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Research article

Assessment of Three-Dimensional Trunk Kinematics and Muscle Activation during Cycling with Independent Cranks

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Abstract
Independent cranks (IC) are recently introduced bicycle cranks that are decoupled; therefore allowing each leg to pedal independent of the other. Despite this introduction, limited research has been conducted assessing biomechanical changes when cycling with IC. Therefore, the purpose of this study was to evaluate and compare trunk kinematics and surface electromyography (sEMG) during IC and normal crank (NC) cycling during a graded exercise test to volitional fatigue. Ten healthy, physically active men performed two tests (IC and NC) on a cycling ergometer on separate days. 3D motion capture data of the trunk and pelvis and sEMG of the latissimus dorsi, tibialis anterior, gastrocnemius lateral head, rectus femoris, vastus lateralis and the biceps femoris were collected bilaterally. The first 30 seconds (beginning) and the last 30 seconds (end) of each trial were analyzed with respect to external load (beginning vs end), crank type (IC vs NC) side (left vs right), and phase of the pedal cycle (push vs recovery). Mean load at volitional fatigue in NC (351 W) was significantly greater than IC (318 W; $p < 0.001$). As external load increased, there was a similar increase in spine flexion angle in the sagittal plane for both NC (8.2°**)** and IC (4.6°**)**. The NC condition demonstrated significantly greater increase in muscle activation from the beginning to the end than the IC condition in the tibialis anterior, rectus femoris and biceps femoris in the push phase, and the rectus femoris and biceps femoris in the recovery phase. As IC demonstrated less increase in activation, they cause less variation in muscular contraction from beginning to end throughout the full pedal range of motion, yet do not alter gross trunk kinematics. Due to altered muscle activation patterns when cycling with IC, they are proposed as a potentially beneficial training tool to increase training diversity.

Key words: Electromyography, kinematics, Powercranks, graded exercise test, ergometer, fatigue.

Introduction

Cycling has been exposed to technological advances that claim to enhance performance. Normally, with regular bicycle cranks, cyclists push down on one pedal with more force and velocity than the contralateral leg can pull up. In effect, some of the downward leg's force is used to push up the contralateral recovering leg, causing a resistance force (Bini et al., 2013). The introduction of clipless pedals, which securely connect the shoe to the pedal, allow cyclists to apply a pulling force on the leg responsible for the resistance, thereby increasing cycling effectiveness (Mornieux et al., 2008). Independent cranks (IC) are recently introduced bicycle cranks that use a one-way clutch design to decouple the cranks, allowing each leg to pedal independently of one another.

The full pedal cycle can be separated into two phases, the push phase and the recovery (pull) phase. Cycling with IC requires active flexion of the hip and knee, as well as dorsiflexion of the ankle for stabilization in the recovery phase. This is necessary for cyclists to maintain an anti-phase cycling pattern, where the cranks remain 180° apart. IC manufacturers have claimed that the active flexion of the hip and knee facilitates training of specific muscles, which results in decreased resistance to the contralateral leg in the push phase when cycling with normal cranks (NC). Previous research on IC has primarily observed physiological factors by studying changes in power output (Böhm et al., 2008), gross efficiency (Burns et al., 2012; Luttrell and Potteiger, 2003) and oxygen uptake (Burns et al., 2012). Improvements to these physiological variables were absent in all of the aforementioned studies except for Luttrell and Potteiger (2003), who found decreased energy expenditure during a 1-hour submaximal ride after 6 weeks of IC training. However, few studies have observed IC cycling from a biomechanical perspective. Burns et al. (2012) observed muscle recruitment patterns of the vastus lateralis, biceps femoris and gastrocnemius after a five week training period with IC, and found no difference in recruitment patterns upon returning to NC cycling (Burns et al., 2012). Their study only collected sEMG of three left leg muscles, and did not include a hip flexor muscle, which is important to observe due to the active flexion requirement during IC cycling. Muscle activation during submaximal IC cycling was also studied by Hug et al. (2013) for 10 muscles of the left leg. Their study considered hip flexor muscles and it was found that during cycling at a 100 watt work rate, there was a significant increase in muscle activity of the tibialis anterior, gastrocnemius medialis, rectus femoris, biceps femoris, semimembranosus, and tensor fascia latae (Hug et al., 2013) when cycling with IC. Their study, although considering hip and knee flexor activation, only observed muscles unilaterally. Due to the independent nature of the cranks, it is important to observe muscles bilaterally in order to consider any asymmetry between legs; a concern for injury risk due to uneven force distribution (Smak et al., 1999). It is also important to observe muscle recruitment between legs at an increased work rate; a 100 watt load is very light and increased muscle fiber recruitment may be seen at higher loads that are more representative of a work rate experienced in training.

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To the best of the authors' knowledge, there has been limited analysis of cycling kinematics when cycling with IC. Previously, cycling kinematics with NC during fatigue was studied, and it was found that the trunk demonstrated increased anterior flexion at the end of an exhaustive cycling test (Dingwell et al., 2010; Sayers and Tweddle, 2012). Conversely, it was found that there was no effect of workload on trunk angles when fatigue was not a factor (Bini et al., 2016). These findings suggest that trunk kinematics are only altered when there is a fatigue effect, or workload and fatigue effects are combined. The aforementioned studies have observed three-dimensional kinematics with NC. However, no studies have compared three-dimensional trunk kinematics between NC and IC cycling. It is important to evaluate the trunk kinematics of IC cycling, as this may have implications with lumbar spine loading while training. Due to cyclists being in a seated position, there is a constant flexion of the lumbar spine, putting them at risk for low back pain (Callaghan and Jarvis, 1996; Manninen and Kallinen, 1996). Additionally, activities that involve repetitive flexion/rotation are associated with flexion pattern pain disorder (O'Sullivan, 2000). Therefore, increased trunk rotation, lateral flexion, forward flexion, and more importantly, overall range of motion (ROM) in the sagittal plane could increase the risk for low back pain in cyclists.

The goal of this study was to gain a greater understanding of the three-dimensional kinematics of the lumbar spine and bilateral muscle recruitment patterns during a graded exercise test during both IC and NC cycling. It was hypothesized that there would be a greater increase in trunk ROM around all three axes during IC cycling. It was also hypothesized that due to reduced assistance from the contralateral leg in the recovery phase, there would be increased muscle activation in the rectus femoris and biceps femoris during IC cycling compared to NC cycling as external load is increased.

Methods

Experimental approach

Kinematics of the trunk and sEMG of six muscles bilaterally were used to evaluate differences between IC and NC cycling during a graded exercise test.

Participants

Ten healthy, physically active male university students were recruited to participate in this study (Table 1), which was approved by the Nipissing University Research Ethics Board (REB: 100669). All participants provided written and informed consent prior to any data collection. The participants were considered recreational cyclists; any previous cycling experience was for leisure purposes (not competition). As IC have been advertised as a training tool for many sports and rehabilitation purposes, elite cyclists were not included in the participant pool to assess the effect on healthy, recreational cyclists. Participants with any previous experience cycling with IC were excluded from the study in order to accurately assess the acute effects of IC without influence of previous training. Participants were instructed to maintain normal daily routines including exercise and diet.

Instrumentation

An indoor Velotron cycle ergometer (RacerMate Inc., WA, USA) using Velotron Coaching Software was used to control resistance. The participants were fitted into appropriate cycling shoes attached to Look Keo clipless pedals (Look Cycle, NV, USA). Visual feedback of the pedalling cadence was provided on a screen in front of the participant.

An original version of Powercranks (Powercranks, CA, USA) was used for the IC condition. For the NC condition, Shimano 105 (Shimano American Corp, CA, USA) cranks were used. Both cranks were 172.5 mm in length and an identical 130 mm radius Velotron Racermate (RacerMate Inc., WA, USA) 62 tooth chainring was used for both conditions.

Kinematic data of the trunk and pelvis were collected at 300 Hz using a 15 camera motion capture system (Qualisys, Gothenburg, Sweden). The motion of the trunk and pelvis were tracked using a marker cluster on the sacrum and lower back (T_{10} – T_{12} vertebrae). The trunk angle was defined as the position of the trunk relative to the position of the pelvis in order to represent the three-dimensional movement of the lumbar spine. sEMG of the latissimus dorsi (LD), tibialis anterior (TA), gastrocnemius lateral head (GL), rectus femoris (RF), vastus lateralis (VL), and biceps femoris (BF) were collected bilaterally and wirelessly at 3000 Hz (Trigno Delsys, MA, USA). The TA, GL, RF, VL and BF sensors were placed according to SENIAM guidelines (Hermens et al. 2000), while the LD was placed on the largest part of the muscle body found via palpation. sEMG data were synchronized with kinematic data through the use of an external trigger.

Table 1. Participants characteristics. Data are means (±SD).

Age (yrs)	22.3 (2.0)
Height (m)	1.78 (.06)
Body Mass (kg)	81.5 (6.6)
Physical Activity (hrs/week)	9.1 (2.7)
Recreational Cycling Experience (hrs/week)	1.5 (1.6)

Procedure

Participants were supplied with appropriately sized cycling shoes, and were fitted to an indoor cycle ergometer with clipless pedals. Seat height was adjusted so that when the participant placed their heel on the pedal, their knee was completely extended (Bini et al., 2011) so that their knee would be slightly flexed at the bottom of a normal pedal stroke. As handlebar height is a subjective measurement (Silberman et al., 2005), the handlebar height was adjusted to the participant's preference. Once fitted on day 1 of testing, the seat and handlebar height were measured to ensure consistency between testing days. After being fitted, participants completed a warm-up and familiarization session, consisting of two 2.5 minute bouts of IC cycling at a resistance of 1.5 watts per kilogram of body mass, separated by a 30 second rest. No participant continued onto the protocol until they could properly cycle with IC. Following the warm-up and familiarization, participants were instructed to maintain movement by walking or stretching while modifications to crank type, or technological preparations occurred if necessary. Participants then completed a graded exercise test cycling with IC or NC depending on random selection, on two testing days separated by 48 hours. The graded exercise test consisted of 60 RPM cycling, beginning at a resistance of 100 W and increased linearly at a rate of 0.667 W/second; participants were instructed to cycle until volitional fatigue. This test was designed to push participants to maximum cardiorespiratory fatigue before localized muscle fatigue forced the participant to stop, which proved to be an issue when cycling with IC during pilot testing. Pilot testing also revealed that a cadence greater than 60 RPM was noticeably difficult to maintain when using IC, thus 60 RPM was chosen as the ideal cadence.

Data analysis

All kinematic data were processed in Visual 3D (C-Motion, MD, USA), to develop an unconstrained skeletal model of the trunk and the pelvis. All further analyses were performed using custom Matlab scripts (The Math Works Inc., MA, USA)

Mean, maximum, and minimum trunk angles from the first thirty seconds of the graded exercise test (beginning) were compared to the final thirty seconds prior to volitional fatigue (end), both within and between cycling conditions. Flexion/extension, lateral flexion and transverse rotation were the measured trunk movements.

All muscle activity was divided into two cycle phases: push (top center to bottom center of the crank position) and recovery (bottom center to top center of the crank position). The beginning and end of the graded exercise test were defined the same as the kinematic data described above. The sEMG root mean squared (RMS) value was calculated in the push and recovery phases for a 30 second window at the beginning and end of the graded exercise test. The change in RMS from the beginning to the end within both phases was expressed as a percentage increase to be compared between conditions.

Statistical analysis

Each trunk angle parameter was statistically analyzed using a two-way repeated measures ANOVA (crank type*time). RMS percentage increase in the push phase between sides of the body and conditions, was evaluated using a two-way ANOVA (crank type*side). RMS percentage increase in the recovery phase was compared using the same methods as the push phase. Within each crank condition, a two-way ANOVA was used to compare the percent increase between sides of the body and phase of the pedal cycle (phase*side). The maximum load (in watts) reached in each condition was compared using a paired sample T-test. SPSS 23 (IBM Corp., NY, USA) was used to complete all statistical calculations. Any differences were considered significant if the critical significance value of $p < 0.05$ was reached. Effect sizes were calculated and reported as partial eta squared (η^2). Effect sizes greater than 0.14 were interpreted as large, effect sizes, between 0.01 and 0.06 were considered medium and effects lower than 0.01 were considered small.

Results

Resistance

Participants were able to cycle to greater resistance in the NC condition ($p < 0.001$). The mean resistance and standard deviation at volitional fatigue in the NC condition was 351 ± 56 watts, compared to 318 ± 55 watts in the IC condition. The time to volitional fatigue corresponded to durations of 377 ± 85 seconds and 327 ± 83 seconds, respectively.

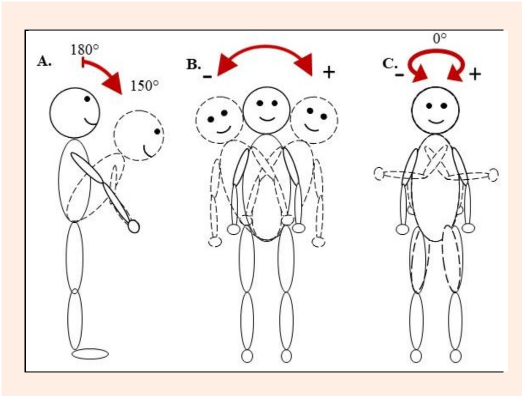

Figure 1. Visual representation of three-dimensional trunk angles. (A) Anterior flexion/extension (B) Lateral flexion left/right (C) Axial rotation left/right.

Trunk kinematics

There was a significant increase in mean sagittal trunk flexion in both conditions ($p = 0.005$, $\eta^2 = 0.606$), at the end compared to the beginning (Table 2). This is represented by a decrease in the measured angle, as 180° represents the trunk aligned with the pelvis, and smaller angles represent forward flexion (Figure 1). There was a similar main effect seen for time when observing maximum trunk angles (flexion ($\eta^2 = 0.815$), extension ($\eta^2 = .576$), lateral flexion left ($\eta^2 = 0.676$), lateral flexion right ($\eta^2 = 0.771$), rotation left ($\eta^2 = 0.551$), rotation right ($\eta^2 = 0.598$), such that ROM increased as resistance increased, in all variables in both conditions ($p < 0.05$). There was a significant interaction seen for condition*time in lateral flexion to the right, with IC showing a larger increase in angle from the beginning to the end ($p = 0.037$, $\eta^2 = 0.398$) (Table 3). There were no significant differences in mean trunk lateral flexion and mean transverse rotation angle between conditions.

Table 2. Mean trunk angle in degrees. Data are means (±SD).

	NC			IC			
Movement Plane	**Beginning**	**End**	**Mean Difference**	**Beginning**	**End**	**Mean Difference**	**η^2**
Sagittal	150.5 (6.4)	143.8 (7.8)	6.7°	148.1 (9.9)	145.4 (6.8)	2.7°	.606 *
Frontal	-1.4 (1.4)	-1.0 (1.4)	-0.4°	-1.6 (2.5)	-1.1 (2.9)	-0.5°	.181
Transverse	-0.5 (2.1)	-0.5 (1.7)	0°	-1.1 (2.0)	-1.0 (2.3)	-0.1°	.008

* significant main effect (time) between beginning and end, η^2 is presented for the main effect of time

Table 3. Mean maximum trunk angle in degrees. Data are means (±SD).

	NC			IC			
Max Angle	Beginning	End	Mean Difference	Beginning	End	Mean Difference	η^2
Flexion	149.2 (6.6)	141.0 (8.2)	8.2	147.8 (7.9)	143.2 (6.7)	4.6	.815*
Extension	151.5 (6.3)	146.7 (7.6)	4.8	149.7 (8.3)	147.4 (6.7)	2.3	.576*
Lateral Flexion Right	6.2 (1.7)	7.2 (1.8)	-1.0	6.4 (2.7)	8.1 (2.7)	-1.7	.771*
Lateral Flexion Left	3.4 (1.4)	5.0 (1.7)	-1.6	2.9 (2.8)	5.0 (3.6)	-2.1	.676*#
Rotation Right	1.8 (2.1)	3.5 (2.5)	-1.7	2.3 (2.1)	3.3 (2.1)	-1.0	.598*
Rotation Left	0.9 (2.0)	2.4 (1.5)	-1.5	0.2 (1.8)	1.6 (2.5)	-1.4	.551*

* significant main effect (time) between beginning and end; # significant interaction (condition*time), η^2 is presented for the main effect of time

Push phase sEMG

RMS percentage change from beginning to end between conditions was significantly greater in the NC condition for the TA ($p = 0.008$, $\eta^2 = 0.557$), RF ($p = 0.002$, $\eta^2 = 0.673$) and BF ($p = 0.032$, $\eta^2 = 0.416$) (Figure 2). There was a significant difference seen between legs in both conditions in the TA ($p = 0.023$, $\eta^2 = 0.452$), VL ($p < 0.001$, $\eta^2 = 0.764$) and BF ($p = 0.008$, $\eta^2 = 0.566$). In this instance there was a greater increase observed in the left leg for the BF, and a greater increase in the right leg for both TA and VL. There was a significant interaction condition*leg only in the TA, showing a greater asymmetry in the NC condition ($p = 0.033$, $\eta^2 = 0.412$). There was no significant difference between conditions in the GL, VL and LD (Figure 2).

Recovery phase sEMG

Results of the recovery phase sEMG are presented in Figure 3. There was significantly greater RMS increase in activation from beginning to end in the RF ($p = 0.002$, $\eta^2 = 0.666$) and BF ($p = 0.024$, $\eta^2 = 0.448$) in the NC condition, compared to IC. There was no significant difference between conditions for the TA, GL, VL, and LD. There was a significant difference between sides for the GL ($p = 0.009$, $\eta^2 = 0.548$), RF ($p = 0.002$, $\eta^2 = 0.682$), VL ($p = 0.006$, $\eta^2 = 0.582$), BF ($p = 0.002$, $\eta^2 = 0.669$), and LD ($p < 0.001$, $\eta^2 = 0.795$), with a greater activation change on the left side for the GL, BF and LD, and right side for RF and VL. There was no significant difference between sides for the TA. There was no significant interaction condition*side in any of the muscles (Figure 3).

Effect of time within conditions between phases

In the NC condition, RF and VL activation significantly increased in the push phase compared to the recovery phase ($p = 0.002$, $\eta^2 = 0.683$ and $p = 0.006$, $\eta^2 = 0.586$ respectively), whereas BF activation was significantly greater in the recovery phase ($p = 0.033$, $\eta^2 = 0.413$). In the IC condition, VL showed a significantly greater increase in the push phase ($p < 0.001$, $\eta^2 = 0.790$). Also in the IC condition, no difference in activation increase was found between the push and recovery phases for the RF and BF, meaning that they increased proportionally in both phases. All other muscles exhibited no significant changes between phases (Figure 4).

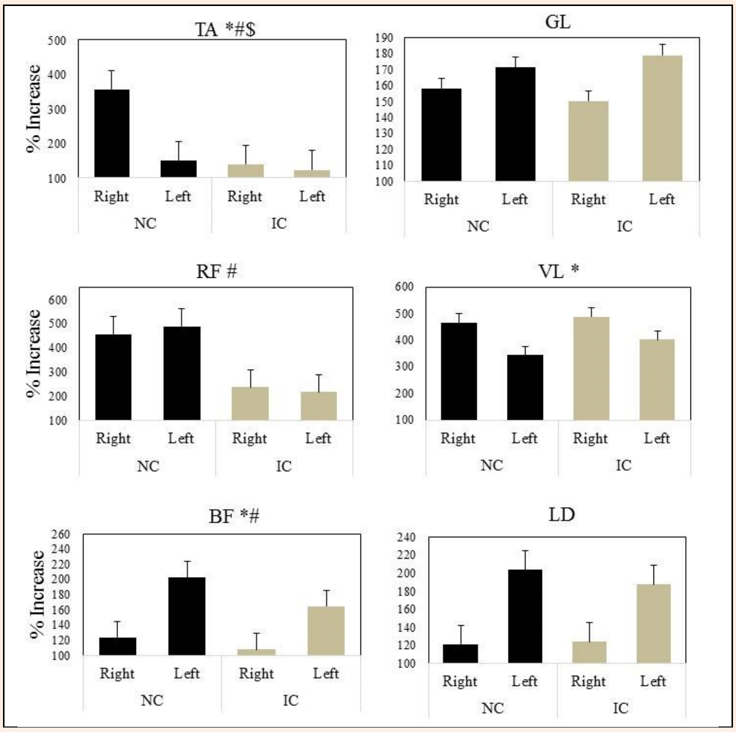

Figure 2. Percentage sEMG increase from beginning to end of a graded exercise test in the push phase. * significant main effect (side of body), # significant main effect between conditions, $ significant interaction (side*condition) ($p < 0.05$). Error bars represent the standard error of the mean.

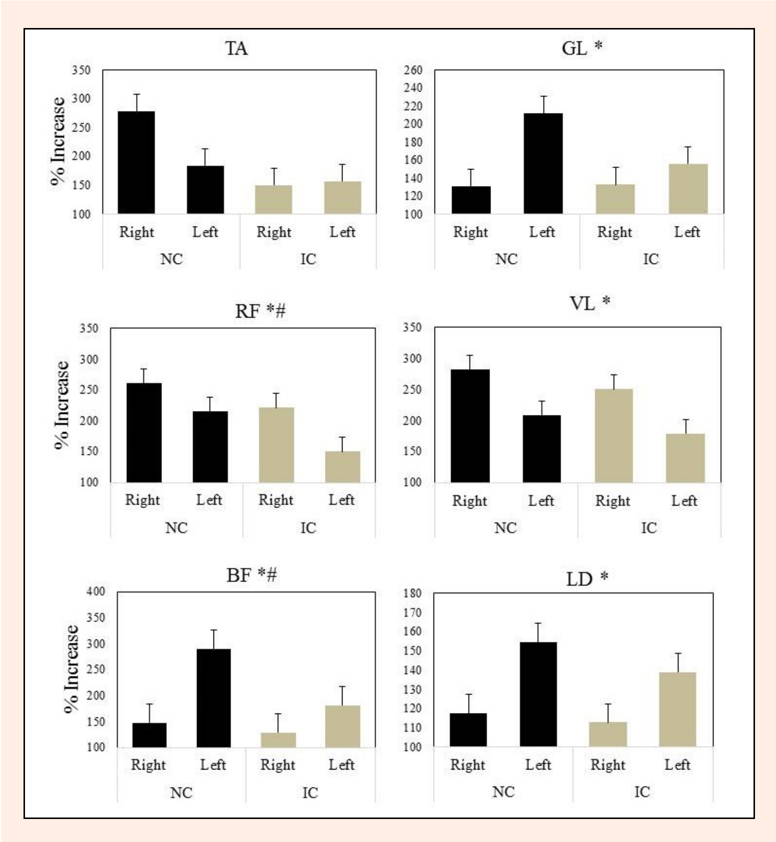

Figure 3. Percentage sEMG increase from beginning to end of the graded exercise test in the recovery phase. * significant main effect (side of body), # significant main effect between conditions, $ significant interaction (side*condition) ($p < 0.05$). Error bars represent the standard error of the mean.

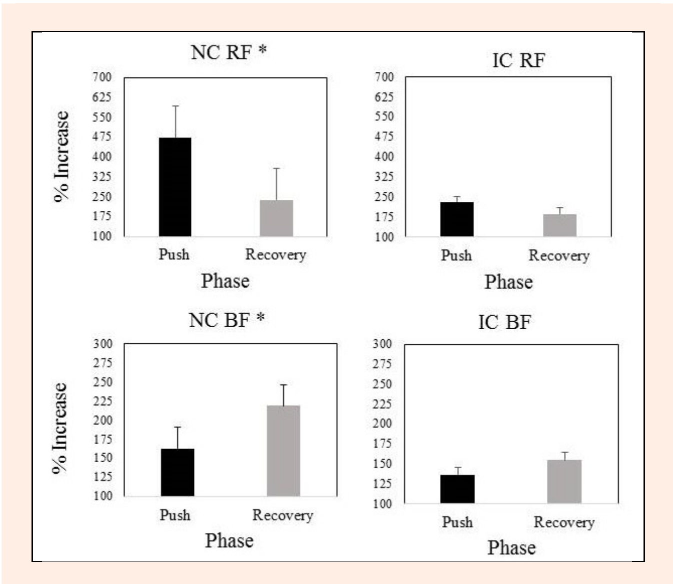

Figure 4. Percentage sEMG increase from beginning to end of the graded exercise test within conditions. * significant main effect between conditions ($p < 0.05$). Error bars represent the standard error of the mean.

Discussion

The purpose of this study was to determine if there are differences in trunk ROM during a graded exercise test when cycling with IC compared to NC, and how these differences change with increasing external load and fatigue. Another purpose of this study was to determine if there are differences in bilateral muscle recruitment patterns between the two conditions during the same graded exercise test. It is important to understand these differences as there may be injury risks associated with them, and it is crucial to evaluate the potential for IC to be used as a safe and effective training tool.

The results from this study indicate that trunk ROM when cycling with IC does not differ from trunk ROM during NC cycling, with the exception of lateral flexion to the right, where a small increase in angle is seen in the IC condition. However, the magnitude of this angle increase is very small (< 1°) and it can be concluded that the overall gross kinematics of the trunk are upheld when cycling with IC. These findings are contrary to the hypothesis that IC cycling would cause an increased ROM in the trunk. This is important, as causation of back pain and injury has been studied in the workplace, and it was concluded that deviations to either side beyond the risk neutral zone (20% deviation from mid-range) is considered hazardous (Kumar, 2001). Therefore, as trunk kinematics when cycling with IC does not differ from NC cycling, there is no greater risk of low back injury or pain when cycling with IC. In addition, as participants could not reach the same external resistance with IC, it is possible that there was less stress on their lumbar spine at volitional fatigue compared to NC cycling.

The findings that sagittal trunk flexion increases with fatigue when cycling with NC agrees with previous research (Dingwell et al., 2010). However, the current study completed a three-dimensional kinematic analysis of the trunk and it can be concluded that, along with increased flexion, there is a greater ROM in terms of lateral flexion to the right (< 1.0°) and left (1.6°), as well as transverse rotation to the right (1.7°) and left (1.5°). This effect can be attributed to increased fatigue levels, as previous research found no effect of increased work load on three-dimensional trunk kinematics (Bini et al., 2016) when fatigue was not a factor. The increased magnitude of trunk angle could cause a greater stress on the lumbar spine, and increase the risk of low back pain in cyclists. This effect on the trunk would be similar to the upper limbs, as it is known that joint moments of the upper limbs increase with external load (Costes et al., 2016). More research on the interaction between external load and low back joint moments is warranted.

In general, muscle activation increased from beginning to end – which was expected since it is well documented that sEMG RMS value increases with oxygen uptake and during incremental cycling (Hug et al., 2003; 2004). However, during IC cycling, although the RF and BF exhibited an overall increase in activation with an increase in external load, there was no significant difference between the push and recovery phases, which suggests that the use of IC causes a more constant and sustained activation of those muscles throughout the entire pedal cycle. This is not consistent with the hypothesis that there would be an increased change in activation in the IC condition; however, the findings from this study can be explained by the constant activation requirement of these muscles needed in the recovery phase, due to the lack of assistance from the contralateral leg in the push phase. A component of muscular endurance is the ability for the muscle to sustain a contraction for an extended duration (Swain and Brawner, 2012). This constant contraction throughout the cycle is demonstrated within the IC condition in the RF and BF, as there was no difference in activation increase from beginning to end between phases. Meanwhile, in the NC condition, there was a greater activation change in the push phase than the recovery phase for the RF, and greater change in the recovery phase for the BF. This indicates that IC cycling more closely represents an isotonic exercise, defined as resistance to the muscles being constant throughout the full ROM (Heyward and Gibsonn, 2014). The effect of isotonic training on muscular endurance during cycling has not been studied, and more information on that topic is needed to describe the performance outcomes that may result from training with IC. Cycling with NC, however, differs from this constant resistance, as each leg is dependent on the other, and a resistance force is applied by the leg in the recovery phase causing resistance to fluctuate around the full ROM (Bini et al., 2013). Building endurance of the RF and BF in particular is important, as they are involved with hip and knee flexion, respectively, which is required in the recovery phase of the cycle. Additionally, this could improve the cyclist's ability to pull backwards on the pedal at the bottom of the pedal cycle, which has been shown to increase the tangential force required to turn the crank (Bini and Diefenthaeler, 2010).

The asymmetry between legs when using IC was the same as NC cycling in all muscles, except for the TA in the push phase of the cycle, where there was greater asymmetry in the NC condition. This means that any asymmetry between limbs that may have occurred was not due to the change in crank type in all muscles except for TA. In fact, IC may reduce asymmetry in the TA in the push phase of the cycle. A reduction of asymmetry would lead to a more consistent training between legs, and could contribute to a more symmetrical cycle pattern, effectively reducing the risk of overuse injuries in training (Smak et al., 1999). This is an issue specifically related to cyclists, as increasing TA activation would exert more force on the ankle, which is a known risk factor for Achilles tendonitis - the most common overuse ankle injury among cyclists (Cohen, 1993).

There were several limitations that existed in this study. First, for sEMG processing, there was no normalizing factor between testing days. This means that sEMG between conditions had to be analyzed as a percentage change from beginning to the end of the trial. Second, although there was a familiarization period during the warm-up, and participants only completed the trial after it was clear they had become accustomed to the independent crank design; there may still have been some changes in cycling due to inexperience with IC. Additionally, the graded exercise test was designed to push the participants to the point of volitional fatigue. This allowed a comparison of movement and sEMG at volitional fatigue. However, comparisons could not be made at specific workloads, as every participant reached volitional fatigue at different times. Future research could include a similar comprehensive analysis of kinematics following a training period with IC compared to NC. This would provide valuable longitudinal information that would be relevant to cyclists using IC as a long term training tool.

Conclusions

Overall, IC are an effective tool for increasing training specificity, because they cause a more isotonic contraction of the RF and BF, thereby allowing the cyclist to target endurance training of these muscles. This allows training programs to be designed much like resistance training programs, where specific contraction types can be modified and targeted depending on the desired outcomes. Also, IC does not modify three-dimensional trunk kinematics from NC, meaning that there is likely no increased risk of low back injury for cyclists.

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Key points

- Independent crank cycling has similar three-dimensional trunk kinematics to normal crank cycling.
- Independent crank cycling with increasing load causes a more constant contraction of the biarticular muscles, rectus femoris and biceps femoris, compared to normal crank cycling.
- Independent cranks may be used as a training tool to supplement normal crank cycling, with no increased risk of injury to the cyclist.

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