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Fatigue effects on the coordinative pattern during cycling: Kinetics and kinematics evaluation

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ABSTRACT

The aim of the present study was to analyze the net joint moment distribution, joint forces and kinematics during cycling to exhaustion. Right pedal forces and lower limb kinematics of ten cyclists were measured throughout a fatigue cycling test at 100% of PO_{MAX}. The absolute net joint moments, resultant force and kinematics were calculated for the hip, knee and ankle joint through inverse dynamics. The contribution of each joint to the total net joint moments was computed. Decreased pedaling cadence was observed followed by a decreased ankle moment contribution to the total joint moments in the end of the test. The total absolute joint moment, and the hip and knee moments has also increased with fatigue. Resultant force was increased, while kinematics has changed in the end of the test for hip, knee and ankle joints. Reduced ankle contribution to the total absolute joint moment combined with higher ankle force and changes in kinematics has indicated a different mechanical function for this joint. Kinetics and kinematics changes observed at hip and knee joint was expected due to their function as power sources. Kinematics changes would be explained as an attempt to overcome decreased contractile properties of muscles during fatigue.

1. Introduction

Fatigue has received attention in sports sciences mainly because of its effects on athletes' performance (Faria et al, 2005). Abiss and Laursen (2005) have presented several models to explain the fatigue process, including biomechanical and neuromuscular model. In the biomechanical model, they have explained the fatigue process by means of economy/efficiency of movement, while in the neuromuscular model, fatigue occurs due to a failure on central control of movement or neuromuscular propagation. Both impairments reduce the force production and could affect movement control (Lepers et al., 2000). However, these models are insufficient to completely explain the occurrence of muscle fatigue or to indicate its effects on cycling performance and movement control.

Coordinative pattern, by means of muscle activity, has been analyzed during cycling to fatigue (Hautier et al., 2000; Lepers et al., 2000; Lepers et al., 2002; Duc et al., 2005; Dingwell et al., in press), competition simulation (Bini et al., 2008), pedaling cadence (Suzuki et al., 1982; Baum and Li, 2003), and workload management (Baum and Li, 2003; Laplaud et al., 2006). Another possible analysis of the coordinative pattern relies on the measurement of net joint moments, which have been computed as an indicative of muscle stress (Kautz and Hull, 1996; Marsh et al., 2000), and have been described to be affected by fatigue process (Sanderson and Black, 2003). It was observed a change at the ankle, knee and hip moments in the final minute of cycling to exhaustion (Sanderson and Black, 2003).

Several studies have analyzed the effects of different mechanical set-ups on the net joint moments, as pedaling cadence (Neptune and Hull, 1999; Marsh et al., 2000), saddle height (Ericson et al., 1986b; Horscroft et al., 2003), workload (Ericson et al., 1986b; Caldwell et al., 1999). Ericson et al. (1986a, 1988) have analyzed the relative joint contribution to the global lower limb activity during cycling task, as an indicative of coordinative pattern. However, only Sanderson and Black (2003) have described the influence of fatigue process on joint moments. Mornieux et al. (2007) have re-analyzed data from Sanderson and Black (2003) and indicated that the contribution of each joint to the total joint moments (i.e. coordinative pattern) have not changed during fatigue process.

Amoroso et al. (1993) observed kinetics and kinematics adaptations as mechanical responses to fatigue process, however they have not analyzed joint moments distribution neither joint forces nor kinematics. Amoroso et al. (1993) have proposed that the analysis of joint moments would be an important approach to understand the reasons for the external forces and kinematics changes throughout fatigue process. Additionally, Mornieux et al. (2007) indicated that ankle joint is tuned to transfer the force produced by the hip and knee joints to the crank. However, it is not well

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understood how does joint forces and kinematics are tuned for the control of the joint moment's distribution during fatigue process.

Herewith, the main purpose of the present study was to analyze the net joint moment distribution in attempt to understand the coordinative pattern during cycling to exhaustion. As indicated by Mornieux et al. (2007)), the fatigue process is not supposed to change joint moment distribution, which is defined as the hypothesis of the present study. Herewith, our secondary purpose was to analyze joint forces and kinematics which would change in the attempt to maximize joint stiffness and optimize force transmission to the crank.

2. Methods

2.1. Subjects

Ten well trained male cyclists (USCF Category 3 or higher level of competition) volunteered to participate in this study. All subjects had competitive experience. All participants signed an Informed Consent Term in agreement with the Committee of Ethics in Research with Humans of the University of Texas at Austin in accordance with the Declaration of Helsinki. The cyclists were asked to avoid high-intensity or exhaustive exercise at least 24 h before the laboratory trials. The mean and standard deviation age, body mass, maximal oxygen uptake (VO_{2MAX}), peak power output and power/mass ratio of the subjects were 30.6 ± 6.8 years, 74.4 ± 7.9 kg, 60 ± 5 ml kg^{-1} min^{-1}, 408.2 ± 38 W, and 5.51 ± 0.45 W kg^{-1}, respectively.

2.2. Protocol

The experiment followed a two days protocol. During the first day each subject performed an incremental exercise bout on a stationary cycle ergometer Lode Excalibur Sport V2.0 (Lode Medical Technology, Groningen, The Netherlands) to determine maximal oxygen consumption (VO_{2MAX}). Throughout the protocol, oxygen uptake (VO_2) and carbon dioxide produced (VCO_2) were measured using an open-circuit indirect gas exchange system (Physiodyne FLO-1B System (Physio-Dyne Instrument Corp., New York, USA). Heart rate (HR) was continuously assessed from a telemetric HR monitor (Polar Electro Oy, Finland). Few minutes after the end of incremental test, the cyclists were familiarized with the fatigue test, at which they had to cycle to voluntary exhaustion at the maximal power output (PO_{MAX}), measured by Lode Excalibur Sport V2.0 control unit (Lode Medical Technology, Groningen, The Netherlands).

During the second evaluation day, the cyclists were submitted to a fatigue cycling test after a 10 min warm up. Subjects cycled at a constant cadence to voluntary exhaustion with workload set at the PO_{MAX} achieved during the incremental cycling test. Vigorous verbal encouragement was given throughout the trials for all subjects.

2.3. Materials

Throughout the fatigue cycling test, VO_2, kinetics and kinematics of cyclists were measured. A bi-dimensional force pedal dynamometer instrumented with *strain gages* was used to measure the normal (Fz) and tangential (Fx) components of the force applied by the cyclists (Newmiller et al., 1988). The pedal force signals passed through a main amplifier 2021B (Vishay Measurements Group, USA) and were acquired by a Vicon 612 analog-to-digital system (Oxford Metrics, England) with 1080 Hz of sample rate.

Kinematics variables were measured by six infrared cameras using Vicon system (Oxford Metrics, England) and were acquired by the aforementioned Vicon 612 analog-to-digital system with 120 Hz of sample rate. As landmarks for the hip, knee and ankle joint axes, reflective markers were placed on the right side of the cyclists at: (1) greater trochanter, (2) lateral femoral condyle, (3) lateral malleolus, (4) pedal spindle, (5) anterior and (6) posterior pedal stick and (7) at the crank axis.

2.4. Data analysis

From physiological data, the VO_{2MAX} was defined as the highest VO_2 value of the last minute of the progressive cycling test (Lucía et al, 2002). During the incremental test, the power output (PO) was continually recorded to determine the PO_{MAX} (Passfield and Doust, 2000).

Pedal forces and kinematic data were smoothed by means of a 4th order Butterworth low-pass digital filter with a cutoff frequency of 10 Hz (Marsh et al., 2000) and 4 Hz (Reiser et al., 2002), respectively. Joint angles were defined as presented in the Fig. 1. Linear and angular velocities and accelerations were computed from smoothed data. Pedal angle in relation to the global coordinate system were calculated to convert the forces on the pedal reference system to forces on the global reference system by means of trigonometric procedures (Marsh et al., 2000). Right lower limb was modeled as three-segment rigid body systems which segment mass, center of mass, and radio of gyration were estimated according to Zatsiorsky and Seluyanov tables (1999) corrected by de Leva (1996). Conventional inverse dynamics, where all forces and moments were calculated based on Newton's second law of motion, were conducted to calculate the net joint moments at the hip, knee and ankle (Smak et al., 1999). This procedure was developed into adapted Matlab codes of van den Bogert and de Koning (1996). The net joint moments at the hip, knee and ankle were normalized to 360 points of each crank angle, and averaged for fifteen complete cycles. All net joint moment values were then turned to positive to compute the single absolute average value that represented each cycle. This procedure was conducted for each joint, which allow, by the sum of hip, knee, and ankle single values, to calculate the total absolute joint moments (Marsh et al., 2000). Each absolute joint moment was then expressed as a percentage of the total absolute joint moments to allow the analysis of the contribution of each joint to the total joint moment (Mornieux et al., 2007). Resultant forces where calculated from the hip, knee and ankle joints. All data analysis was conducted off-

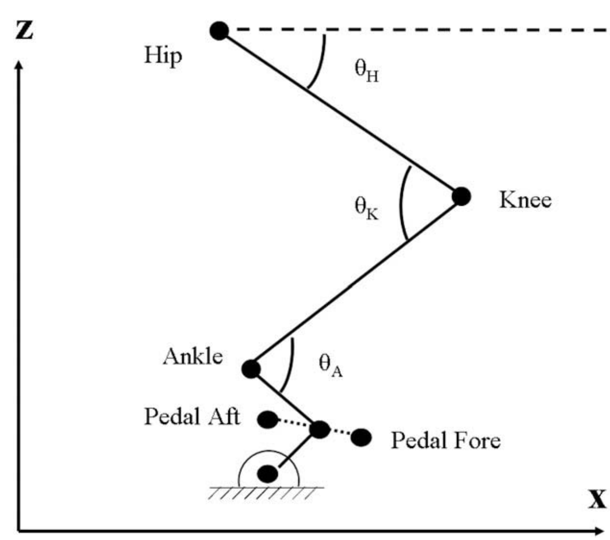

Fig. 1. Schematic illustration of angle definitions for kinematic analysis. Angle definitions: θ_H – Hip angle; θ_K – Knee angle; θ_A – Ankle angle.

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line for fifteen complete crank cycles at four instants of the fatigue cycling test (10, 40, 70, and 90% of total time of cyclists' test). For data analysis, proper codes were developed using software MATLAB® 7.3 (MathWorks Inc., USA).

2.5. Statistical analysis

Descriptive statistics were used to report the results for mean and standard deviation. Data normality was evaluated by Shapiro–Wilk's test then a one-way ANOVA for repeated measures was employed. The analyzed variables were: (1) mean value of resultant joint forces, (2) mean value and (3) the range of motion (ROM) of joint angles, (4) pedaling cadence, (5) mean value of the total average net joint moments, (6) mean value of average absolute net moment of each joint, and (7) the contribution of each joint to the total joint moment. *Post hoc* were performed using Bonferroni when main effects or interactions were significant after ANOVA. The level of statistical significance for all analyses was set at $p < 0.05$. For statistical procedures the SPSS 12.0 package (SPSS Inc., USA) was used.

3. Results

Cyclists' performance at fatigue test was measured by the total time of fatigue cycling test (TT), which was 405 ± 90 s. Pedaling cadence was compared in attempt to verify if it was influenced by fatigue process, and it is depicted in Fig. 2.

It was observed a significant reduction in pedaling cadence at 90% of TT in relation to 10 ($p = 0.01$), 40 ($p < 0.01$), and 70% of TT ($p < 0.01$) for $F_{1.118,\ 10.062} = 19.315$, $p < 0.01$.

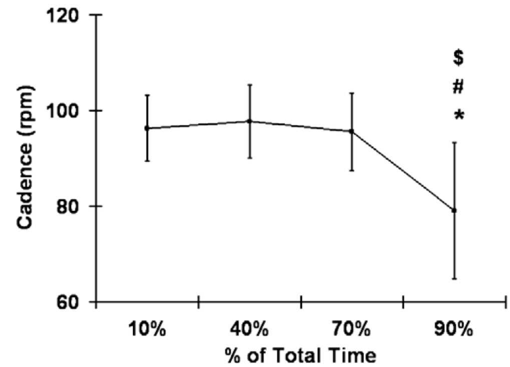

Fig. 2. Mean + SD of pedaling cadence at the four instants of the fatigue cycling test ($N = 10$). * Significant difference related to 10% of total time ($p < 0.05$). # Significant difference related to 40% of total time ($p < 0.05$). $ Significant difference related to 70% of total time.

Representative data of the net joint moments from a subject are presented in Fig. 3.

The absolute net joint moments averaged over the pedaling cycle were summed to compute the total net joint moments. Relative contribution of each joint in relation to the total net joint moments is presented in Fig. 4.

The percentage of ankle moment to the total joint moments has decreased at 90% (13 ± 0.03%) in relation to 70% of TT (17 ± 0.04%, $p = 0.01$) for $F_{3,\ 24} = 5.832$, $p < 0.01$. Hip and knee relative moments did not differ.

The results of the absolute net joint moments are presented in Fig. 5.

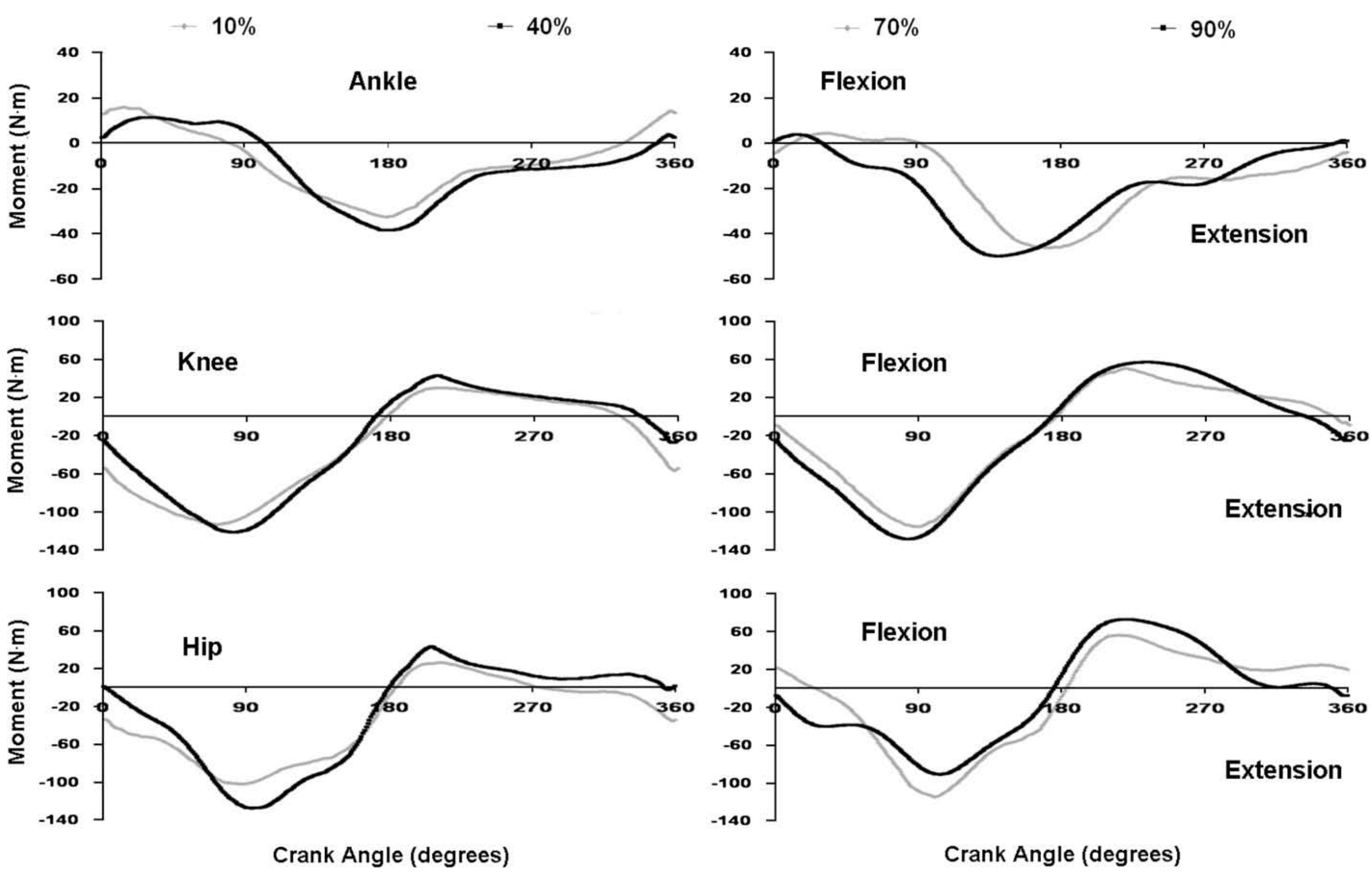

Fig. 3. Representative data of the net joint moments from a single subject performing at 10%, 40%, 70%, and 90% of total time of fatigue test. Ankle, knee and hip moments are plotted against crank angle. Negative values indicate plantar flexor moment at the ankle, and extensor moment at the knee and hip joints.

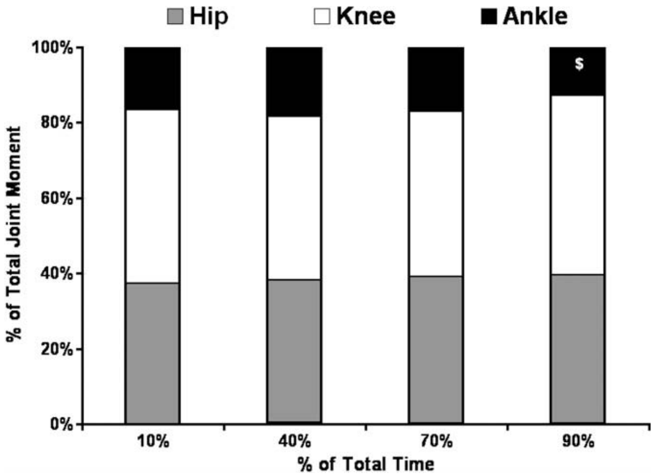

Fig. 4. Percentage of each joint to the total average absolute net joint moments in every four instants of cycling fatigue test. ($N = 10$). $Significant difference related to 70% of total time ($p < 0.05$). No significant differences in relation to 10% and 40% of total time.

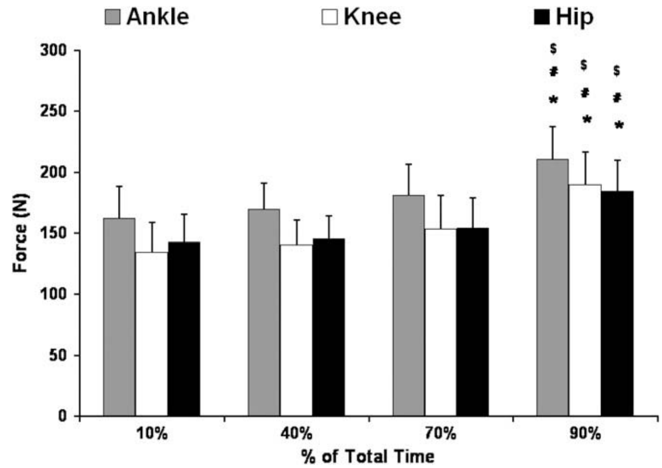

Fig. 6. Mean + SD of the ankle, knee and hip resultant forces at the four instants of the fatigue cycling test ($N = 10$). * Significant difference related to 10% instant ($p < 0.05$). # Significant difference related to 40% instant ($p < 0.05$). $ Significant difference related to 70% instant ($p < 0.05$).

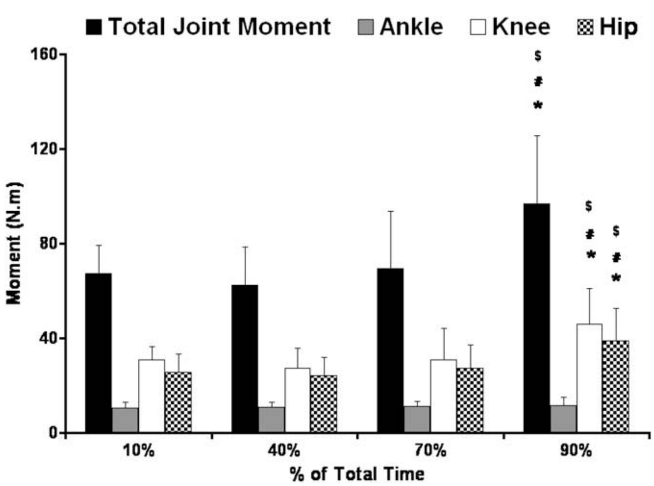

Fig. 5. Mean + SD of the total net joint moment, ankle, knee and hip moments at the four instants of the fatigue cycling test ($N = 10$). * Significant difference related to 10% of total time ($p < 0.05$). # Significant difference related to 40% of total time ($p < 0.05$). $ Significant difference related to 70% of total time ($p < 0.05$).

Table 1
Mean **+** SD results of the mean value and the range of motion (ROM) of the ankle, knee and hip angle. The results are presented for each instant of the total time of fatigue cycling test (% of total time).

Kinematics variables (deg.)	10% of total time	40% of total time	70% of total time	90% of total time
Ankle mean	158 + 5	157 + 5	155 + 6*	153 + 7*,**,***
Ankle ROM	19 + 4	20 + 6	20 + 6	26 + 9*,**,***
Knee mean	131 + 3	132 + 2	133 + 2	134 + 3*
Knee ROM	69 + 4	70 + 4	71 + 5	70 + 5
Hip mean	53 + 4	54 + 4	55 + 4*	58 + 5*,**,***
Hip ROM	54 + 4	54 + 4	53 + 4	51 + 5*,**,***

* Significant difference related to the 10% of total time ($p < 0.05$).
** Significant difference related to the 40% of total time ($p < 0.05$).
*** Significant difference related to the 90% of total time ($p < 0.05$).

The total average absolute joint moment increased at 90% of TT in comparison to 10% ($p = 0.01$), 40% ($p < 0.01$), and 70% of TT ($p = 0.03$), for $F_{1.819,\ 14.549} = 7.565$, $p < 0.01$. Hip average absolute joint moment has also increased at 90% of TT in comparison to 10% ($p = 0.01$), 40% ($p = 0.01$), and 70% of TT ($p = 0.03$), for $F_{3,\ 24} = 7.114$, $p < 0.01$. For knee average joint moment, an increase was observed at 90% of TT in comparison to 10% ($p = 0.02$), 40% ($p < 0.01$), and 70% of TT ($p = 0.02$), for $F_{1.89,\ 15.12} = 7.337$, $p < 0.01$.

In Fig. 6, the results of the resultant joint forces are presented at the four instants of the total time of fatigue cycling test.

The ANOVA for repeated measures reported $F_{1.628,\ 13.021} = 12.48$, $p < 0.01$, for the ankle joint force, $F_{1.698,\ 13.587} = 16.835$, $p < 0.01$, for the knee joint force, and $F_{1.523,\ 12.186} = 16.674$, $p < 0.01$ for the hip joint force. Herewith, there were significant differences for the resultant forces at the ankle, knee and hip joints in 90% in comparison to 10% ($p = 0.02$ for ankle, $p = 0.01$ for knee, and $p = 0.01$ for hip), 40% ($p < 0.01$ for ankle, $p < 0.01$ for knee, and $p < 0.01$ for hip) and 70% of TT ($p = 0.02$ for ankle, $p = 0.01$ for knee, and $p < 0.01$ for hip)..

Kinematics was analyzed in relation to the mean value and the range of motion (ROM) of each joint angle. These results are presented in Table 1.

There was a significant decrease of mean ankle angle at 70% of TT in relation to 10% of TT ($p = 0.025$), and at 90% of TT in relation to 10% of TT ($p = 0.018$), 40% of TT ($p = 0.032$), and 70% of TT ($p = 0.034$), for $F_{1.247,\ 9.979} = 15.04$, $p < 0.01$. There was also an increase of ankle ROM at 90% of TT in relation to other instants of fatigue test ($p = 0.014$ in relation to 10% of TT, $p = 0.012$ in relation to 40% of TT, and $p < 0.01$ in relation to 70% of TT), for $F_{1.408,\ 11.267} = 9.146$, $p < 0.01$. For knee joint, there was an increase of mean angle at 90% of total time in relation to 10% of TT ($p = 0.028$), for $F_{1.275,\ 10.202} = 5.858$, $p = 0.03$. There was no significant differences for knee ROM during fatigue cycling test, for $F_{1.552,\ 12.42} = 3.314$, $p = 0.079$. For the hip joint, there was an increase of mean angle at 90% of TT in relation to 10% of TT ($p = 0.013$), 40% of TT ($p = 0.012$), and 70% of TT ($p = 0.012$), for $F_{1.168,\ 9.341} = 18.234$, $p < 0.01$. Hip joint ROM have reduced at 90% of TT in relation to 10% of TT ($p = 0.016$), 40% of TT ($p = 0.038$), and 70% of TT (0.023), for $F_{1.397,\ 11.172} = 6.508$, $p = 0.02$.

4. Discussion

The main purpose of the present study was to analyze the net joint moment distribution in attempt to understand the coordinative pattern during cycling to exhaustion. The main result of the present study was that only ankle joint have a reduced contributed to the total joint moments during fatigue test. It was different from our hypothesis, which suggested the maintenance on joint moment distribution during fatigue, as previously reported by Mornieux et al. (2007). At this regard, our secondary purpose was the novelty of the present study, which was the measurement of joint forces and kinematics as an attempt to understand how the joint moments are tuned during cycling task, and how joint forces and kinematics' change during fatigue process.

Regarding methodological approach conducted in the present study we must concern on two aspects. The first regards the choice of the greater trochanter method to compute hip joint center in relation to iliospinale method, which was based on the maximal error of 4.9% for hip joint power, as its small effects on hip joint angular velocity and hip joint moment reported by Neptune and Hull (1995). The second regards on the choice of conventional inverse dynamics without concern on the forces on the seat, which was based on the reporting that the force produced by the upper body, which is transferred through the hip joint, has been reported to input only 4.5% to total joint mechanical work during seated cycling (Kautz et al., 1994).

An expected result observed in the present study was the reduction of pedaling cadence with fatigue, which has been previously described (Lepers et al., 2000, and Lepers et al., 2002). Some mechanical changes were observed related to the decrease of pedaling cadence. These alterations were expected to occur due to the set-up of the cycle ergometer, which was supposed to increase the resistance with the decrease of pedaling cadence for the same PO (constant workload mode). It would be defined as a limitation of the present study, however if the pedaling cadence were kept constant we were supposed to observe a decrease on PO due to fatigue process (St. Clair Gibson et al., 2001; Abiss et al., 2006). When we analyze cycling on the road, we would expect a significant decrease on pedaling cadence with the increase of workload (Caldwell et al., 1999) and fatigue effects (Abiss et al., 2006), which reinforces the chosen set-up of PO on the cycle ergometer. Abiss et al. (2006) observed in outdoor measurements of crank torque and pedaling cadence, that during time trial, the cyclists have attempted to keep the crank torque instead of pedaling cadence.

The mechanical implications of fatigue effects and the reduced pedaling cadence have affected joint moments' distribution, joint forces and kinematics. The reduced contribution of ankle joint to the total joint moments suggests a different mechanical function for this joint. Kautz and Neptune (2002) and Zajac (2002) reported that ankle joint muscles (e.g. soleus and gastrocnemius) are described to work mainly as force transfer from the legs to the crank. Mornieux et al. (2007) have proposed that ankle joint muscles are tuned to maximize joint stiffness improving the force transmission. Herewith, we would expect that with the increase of pedal reaction force and crank torque (e.g. 90% of total time of fatigue cycling test in the present study) there would have an increase of ankle joint force, but not necessarily the average ankle net joint moment. The reduction observed at ankle moment contribution would be related to a decreased capability of ankle joint muscles to transfer force to the pedal as a consequence of fatigue effects. Bini et al. (2008) indicated that *Gastrocnemius Medialis* might be inversely related to PO, as an attempt to minimize fatigue process.

The increase of ankle joint force was necessary to improve the stiffness of the foot-pedal complex at higher pedal reaction forces, which allows the cyclist to transfer the force produced by hip and knee joint to the crank. If the ankle joint muscles have not transferred the force from the upper joints, this would probably be wasted, reducing cycling effectiveness. At this regard, the balance between pedal reaction force and ankle joint force seems to be a strategy for the maintenance of the average ankle joint moment, and overcoming higher workloads during cycling.

In relation to ankle joint stiffness, the combined effects of higher pedal reaction force (due to reduced pedaling cadence) and fatigue resulted in the increase of ankle joint resultant force and increased ankle joint range of motion (ROM). Kinematics' change has been previously described during fatigue cycling test (Amoroso et al., 1993; Sanderson and Black, 2003; Dingwell et al., in press) and at higher workloads (Kautz et al., 1991; Black et al., 1993). However, neither Amoroso et al. (1993) nor Sanderson and Black (2003) have observed increased ankle ROM at the final minutes of fatigue cycling test, as the results of the present study.

However, the reduced mean ankle angle joint towards dorsiflexion, which was observed in the present study, have already been described as fatigue effect (Amoroso et al., 1993) and at higher workloads (Kautz et al., 1991; Black et al., 1993; Sanderson et al., 2006). The combined increase of ROM and the reduced mean ankle angle would be explained by an attempt to overcome the decrease of contractile properties of ankle muscles, as an effect of fatigue process. If we consider muscles' need of producing power at the joints, and that fatigue reduces the force production capability of these muscles, they would be expected to increase their lengthening velocity (Sanderson et al., 2006).

Kinetics and kinematics change observed at hip and knee joint was expected due to their function as power sources (Mornieux et al., 2007). Hip and knee joint moments have been increased while mean angle have been changed towards extension, as previously reported by Sanderson and Black (2003), who have suggested an attempt of the cyclists to keep performance. The attempt of increasing the intersegmental joint force transmission needs to be addressed, since hip and knee joints were moved to less extended angles at 90% of total time of fatigue cycling test. Increasing moments and forces, and changing hip and knee joint kinematics would be addressed as a strategy to overcome higher pedal reaction force at the 90% of total time of fatigue cycling test.

The change of hip and knee joint muscles towards lower length could probably affect force–length relation of these muscles (Sanderson et al., 2006). However, it is not known how do the shift of hip and knee joint mean angle could affect muscles force production and how this is an advantage in relation to intersegmental joint force transmission.

By the evidences observed in the present study, we can conclude that fatigue effects should change the coordinative pattern during cycling. The main compensatory effect seams to occur at the ankle joint, with changes on the ankle joint contribution to the total joint moments. Knee and hip joints were observed to be related to power production, while ankle joint should be related to mechanical energy transfer to the crank.

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