Lower Extremity Joint Moments During Uphill Cycling

Graham E. Caldwell, James M. Hagberg, Steve D. McCoile, and Li Li

Lower extremity joint moments were investigated in three cycling conditions: level seated, uphill seated and uphill standing. Based on a previous study (Caldwell, Li, McCoile, & Hagberg, 1998), it was hypothesized that joint moments in the uphill standing condition would be altered in both magnitude and pattern. Eight national caliber cyclists were filmed while riding their own bicycles mounted to a computerized ergometer. Applied forces were measured with an instrumented pedal, and inverse dynamics were used to calculate joint moments. In the uphill seated condition the joint moments were similar in profile to the level seated but with a modest increase in magnitude. In the uphill standing condition the peak ankle plantarflexor moment was much larger and occurred later in the downstroke than in the seated conditions. The extensor knee moment that marked the first portion of the downstroke for the seated trials was extended much further into the downstroke while standing, and the subsequent knee flexor moment period was of lower magnitude and shorter duration. These moment changes in the standing condition can be explained by a combination of more forward hip and knee positions, increased magnitude of pedal force, and an altered pedal force vector direction. The data support the notion of an altered contribution of both muscular and non-muscular sources to the applied pedal force. Muscle length estimates and muscle activity data from an earlier study (Li & Caldwell, 1996) support the unique roles of mono-articular muscles for energy generation and bi-articular muscles for balancing of adjacent joint moments in the control of pedal force vector direction.

Key Words: cycling, kinetics, joint moments, muscle kinematics

The biomechanics of cycling have been examined extensively from different perspectives. Using an instrumented pedal it has been possible to measure the forces applied by the cyclist to the pedal and crank of the bicycle (Broker & Gregor, 1990; Hull & Davis, 1981; Soden & Adeyefa, 1979). The measured pedal forces and kinematic descriptions of lower extremity motion have been combined to estimate loading at the joints (Gregor, Cavanagh, & Lafortune, 1985; van Ingen Schenau, van Woensel, Boots, 1986).

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Snackers, & de Groot 1990). Further insight has been gained by monitoring muscle activity patterns with electromyography (EMG) to examine which muscles contribute to the moments of force seen at each joint (Ryan & Gregor, 1992). These analysis techniques have been used to examine the effect of changing cadence (Marsh & Martin, 1993; Patterson & Moreno, 1990; Redfield & Hull, 1986), varied power output (Ericson, Bratt, Nisell, Arborelius, & Ekblom, 1986; Kautz, Feltner, Coyle, & Baylor, 1991) and rider skill level (Coyle et al., 1991).

Most of these studies have focused on cyclists riding on level terrain, while there is much less research found on cycling in either uphill or downhill conditions. In competitive cycling road races, performance on the graded sections of a course is often the major determinant of success in a race. Despite this importance, the biomechanical responses to changing grade are relatively unknown. Basic physical principles suggest that lighter riders would have an advantage during uphill climbs. On level terrain the two major impediments to forward velocity are air resistance and friction, while on uphill terrain there is an additional gravitational load. However, it remains to be seen whether other biomechanical factors such as muscle activation or joint moment profiles are altered in response to the change in incline. A further interesting aspect to uphill climbing is the fact that cyclists use various postural strategies (different relative periods of sitting and standing) to overcome the higher loads and subsequent fatigue associated with uphill terrain.

Cycling on graded surfaces also has implications from a motor control perspective, as the basic cycling motion found during level riding is now performed with an altered environmental constraint. Recent motor control theories suggest an important role for environmental constraints in shaping motor behavior through their integration with biophysical properties of the human actor (Newell, 1986). In the case of uphill cycling, the new orientation of the rider and bicycle with respect to both the surface and direction of the gravitational load may evoke subtle but important changes in the rider’s control strategy. Indeed, Brown, Kautz, and Dairaghi (1996) found that altering the incline angle in a cycling task caused changes in both joint moments and individual muscle activity patterns, even without the addition of a gravitational load as in actual uphill cycling.

Recently, several studies have explored the biomechanics of uphill cycling. Stone and Hull (1993) examined both pedal and handlebar forces in uphill standing cycling on an inclined treadmill for three subjects, and Alvarez and Vinyolas (1996) reported exemplar pedal force profiles from an instrumented bicycle in an actual hill climbing trial. The largest data set was reported by Caldwell and colleagues (1998), who compared pedal and crank kinetics for eight elite riders in level seated, uphill seated, and uphill standing conditions on a computerized cycle ergometer. Their results indicated that pedal force profiles were very similar between level and uphill seated conditions. However, pedal forces and crank torque in the uphill standing condition were significantly altered from the seated trials (Figure 1). In particular, the peak resultant pedal force and peak crank torque were of much higher magnitude, increasing by 200% and 130%, respectively, over the uphill seated condition. Further, these peak values occurred later in the crank cycle for the standing condition (155° [force] and 130° [torque] of crank angle) than in the seated condition (100° and 85°). The higher peak kinetic values were linked to changes in pedal orientation and pedal force vector direction throughout the crank cycle, and were associated with the upward and forward movement of the center of mass as the pelvis came off the saddle. Removing the saddle as a base of support requires increased weight support by the pedals and handlebars, leading to the possibil-
ity of altered muscular and non-muscular contributions to pedal force (Caldwell et al., 1998; Kautz & Hull, 1993).

The changes in pedal and crank loading in uphill cycling lead one to question the nature of joint loading within the lower extremity. Therefore, the purpose of the present study was to examine the joint moment patterns of the lower extremity in uphill cycling and compare them to those in level cycling. Because of the modified magnitude, timing, and direction of applied pedal forces, our hypothesis was that the uphill standing condition would result in alterations in the joint moment profiles at each of the three lower extremity joints, particularly in the late downstroke phase of the crank cycle. In this paper we report the results of laboratory cycling trials using a computerized ergometer to simulate both level and uphill cycling conditions. The kinematics of lower extremity segments are also reported to aid in our interpretation of the joint moments.
Methods

Eight elite (USCF category 1 or 2) male cyclists were recruited from cycling clubs in the Baltimore/Washington area. Subjects were free of any neural or musculoskeletal conditions that might preclude their participation. Each subject read and signed informed consent documents after the experimental procedures were described. The subjects rode a simulated course on their own bicycles mounted on a computerized Velodyne ergometer capable of replicating the resistance associated with cycling at different intensities and grades. The ergometer required the removal of the front wheel so that the forks could be attached firmly to the front mounting bracket. This attachment limited the mediolateral motion of the bicycle, which is commonly observed in normal uphill cycling. However, the pedal force and crank torque patterns from subjects riding on this ergometer (Figure 1 and Caldwell et al., 1998) were very similar to the patterns observed cycling uphill on a treadmill (Stone & Hull, 1993) or in actual hill climbing trials (Alvarez & Vinyolas, 1996).

After a warm-up of 5 min of level cycling at a self-selected pace, the subjects were required to ride a preset simulated course on the ergometer. The cycling “course” consisted of an initial 5-min level grade, followed by a 10-min simulated uphill section (8% grade) and a final 5-min level portion (see Figure 2). The ergometer was secured to a standard exercise treadmill capable of grade adjustments to physically match the 8% grade of the uphill section. For the entire 20-min trial the subjects rode at a power output corresponding to 80% of their predetermined cycling VO₂max. To achieve constant power output on the level and uphill parts of the course, the subjects selected their preferred gear ratios and cadences, and the ergometer speed was adjusted by the experimenter to produce the correct power output as indicated by the Velodyne control panel. Each subject rode four trials on the simulated course, conducted on separate days and in random order. On all level sections the riders remained seated, while different hill-climbing strategies (climbing in seated or standing posture) were used in each trial. In the present paper, we contrast the joint moment profiles associated with three conditions: seated level grade, seated uphill grade, and standing uphill grade.

Sagittal plane kinematics of reflective markers attached to the subject and bicycle were determined with a high speed (50 fps) cine camera. Markers on the bicycle were located on the crank spindle, pedal spindle, and a pin marking the back of the pedal. The joint centers and segmental alignment were identified by markers on the fifth metatarsal, lateral malleolus, lateral femoral epicondyle, greater trochanter, and humeral head. A clip-in Look pedal instrumented with two piezoelectric load washers (Broker & Gregor, 1990) was attached to the subject’s bicycle on the side facing the camera. Film and force data were collected at the midpoint of each 5-min level portion and at the 3rd- and 7th-minute marks on the uphill portion of the course. Force pedal data were collected at 100 samples per second using a 12-bit analog-to-digital converter interfaced with a microcomputer. The force and film data were synchronized using a switch that illuminated an LED in the field of view and provided a 3V pulse to a channel on the force A/D record. The digitized force data were converted from A/D computer units to Newtons using a zero-force bias trial and predetermined calibration coefficients, while the film data were scaled using a known calibration length filmed in the plane of the cyclist’s leg. A recursive low-pass Butterworth filter was used to smooth the film (4 Hz cutoff) and force (10 Hz cutoff) data. The pedal forces and marker kinematics were output at 1° increments of crank angle using quintic spline techniques (Dierkx, 1975). Data were expressed as functions of crank angle referenced to top-dead-center (0° and 360°, TDC) and bottom-dead-center (180°, BDC).
The kinetic and kinematic data were used in a linked-segment inverse dynamics analysis (Gregor et al., 1985; Redfield & Hull, 1986). The rider’s body was modeled as four rigid segments consisting of the foot, leg, thigh, and trunk (Figure 2). The linear and angular kinematics of each segment were calculated using standard planar techniques (Winter, 1990). The pedal reaction forces acting on the foot were considered external forces and were resolved into vertical and horizontal components. These pedal reaction forces were combined with the foot kinematics to calculate the ankle joint reaction forces and moments, using Newtonian equations of motion. Similar calculations using the ankle and knee joint kinetics as inputs to leg and thigh segment models, respectively, permitted calculation of kinetics at the knee and hip joints.

Patterns of the segmental and joint kinematics and moments were plotted for each trial as a function of crank angle from $0^\circ$ to $360^\circ$, referenced to TDC in the global reference system. The individual trial patterns were combined to form ensemble average curves and used to calculate the variability around these ensembles. Variability of kinematic and kinetic profiles across subjects and conditions was assessed through the coefficient of variation (CV), defined as the standard deviation expressed as a percentage of the range of values observed for the variable throughout the crank cycle (Whitall &
Table 1 Subject (n = 8) and Cadence Characteristics

<table>
<thead>
<tr>
<th>Variable</th>
<th>M</th>
<th>SD</th>
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</thead>
<tbody>
<tr>
<td>Age (yrs)</td>
<td>28</td>
<td>5</td>
</tr>
<tr>
<td>Mass (kg)</td>
<td>78.9</td>
<td>5.3</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>185.4</td>
<td>6.0</td>
</tr>
<tr>
<td>Power output (W)</td>
<td>294</td>
<td>17</td>
</tr>
<tr>
<td>Cadence (rpm)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Level seated</td>
<td>82</td>
<td>15</td>
</tr>
<tr>
<td>Uphill seated</td>
<td>65</td>
<td>5</td>
</tr>
<tr>
<td>Uphill standing</td>
<td>64</td>
<td>5</td>
</tr>
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</table>

Caldwell, 1992). Discrete measures of interest from the kinetic patterns were the peak moments at the ankle (plantarflexor), knee (extensor and flexor), and hip (extensor), and the crank angles at which these peaks occurred. The ensemble curves were used as the basis of a qualitative analysis among conditions, while the discrete measures were analyzed with a repeated measures, within-subjects analysis of variance (ANOVA) to illustrate quantitative similarities and differences among conditions. The significance level for the ANOVA was set at $\alpha \leq .05$.

Results

Rider characteristics, cycling cadences, and 80% VO\textsubscript{2}max power outputs are given in Table 1. Despite the differences in subject mass, height, and cadence, relatively stereotypical kinematic patterns were evident for each condition (Table 2). CV values for segment angular displacement were low in all cases, ranging from 8.24% to 13.01%, with similar variability levels across cycling conditions. These variability data indicate that stable kinematic patterns were in evidence for our subjects in all three conditions.

A kinematic comparison of the mean segment and joint angles for all conditions is shown in Figure 3. For the segment data, the level and uphill seated conditions show very similar patterns, with a slight offset due mainly to the altered bicycle orientation in the global reference system. The uphill standing condition has greater differences in both offset and relative timing of minima and maxima. The offset from the level condition is not due to the bicycle orientation, as it is in the opposite direction from the uphill seated offset. In this case the change in posture is responsible for the angular offset. The thigh and leg patterns both show a shift in their reversal points, from just prior to BDC in the seated conditions to just after BDC in the standing condition. The foot segment data also illustrate a shift in the maximum and minimum, near the 90° and 270° points in the crank cycle, respectively.

Table 2 Coefficients of Variation (CV) for Segmental Angular Displacement

<table>
<thead>
<tr>
<th>Segment</th>
<th>Level seated</th>
<th>Uphill seated</th>
<th>Uphill standing</th>
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</thead>
<tbody>
<tr>
<td>Thigh</td>
<td>8.24</td>
<td>8.69</td>
<td>9.57</td>
</tr>
<tr>
<td>Leg</td>
<td>10.38</td>
<td>9.11</td>
<td>9.26</td>
</tr>
<tr>
<td>Foot</td>
<td>11.96</td>
<td>13.01</td>
<td>9.83</td>
</tr>
</tbody>
</table>
Figure 3 — Segmental and joint angular kinematics expressed as a function of crank rotation from TDC (0°) through BDC (180°) to TDC (360°). Mean ensemble averages are shown for the thigh, leg and foot segments, and for the hip, knee and ankle joints. The three cycling conditions are: level seated (thin solid line), uphill seated (thin broken line), and uphill standing (thick solid line).

The joint kinematics (also shown in Figure 3) confirm this interpretation of the segmental kinematics. For the two seated conditions, the offset due to the bicycle orientation is no longer apparent because the subtraction of segment angles removes it, and the level and uphill data are very similar. For the uphill standing condition the effects of the different posture are still in evidence, especially at the hip where a 0.5 radian offset towards a more extended position is seen. At both the hip and knee joints, there is a shift in the timing of the most extended positions to after BDC, whereas the seated conditions display these peaks just prior to BDC. The ankle joint also displays this shift in peak, but the overall patterns for the three conditions are more similar than for the hip and knee.

Variability data for the resultant joint moments are shown in Table 3. The CV values ranged from 8.87% to 31.83%, with lower values (8.87 to 16.52) at the ankle and

<table>
<thead>
<tr>
<th>Segment</th>
<th>Level seated</th>
<th>Uphill seated</th>
<th>Uphill standing</th>
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<tbody>
<tr>
<td>Hip</td>
<td>28.49</td>
<td>23.56</td>
<td>31.83</td>
</tr>
<tr>
<td>Knee</td>
<td>12.62</td>
<td>12.81</td>
<td>16.52</td>
</tr>
<tr>
<td>Ankle</td>
<td>15.63</td>
<td>14.99</td>
<td>8.87</td>
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</table>
Figure 4 — Ankle joint moments expressed as a function of crank rotation. Mean ensemble averages are shown for level seated (thin solid line), uphill seated (thin broken line), and uphill standing (thick solid line) cycling conditions.

knee joints but higher values (23.56 to 31.83) at the hip. Across conditions, the uphill standing data show higher variability than the two seated conditions for the hip and knee, but less variation for the ankle joint moments. The higher CVs for the hip and knee moments are likely related to the greater freedom of motion afforded by the removal of the saddle as a base of support, although because of the bicycle attachment to the Velodyne, this motion was restricted mostly to the sagittal plane.

Ensemble patterns of the ankle moments for each condition are illustrated in Figure 4. The ankle moment profiles for all conditions illustrate exclusively plantarflexor torque throughout the crank cycle, with the highest values after 90° in the latter part of the downstroke. For the two seated conditions, the profiles have similar shapes but the peak moment is significantly higher in the uphill seated (−65 Nm) than in the level (−52 Nm) condition (Table 4). In addition, the peak moment occurs slightly earlier in the crank cycle for the uphill seated condition (108° vs. 120°). For the uphill stand-

<table>
<thead>
<tr>
<th>Joint/angle</th>
<th>Level seated</th>
<th>Uphill seated</th>
<th>Uphill seated</th>
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<tr>
<td></td>
<td>M</td>
<td>SD</td>
<td>M</td>
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<tr>
<td>Hip extensor (Nm)</td>
<td>−68&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>(13)</td>
<td>−75&lt;sup&gt;a&lt;/sup&gt;</td>
</tr>
<tr>
<td>Crank angle (°)</td>
<td>124&lt;sup&gt;ab&lt;/sup&gt;</td>
<td>(30)</td>
<td>118&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>Knee extensor (Nm)</td>
<td>66&lt;sup&gt;b&lt;/sup&gt;</td>
<td>(14)</td>
<td>76&lt;sup&gt;a&lt;/sup&gt;</td>
</tr>
<tr>
<td>Crank angle (°)</td>
<td>50&lt;sup&gt;a&lt;/sup&gt;</td>
<td>(16)</td>
<td>35&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>Knee flexor (Nm)</td>
<td>−42&lt;sup&gt;a&lt;/sup&gt;</td>
<td>(11)</td>
<td>−45&lt;sup&gt;a&lt;/sup&gt;</td>
</tr>
<tr>
<td>Crank angle (°)</td>
<td>179&lt;sup&gt;a&lt;/sup&gt;</td>
<td>(28)</td>
<td>187&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>Ankle plantarflexor (Nm)</td>
<td>−52&lt;sup&gt;c&lt;/sup&gt;</td>
<td>(12)</td>
<td>−65&lt;sup&gt;b&lt;/sup&gt;</td>
</tr>
<tr>
<td>Crank angle (°)</td>
<td>120&lt;sup&gt;b&lt;/sup&gt;</td>
<td>(11)</td>
<td>108&lt;sup&gt;c&lt;/sup&gt;</td>
</tr>
</tbody>
</table>

<sup>Note</sup>. Statistically significant differences between conditions are indicated by the superscript letters; condition means with the same superscript letter are not significantly different.
ing condition the peak moment is much higher in magnitude (-104 Nm) and is shifted to much later in the downstroke portion of the crank cycle (156°).

The ensemble knee joint moment profiles are shown in Figure 5. The knee moment patterns follow a similar profile in the two seated conditions, with an extensor period that begins near 270° before TDC and continues through the initial portion of the downstroke. The uphill standing condition demonstrates an extended, bimodal knee extensor phase. In the two seated conditions, the extensor phase ends just past 90°, while for the uphill standing condition the extensor activity is prolonged by the second local maximum until near BDC. The knee moment is reversed from extensor to flexor for the mid-portion of the crank cycle (90° to 270°) for the seated conditions. Because of the extended extensor period, this flexor period is more restricted, from ≈180° to 270°, in the uphill standing condition.

For the extensor knee moment phase, the uphill seated condition displayed a peak value (76 Nm) which was significantly greater than the level seated condition (66 Nm). The uphill standing peak was between these two values (68 Nm) but was not statistically different than either. The two uphill conditions exhibited the extensor peak earlier in the crank cycle (35° and 32° for the uphill seated and standing, respectively) than the level seated (50°). The large variability in the uphill standing condition is due to the second extensor peak (near 135°) being larger than the first in a few trials. For the flexor knee moment period, the seated conditions both displayed a significantly higher peak value (-42 Nm and -45 Nm for the level and uphill seated, respectively) than the uphill standing condition (-26 Nm). This lower flexor peak moment occurred significantly later in the crank cycle for uphill standing (204°) than for the level seated condition (179°), while the uphill seated value (187°) was not statistically different than either.

Of the three lower extremity joint moments, the patterns for the hip (Figure 6) display the most similarity across conditions. The profiles are predominantly extensor, with a brief, low magnitude flexor burst at the end of recovery from 270° to TDC. The peak extensor moment is highest (-75 Nm) and occurs earliest (118°) in the uphill seated condition. These values are significantly different than the uphill standing condition (-60 Nm at 139°) but not different than the level seated condition (-68 Nm at 124°).
Figure 6 — Hip joint moments expressed as a function of crank rotation. Mean ensemble averages are shown for level seated (thin solid line), uphill seated (thin broken line), and uphill standing (thick solid line) cycling conditions.

Discussion

This study compared kinematic and kinetic responses of the lower extremity to changing both grade and posture while cycling. The kinematic profiles showed relatively little alteration with the change in grade alone but larger changes with the uphill standing posture. For the joint moments, the largest modifications in the magnitude and timing of peak values were evident at the knee and ankle in the uphill standing condition. These data support our hypothesis that the joint kinetic patterns would be altered in the late downstroke phase of the uphill standing condition.

Several studies (Ericson et al., 1986; Francis, 1986; Hull & Hawkins, 1990) report joint angular displacement profiles that are similar to the level seated condition of the present study. The joint moment patterns for the level condition in the present study also resemble those reported by others (Ericson et al., 1986; Gregor et al., 1985; Redfield & Hull, 1986), and the peak joint moment values are similar to other studies if the effects of cadence and power output are taken into account (Redfield & Hull, 1986). For the uphill conditions, there are no studies of which we are aware that describe joint kinetics. However, our pedal and crank kinetics (Figure 1) for the uphill conditions were similar in shape and magnitude to other studies (Alvarez & Vinyolas, 1996; Stone & Hull, 1993), which lends support to the validity of our uphill joint moment profiles.

While the effects of changing grade alone (seated level vs. seated uphill) were minimal, it is important to remember that the subjects chose a lower cadence (65 rpm) in the uphill trials as compared to the level (82 rpm). The uphill seated condition showed an increase in the magnitude of peak ankle plantarflexor and knee extensor moments, and a shift in these peak moments to slightly earlier in the crank cycle. These changes were likely related to the increase in total work done per crank revolution, a consequence of holding power output constant while cadence decreased. In level cycling, decreased cadence has been shown to increase the magnitude of peak extensor moments at both the hip and knee (Redfield & Hull, 1986).

The larger changes seen in the standing condition cannot be explained as easily, because cadence, gearing, and power output were the same in the two uphill conditions. The most distinct kinetic modifications were seen at the ankle and knee joints, with peak
ankle plantarflexor moment increased by 160% and shifted by roughly 45° in the crank cycle compared to the uphill seated condition. The knee extensor profile also showed a shift towards the late downstroke period, with a bimodal extensor pattern that exhibited its second peak at about 135° crank angle. This second peak delayed the onset of the flexor moment to just before BDC, and resulted in reduced flexor moment magnitude and duration. Finally, the uphill standing condition exhibited a decreased and later peak hip extensor moment, although this change was much more modest than the changes at the knee and ankle.

Undoubtedly these changes are associated with removal of the saddle as a base of support for the rider. In the standing posture the cyclist’s hip joint and mass center move forward and upward. The hip joint angle becomes more extended, and the thigh segment is oriented more vertically. The leg orientation also changes so that knee joint angles are similar to the seated conditions during the downstroke. Unlike the seated conditions, the knee extension phase persists into the first portion of the upstroke past BDC. The ankle joint also exhibits extended plantarflexion past BDC but to a much smaller extent. These kinematic changes result in an alteration in the pedal orientation to a more toe down position throughout the crank cycle and modification of the direction of the applied pedal force vector (Caldwell et al., 1998).

Although resultant joint moments must balance gravitational, inertial and external force effects, the magnitude and direction of the pedal force vector is a major determinant of the required lower extremity joint moments (van Ingen Schenau, Boots, de Groot, Snackes, & van Woensel, 1992). The changes in joint moments from seated to standing are associated with three factors: higher pedal forces, particularly late in the downstroke, the forward (toe down) shift in pedal orientation, and the more forward hip and knee positions (Caldwell et al., 1998). Figure 7 illustrates these effects on the pedal force vector in the latter part of the downstroke at the 115°, 140° and 165° crank angle positions for an exemplar subject. In the standing condition, the line of action of the force vector is closer to the ankle joint center, but the pedal force is so much larger in magnitude that the ankle joint moment is increased in this part of the downstroke. For the seated condition, the force line of action is posterior to the knee joint at 115° but anterior at 140°, indicating a switch from knee extensor to flexor moment. In the standing condition, the force line of action is still posterior to the knee joint at 140° and moves in front of the knee later in the downstroke (near 160° for this trial). This explains the extended knee extensor moment and delayed knee flexor moment in the standing condition. The hip extensor moments are more similar in the seated and standing conditions due to interactions among the three factors. In the standing position the hip joint center is positioned further forward with respect to the crank center. In combination with the altered pedal force direction, the force moment arm at the hip is smaller in the standing condition. However, the much larger force magnitude offsets this to keep the hip extensor moment similar to the seated condition.

Kautz and Hull (1993) have shown that both muscular and non-muscular factors dictate the magnitude and profile of the pedal force vector throughout the crank cycle in level seated cycling. When standing the removal of the saddle support results in an increased contribution of gravitational forces to the measured pedal forces as a larger proportion of the weight is borne by the pedal during the downstroke (Caldwell et al., 1998). In the late part of the downstroke, the rider can make effective use of this gravitational component to generate propulsive crank torque by angling the pedal to increase the horizontal pedal force in the backward direction. Because of the extended leg position, the more forward hip position and the altered force vector direction, this is accom-
plished using a knee extensor moment. In contrast, during seated cycling the backward directed horizontal component can only be generated by pulling back on the pedal with knee flexor moment. The increased gravitational load in late downstroke also is a major reason for the large increase in ankle plantarflexion moment while standing.

These ankle and knee moment alterations indicate that muscular changes are involved in the switch from seated to standing posture. However, the inverse dynamics analysis assumes the use of one joint muscles and the absence of antagonist cocontraction. It has been suggested that mono- and bi-articular muscles play different roles in the production of multi-segment actions such as in cycling (van Ingen Schenau et al., 1992; van Ingen Schenau, Dorssers, Welter, Beelen, de Groot, & Jacobs, 1995). These authors suggest that mono-articular muscles produce the majority of the muscular work at each
joint, while the bi-articular muscles act to redistribute energy across adjacent joints. Bi-articular muscles are also thought to control the direction of applied external forces by balancing the relative moments of the adjacent joints they cross (Jacobs & van Ingen Schenau, 1992; van Ingen Schenau et al., 1992). The changes in pedal force direction in the standing condition may therefore indicate different usage of bi-articular muscles.

Figure 8 illustrates length changes for six different muscles of the lower extremity, modeled from regression equations (see the appendix) based on the ensemble joint kinematics from each condition (Bobbert, Huijing, & van Ingen Schenau, 1986; Grieve, Pheasant, & Cavanagh, 1978; Visser, Hoogkamer, Bobbert, & Huijing, 1990). In the standing condition the single joint extensors GM, VA, and SO each display a longer shortening (concentric) phase during the downstroke that lasts past BDC, in contrast to the seated trials in which the concentric phase ends prior to BDC. Electromyographical data (horizontal bars in Figure 8) from a different group of riders undergoing similar cycling conditions (Li & Caldwell, 1996, 1998) indicate that the single joint GM and VA both have extended periods of muscle activity that match these extended shortening
phases. Although SO activity was not recorded, the increase in plantarflexor moment without changes in GA activity indicate that SO also would have an extended period of activation (van Ingen Schenau et al., 1992).

These data support the suggestion that mono-articular muscles act mainly as generators of mechanical energy (van Ingen Schenau et al., 1995). In the standing condition the three mono-articular muscles demonstrated extended periods of concentric activity resulting in energy generation late in the downstroke. This phase coincided with the much higher plantarflexion moment at the ankle and the extended period of knee extensor activity. This also supports the suggestion that single joint extensors would have an increased role when gravitational support is necessary, such as in the standing condition (van Ingen Schenau et al., 1992). This gravitational support function in the standing condition is likely another reason for the longer period of knee extensor moment in the late downstroke.

The two joint muscles demonstrate more variable responses in the standing condition. Unlike the single joint muscles, the bi-articular muscles each had unique patterns of length change and activity. BF displayed a longer period of shortening but without a change in its period of activity from the seated condition, while RF had extended periods of both shortening and activity. GA showed the smallest changes from seated to standing, with similar kinematic and activity patterns. These varied bi-articular muscle responses are likely related to alterations in the balance between joint moments needed to control the direction of the pedal force vector (van Ingen Schenau et al., 1992). For example, the increased RF activity in late downstroke would reduce the hip extensor moment and increase the knee extensor moment. By itself, this change would alter the pedal force vector to a more forward orientation. However, the increased pedal angle and higher plantarflexor moment counteract this tendency so that the pedal force vector is angled in a more backward direction. This allows RF to add to the energy generation at the knee and contribute needed gravitational support, and leads to a resultant knee extensor rather than flexor moment in late downstroke. The extended RF activity may also be associated with a transfer of energy from the proximal hip to the knee and then to the distal ankle through the action of the bi-articular GA (van Ingen Schenau, 1989).

In summary, there are many requirements for the lower extremity muscles during cycling, and these requirements are altered with the switch from seated to standing posture. Among these are the control of the pedal force vector direction, generation and distribution of mechanical energy, and gravitational support. The key to the modifications that occur in the standing posture is the forward shift of the hip and knee as the pelvis is removed from the saddle. This results in a more vertical position for the thigh segment, and the hip and knee positions being further forward relative to the crank and pedal. This means that a knee extensor moment can be used throughout the entire downstroke, providing both propulsive crank torque and gravitational support.

References


Appendix

The changes in muscle length shown in Figure 8 were computed from regression equations based on cadaver data from different studies (Bobbert, Huijing, & van Ingen Schenau, 1986; Grieve, Pheasant, & Cavanagh, 1978; Visser, Hoogkamer, Bobbert, & Huijing, 1990). Muscle length changes were estimated from sagittal plane angles of joints that they cross. The equations defining the length changes for each muscle were of the form:

$$\Delta L = a_0 + (a_1 \times \theta) + (a_2 \times \theta^2)$$

for mono-articular muscles, and

$$\Delta L = a_0 + (a_1 \times \theta) + (a_2 \times \psi^2) + b_0 + (b_1 \times \psi) + (b_2 \times \psi^2)$$

for bi-articular muscles.

In these equations ($a_0$, $a_1$, $a_2$) and ($b_0$, $b_1$, $b_2$) are coefficients associated with the joint angles $\theta$ and $\psi$, respectively. Note that these equations compute changes in muscle length as a percentage of segment length from some given reference angle. The reader is referred to Bobbert and colleagues’ (1986), Grieve and colleagues’ (1978), and Visser and colleagues’ (1990) papers for the actual coefficients.

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