PEDAL FORCE DIRECTION CONTROL IN CYCLING

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The purpose of this study was to measure the pedal force magnitude and direction in the power phase of cycling for increasing force application in unconstrained (force direction was not constrained) and for constrained (force direction had to be perpendicular to the crank) conditions. Participants exerted forces on the pedal while the resultant force magnitude and direction, and the associated electromyographic activity of major lower limb muscles, were measured. Force direction for the constrained situation was maintained with a single muscle synergy across all levels of effort, suggesting that force direction is a strong regulator of muscle synergies. In contrast, for the unconstrained condition, muscle synergies switched to an ankle extensor strategy with muscular effort, and force direction typically changed significantly, suggesting that increasing muscle effort is associated with variable muscle synergies. We speculate that with increasing muscle effort, the preferred muscle synergies take advantage of the functional capabilities of the synergistic muscles.

KEYWORDS: muscle synergies, muscle coordination, force-sharing, distribution problem, cycling, motor control

INTRODUCTION: Cycling performance is determined by the torque and power exerted by the leg muscles. The torque exerted by the leg muscles, in turn, depends on the length of the crank and the magnitude and direction of the forces exerted on the pedals. The crank length is constant and typically 165-172 mm (Yoshihuku & Herzog, 1990). The pedal force direction is important as only forces perpendicular to the crank contribute to the propulsion of the bicycle-rider system. Therefore, it has been suggested that athletes should apply pedal forces in a direction perpendicular to the crank (Cavanagh & Sanderson, 1986).
In a recent study, aimed at identifying the force potential of athletes when exerting forces on the pedal during the power phase of cycling without constraining the force direction (unconstrained test), and while constraining the force direction to be perpendicular to the crank (constrained test), we found that the force component perpendicular to the crank was substantially greater in the unconstrained compared to the constrained testing. We speculated that forces in the unconstrained case were greater because subjects could use all leg extensor muscles to their full potential. In contrast, in the constrained case, we found that the leg extensor muscles were not fully activated, presumably because they would change the direction of the resultant pedal force away from the perpendicular direction.
In this previous study, all testing was done using maximal effort muscle contractions. However, human leg extensor muscles have different force capabilities, different sizes, different moment arms, and different fibre types. Therefore, one would expect that the force sharing between muscles depends on the effort, as observed for force sharing in leg muscles cats (Walmsley et al. 1978, Herzog & Leonard, 1991). In cats, soleus forces dominate the gastrocnemius forces for standing still and slow walking, while the reverse is true for running and jumping (e.g. Walmsley et al, 1978; Kaya et al. 2006).
In cycling, the direction of force application is crucial for performance. Furthermore, cycling is primarily done at submaximal levels. Therefore, we wanted to know if the force sharing among leg extensor muscles changes for different efforts, and how effort may affect the direction of force application. Therefore, the purpose of this study was to measure the force magnitude and direction in the power phase of cycling for increasing force application in unconstrained and for constrained conditions. We hypothesized that for the constrained situation, muscular synergies would be maintained. Conversely, for the unconstrained situation, we expected changes in force direction with increasing efforts of pedal force application, indicating changes in muscle synergies.

METHODS: Subjects (n=4, to date) were asked to exert pedal forces on a bicycle at five crank angles (30, 60, 90, 120 and 150° from top dead centre). Pedal forces were applied statically for two conditions; constrained and unconstrained. For the constrained conditions, subjects were asked to push against the pedal in a direction perpendicular to the crank arm, and visual force directional feedback was provided. For a successful trial, subjects had to maintain the force direction within 5° from perpendicular to the crank throughout the constrained test. For the unconstrained condition, subjects were not given any instructions regarding the force direction. Participants were asked to increase forces from zero to maximal effort over a period of about 10s for both conditions (Figure 1).

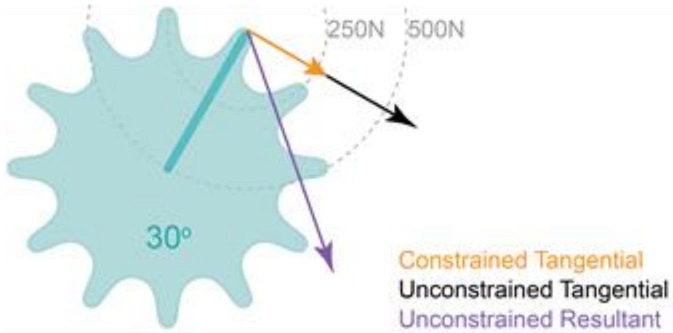

Figure 1: Unconstrained resultant force, corresponding tangential component of the resultant force, and constrained tangential force applied statically on the pedal at a 30° crank angle. Note the substantially greater force and tangential force component for the unconstrained compared to the constrained condition.

Pedal forces were measured using a velo® ergoselect bicycle ergometer (Ergoline, Bitz, Germany) using "I-Crank" software (250 Hz; Sensix, Poitiers, France). Surface EMG (Kendall™ 100 Foam Electrodes, Covidien, Mansfield, USA) activity was measured from the tibialis anterior (TA), vastus lateralis (VL), rectus femoris (RF), gluteus maximus (GMAX), biceps femoris (BF), gastrocnemius medialis (GM), and soleus (SO). EMG signals were pre-amplified (Biovision, Wehrheim, Germany) and recorded at 2000 Hz, using WinDaq software (DATAQ® Instruments, Akron, Ohio). The EMG data of all seven muscles were band pass filtered (first order, recursive Butterworth, 10-500 Hz) and rectified. Next, the Root Mean Square (RMS) values of a 500ms moving window were calculated for each trial. Finally, to allow for comparison between the constrained and non-constrained conditions, EMG data were normalised relative to the maximum root mean square (RMS) value produced by each subject across all test conditions. Data processing and analysis were conducted with Matlab (R2016a, The Mathworks, Natik, MA, USA). Continuous 2D force magnitude and direction were recorded and graphed throughout the experimental trials. Normality of the data was assessed using Kolmogorov-Smirnov testing and differences in force and EMG between constrained and non-constrained conditions were determined using Wilcoxon testing for paired samples. In order to account for multiple post-hoc comparisons, a Bonferroni correction was performed with a level of significance for each test of $\alpha = 0.01$.

RESULTS: Maximal effort forces for the constrained condition were substantially smaller than for the unconstrained condition (821 vs 334 N, 957 vs 589 N; 930 vs 527 N; 873 vs 301 N; and 631 vs 204 N for crank angles of 30, 60, 90, 120 and 150°, respectively). Furthermore, the force component perpendicular to the crank for the unconstrained conditions was always greater for the unconstrained compared to the constrained condition (e.g. Figure 1).
When increasing forces from zero to maximal effort for the constrained condition, pedal force direction remained virtually constant, as designed (Figure 2). For this condition, EMG signals of all muscles increased “linearly” and “in proportion” to each other (Figure 2). In the

unconstrained tests, pedal force direction changed with effort (Figure 3), and EMG signals increased non-linearly and disproportionally. When the direction of pedal force application changed, so did the relative EMG signals of the leg muscles.

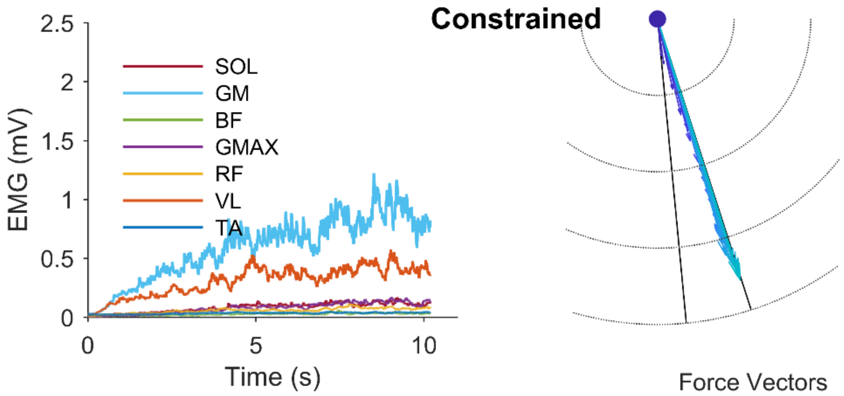

Figure 2: Exemplar EMG activity of seven lower limb muscles and corresponding pedal forces (magnitude and direction) for the constrained condition at a 60° crank angle. Note the small variation in force direction and the "linear" and "proportional" increase in EMG activity.

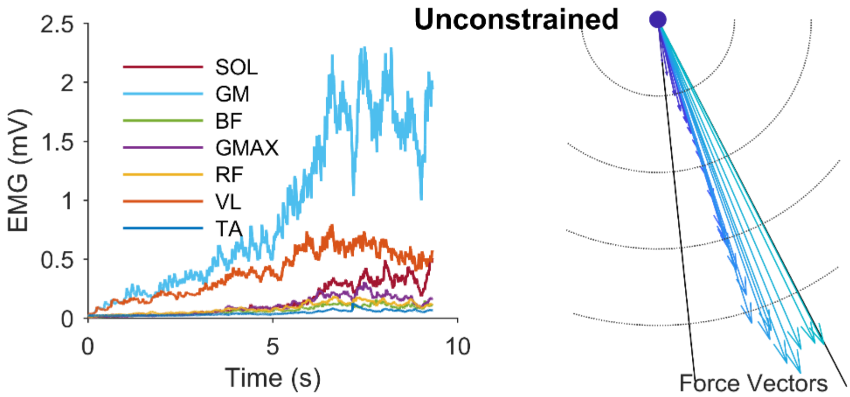

Figure 3: Exemplar EMG activity of seven lower limb muscles and corresponding pedal forces (magnitude and direction) for the unconstrained condition at a 60° crank angle. Note the much greater variation in pedal force direction, and the sudden "non-linear" and "non-proportional" increase in some EMGs relative to others when force direction changed at about 5 s.

DISCUSSION: It has been argued that cyclists should apply pedal forces perpendicular to the crank, as any component parallel to the crank does not contribute to propulsion (e.g. Cavanagh & Sanderson, 1986). However, cyclists typically apply pedal forces with a distinct radial component (e.g. Kautz et al., 1991; Bini et al. 2013). We found that applying a force perpendicular to the crank results in a vastly decreased force magnitude compared to conditions without restriction on the force direction. For a given lower limb configuration relative to the crank, a given muscle will produce a force in a given direction. Increased activation of that muscle will increase the magnitude but not the direction of the pedal force. A change in pedal force direction can only be achieved by involving a second muscle (Liang and Brown, 2014). Constraining the pedal force to be perpendicular to the crank may limit muscle activation and force. This theory was supported in previous work where maximal effort constrained testing was associated with decreased EMG activities compared to the unconstrained condition. Previous studies revealed that pedalling rate also affects the direction of pedal force application (Sanderson, 1991). This result was explained with changes in the inertial forces of the rotating crank with changing angular speeds. However, different muscles have different force-velocity characteristics that are affected by pedalling speed. Therefore, it is perceivable that the force potential among muscles changes for different speeds of pedalling, thus affecting the relative contributions of the muscles to the pedal force, which, in turn, results in changes in the direction of the pedal force.

In our study, we eliminated inertial effects and changes in force-velocity relationships by executing all experiments statically. For the constrained situation, we found that the force direction was maintained, as required by protocol, and this was associated with an essentially proportional increase in EMG activities across muscles (Figure 2). This result suggests that when the force direction is constant, the relative contributions of muscles stays proportional, independent of the effort, and there is no change in muscle synergy. Although this result may appear trivial, a “constant” force direction can be achieved with different muscle synergies, but that was not done here. Rather, subjects maintained a given synergy across all levels of effort. This result suggests that once a muscle synergy was chosen for the constrained experiment, it was difficult to change that strategy with increasing effort (pedal force magnitude).

In contrast, for the unconstrained experiments, the resultant pedal force direction changed significantly with increasing levels of effort (Figure 3). At the instant when pedal force direction changed, so did the relative EMG signal magnitudes, indicating that the change in force direction was associated with a change in muscle synergies. Therefore, it appears that depending on the magnitude of the resultant pedal force, the force-sharing strategy among the synergistic muscles of the lower limb changed as well. This result is in agreement with studies in which muscle force synergies were directly measured, and where increases in “effort” were associated with changed relative contributions of muscle forces. This was illustrated convincingly in force sharing patterns published for the cat ankle plantar flexor muscles. For example, for standing still, soleus forces were found to be great while gastrocnemius forces were minimal, while on the other end of the spectrum, for paw shaking, scratching and jumping, gastrocnemius forces were high, while soleus forces were low (often zero) (e.g. Smith et al. 1985). It will be of interest to identify which muscles are recruited to greater extents for low level and high level efforts of pedal force application, thereby gaining new insights into the force sharing patterns and muscle synergies in human movement, and how they may depend on the control of the direction of external forces.

CONCLUSION: We conclude from the results of this study that maintaining external force direction in the presence of increasing muscular efforts is accomplished with a single muscle synergy, despite different available possibilities. In contrast, increasing the level of muscular effort in the absence of direction control is associated with changing muscle synergies and changing external force directions, presumably to take full advantage of the changing functional capacity and properties of muscles with increasing levels of effort.

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