is oscillatory and 180° out-of-phase under low and high inertia conditions (see Fig. 2(b)). To investigate within-subject variability, the ten consecutive cycles used in the ensemble averaging were reprocessed to compute the standard deviations associated with two mean values: (i) maximum tangential crank force (F_t max SD), and (ii) cadence (*Cadence* SD).

The statistical analysis was performed using two- and three- factor factorial analysis of variance (ANOVA) models, where the level of statistical significance was set at $p < 0.05$. Inertial load (low or high) was the primary factor of interest, and the experimental subject (1–10) was used as a noninteractive blocking factor to control for between-subject variability (Neter *et al.*, 1985). For data existing on both sides of the body (e.g. crank forces), side (right or left) was added as a third factor to the model to allow more general bilateral rather than unilateral conclusions to be drawn. Using the trajectory characterization process described above, a total of 33 individual analyses were performed (see Tables 1 and 2).

RESULTS

The subjects pedaled against the low inertia load with steady-state kinematics and kinetics that agree with pre-

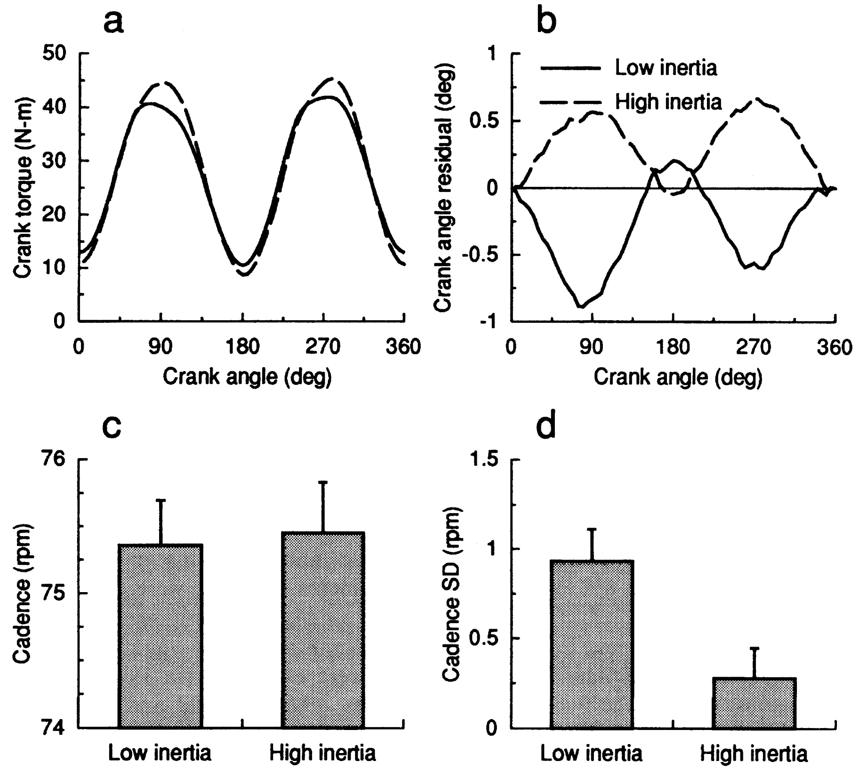

Fig. 2. Mean crank torque from both legs (a), deviation of the crank angle within a cycle from the profile which corresponds to a constant forward crank speed (b), cadence (c), and standard deviation (SD) of cadence (d) when pedaling against a low and a high inertial load. Mean values were calculated using 100 pedaling cycles (i.e. 10 consecutive cycles from each of 10 subjects). (a) Crank torque is the sum of the left and right tangential crank forces (Fig. 3(b)) multiplied by the crank arm length. Positive torque propels the crank forward. Solid line is trajectory for low inertial load; dashed line for high inertial load. (b) Positive crank angle residual implies crank rotation leads a constant crank angular speed. The 180° phase change in the residual occurs because the selected crank angular speed is below the natural frequency of the drive system for the low inertial load and above it for the high inertial load (Fregly, 1993). (c) and (d) The error bars indicate + 1SD from the mean and account for between-subject variability. Statistically significant changes in variables associated with the crank torque (< 22%), crank angle residual (40%), and cadence SD (67%) were found (Table 1). Note that the crank angle is measured with respect to the vertical, where the downstroke is defined to be between 0 and 180°, the upstroke between 180 and 360°, top dead center (TDC) at 0°, and bottom dead center (BDC) at 180°.

vious studies. Subjects propelled the crank during the downstroke (Fig. 3(b), solid line, 0–180°), retarded the crank during the upstroke (Fig. 3(b), solid line, 180–360°), and tilted the front of the pedal downward throughout the cycle, though the tilt itself was cyclical (Fig. 3(d), solid line) (Davis and Hull, 1981; Hull and Jorge, 1985; Kautz *et al.*, 1991; Newmiller *et al.*, 1988; Redfield and Hull, 1986). The radial component of crank force was directed away from the crank axis of rotation over half of the cycle through bottom dead center (BDC, 180°) (Fig. 3(a), solid line) (Kautz *et al.*, 1991). Over much of the cycle, the crank angle lagged somewhat behind the profile which corresponds to a constant crank angular speed (Fig. 2(b), solid line, which is negative almost everywhere). Consistent with the crank angle residual being oscillatory and having negative peaks at 90 and 270° (Fig. 2(b), solid line), the crank torque developed by both legs was also oscillatory with positive peaks at 90 and 270° (Fig. 2(a), solid line). The net hip and ankle muscle joint torques had similar shapes and peaked in extension near 120° (Fig. 4(a) and (c), respectively, solid lines), while the net knee joint torque exhibited a peak in extension near 45° and a peak in flexion near 180° (Fig. 4(b), solid line) (Ericson *et al.*, 1986a, b; Gregor *et al.*, 1985; Hull and Jorge, 1985).

The net hip, knee, and ankle joint torques produced by muscles were generally unaffected when subjects pedaled against the high inertial load (Fig. 4). Even for those joint torque variables which showed statistically significant changes (i.e. all the net ankle joint torque variables but one, and one net knee joint torque variable; Table 2), the changes were small (< 11%). The radial crank force and the pedal angle were also generally unaffected by the higher inertia (Table 2, *Radial crank force* and *Pedal angle*; Fig. 3(a) and (d)). The statistically significant change in the maximum value of the pedal angle (θ_p max) was small compared to the excursion of the pedal angle over the crank cycle (Fig. 3(d)).

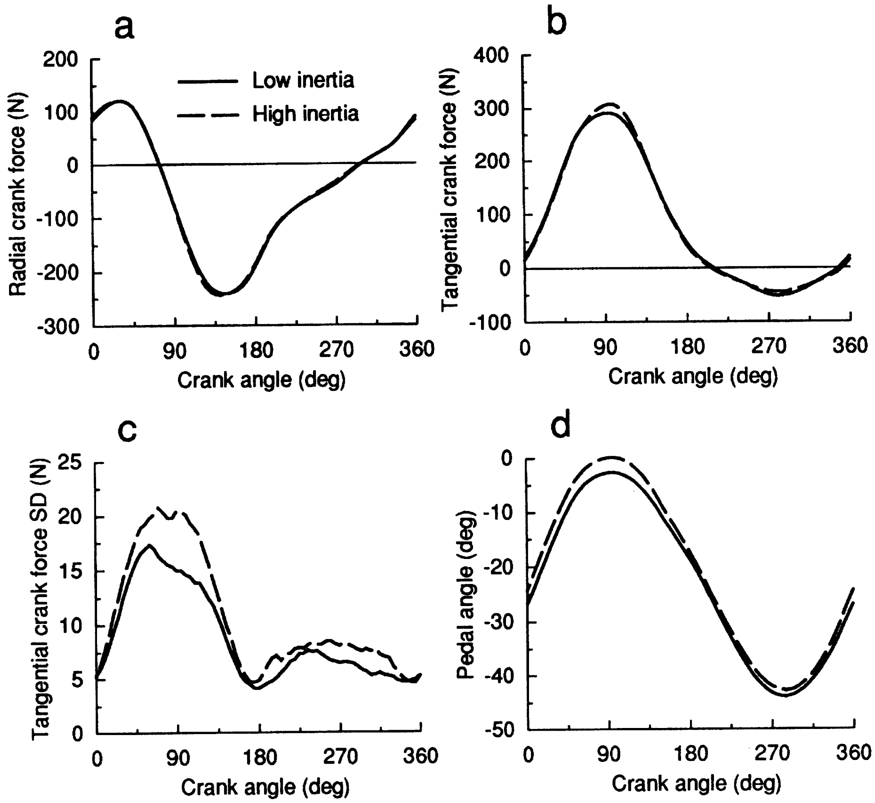

Fig. 3. Mean radial (a) and tangential (b) crank force, standard deviation of tangential crank force (c), and pedal angle (d) when pedaling against a low inertia (solid lines) and a high inertia (dashed lines). (a) Positive radial crank force is directed toward the crank axis of rotation. (b) Positive tangential crank force propels the crank forward. (c) Tangential crank force SD shows the within-subject variability in the tangential crank force. (d) Positive pedal angle corresponds to toe up. Statistically significant changes in variables associated with the tangential crank force (< 12%) and tangential crank force SD (41%) were found (Table 2).

Even though the frictional load was the same at the two inertial loads (Table 1, T_c ave), statistically significant changes were still found in variables related to the tangential crank force (i.e. one-legged crank torque). Specifically, the maximum and minimum values of tangential crank force increased despite no change in the range and average values (Table 2, *Tangential crank force*; Fig. 3(b)). With respect to the two-legged crank torque, its maximum value and range increased while its minimum value decreased (Table 1, *Crank torque*; Fig. 2(a)). Nevertheless, all of these changes were small (< 13% except T_c min = 22%, Tables 1 and 2; Figs 2(a) and 3(b)).

In addition, both within- and between-cycle variations in crank kinematics were smaller when pedaling against the higher inertia. The crank angle residual, which was used as an indicator of within-cycle kinematic variations (i.e. how nonsmooth the pedaling was), exhibited decreases of 40 and 36% in its range and maximum absolute value, respectively (Table 1, θ_r rng and $|\theta_r|$ max; Fig. 2(b)). When the compliance of the apparatus is taken into account, this decrease in range is reasonably consistent with the 25% expected for a 20-fold increase in inertia (see Appendix B in Fregly (1993), and use equation (B.4); compare the 25% decrease predicted when including compliance with the 20- fold decrease predicted without it). Similarly, the cadence standard deviation, which was used to measure between-cycle variations in crank kinematics (i.e. how unsteady the pedaling was), declined by 67% (Table 1, *Cadence* SD; Fig. 2(d)), with the subjects maintaining the same mean cadence as instructed (Table 1, *Cadence*; Fig. 2(c)). At the same time, variability in maximum tangential crank force increased (Table 2, F_t max SD; Fig. 3(c)).

DISCUSSION

The crank inertial load encountered during steady-state pedaling varies widely between a standard ergometer and road bicycle in high gear (more than a factor of 20, Fregly (1993)), as well as between a road bicycle in its lowest and in its highest gear (roughly a factor of 7, Fregly, 1993). Even so, the nervous system could, theoretically, use identical muscle coordination to pedal steadily at any given cadence and workload. We hypothesized that the net joint torques produced by the muscles at the ankle, knee, and hip would be unaffected by inertial load

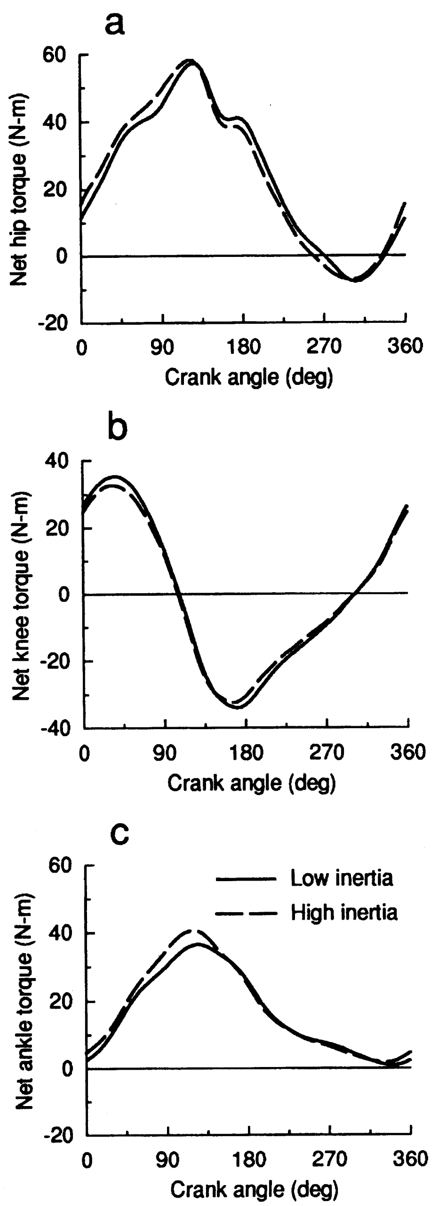

Fig. 4. Mean net muscle joint torques for the hip (a), knee (b), and ankle (c) when pedaling against a low inertia (solid lines) and a high inertia (dashed lines). Positive joint torque is in the extensor direction. All three joint torques resemble those reported by others (see text). Notice that the net joint torques are very similar when pedaling against the low and high inertial loads. Nonetheless, statistically significant changes were found in some variables associated with these net joint torques, especially the net ankle joint torque (Table 2). These changes were, however, small (< 11%).

during steady pedaling. We also hypothesized that crank kinematics would vary less at the higher inertia. Our investigation involved studying pedaling biomechanics, and muscle coordination as evidenced by the net muscle joint torques. Thus, even if our data should support these hypotheses, we would be unable to conclude decisively that coordination of individual muscles is unaltered with inertial load (i.e. muscle forces could change with high inertia pedaling even though the net muscle joint torques do not).

On the whole, our data support the hypothesis that the net joint torques produced by muscles are unaffected by inertial load. Many kinetic variables, including those associated with net muscle joint torques, did not exhibit statistically significant changes. For those few that did, the changes were small (< 13%, except T_c min = 22%, e.g. Figs 2(a), 3(b), and 4(c)). Importantly, the variables associated with the net hip and knee muscle joint torques, which produce most of the energy needed to overcome the effective frictional torque at the crank (Ericson, 1988; Ericson *et al.*, 1986a), hardly changed. Also, the net hip and knee joint torque trajectories at the higher inertial load are extremely similar to those at the lower inertia (Fig. 4(a) and (b)). Even the net ankle joint torque, which showed statistically significant increases at the higher inertia, changed little in amplitude (< 11%) and shape (Fig. 4(c)).

Since the net ankle joint torque functions in the power stroke primarily to transfer to the crank much of the energy generated by the proximal joint torques (Fregly and Zajac, 1996; Raasch *et al.*, 1996), the net ankle joint torque would not be expected to change (much) unless the net hip and knee joint torques changed. Although we found only one significant change in the net hip and knee joint torque variables, other variables not investigated statistically might show significant changes with inertial load (e.g. net hip joint torque at 90°, Fig. 4(a)). Even so, the effects of inertia on the net hip and knee joint torques would be small (Fig. 4(a) and (b)), just as they were found to be (with statistical significance) for the net ankle joint torque (Fig. 4(c)). It should also be noted that the small changes in kinetics with a 20-fold increase in inertial load are very much less than the changes in maximum tangential crank force (35%) and net ankle joint torque (63%) observed for much smaller increases in workload (1.6 and 2 times, respectively; Ericson, 1986; Kautz *et al.*, 1991).

The absence of statistically significant changes, or the presence of small changes, in net muscle joint torques (or kinetics, in general) is probably not due to inaccuracies in our model or to model simplicity. Since identical linkage models were used for both inertial loads to estimate net joint torques, any error introduced by modeling assumptions (e.g. revolute joints) should be systematic, having little effect on statistical analyses which investigate relative differences (Hull and Gonzalez, 1990). Also, Neptune and Hull (1995) found that for our cadence/workrate combination (and others), the assumption of a fixed hip is very reasonable for calculating the net hip joint torque via inverse dynamics. Furthermore, ensuring that the model matches the specified minimum knee flexion angle near BDC greatly reduces potential errors in the calculated kinematics and hence in the calculated joint torques. We believe, therefore, that the main findings of this study are unaffected by our modeling assumptions.

The small effect of inertial load on propulsion biomechanics (e.g. as evidenced by crank torque and tangential crank force, Figs 2(a) and 3(b)) and muscle

Table 1. Two-factor ANOVA statistical results for biomechanical quantities calculated from the combined influence of both legs. The standard-deviation dependent variable *Cadence* SD represents within-subject variability, while the standard deviation columns (SD) represent between-subject variability. Compared to the statistically significant increases ($p < 0.05$) observed for the crank torque ($\leqslant 22\%$), the decreases related to within-subject variability (i.e. *Cadence* SD, $|\theta_r|$ max, and θ_r rng) tended to be much larger ($\geqslant 36\%$). Note that because the maximum values for each subject did not occur at exactly the same location in the crank cycle, the mean columns (Mean) calculated by the ANOVA model may not agree exactly with the values indicated in Fig. 2

		Low inertia		High inertia			
Biomechanical quantity	Dependent variable	Mean	(SD)	Mean	(SD)	p-value	Change in Mean
Crank torque (N m)	T_c max	45.2	(2.8)	48.2	(3.6)	0.0024	7%
	T_c min	7.9	(3.4)	6.2	(2.5)	0.0421	− 22%
	T_c rng	37.3	(5.6)	42.0	(5.6)	0.0004	13%
	T_c ave	28.1	(0.7)	28.5	(1.8)	0.5125	—
Crank angle residual (deg)	$\lvert\theta_r\rvert$ max	1.1	(0.3)	0.7	(0.2)	0.0001	− 36%
	θ_r rng	1.5	(0.3)	0.9	(0.1)	0.0001	− 40%
Cadence (rpm)	Cadence	75.4	(0.3)	75.5	(0.4)	0.5516	—
	Cadence SD	0.9	(0.2)	0.3	(0.2)	0.0001	− 67%

Table 2. Three-factor ANOVA statistical results for biomechanical quantities calculated separately for each leg. The standard-deviation dependent variable F_t max SD represents within-subject variability, while the standard deviation columns (SD) represent between-subject and between-side variability. Statistically significant changes ($p < 0.05$) between subjects were found in some variables associated with the tangential crank force (i.e. one-legged crank torque) and the net ankle joint torque, though the changes were small ($\leqslant 12\%$). No statistically significant changes were found in the variables associated with the net hip and knee joint torques (except for the small decrease in T_k rng; 6%). The statistically significant change in the maximum pedal angle (θ_p max) was small compared to the excursion of the pedal angle over the cycle (Fig. 3(d); thus no change is given in the table). The within-subject variability of the tangential crank force (F_t max SD) did, however, show a large statistically significant increase (41%). Because the maximum values for each subject did not occur at exactly the same location in the crank cycle, the mean columns (Mean) calculated by the ANOVA model may not agree eactly with the values indicated in Figs 3 and 4

		Low inertia		High inertia			
Biomechanical quantity	Dependent variable	Mean	(SD)	Mean	(SD)	p-value	Change in mean
Radial crank force (N)	F_r max	126.8	(22.6)	129.3	(20.8)	0.5333	—
	F_r min	− 247.2	(40.6)	− 249.1	(30.0)	0.7084	—
	F_r rng	374.0	(32.8)	378.3	(31.0)	0.4106	—
	F_r ave	− 51.6	(17.4)	− 51.0	(14.1)	0.7708	—
Tangential crank force (N)	F_t max	296.1	(26.2)	312.8	(27.2)	0.0001	6%
	F_t min	− 54.6	(16.4)	− 47.9	(21.6)	0.0201	12%
	F_t rng	350.8	(40.6)	360.7	(43.2)	0.0757	—
	F_t ave	82.7	(4.6)	83.9	(6.5)	0.4096	—
	F_t max SD	14.8	(5.2)	20.9	(6.2)	0.0001	41%
Pedal angle (deg)	θ_p max	− 2.3	(9.3)	0.5	(7.5)	0.0202	—
	θ_p min	− 44.4	(8.2)	− 43.2	(7.6)	0.4171	—
	θ_p rng	42.1	(4.8)	43.7	(5.0)	0.2392	—
	θ_p ave	− 22.7	(8.1)	− 20.8	(6.7)	0.1043	—
Hip torque (N m)	T_h max	66.2	(21.5)	68.1	(21.1)	0.6564	—
	T_h min	− 7.7	(5.1)	− 8.8	(7.1)	0.3942	—
	T_h rng	73.9	(21.0)	76.9	(19.9)	0.5077	—
	T_h ave	27.4	(10.8)	27.1	(10.8)	0.9059	—
Knee torque (N m)	T_k max	36.8	(7.6)	34.1	(7.4)	0.1415	—
	T_k min	− 34.8	(10.4)	− 33.0	(10.3)	0.2939	—
	T_k rng	71.6	(8.6)	67.1	(10.5)	0.0116	− 6%
	T_k ave	− 0.8	(6.6)	− 1.2	(6.2)	0.7451	—
Ankle torque (N m)	T_a max	37.4	(6.2)	41.4	(5.6)	0.0004	11%
	T_a min	0.3	(2.7)	1.0	(2.9)	0.1062	—
	T_a rng	37.0	(5.8)	40.3	(5.9)	0.0016	9%
	T_a ave	16.5	(3.1)	17.7	(2.8)	0.0103	7%

coordination (e.g. as evidenced by net muscle joint torques, Fig. 4) observed here for smooth, steady pedaling may not apply to variable-speed pedaling where large changes in crank angular acceleration occur, especially during pedaling initiation. Clearly, much higher tangential crank forces and mechanical energy must be developed by muscles to accelerate a higher inertia the same distance in the same amount of time. For example, assuming no frictional resistance and a two-legged crank torque which is constant over the cycle, muscles would have to produce 20 times (i.e. the ratio of the two inertial loads) more crank torque and deliver 20 times more energy to the crank with the higher inertia. Also, whether the small effects observed here apply to cadence/workrate combinations other than 75 rpm/225 W is unknown (however, the effects appear to be similar for the lower cadence/workrate combination of 60 rpm/130 W; Fregly, 1993).

Our data also support the hypothesis that crank kinematics vary less at the higher inertia. Both the cycle-to-cycle cadence variability and the within-cycle, peak-to-peak change in crank angular acceleration (as evidenced by the peak-to-peak change in crank angle residual) are reduced when pedaling against the higher inertia (about 67 and 40% reductions, respectively, Table 1; Fig. 2(b) and (d); see Results). When evaluating such results for biomechanical variables related to variability (i.e. $|\theta_r|$ max, θ_r rng, *Cadence* SD, and F_t max SD), it is important to consider whether the observed changes could have been caused by differences in cycle-to-cycle variability of the frictional load. To investigate this possibility, we developed analysis of covariance (ANCOVA) models by adding the standard deviation of the frictional load (i.e. T_c ave SD) as an independent continuous variable to the original ANOVA model for each dependent variable above. The ANCOVA results ($p < 0.05$) revealed that cycle-to-cycle variations in frictional load cannot account for the observed changes in any of these dependent variables.

A possible explanation for the reduction in crank kinematic variations at the high inertial load is the large corresponding reduction in crank load speed of response. Assuming no change in tangential crank force variability, the expected reduction in cadence variability can be estimated by using (1/rise time) as an indication of crank load speed of response. For each inertial load, the rise time was estimated from the crank load natural frequency (i.e. rise time $= \pi/(2\omega_n)$, where ω_n is the crank load natural frequency; Ogata, 1970), which was determined experimentally by having one additional subject pedal at a range of cadences between 20 and 100 rpm (see Fregly (1993) for frequency response plots). For the higher inertia used in our experiments, the rise time was about 630 ms; for the lower inertia, 140 ms. Thus, because the dynamics of the high inertial load was about 450% slower than those of the low inertial load, cadence variability would have been $4\frac{1}{2}$ times lower at the higher inertia had the tangential crank force variability been unchanged. Even with the increased tangential crank force variability (F_t max SD, Table 2), cadence variability was 3 times lower (*Cadence* SD, Table 1).

Subjects were able, therefore, to pedal against a higher inertia not only with less variation in crank kinematics but also with less regulation of the (peak) tangential crank force. Since peak tangential crank force is dominated by contributions from the net muscle joint torques (Fregly and Zajac, 1996; Kautz and Hull, 1993), the increase in peak tangential crank force variability is probably due to higher variability in the generation of muscle forces. Less regulation may be preferred by subjects at the higher inertia because more mental effort might otherwise be required and, clearly, the perceived requirements of the task do not demand more regulation. Indeed, Patterson *et al.* (1983) reported that subjects found it harder to maintain a constant cadence with a lower inertial load.

We have shown that pedaling against widely different inertial loads, typical of the range encountered during road riding and ergometry pedaling (Fregly, 1993), is accomplished with very similar net muscle joint torques at the ankle, knee, and hip, at least for 75 rpm pedaling at a 225 W workrate. As a consequence of the near identical kinetics of the legs, crank kinematic variability both within a cycle and between cycles is reduced when pedaling against a high inertia. However, because the variability in the one-legged crank torque (i.e tangential crank force) is increased, the reduction in cadence variability is not as large as it would have been had the variability in the one-legged crank torque been unaffected by inertial load. We suggest, therefore, that muscle coordination during steady-state pedaling is largely unaffected, though less well regulated, when crank inertial load is increased.

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