Influence of different racing positions on mechanical and electromyographic patterns during pedalling

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The aim of this study was to test the hypothesis that, in comparison with standard postures, aero posture (AP) would modify the coordination of lower limb muscles during pedalling and consequently would influence the pedal force production. Twelve triathletes were asked to pedal at an intensity near the ventilatory threshold (VT1D20%) and at an intenisty corresponding to the respiratory compensation point (RCP). For each intensity, subjects were tested under three positions: (1) upright posture (UP), (2) dropped posture (DP), and (3) AP. Gas exchanges, surface electromyography and pedal effective force were continuously recorded. No significant difference was found for the gas-exchange variables among the three positions. Data illustrate a significant increase [*gluteus maximus* (GMax), *vastus medialis* (VM)] and decrease [*rectus femoris* (RF)] in electromyography (EMG) activity level in AP compared with UP at RCP. A significant shift forward of the EMG patterns (i.e. later onset of activation) was observed for RF (at VT1D20% and RCP), GMax, VL, and VM (at RCP) in AP compared with UP. These EMG changes are closely related to alteration of force profile in AP (higher downstroke positive peak force, lower upstroke negative peak force, and later occurrence of these peaks along the crank cycle).

Air resistance is the dominant force-resisting motion of cyclists on flat terrain. It depends on different external factors (e.g. ambient air velocity or air density) and biomechanical-anthropometric factors (e.g. drag coefficient and frontal area of rider1bicycle system). In an effort to reduce this force, the cyclist's body positioning has received much attention in the past two decades. From a mechanical point of view, a more crouched upper body position (i.e. areo posture, AP), by decreasing the frontal area, allows a lower wind resistance (Capelli et al., 1993) compared with conventional postures (upright posture, UP or dropped posture, DP) and hence a higher riding speed for a given power ouput. In parallel, some research has focused on the effects of this AP on the ventilatory and metabolic responses during submaximal and/or maximal exercises (Ryschon & Stray-Gundersen, 1991; Origenes et al., 1993; Gnehm et al., 1997; Grappe et al., 1998). Although some authors have failed to detect any significant differences in several cardiorespiratory variables between standard and APs (Origenes et al., 1993; Grappe et al., 1998), others concluded that aero position increases the metabolic cost of cycling (Gnehm et al., 1997). It is surprising that the effects of this cycling position on both mechanical aspects of force production on pedals and lower limb muscles activation pattern have not been investigated.

Surface electromyography (EMG) records have been widely used to study muscle activity and neuromuscular coordination during pedalling (Houtz & Fischer, 1959; Ericson, 1986; Gregor et al., 1991; For review see Hug & Dorel, in press). Other investigators have measured the forces exerted on pedals and/or cranks (Hoes et al., 1968; Sanderson et al., 2000; Sanderson & Black, 2003; Bertucci et al., 2005) and computed their effective component (i.e. effective force, tangential to the crank displacement). In order to maintain a given level of performance, the EMG and mechanical patterns vary in timing and magnitude as pedalling conditions change. Along this line, some studies have demonstrated that the pattern of muscles activation and/or of forces exerted on pedals could be modified by factors such as workload (Ericson, 1986; Jorge & Hull, 1986), pedalling rate (Suzuki et al., 1982; Ericson, 1986), shoe–pedal interface (Ericson, 1986; Cruz & Bankoff, 2001) and saddle height (Houtz & Fischer, 1959; Ericson, 1986). Literature concerning the influence of body position on mechanical and EMG patterns is less profuse. Two studies have showed that standing and seated postures lead to different patterns (Li & Caldwell, 1998; Duc et al., in press). To the best of our knowledge, only one study has focused on the effects of trunk orientation on the activation pattern of lower limb muscles during pedalling (Savelberg et al., 2003b). The authors reported that trunk angle influences the EMG patterns. However, the tested positions (i.e. 20l forward and backward of the vertical) were not comparable to the standard postures used by competitive cyclists (and especially far from the AP). Furthermore, the pedal force production was not measured.

Thus, we designed the present study to test the hypothesis that, in comparison with standard postures, aero position would modify the coordination of lower limb muscles during pedaling and consequently would influence the mechanical aspects of pedal force production. A population of trained triathletes was tested in three different positions (i.e. aero, upright and dropped posture (DP)). Each subject was tested at an intensity near the ventilatory threshold and at an intenisty corresponding to the respiratory compensation point (RCP) with the ulterior motive of transposing our observations to a time trial event.

Materials and methods

Subjects

Twelve male triathletes whose anthropometrical and physiological characteristics are presented in Table 1 volunteered to participate in this study. The subjects had a 9 ± 5 years of competitive experience. None of them had recent or ancient pathology of lower limb muscles or joints. They were informed of the nature of the study and the possible risk and discomfort associated with the experimental procedures before they gave their written consent to participate. The experimental design of the study was approved by the Ethical Committee of Saint-Germain-en-Laye (acceptance no 06016) and was carried out in accordance with the Declaration of Helsinki. All subjects were instructed to refrain from intense physical activities during the 2 days before testing.

Table 1. Anthropometrical and physiological characteristics of the population of triathletes (n 5 12)

	Mean ± SD
Age (years)	31.1 ± 8.4
Height (m)	1.81 ± 0.08
Body mass (kg)	72.2 ± 6.8
BMI (kg/m)	22.1 ± 1.7
$\dot{V}O_{2max}$ (mL/min/kg)	63.5 ± 9.1
$\dot{V}_E$ max (L/min)	181 ± 19
MAP (W)	392 ± 31
MAP (W/kg)	5.5 ± 0.7
VT (% MAP)	56.9 ± 4.6
RCP (% MAP)	83.3 ± 3.9

BMI, body mass index; MAP, maximal aerobic power; VT and RCP, ventilatory threshold and respiratory compensation point (expressed in percentage of MAP).

Exercise protocol

The testing protocol consisted of two sessions conducted in the following order: (1) anthropometric measurements and incremental cycling exercise performed until exhaustion in order to characterize our population in terms of physical and physiological capacities; (2) experimental session consisting of two submaximal cycling exercises, each performed in three different upper body postures.

During the first visit, 2 weeks before the experimental session, each subject performed an incremental cycling exercise (workload increments of 25 W/min; starting at 100 W) during which the usual gas-exchange and ventilatory variables were measured. Throughout this exercise trial, the device (K4B2, Cosmed s, Rome, Italy) computed breath-by-breath data of $\dot{V}_E$, VO_2, VCO_2, and the ventilatory equivalents for O_2 ($\dot{V}_E$ VO_2^{-1}) and CO_2 ($\dot{V}_E$ VCO_2^{-1}). The VT and RCP, respectively, were determined with the method based on the ventilatory equivalents for O_2 and CO_2 (Reinhard et al., 1979). Two independent observers detected VT and RCP following the criteria previously described. If they did not agree, the opinion of a third investigator was included. The first power achieved when the maximal oxygen uptake was reached (VO_{2max}) was referred as the maximal aerobic power. All these variables were determined to further adjust the two constant workloads tested during the experimental session. Finally, after a 15-min recovery period, the subjects were acclimatized to the three upper body positions that will be adopted during the subsequent visit.

During the second session, subjects were asked, after a 10 min warm-up at 100 W, to pedal at two different intensities corresponding to the power associated to: (1) VT plus 20% of the difference between power outputs measured at VT and RCP (VT1D20%) and (2) RCP. For each of these two intensities, subjects were tested, in a randomized order, under three body positions (detailed below): (1) the UP, (2) the DP, and (3) the AP (Fig. 1(a)). Each position was tested during 6 min at VT1D20% (5 min of active recovery between bouts) and 2 min at RCP (7 min of active recovery between bouts, Fig. 1(b)). The reason for choosing short test periods was to avoid fatigue throughout the session. Subjects were asked to keep the same constant preferred pedalling rate chosen at the end of the warm-up period throughout the session (± 2 r.p.m.). Gas exchanges, surface EMG and mechanical variables were continuously recorded during this protocol.

Stationary bicycle and body positions

Subjects exercised on an electronically braked cycle ergometer (Excalibur Sport, Lode s, Groningen, the Netherlands) equipped with standard crank (length 5 170 mm) and with their own clipless pedals. During both sessions, vertical and horizontal positions of the saddle, handlebar height and stem length were set to match the two usual positions of the participants: (1) UP, with hands on top of the handlebars, near the stem and elbow angle between 160l and 180l; (2) DP, the traditional racing position with the torso partially to fully bent-over, hands on the drops portion of the handlebars and elbows partially flexed (elbow angle less than 160l). To obtain an AP during the second session clip-on aerobars (Aero II, Profile, Los Angeles, USA) were added on the classical handlebar. Aero-handlebars and stem were adjusted (height and length) to custom fit each subject into his own aerodynamic position with the following restrictions: elbows on the pads of aero-handlebars with elbow angle close to 90l and the

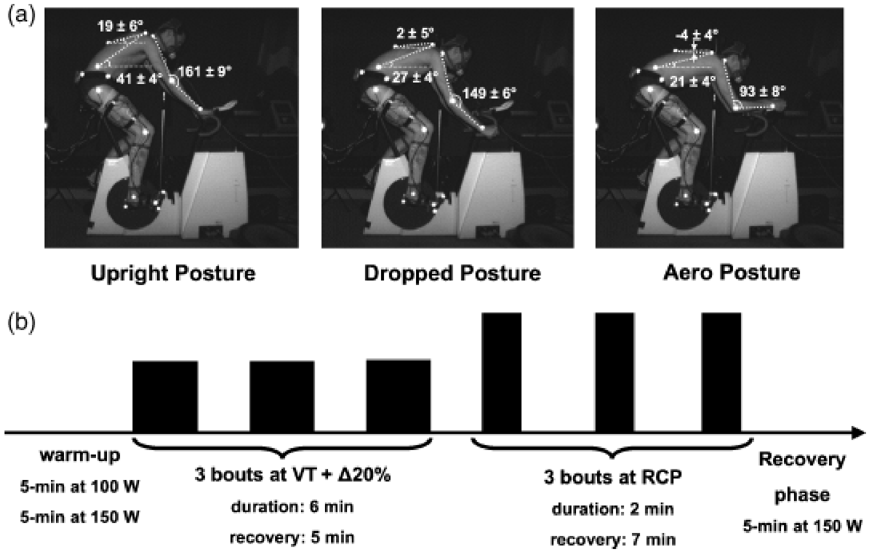

Fig. 1. A/Illustration of the three different positions [i.e. upright, dropped and aero postures (AP)] as defined by the placement of the hands and torso for one subject. Values of angle between (1) the trunk and horizontal (2) the upper part of the torso and horizontal and (3) arm and forearm reported on the photograph are the mean values obtained on the entire population. B/Schematic representation of the protocol.

upper part of the torso (line T10–C7) held parallel to the ground (with 10l tolerance, Fig. 1(a)). All these angles were measured with a goniometer. Video recording was used to check these standardization criterions in each position during the experimental session. Briefly, reflective markers placed on the right side of each subject and video analysis (2D sagittal plane, f 5 250 Hz) were used to approximate (1) the positions of the trunk (line from iliac crest to C7) and the upper part of the torso (line from T10 to C7) in reference to a horizontal line and (2) the elbow angle (acromion process, olecranon process, and wrist). Figure 1(a) depicts the mean angle values obtained for the group in each position.

Material and data collection

The torque exerted on the left and right cranks was measured by strain gauges located in the crank arms of the cycle ergometer. Before the experiments, a classical calibration procedure (i.e. with known masses) was performed and a zero adjustment was done before each session. Effective force (i.e. the propulsive force applied perpendicularly to the crank arm) was determined by the ratio between torque and the constant length of the crank arm. The crank angle and the angular velocity were calculated (by derivative) based on TTL rectangular pulses delivered each 2l by the cycle ergometer. Additional TTL rectangular pulse permitted to detect the bottom dead center of the right pedal (i.e. BDC: lowest position of the right pedal with crank arm angle 5 180l). All these data were digitized at a sampling rate of 2 kHz (USB data acquisition DT9800, Data translation s, Malboro, USA).

Surface EMG activity was continuously recorded for the following 10 muscles of the right lower limb: *gluteus maximus* (GMax), *semimembranosus* (SM), *Biceps femoris* (BF), *vastus medialis* (VM), *rectus femoris* (RF), *vastus lateralis* (VL), *gastrocnemius medialis* (GM) and *lateralis* (GL), *soleus* (SOL) and *tibialis anterior* (TA). A pair of surface Ag/AgCl electrodes (Blue sensor, Ambu s, Ballerup, Denmark) was attached to the skin with a 2 cm interelectrode distance. The electrodes were placed longitudinally with respect to the underlying muscle fibres arrangement and located according to the recommendations by Surface EMG for Non-Invasive Assessment of Muscles (SENIAM) (Hermens et al., 2000). Before electrode application, the skin was shaved and cleaned with alcohol in order to minimize impedance. The wires connected to the electrodes were well secured with adhesive tape to avoid movement-induced artifacts. Raw EMG signals were pre-amplified close to the electrodes (gain 375, bandwidth 8–500 Hz), and digitized at a sampling rate of 1 kHz (ME6000P16, Mega Electronics Ltd s, Kuopio, Finland). BDC TTL rectangular pulses were simultaneously digitized for further synchronization with cycle ergometer data. In order to diminish movement artefacts, a high pass filter (20 Hz) was further applied on the EMG signals (Chart 5.4, AD instruments s, Hasting, UK).

Data processing

All data were analyzed with custom written scripts (Origin 6.1, OriginLab Corporation, Northampton, USA). The BDC TTL rectangular pulses were used to synchronize signals of the right pedal effective force, right crank angle and EMG data. Raw EMG data were root mean squared (RMS) with a time averaging period of 25 ms to produce linear envelope for each muscle activity pattern. Effective force and EMG RMS were then re-sampled in order to obtain one value each 2 degrees of crank displacement. Prior re-sampling, all data were filtered with an anti-aliasing filter which cutoff frequency was dynamically computed according to Shannon Theorem (i.e. 2 degrees TTL pulses half mean frequency). Linear interpolation technique was then used to obtain a mean value of force and EMG RMS each degree of rotation. Finally, these data were respectively averaged over 90 and 45 consecutive pedalling cycles for VT1D20% and RCP conditions in order to get a representative effective force profile and an EMG RMS linear envelope for each muscle, each subject and each condition. These values were expressed as a function of the crank arm angle as it rotated from the highest pedal position (0l, top dead center: TDC) to the lowest (180l, bottom dead center, BDC) and back to TDC to complete a 360l crank cycle.

The following mechanical variables were calculated or identified from the effective force profile: the mean values of effective force (F_{cycle}, N), pedalling rate (f_{cycle}, r.p.m.) and power output (P_{cycle}, W) over one complete cycle, the maximal (peak) value of the effective force exerted during the downstroke (F_{max}, N) and minimal value exerted during the upstroke (F_{min}, N) and the arm crank angle corresponding, respectively, to F_{max} ($Angle_{max}$, 1) and F_{min} ($Angle_{min}$, 1). To quantify the muscle activity pattern, a series of classical variables were calculated from the EMG RMS linear envelope. The overall activity level was identified by the mean EMG

RMS magnitude over one complete cycle (RMS_{cycle}). The EMG timing analysis consisted in determining onset and offset of the burst of muscle activation which was defined as the period of higher activity phase where the signal was above a threshold of 20% of the difference between peak and baseline EMG (Li & Caldwell, 1999). The technique of cross-correlation was also used to measure the relative change in the temporal characteristics of neuromuscular activity (Li & Caldwell, 1998, 1999). Firstly, the cross correlation coefficients of EMG RMS curves between the three body configurations for each muscle were calculated according to the equation proposed by Li and Caldwell (1998) with lag time equal to zero (r_0). Then, the magnitude of a significant time shift between signals obtained in two different body positions was found by assessing the k value (k_{max}) at which the cross correlation coefficient was maximized (r_{max}). The purpose was to estimate the effects of body position on linear envelope patterns of the different muscles by using this more recent and objective approach for comparing signals.

Statistical analysis

All analyses were performed with the statistical package SPPS 11.0 and ORIGIN 6.1 software for Windows. Data were first tested for normality using Kolmogorov–Smirnov test. Because the normality condition was verified, the results are expressed as mean ± standard deviation (± SD). One-way analysis of variance (ANOVA) with repeated measures was employed to test the effect of the three body postures on all the ventilatory, gas-exchange, mechanical and EMG variables at both exercise intensities. When significant F ratios were found, all the means were compared using a Tukey's *post hoc* test. Differences were considered significant when probability (P) of a type I error was ::: 5%.

Results

Ventilatory and gas-exchange variables

No significant difference was found for the ventilatory and gas-exchange variables measured at VT1D20% among the three body positions (Table 2). Because no metabolic steady state was achieved during the exercise bouts at RCP, results are not reported for this intensity.

Effective force profile

To achieve the power outputs corresponding to VT1D20% and RCP, similar mean effective forces and pedalling rates were maintained among the three body positions (Table 3). Figure 2 depicts the pattern of the mean effective force over the crank cycle for all conditions. F_{min} was significantly lower for UP than for DP and AP at VT1D20% and F_{max} was significantly higher for AP than for UP at RCP (Table 3). For sake of clarity, mean values of effective force were calculated for four angular sectors (Fig. 3): Sector 1 represented 330–30l; Sector 2, 30–150l; Sector 3, 150–210l and Sector 4, 210–330l. Sectors 1 and 3 correspond respectively to the top and bottom dead centers; Sectors 2 and 4 correspond respectively to the main propulsive and recovery phases. For AP, effective force at VT1D20% was significantly higher in Sector 3 and lower in Sectors 1 and 4 compared with the other positions. In addition to these differences which persist at RCP, effective force for AP was significantly higher in Sector 2 than for UP at RCP (Fig. 3). Furthermore, angles corresponding to minimal ($Angle_{min}$) and maximal ($Angle_{max}$) effective forces were significantly higher for AP than for UP and DP (only for $Angle_{max}$) at RCP (Table 3).

Table 2. Ventilatory and gas-exchange variables (mean ± SD) measured at VT1D20% for the three body positions. All the values were averaged during the last 2 min of each bout

	Aero posture (AP)	Dropped posture (DP)	Upright posture (UP)
VO_2 (mL/min)	3458 ± 297	3368 ± 270	3394 ± 234
VE (L/min)	88.7 ± 11.6	86.7 ± 10.9	84.4 ± 9.4
RER	0.94 ± 0.03	0.93 ± 0.04	0.92 ± 0.03
VT (L)	2.83 ± 0.45	2.73 ± 0.41	2.86 ± 0.41
Bf (cycles/min)	32.0 ± 5.7	32.8 ± 6.3	30.2 ± 6.1

VO_2, oxygen uptake; VE, ventilatory flow; RER, respiratory exchange ratio; VT, tidal volume; Bf, breathing frequency.

Muscle activity level

At VT1D20%, AP and DP induced significantly higher muscle activity level (RMS_{cycle}) compared with UP for GMax. RMS_{cycle} for SOL was also higher for AP compared with the other positions (Fig. 4). At RCP, these differences persisted only for GMax, while other differences of muscle activation appeared: RMS_{cycle} was significantly higher for AP than for DP and UP for VM, and in AP than for DP for VL. In contrast, UP induced a higher activity level of RF than did any of the other positions.

Muscle activation timing

Muscle activity patterns from the three cycling positions are represented with ensemble linear envelopes (45 or 90 consecutive cycles x 12 subjects per condition and per intensity) of the EMG RMS data (Fig. 5(a) and (b) for VT1D20% and RCP, respectively). Figure 6 depicts the mean values of the muscle activation timing variables obtained at VT1D20% and RCP. The onset of activation occurred later for RF in AP condition than it did for UP at both VT1D20% and RCP. GMax, VL and VM were also recruited later in AP and DP than for UP, only at RCP. The offset of activation of SM occurred later in AP compared with DP. However, no significant difference was found among the three body positions in muscle activation timing for the four leg muscles. All these results were confirmed and emphasized by the cross-correlation analysis (Table 4). The k_{max} values demonstrate a significant pattern shift (i) among the three conditions for GMax, RF and VL,

Table 3. Mechanical variables (mean ± SD) measured at VT1D20% and RCP for the three body positions

	VT1D20%				RCP			
	AP	DP	UP		AP	DP	UP	
P_{cycle} (W)	238.2 (22.9)	238.0 (22.6)	238.2 (22.8)	NS	318.6 (27.6)	318.0 (28.1)	318.5 (27.7)	NS
f_{cycle} (rpm)	89.2 (8.1)	89.6 (10.2)	89.6 (9.3)	NS	90.3 (8.9)	90.3 (9.1)	90.2 (9.6)	NS
F_{cycle} (N)	75.9 (11.7)	75.8 (12.8)	75.8 (12.8)	NS	100.1 (14.1)	99.9 (13.7)	100.4 (14.6)	NS
F_{max} (N)	324.8 (31.8)	317.0 (33.9)	320.0 (44.3)	NS	386.9 (45.2)	380.4 (44.5)	375.4 (41.0)	*
$Angle_{max}$ (l)	95.7 (7.3)	94.0 (6.0)	92.7 (4.6)	NS	95.8 (5.5)	90.8 (5.7)	90.3 (5.6)	# and *
F_{min} (N)	−70.0 (26.1)	−65.6 (27.2)	−63.9 (26.6)	# **	−53.2 (31.7)	−52.6 (29.6)	−49.1 (30.4)	NS
$Angle_{min}$ (l)	268.0 (20.6)	267.7 (22.9)	265.2 (23.2)	NS	284.5 (25.3)	275.8 (25.0)	271.4 (25.6)	**

AP, aero posture; DP, dropped posture; UP, upright posture.

P_{cycle}, f_{cycle}, and F_{cycle}: mean values of power output, pedalling rate and right pedal effective force calculated over one complete cycle; F_{max} and F_{min}: maximal and minimal value of the effective force exerted during the cycle; $Angle_{max}$ and $Angle_{min}$: respective arm crank angle corresponding to F_{max} and F_{min}.

#Significant difference between AP and DP (*P*00.05).

*Significant difference between AP and UP (*P*00.05).

**Significant difference between AP and UP (*P*00.01).

(ii) between UP and the other conditions for VM, (iii) between AP and DP for SM and BF and AP and UP for SM. The maximal values of cross correlation coefficients (r_{max}) indicate a very high degree of similarity in the EMG activation patterns among the three conditions for all muscles (r_{max}: 0.960–0.997, Table 4).

Discussion

The present investigation is the first to focus simultaneously on both the mechanical aspects of pedal force production and lower limb muscle activation patterns in response to changes in upper body position. Despite the stability of both ventilatory and gas-exchange variables, the results of this study demonstrate significant alterations in pedal effective force and in the level and timing of activation of some lower limb muscles when the upper body configuration is changed, especially from the upright to the AP.

Methodological and general considerations

In addition to randomization, 5-min submaximal exercise at 150 W was performed just before and after the experimental protocol in order to verify the repeatability of the lower limb muscles activation patterns. These results have been previously published (Dorel et al., in press) and show a very good repeatability of the EMG patterns.

The two exercise intensities were chosen in order to reproduce typical training intensities (i.e. VT1 D20%; corresponding to 238.1 ± 22.8 W) and specific intensities maintained during competitive time trial events (i.e. RCP: corresponding to 318.4 ± 27.8 W). Moreover, VT1D20% intensity ensured that a true metabolic steady state was achieved allowing the comparison of ventilatory and gas-exchange variables among the different body configurations. Unlike Gnehm et al. (1997), the present investigation failed to show significant modifications in oxygen consumption or other ventilatory/gas-exchange variables among the three body positions. These results are consistent with the majority of previous studies which also reported no significant posture effect despite some tendencies towards higher metabolic cost and ventilatory flow in the AP (Origenes et al., 1993). Overall, these results confirm the difficulty of demonstrating a true inconvenience in term of metabolic cost in the AP, especially in well-trained cyclists. Indeed, the triathletes serving as subjects in the present work were accustomed to using the three tested positions during training and competition. Moreover, it can be argued that the criteria selected to standardize the positions, and especially the aero position, were voluntarily chosen to not induce larger modifications than those that are common on the field. The question remains to be answered whether a more extremely aerodynamic position used by elite time-trial cyclists (i.e. with a greater torso inclination) and reported in previous studies (Gnehm et al., 1997) would induce greater

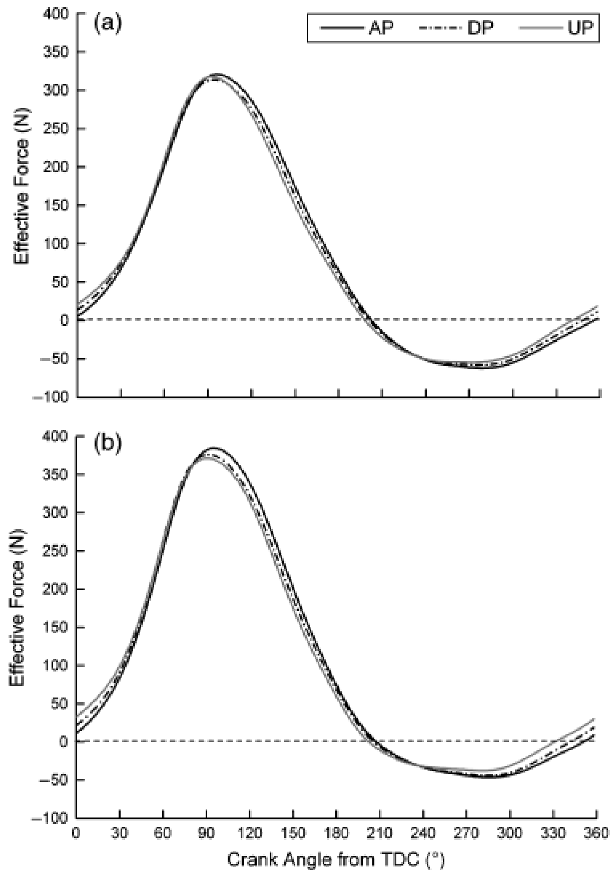

Fig. 2. Mean profile of the effective pedal force measured during the exercises at VT1D20% (a) and respiratory compensation point (RCP) (b) in the three upper body positions for the entire population. AP, aero posture; DP, dropped posture; UP, upright posture. Note the higher peak value of force produced during the second part of the downstroke phase and the lower value during the last part of the upstroke and the beginning of the downstroke for AP compared with UP.

alterations in ventilatory response as well as in mechanical and neuromuscular responses.

The results concerning the influence of body position on force and EMG variables do not noticeably differ among the two intensities conditions. In a great majority, the significant changes or tendencies observed at VT1D20% are respectively amplified or begin significant at RCP. As a consequence, considering these similarities and the purpose of the study, the influence of exercise intensity will not be extensively detailed afterward so that the following discussion will focus on the results obtained at RCP.

Effective force

As expected, mean power output, pedalling rate and hence mean pedal effective force were kept constant among the three conditions for the two exercise intensities (Table 3), allowing the comparison of mechanical and EMG variables in respect to the

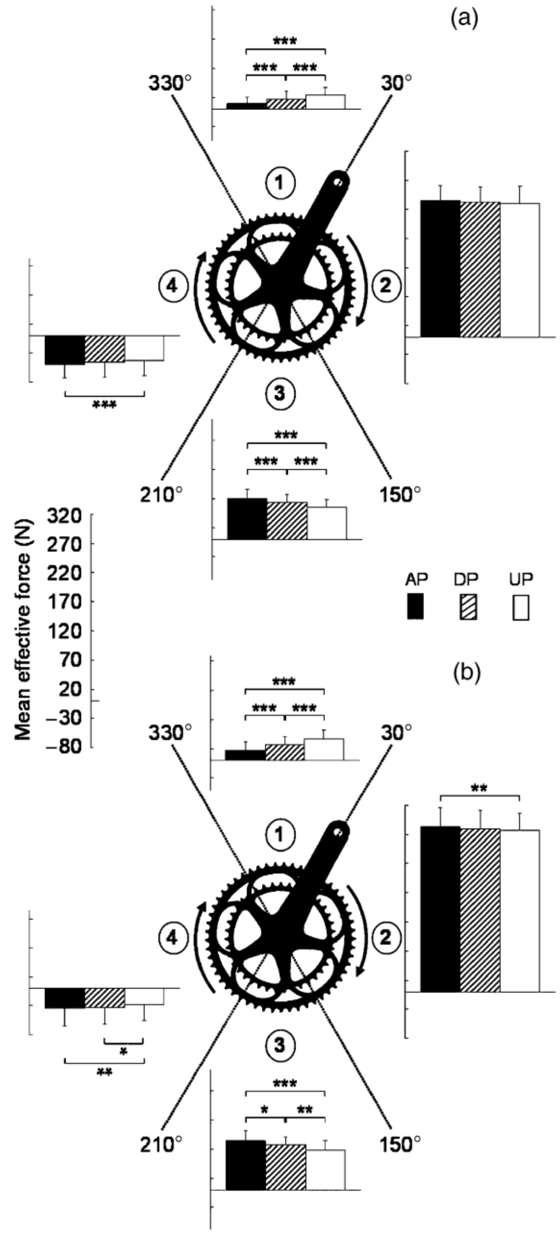

Fig. 3. Mean values of effective force calculated for four angular sectors as a function of crank position over the complete cycle in the three body configurations [(a) at VT1D20% and (b) at respiratory compensation point (RCP)]. Sector 1, 330–30l; Sector 2, 30–150l; Sector 3, 150–210l and Sector 4, 210–330l. AP, aero posture; DP, dropped posture; UP, upright posture. ****P*o0.001, ***P*o0.01, **P*o0.05 significant difference between two conditions.

body positions. Whereas several studies have reported changes in the torque profile in response to alterations of pedalling rate, power output, surface grade, and/or the occurrence of fatigue (Patterson & Moreno, 1990; Li & Caldwell, 1998; Sanderson et al., 2000; Sanderson & Black, 2003; Bertucci et al., 2005), few of them focused on the effects of different body configurations (Brown et al., 1996; Li & Caldwell,

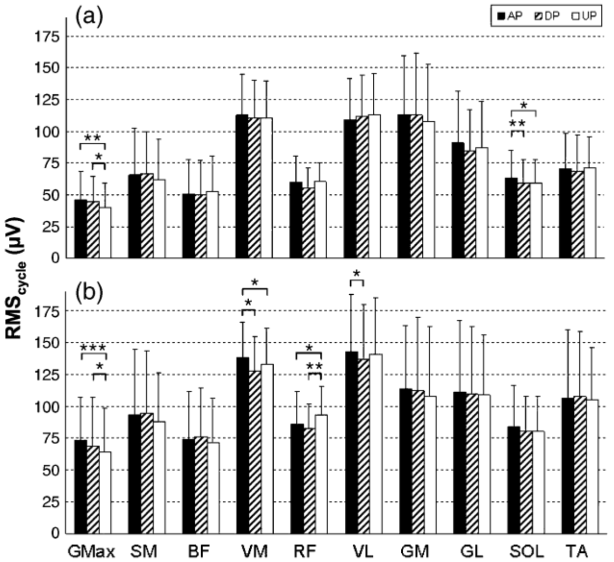

Fig. 4. Mean values of the root mean squared (RMS) magnitude for the complete cycle (RMS_{cycle}, i.e. 0 to 360l) for the three body positions (AP, aero posture; DP, dropped posture; UP, upright posture) at both constant submaximal power outputs [VT1D20% and respiratory compensation point (RCP)] for 10 lower limb muscles implied in pedalling. GMax, *gluteus maximus*; SM, *semimembranosus*, BF, *Biceps femoris*; VM, *vastus medialis*; RF, *rectus femoris*; VL, *vastus lateralis*; GM, *gastrocnemius medialis*; GL, *gastrocnemius lateralis*; SOL, *soleus* and TA, *tibialis anterior*.

1998). Moreover, they concerned extreme postural adjustments such as standing vs seated (Li & Caldwell, 1998) or recumbent vs traditional posture (Brown et al., 1996). Our results clearly demonstrate that AP, compared with DP and to a large extent to UP, leads to an increase in the magnitude of the negative effective force during the upstroke (Sector 4 and F_{min} more negative) and a decrease of the slight positive force produced just before and after TDC (Sector 1) (Figs 2 and 3). In order to counterbalance these larger counterproductive regions for AP (i.e. mean effective force at RCP in Sector 4 and 1 amounts – 8.6% and – 3.2% of F_{cycle} for AP and UP, respectively), a greater force would be required during the downstroke for AP (i.e. mean effective force in Sectors 2 and 3 amounted 108.6% and 103.2% of F_{cycle} for AP and UP, respectively). In this line, a significantly higher F_{max} of 3.1% was observed for AP compared with UP at RCP. These results are the first to confirm the pioneering observations of Broker (2002) on cyclists and triathletes. Indeed, this author also reported effective force profiles in AP that are similar to those observed in the present study (i.e. with greater peak-to-peak oscillations). As argued by several authors, the lower hip angle towards more flexion (from almost – 20l from UP to AP) alters the mean working length of muscles crossing this articulation (RF, BF, and GMax) which could be responsible for the modification in the amount of force produced by these muscles (Gnehm et al., 1997; Ashe et al., 2003) and finally could change the force profile. It is noteworthy that these force profile adaptations in AP position could have not been detected by measuring only net crank torque (Bertucci et al., 2005). The direct measurement of each pedal effective force allowed a further analysis of the interactions among the specific changes in body configuration, these kinetics parameters and finally the activation patterns of the corresponding lower limb muscles.

EMG patterns

Comparing the EMG changes observed in the present study to those reported on more drastic body orientation manipulations (Brown et al., 1996; Savelberg et al., 2003a) or on differences between standing *vs*. seated positions (Li & Caldwell, 1998; Duc et al., in pres), it becomes clear that the effects of the aero position are less important. However, it is noteworthy that significant modifications of EMG patterns reported in our study seem to be directly linked to changes of the effective force pattern. The absence of change in both the intensity and the timing of the EMG activity of muscles crossing the ankle joint (GM, GL, TA, SOL) confirmed the relative stability of EMG pattern of these muscles in different body postures already reported in the literature: seated vs standing (Duc et al., in press), flexed forward vs upright (Savelberg et al., 2003a). Only one study has focused on the effects of trunk inclination on the activation pattern of lower limb muscles (Savelberg et al., 2003a). Although the body configurations tested in this later study are far away from ecological conditions (i.e. racing conditions), it is interesting to note that our results are consistent with these authors (Savelberg et al., 2003a) who reported a higher activation for GMax and a lower activation for RF in a flexed forward compared with an extended backward configuration. As a consequence, the decrease of pedal force found in AP during the upstroke and TDC phases could be explained by this lower activation of RF. Although this biarticular muscle both acts as knee extensor and hip flexor, a detailed analysis of RF EMG patterns (Fig. 5(b)) confirms that its activation is affected during the upstroke phase (i.e. while acting as hip flexor). However, it should be kept in mind that the relation between muscle activation and force production is not so simple because the increase of trunk inclination (i.e. flexed forward), by modifying pelvis orientation, has been showed to induce alteration of RF and GMax muscle lengths (Savelberg et al., 2003a). Then, it could be hypothesized that in aero position disadvantage due to more marked stretching of GMax (which induces passive resistance) and more

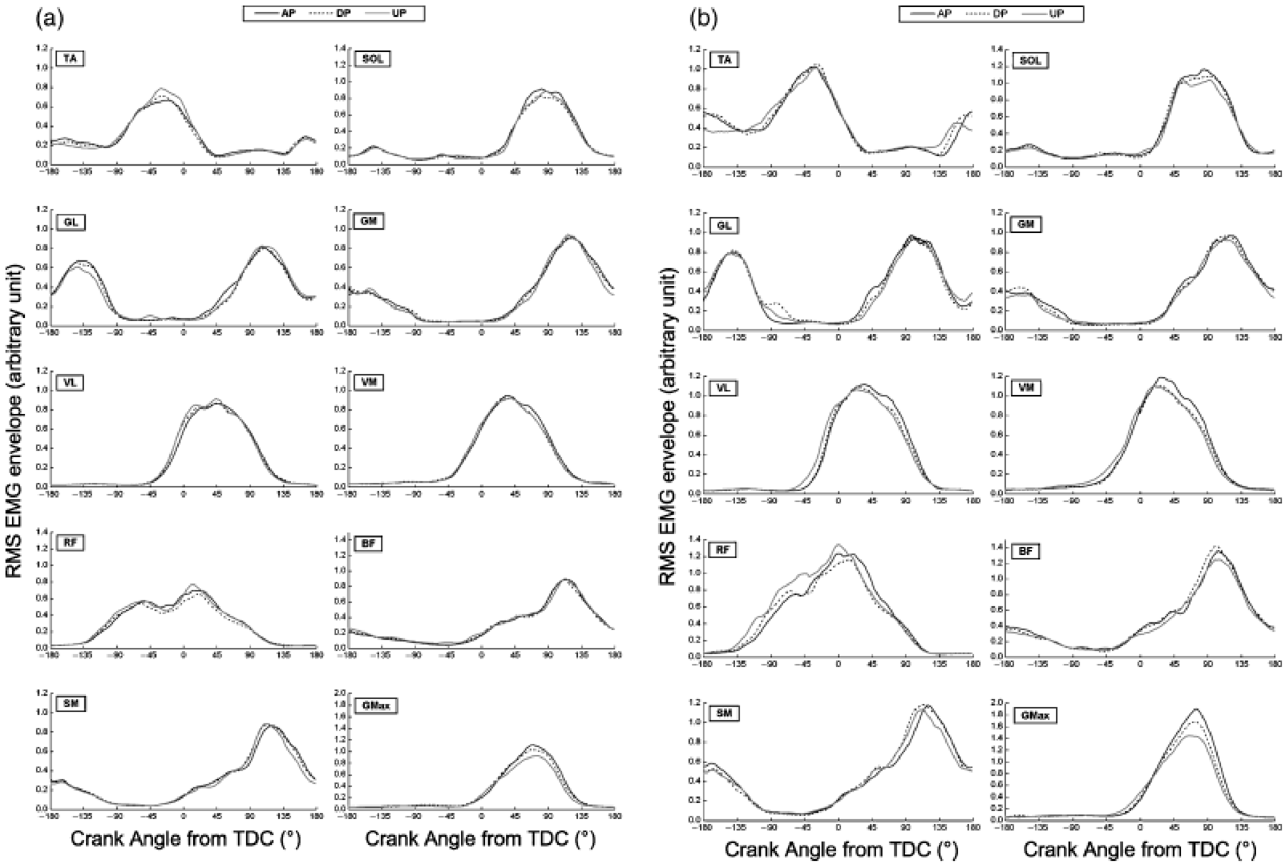

Fig. 5. Ensemble curves of electromyography (EMG) root mean squared (RMS) linear envelope for 10 lower limb muscles for all body positions [A/at VT1D20% and B/at respiratory compensation point (RCP)]. AP, aero posture; DP, dropped posture; UP, upright posture. For sake of clarity, all the curves on one panel use same arbitrary unit on vertical axes (AP and DP tests are normalized to maximal EMG obtained during UP). TDC, top dead centre (0l); BDC, bottom dead centre (180l).

marked shortening of RF (which affects its ability to generate force) partly explain the decrease of pedal force during the upstroke and TDC phases.

Considering that the knee joint kinematics is not notably affected by upper body position in seated position (Li & Caldwell, 1998; Savelberg et al., 2003a), the greater effective force found for AP during the downstroke phase could partly be explained by the higher activation of VM (in AP compared with other conditions) and VL (in AP compared with DP). Moreover, even if it is more difficult to conclude on the mechanical advantage/disadvantage of modification of GMax muscle length in AP condition, it is reasonable to think that the significant increase of GMax activity (113.6% from UP to AP) also contributes to the increase of effective force during the propulsive phase (i.e. during downstroke). Then, because hip flexors was reported to only contribute to 4% of the total positive work (Ericson, 1986), it could be surmized that the subjects spontaneously adopted a compensatory strategy with these monoarticular hip and knee extensors rather than an increase of RF activity in AP condition. To confirm this assumption, it could have been interesting to record other hip flexor muscles as the Psoas major. However, this deep muscle is difficult to record using surface EMG.

Another variable of interest in examining the EMG patterns is the muscles' activation timing. As done in previous studies (Li & Caldwell, 1998; Dorel et al., in press), both onset/offset of EMG bursts and cross-correlation coefficients were used to give an objective estimation of similarities and temporal characteristics of muscle activity patterns. Despite similar patterns among the three riding positions, timing analysis showed: a phase shift (i.e. later activation) of EMG activity for the hip extensor (GMax), flexor (RF) and two knee extensors (VL and VM) from upright to drop and AP (Table 4, Fig. 6). The discrepancy in the values of shift crank angle observed between the two methods can be explained by methodological considerations (Li & Caldwell, 1998; Dorel et al., in press). In fact, the shift reported using ON-OFF method only concerned the onset of the burst (except for VL) whereas the phase shift obtained from the cross correlation method took the entire EMG profile into account. Overall, it is noteworthy that this result could also

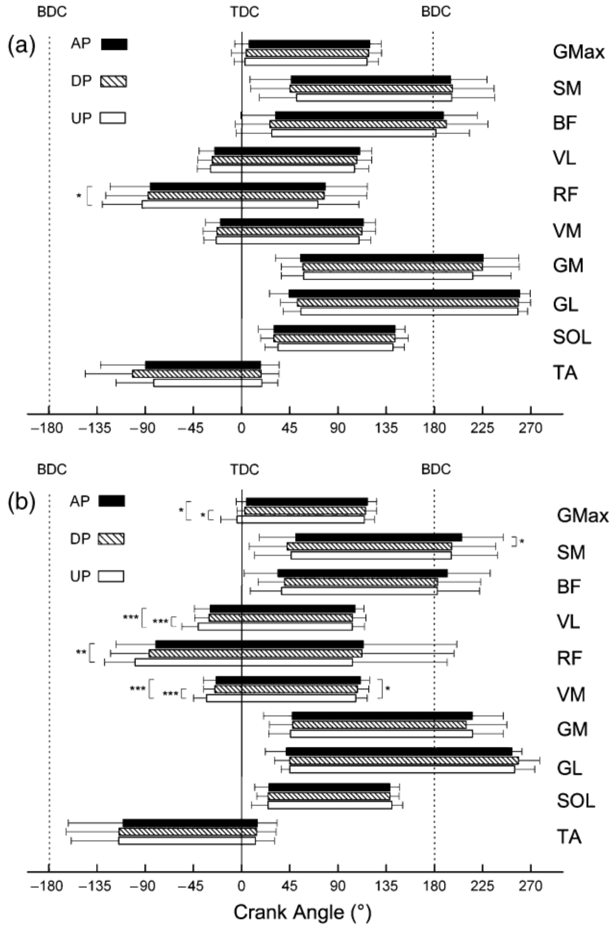

Fig. 6. Mean onset, offset and duration of higher electromyography (EMG) activity phase indicated by bars for the 10 muscles, displayed as a function of crank position. Only the main burst is represented when two bursts were observed for some muscles and some subjects. A/at VT1D20% and B/ at respiratory compensation point (RCP). TDC, top dead centre (0l); BDC, bottom dead centre (180l). ****P*o0.001, ***P*o0.01, **P*o0.05 significant difference between two conditions.

Table 4. Coefficient of cross correlation and shift angle for the EMG RMS linear envelopes of each muscle for aero vs dropped positions (AP vs DP), for dropped vs upright positions (DP vs UP) and for aero vs upright positions (AP vs UP) at VT1D20% (A) and RCP (B)

	AP vs DP			DP vs UP			AP vs UP		
	r_0	r_{max}	k_{max} (deg.)	r_0	r_{max}	k_{max} (deg.)	r_0	r_{max}	k_{max} (deg.)
(A)									
GMax	0.990	0.996	**2.4***	0.990	0.995	**1.5***	0.987	0.995	**3.8***
SM	0.980	0.993	**3.2***	0.987	0.992	0.8	0.981	0.990	**3.7***
BF	0.980	0.993	2.1	0.984	0.985	−1.4	0.973	0.981	0.9
VM	0.990	0.996	**2.3***	0.992	0.996	0.5	0.989	0.996	**3.2***
RF	0.977	0.982	1.6	0.977	0.989	**2.3***	0.969	0.983	**3.8***
VL	0.990	0.997	**2.7***	0.992	0.998	**1.6***	0.989	0.997	**4.1***
GM	0.986	0.996	0.0	0.988	0.992	1.8	0.982	0.988	2.2
GL	0.975	0.984	0.0	0.987	0.990	0.5	0.965	0.972	0.7
SOL	0.982	0.994	0.0	0.989	0.995	−1.0	0.983	0.992	−0.6
TA	0.968	0.987	0.2	0.969	0.979	−1.9	0.949	0.972	−1.7
(B)									
GMax	0.989	0.996	**2.6***	0.984	0.993	**2.4***	0.974	0.990	**5.2***
SM	0.978	0.988	**3.3***	0.986	0.989	1.3	0.967	0.982	**5.0***
BF	0.983	0.993	**3.6***	0.979	0.988	−0.9	0.968	0.982	2.3
VM	0.992	0.996	1.5	0.988	0.995	**3.8***	0.983	0.993	**5.7***
RF	0.979	0.988	**4.4***	0.975	0.985	**4.9***	0.949	0.977	**9.7***
VL	0.993	0.997	**2.1***	0.988	0.995	**4.3***	0.979	0.993	**6.4***
GM	0.981	0.990	1.3	0.987	0.991	1.8	0.971	0.986	**3.6***
GL	0.956	0.963	1.6	0.969	0.976	−0.4	0.943	0.960	1.6
SOL	0.989	0.994	1.0	0.982	0.989	−0.9	0.979	0.988	1.0
TA	0.972	0.986	−0.4	0.972	0.984	1.8	0.945	0.978	1.1

Values are cross correlation coefficients (r_0) with no shift angle (k50), and maximal value of cross correlation coefficients (r_{max}) obtained with a shift angle k5k_{max}. Only the shift angle values with asterisk (and in bold) are considered as significant according to the method (95% of confidence interval) proposed by Li and Clawell (1999). Positive *k*-value corresponds to a backward shift in crank angle and indicates EMG activity of the second condition cited is shifted *k* degree(s) earlier in the crank cycle. GMax, *gluteus maximus*; SM, *semimembranosus*; BF, *Biceps femoris*; VM, *vastus medialis*; RF, *rectus femoris*; VL, *vastus lateralis*; GM, *gastrocnemius medialis*; GL, *gastrocnemius lateralis*; SOL, *soleus*; TA, *tibialis anterior*.

explain the differences observed in effective force distribution along the crank cycle (Figs 2 and 3) and is completely consistent with the shift of the Angle$_{max}$ corresponding to a later application of the maximal effective force during the downstroke phase (i.e. 5.5l from UP to AP; Table 3). Finally, the later burst offset observed for SM in AP compared with UP (Fig. 6) was confirmed by the phase shift of 5l obtained by the cross correlation method. This result could be induced by the fact that, as argued by Van Ingen Schenau et al. (1992), hamstring activity might be employed to transfer the power produced by monoarticular muscles (GMax, VM, VL). As a consequence, these muscles could be activated later in the way similar to the monoarticular muscles. It this case, it remains unclear why BF activity seems not to be affected in the same way. However, in line with hypotheses of Ericson (1988) according to which SM acts more as knee flexor than as hip extensor, it is also possible that the shift of SM allows subjects to generate a higher propulsive force in the BDC phase (i.e. Sector 3, Fig. 3).

Conclusion

This study shows that riding in AP induces significant alteration of both intensity and timing of EMG activity of lower limb muscles crossing hip (GMax, RF) and knee joints (VM, VL, and to a lower extent SM) whereas muscles crossing the ankle joint were unaffected. The EMG activity modifications (increased activity for GMax, VL, VM, and decreased activity for RF) and the shift forward reported when the upper body configuration is changed (especially from UP to AP) is strongly related to the changes of pedal effective force profile: i.e. higher downstroke peak positive force, lower upstroke peak negative force and later occurrences of these values along the crank cycle. Further investigations are needed to evidence the necessity or not of training in this specific AP and to clarify evolution of these adaptations during an exhausting exercise such a time trial event.

Perspectives

It is obvious that the benefits of reducing air drag in AP (and to a lesser extent in DP) on the field (Capelli et al., 1993), far outweigh the disadvantage of a slightly increased activity of some lower limb muscles. Nevertheless, this higher activity in AP seems not to be negligible in the course of a time trial event since it would induce greater neuromuscular fatigue. Thus, it questions on the pertinence to perform a specific training program to improve mechanical patterns for cyclists or triathletes using an aero-position in competition. Along this line, Lucia et al. (2000) suggested that professional cyclists who show best performance in time trials may have a greater ability to pull the pedal up slightly, resulting in less work by the knee and hip extensors. Although we focused essentially on the differences between AP and UP, our results also showed significant higher activation of knee and hip extensors and lower activation of hip flexors in DP compared with UP (at RCP). The fact that cyclists naturally adopt UP as an alternative to AP (during a time trial) or DP (during a race) when air drag is less important (e.g. climbing) could therefore be an interesting strategy for optimizing the EMG pattern in order to pull the pedal better during the upstroke and hence delay fatigue in hip and knee extensor muscles.

Key words: body position, aero position, dropped position, upright position, electromyography, torque, cycling, triathlon.

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