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The influence of cadence and power output on force application and in-shoe pressure distribution during cycling by competitive and recreational cyclists

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The aim of this study was to determine the response of cyclists to manipulations of cadence and power output in terms of force application and plantar pressure distribution. Two groups of cyclists, 17 recreational and 12 competitive, rode at three nominal cadences (60, 80, 100 rev·min⁻¹) and four power outputs (100, 200, 300, 400 W) while simultaneous force and in-shoe pressure data were collected. Two piezoelectric triaxial force transducers mounted in the right pedal measured components of the pedal force and orientation, and a discrete transducer system with 12 transducers recorded the in-shoe pressures. Force application was characterized by calculating peak resultant and peak effective pedal forces and positive and negative impulses. In-shoe pressures were analysed as peak pressures and as the percent relative load. The force data showed no significant group effect but there was a cadence and power main effect. The impulse data showed a significant three-way interaction. Increased cadence resulted in a decreased positive impulse, while increased power output resulted in an increased impulse. The competitive group produced less positive impulse but the difference became less at higher cadences. Few between-group differences were found in pressure, notable only in the pressure under the first metatarsal region. This showed a consistent pattern of in-shoe pressure distribution, where the primary loading structures were the first metatarsal and hallux. There was no indication that pressure at specific sites influenced the pedal force application. The absence of group differences indicated that pressure distribution was not the result of training, but reflected the intrinsic relationship between the foot, the shoe and the pedal.

Keywords: cycling, effectiveness, force, pressure.

Introduction

The mechanics of bicycle pedalling involve the transfer of force from the muscles of the legs through the feet and onto the pedal surfaces. Ruby and Hall (1993) examined in detail the link between the foot–pedal interface and loads at the knee joint. They reported that design modifications of the pedal could lead to a reduction of forces at the knee joint. Also worthy of examination is the relationship between the load on the structures of the foot and on the pedal.

The forces applied to the pedal have been examined by many researchers (Hoes et al., 1968; Davis and Hull, 1981; Lafortune and Cavanagh, 1983; Lafortune et al., 1983; Broker and Gregor, 1990; Sanderson, 1991). Across a range of rider characteristics and riding conditions, a profile of the force application during steady-rate cycling has been generated. This profile showed that, for steady-rate riding, the resultant pedal force rose to a maximum around 110° after top dead centre and decayed by bottom dead centre but not to zero. The recovery phase, or second 180° of crank rotation, was often characterized by a force that did not contribute to the positive angular impulse, unless the power output demands were very high.

Although the foot is the link between the bicycle and the rider, only limited research has been published on the relationship between pedal force and in-shoe pressure distribution. Sanderson and Cavanagh (1987) showed that there were some changes in the distribution of pressure when cadence was increased from 45 to 90 rev·min⁻¹ at a power output of 400 W. These changes...
indicated a medial shift in the peak pressures as the cadence was increased. Hennig and Sanderson (1995) examined individual sites of pressure application for a series of increasing power outputs at 80 rev·min⁻¹. They showed elevated fifth metatarsal and first metatarsal pressures with respect to the second, third and fourth metatarsals, which gave the appearance of a ‘transversal forefoot arch’. A limitation of that study was that the pressures were determined at only one cadence.

The aim of the present study was to determine the effect that manipulations of power output and cadence have on pedal forces and in-shoe pressure distribution among two groups of cyclists. We were interested in whether there was a relationship between variations in pedal force and plantar pressures that could be attributed to some global response of the cyclists to cadence and power output manipulations. Lower limb injuries have been associated with foot position (Francis, 1988), which may be affected by in-shoe pressure distribution. By manipulating cadence through gearing, cyclists strive for an optimal pattern of force application. It has been suggested that the perception of loading may play a role in this selection of cadence (Pandolf and Noble, 1973).

As plantar-pressure distribution changes in response to power output have been noted (Hennig and Sanderson, 1995), combining cadence changes with power output manipulations would add to our understanding of cycling mechanics. By using experienced and inexperienced cyclists, we were able to determine whether this response could be attributed to practice, as in experienced cyclists, or whether it arose as a consequence of the mechanical relationship between the foot and the bicycle pedal.

Methods

Participants

The participants were 29 males, who were grouped into one of two categories based upon their cycling experience. The competitive group (n = 12; mean ± s: age 26 ± 8 years, body mass 68 ± 10 kg, height 173 ± 8 cm) comprised cyclists with a current category 1 or higher competitive licence, according to Canadian Cycling Association standards. The recreational group (n = 17; age 25 ± 3 years, body mass 75 ± 11 kg, height 175 ± 8 cm) comprised cyclists from the local community who were active but not competitive. All participants signed an informed consent form before participating in the study.

Instrumentation

An instrumented bicycle mounted on a Velodyne stationary trainer (Schwinn, Chicago, IL) was used to record data in this study. The bicycle crank was instrumented such that the vertical position, or top dead centre, of the right crank, as well as continuous crank position, was recorded as an analog signal. The right pedal was instrumented with two Kistler type load cells (9251 A), which recorded the vertical and two components of the shear force as described by Broker and Gregor (1990). The left pedal was constructed in a similar fashion, but without the associated electronics and matched in mass to the right pedal. A pressure distribution unit (PD-16, halm, Frankfurt, Germany) with 12 discrete piezoceramic transducers was used to measure the pressures under the central heel, lateral midfoot, each of the metatarsal heads and all toes. The physical properties of the transducers have previously been reported (Hennig et al., 1982). The transducers have an active area of 3 mm² and a thickness of 2 mm. After palpation of the plantar foot, the transducers were individually fastened for each participant under the specific anatomical regions using surgical tape. This method has previously been used for the measurement of in-shoe pressures during various sports (Hennig and Milani, 1989), and there was no evidence of movement of the transducers upon removal. Because the presence of discrete transducers under the foot may have an effect on riding style, a questionnaire regarding the feel of each transducer to the rider was administered. All but one participant reported that the transducers had no effect on their riding performance or style. Most participants were unable to feel the sensors under their feet, even when they were asked to concentrate on the presence of the transducers. These observations confirm findings with similar transducers for sports with large ground reaction forces (e.g. triple jumping; Milani and Hennig, 1992).

Protocol

Upon arrival in the laboratory, height, weight and age were recorded and each participant arranged the seat position to match their own set-up. The frame size and crank length were in most cases the same as the rider’s own bike. In cases where these were different, differences were small and deemed to have no effect on the variables of interest. Position during riding was standardized, with the cyclists’ hands on the drops. They were instructed to remain seated throughout each ride. They wore conventional cycling shoes of identical manufacture (Nike, Beaverton, USA), which were attached to pedals with toeclips and straps during the rides. After a warm-up ride, the cyclists were requested to perform each of 12 rides presented in random order. There were three pedalling rates (60, 80 and 100 rev·min⁻¹) and four nominal power outputs (100, 200, 300 and 400 W). For each ride, there was a brief
warm-up and then a ride of about 2 min duration. The data were recorded for 12 consecutive pedalling cycles (from top dead centre to top dead centre) during the final 30 s of this interval. The participants rested between rides for at least 2 min to ensure an adequate recovery.

Data collection and evaluation

All data were sampled through a 12-bit data acquisition system at a sample rate of 1000 Hz. Cadence was presented to the rider via a digital display on the handlebars. Power output was set by the Velodyne, but actual rider power output was determined from the right pedal force measurements, assuming symmetry of force application from both legs. Sanderson (1990) reported that force asymmetry was insensitive to manipulations of cadence and power output. The values of crank and pedal angle, normal and antero-posterior shear forces, and plantar pressures were averaged across the 12 consecutive revolutions for each participant and each condition. Force application was evaluated by examining the force-time curves and quantified by determining the peak resultant effective component of the resultant force, positive and negative impulses. The effective component of the resultant force is that component perpendicular to the crank axis (Sanderson, 1990). For the assessment of pressure distribution, the peak pressure and relative load were determined. Relative load on each pressure transducer was calculated as the percent contribution of the force-time integral from each transducer with respect to the total force integral across all 12 transducers. These data showed whether there were relative shifts in the proportion of loading on the foot structures as the power output and cadence were manipulated.

The data were then analysed in a one between-group (rider) and a two-factor (power and cadence) analysis of variance with repeated measures on both factors. Systat 5.0 (SAS Institute, Cary, NC) was used to perform the analysis, as it accommodated the unequal group sizes in this study. In the case of a significant F-ratio, follow-up was performed using the Scheffé test. All statistical inferences were made on the basis of $P < 0.05$. Dependent variables included peak resultant and effective pedal force, positive or propulsive impulse, negative or retarding impulse, peak plantar pressure and percent relative load pressure.

Results

The between-group differences for maximum resultant force were not significant ($F_{127} = 0.001, P = 0.93$); similarly, there were no significant between-groups differences ($F_{127} = 0.348, P = 0.54$) in peak effective force. For both peak resultant and peak effective force, there was a significant interaction between cadence and power output ($F_{11297} = 351, P < 0.0001; F_{11297} = 464, P < 0.0001$, respectively). The main effect of increasing power output, while maintaining a constant cadence, was to significantly increase the maximum force (both effective and resultant) applied to the pedal. Conversely, when cadence was increased from 60 to $100 \text{ rev.min}^{-1}$ and power output was held constant, a significant decrease in both peak resultant and effective force occurred. The effective force data plotted as a function of changes in power output and cadence manipulations with respect to crank angle are summarized in Figs 1a and 1b, respectively.

There was a significant three-way interaction among groups, cadence and power output for positive and negative impulses ($F_{11297} = 5.12, P < 0.001; F_{11297} = 4.43, P < 0.001$, respectively). The cell means are plotted
Sanderson et al. in Figs 2a and 2b. With increased power output, there was an increase in the positive impulse for both groups. The increased power output also led to a negative impulse that became less negative. Increased cadence resulted in a decrease in the positive impulse and an increase in the negative impulse for most conditions. There was a clear between-groups trend – the recreational group generated a greater positive impulse (1115 N·s) for each condition and a larger negative impulse (−77 N·s) for each condition than the competitive group (772 N·s and −47 N·s, respectively).

To analyse the pressure data, an analysis of variance was applied to the data for each transducer location for the two dependent variables, peak pressure and percent relative load. These results are summarized in Tables 1 and 2. There were two general observations from these data: in only one case was there a significant group effect – the first metatarsal head location – and there was a significant cadence × power output interaction. These results are consistent with changes in the force data. With increased power output, the peak pressure increased, whereas with increased cadence, the peak pressure decreased. The relative loading, on the other hand, showed little change over most of the foot, with the exception of the first metatarsal and hallux regions. In the latter, relative load increased significantly with increased power output.

For each group, the peak and percent relative load pressures were collapsed across selected locations, as described by Hennig and Sanderson (1995), and plotted as a function of cadence and power output manipulations. These regions were selected to facilitate comparison with the earlier publications. Furthermore, the regions were selected because of the similarly of their response to the independent variables. The peak pressure and percent relative load under the first metatarsal and hallux were higher than in any of the other regions; as they were the major loading structures, they were loaded as a unit. The results for peak pressure

Fig. 2. Mean impulse (± s) for (a) positive and (b) negative impulse averaged for each group and condition. (○) competitive group, (●) recreational group.

Table 1. The F-scores and associated probabilities (P) for each analysis of variance applied to the percent relative load pressure data for the group effect, the cadence × power interaction and the cadence × power × group interaction

<table>
<thead>
<tr>
<th>Transducer location</th>
<th>Group F</th>
<th>Group P</th>
<th>Cadence × power F</th>
<th>Cadence × power P</th>
<th>Cadence × power × group F</th>
<th>Cadence × power × group P</th>
</tr>
</thead>
<tbody>
<tr>
<td>1st metatarsal</td>
<td>12.22</td>
<td>0.002*</td>
<td>8.748</td>
<td>0.000*</td>
<td>3.016</td>
<td>0.001*</td>
</tr>
<tr>
<td>1st toe</td>
<td>1.367</td>
<td>0.253</td>
<td>17.87</td>
<td>0.000*</td>
<td>0.153</td>
<td>0.999</td>
</tr>
<tr>
<td>2nd metatarsal</td>
<td>0.888</td>
<td>0.354</td>
<td>1.138</td>
<td>0.331</td>
<td>1.157</td>
<td>0.317</td>
</tr>
<tr>
<td>2nd toe</td>
<td>0.809</td>
<td>0.377</td>
<td>0.625</td>
<td>0.807</td>
<td>1.023</td>
<td>0.426</td>
</tr>
<tr>
<td>3rd metatarsal</td>
<td>0.135</td>
<td>0.716</td>
<td>2.581</td>
<td>0.004*</td>
<td>1.628</td>
<td>0.090</td>
</tr>
<tr>
<td>3rd toe</td>
<td>1.028</td>
<td>0.320</td>
<td>1.693</td>
<td>0.074</td>
<td>0.094</td>
<td>0.503</td>
</tr>
<tr>
<td>4th metatarsal</td>
<td>0.295</td>
<td>0.592</td>
<td>2.948</td>
<td>0.001*</td>
<td>0.682</td>
<td>0.756</td>
</tr>
<tr>
<td>4th toe</td>
<td>1.583</td>
<td>0.219</td>
<td>1.846</td>
<td>0.046*</td>
<td>1.016</td>
<td>0.432</td>
</tr>
<tr>
<td>5th metatarsal</td>
<td>0.211</td>
<td>0.650</td>
<td>13.67</td>
<td>0.000*</td>
<td>0.205</td>
<td>0.997</td>
</tr>
<tr>
<td>5th toe</td>
<td>3.069</td>
<td>0.091</td>
<td>18.76</td>
<td>0.000*</td>
<td>0.531</td>
<td>0.882</td>
</tr>
<tr>
<td>Midfoot</td>
<td>1.980</td>
<td>0.171</td>
<td>15.66</td>
<td>0.000*</td>
<td>0.806</td>
<td>0.634</td>
</tr>
<tr>
<td>Heel</td>
<td>0.825</td>
<td>0.372</td>
<td>12.45</td>
<td>0.000*</td>
<td>1.153</td>
<td>0.320</td>
</tr>
</tbody>
</table>

*Significant at P < 0.05.
Table 2. The $F$-scores and associated probabilities ($P$) for each analysis of variance applied to the peak pressure data for the group effect, the cadence $\times$ power interaction and the cadence $\times$ power $\times$ group interaction

<table>
<thead>
<tr>
<th>Transducer location</th>
<th>Group</th>
<th>$F$</th>
<th>$P$</th>
<th>Cadence $\times$ power</th>
<th>$F$</th>
<th>$P$</th>
<th>Cadence $\times$ power $\times$ group</th>
<th>$F$</th>
<th>$P$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1st metatarsal</td>
<td></td>
<td>11.50</td>
<td>0.002*</td>
<td>48.11</td>
<td>0.000*</td>
<td>6.555</td>
<td>0.000*</td>
<td></td>
<td></td>
</tr>
<tr>
<td>1st toe</td>
<td></td>
<td>0.001</td>
<td>0.982</td>
<td>133.2</td>
<td>0.000*</td>
<td>0.174</td>
<td>0.999</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2nd metatarsal</td>
<td></td>
<td>0.028</td>
<td>0.869</td>
<td>12.72</td>
<td>0.000*</td>
<td>0.926</td>
<td>0.515</td>
<td></td>
<td></td>
</tr>
<tr>
<td>2nd toe</td>
<td></td>
<td>0.012</td>
<td>0.914</td>
<td>37.01</td>
<td>0.000*</td>
<td>0.564</td>
<td>0.857</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3rd metatarsal</td>
<td></td>
<td>0.117</td>
<td>0.735</td>
<td>50.41</td>
<td>0.000*</td>
<td>1.043</td>
<td>0.408</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3rd toe</td>
<td></td>
<td>0.129</td>
<td>0.723</td>
<td>32.74</td>
<td>0.000*</td>
<td>0.616</td>
<td>0.815</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4th metatarsal</td>
<td></td>
<td>1.648</td>
<td>0.210</td>
<td>40.99</td>
<td>0.000*</td>
<td>0.605</td>
<td>0.825</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4th toe</td>
<td></td>
<td>0.073</td>
<td>0.790</td>
<td>24.90</td>
<td>0.000*</td>
<td>0.892</td>
<td>0.549</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5th metatarsal</td>
<td></td>
<td>0.048</td>
<td>0.828</td>
<td>24.44</td>
<td>0.000*</td>
<td>0.259</td>
<td>0.992</td>
<td></td>
<td></td>
</tr>
<tr>
<td>5th toe</td>
<td></td>
<td>1.249</td>
<td>0.274</td>
<td>26.07</td>
<td>0.000*</td>
<td>0.307</td>
<td>0.984</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Midfoot</td>
<td></td>
<td>0.028</td>
<td>0.869</td>
<td>19.84</td>
<td>0.000*</td>
<td>0.872</td>
<td>0.569</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Heel</td>
<td></td>
<td>1.494</td>
<td>0.232</td>
<td>48.85</td>
<td>0.000*</td>
<td>0.493</td>
<td>0.907</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

*Significant at $P < 0.05$.

are presented in Figs 3a and 3b and those for percent relative load in Figs 4a and 4b. The significant three-way interaction data for the first metatarsal location are plotted in Figs 5a and 5b.

**Discussion**

The aim of this study was to assess the response of two groups of cyclists to manipulations of cadence and power output in terms of force application and in-shoe plantar pressure distribution. We were interested in whether there was a relationship between variations in pedal force and plantar pressures that could be attributed to some global response of the cyclists to cadence and power output manipulations. By using experienced and inexperienced cyclists, we were able to determine whether this response could be attributed to experience or whether it arose as a consequence of a mechanical relationship between the foot and the pedal.

**Power output effects**

As the power output demands increased, the riders responded by increasing the forces applied. This is shown very clearly in Fig. 1a, where there appears to be a consistent increase in the peak effective force with increased power output. There was also a decrease in the peak negative force with increased power output. Positive impulses increased with increased power output. The increase depended on the cadence, with the largest increase seen at the lowest cadence. With increased power output, there was also a reduction in the negative impulse; that is, as the intensity of the rides increased, the riders in both groups generated less of a

![Fig. 3. Mean peak pressure averaged across selected transducer locations for each group. (a) Means plotted as a function of changes in power output; (b) the changes plotted as a function of changes in cadence. The open symbols are for the competitive group and the closed symbols are for the recreational group. MH = metatarsal head.](image)
retarding force. This reflected the strategy of the rider to improve the effective application of force by reducing the need for the propulsive leg to overcome the recovery leg.

For the most part, the pressure data followed the changes indicated by the resultant and effective forces. As the power output demands increased, the response of the riders was to apply greater forces through the pedal, resulting in increased in-shoe pressures. Examination of the peak pressures (Fig. 3a) show a general increase in pressure at all transducer locations. However, when examining the percent relative load (Fig. 4a), it is clear that the distribution of pressure did not change substantially. Only under the first metatarsal and hallux was there a demonstrative increase. That the percent relative load on the other regions decreased with increased power output, indicated that the rider responded to the increased demands with an anterior–medial shift in applied force. That is, there was an increased reliance on the first metatarsal and hallux as load-bearing structures. This effect was consistent in both groups and thus did not appear to represent the result of riding experience. This concentration of pressure was consistent with the medial loading described by Sanderson and Cavanagh (1987) and by Hennig and Sanderson (1995). Whereas Hennig and Sanderson (1995) looked at only one cadence (i.e. 80 rev·min⁻¹), we have shown that the effect appears regardless of cadence in the range 60–100 rev·min⁻¹; we suggest that it is a robust effect and independent of cycling experience.

For the pressure data, only the transducer located beneath the first metatarsal showed a three-way interaction. These results, plotted in Figs 5a and 5b, show that the competitive group had consistently higher peak and percent relative load pressure than the recreational group. The reason for this was not clear. The peak pressures and percent relative loads under the other transducer locations were not significantly different, and post-hoc assessment did not reveal a consistent trend to explain these differences beneath the first metatarsal and hallux.

Our results do not show the longitudinal or transverse arch suggested by Hennig and Sanderson (1995). The
loading on the fifth metatarsal was relatively low and did not change in response to cadence and power output manipulations. The pressure beneath the hallux and first metatarsal appeared to be sensitive to manipulations of cadence and power output, whereas the pressure remained almost uniform beneath the other regions of the foot. The manipulations in power output resulted in a shift in the relative load, such that the reliance on the first ray increased with increased power output and decreased with increasing cadence. This supports and extends the conclusions of Hennig and Sanderson (1995) and indicates that there was a medial and anterior shift in load-bearing as power output demands were increased. This effect was independent of the grouping of the riders.

The higher power output and lower cadence conditions were deemed similar to a long hill climb. Because the riders remained seated and the bike was fixed to the Velodyne, the riders were limited in their ability to use body and bike motion as they rode. Such body and bike motion – rocking from side-to-side – might have resulted in higher pressures than reported here or perhaps a lateral shift in the pressure distribution. The present experimental conditions did not allow such an evaluation; it will be the focus of an upcoming publication. We believe that riders generally spend more time seated than standing and that these data are of interest in the broader sense.

**Cadence effects**

The effect of increased cadence, while power output is held constant, has been shown previously to reduce the peak forces applied to the pedal (Sanderson, 1991). The results presented in Fig. 1b are consistent with those earlier observations. Both the resultant and the effective forces decreased consistently with increased cadence, but the decrease was limited to the phase where the forces were positive. During the recovery phase, the peak effective force became more negative with increased cadence. Furthermore, the range of crank angles where the effective force was negative increased with increased cadence. Computing the linear impulse allowed us to quantify this effect. With the exception of the 100 W condition, increased cadence resulted in an increase in the negative impulse. This increased negative impulse demanded an increase in the positive impulse because the riders were not ‘pulling up’ on the pedal and thus the contralateral leg had to do some work to raise the recovery leg. The exception in the 100 W condition is puzzling. However, all cyclists commented on the difficulty of maintaining a smooth stroking pattern in this high cadence, low power output condition. Sanderson and Cavanagh (1989) showed that recovery forces can be manipulated; perhaps this unique combination resulted in changes in force application because of the rider’s need to balance himself on the bike. On the other hand, this effect could represent a basement effect in terms of force application. There is a minimum to which force can be reduced and this would be evident in the low power, high cadence condition.

Although there were no significant differences between the groups for forces or impulses, there was a clear trend for the recreational group to generate a larger negative impulse. To overcome this, this group had to generate a larger positive impulse and, in doing so, a large effective and resultant force. There are two probable contributing factors. First, the cyclists in the recreational group were somewhat heavier than in the competitive group, although this difference was not significant ($P = 0.294$). This increase in size could have resulted in a larger retarding or downward push on the pedals during recovery as a consequence of the larger inertial effect of the legs. The second factor could well be attributed to a skill difference between the groups. The competitive group would have learned to minimize this braking or retarding force to achieve a more economical motion. Perhaps the difficulty of maintaining a smooth stroke at the high cadences was more readily achieved by simultaneously pushing down in the recovery phase to maintain balance in the upper body.

When a rider applies a negative effective force, the opposite leg must overcome that force during its propulsion phase to maintain the riding conditions. That is, the leg in propulsion pushing down must lift the leg in recovery. A reduction of the recovery leg forces impacts on the propulsive leg forces, such that a smaller component of the propulsive force would be required to raise the recovery leg. This was noted by Sanderson and Cavanagh (1989) when they manipulated the recovery forces through a training regimen. By reducing the recovery forces to near zero, they revealed a reduction of proportional magnitude in the propulsive force. The riders in the present study responded similarly.

In a parallel pattern, the in-shoe peak pressures decreased as the cadence was increased (Fig. 3b). This decrease, however, was not evenly distributed over the surface of the foot (Fig. 4b). The largest differences were in the first metatarsal and toe region, with the smallest decreases beneath the heel, midfoot and fifth metatarsal regions. In fact, the relative loadings show that the first metatarsal and hallux regions were responsible for most of the unloading, while there was a slight increase or no change in the relative loading of the other locations. Comparisons between groups showed that the competitive group had higher pressures, both relative and peak, over this first metatarsal and hallux region than the recreational group.

Thus, the decreased relative load with increased cadence may help to determine whether riders will
pedal at a high or low cadence. In competitive cycling, the optimal cadence has been reported to be much higher than expected by an examination of oxygen uptake data. If pedal forces were high and cadence were low, the eversion of the foot with inward rotation of the tibia through the cycle would lead to this medial loading pattern. Reduction of these forces may be one means by which the rider reduces the load on the knee, even though this may result in higher oxygen costs. Pedalling at these different cadence and power output combinations leads to the supposition that there may be differing pedalling strategies. For example, riding at 60 rev·min⁻¹ at 400 W would lead to a localized fatigue, whereas pedalling at 100 rev·min⁻¹ at 200 W may lead to more central effects. The results presented here show a consistent pattern of change in pedal force and pressure distribution, and suggest that the limited degrees of freedom offered by the conventional bicycle determined the response of the riders.

Increased medial loads at the foot may lead to increased tibial torsion via inversion (Francis, 1988) and subsequent knee pain (Francis, 1988) because the foot cannot pronate on the pedal without longitudinal twisting of the shoe, which itself provides an increased resistance. Therefore, high loads at low cadences may lead to a tendency for pronation that manifests itself at the knee joint as chondromalacia patellae (Pruitt, 1988), resulting from torque about the tibial axis. This is not to suggest that foot position determines cadence, but that it might be a factor in a complex series of interactions which play a role in determining cycling motions. In a study of foot orientation, no suggestion was made that manipulating the orientation of the foot had an impact on cadence (Sanderson et al., 1994).

It appears that cyclists, regardless of their riding experience, respond to increased intensity or power output by two strategies: increasing the normal force component of the pedal force and altering its orientation during the recovery phase to become more effective. Although the recovery forces were small, they could nonetheless accumulate to become a significant factor in terms of the rider’s overall performance, be it competitive or recreational. Medial plantar foot loading, which was probably linked to the amount of foot pronation, increased relatively with increased power output and decreased with increased cadence. A reduction in foot pronation – as shown by reduced first metatarsal loading with increased cadence – to limit knee forces may contribute to less effective cycling with higher than optimal cadences. What is now required is a measure of the internal and external rotation of the tibia in response to similar variations in cadence and power output. The present results offer further insights into riders’ response to variations in the riding environment and form the basis for future work.

References


