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Influence of Position and Power Output on Upper Limb Kinetics in Cycling

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Several suggestions on the upper limb involvement in cycling exist but, to date, no study has quantified upper limb kinetics in this task. The aim of this study was to determine how crank power and pedaling position (seated or standing) affect upper limb kinetics. Handlebar loadings and upper limb kinematics were collected from 17 participants performing seated or standing pedaling trials in a random order at 6 crank powers ranging from 20% (112 ± 19 W) to 120% (675 ± 113 W) of their spontaneous sit-to-stand transition power. An inverse dynamics approach was used to compute 3D moments, powers, and works at the wrist, elbow, and shoulder joints. Over 29 parameters investigated, increases in crank power were associated with increases in the magnitudes of 23 and 20 of the kinetic variables assessed in seated and standing positions, respectively. The standing position was associated with higher magnitudes of upper limb kinetics. These results suggest that both upper and lower limbs should be considered in future models to better understand whole body coordination in cycling.

Keywords: inverse dynamics, joint torques, powers, works, standing

Previous work dedicated to pedaling performance and injury prevention in cycling mainly focused on lower limb biomechanics.¹ However, because the upper limbs may provide significant contributions in trunk support, stabilization, or power transfer to the cranks,² knowledge in regard to upper limb kinetics in cycling could be useful for both performance improvement and clinical applications.

To date, only indirect insights into upper limb actions have been available. Previous studies assessed rider-induced forces and moments on the handlebar and showed that both pattern and magnitude of the crank power, and position (ie, seated or standing), influenced handlebar loadings. At crank power corresponding to ≈ 200 W, the handlebar forces were always oriented in both the forward and downward directions in a seated position, suggesting that the role of the arms is mainly to support the torso weight.³ Conversely, at crank power higher than 200 W, alternation between pulling and pushing forces on the handlebar have been observed both in seated and standing positions, suggesting that the arms play a different role as crank power increases.^{4,5}

Upper limb contribution to crank power has never been assessed directly but its role has been suggested by Baker and colleagues.⁶ They showed that maximal crank power was lower without handgrip on the handlebar (1136 ± 88 W) in comparison with handgrip (1461 ± 94 W), suggesting that the upper limb may influence the

crank power. Nevertheless, hand gripping could influence the lower limbs' performance by inducing greater muscular activations,^{7,8} suggesting that muscular/mechanical upper limbs' contribution to the crank power needs further verification.


Further insights about the contribution of the upper limbs in cycling have been highlighted through the quantification of energy transfer through the hip joint. Indeed, inverse dynamics models of cycling have shown that crank power could be decomposed into the sum of ankles, knees, hip powers, and hip power transfer. This last term, calculated as the dot product between hip joint linear velocity and hip reaction force, is presumably partly generated by the upper limbs.² One study compared the seated and standing positions and showed higher magnitudes of hip power transfer in a standing position.⁹ In a seated position, power transfer through the hip joint has been reported to represent 5% of the crank power at 540 W,^{9,10} a percentage that could increase with power output (ie, from 3% at 250 W to 8% at about 1000W¹¹), suggesting that the upper limbs may contribute to crank power.

These indirect evidences prove the importance of the upper limbs' involvement in the cycling movement, however knowledge of the level of upper limb joint moments or upper limb power production in cycling is still lacking. Therefore, the aim of this study was to test the effect of power output and position on the magnitudes of upper limb joint moments, powers, and works. Our hypothesis is that the magnitude of upper limb kinetics could be increased by an increase in crank power output and/or the use of a standing position instead of a seated one.

Methods

Seventeen male students (age: 23.2 ± 3.4 y, height: 1.77 ± 0.06 m, body mass: 73.7 ± 8.8 kg) volunteered to participate in the study. These participants were chosen to avoid expertise effects and were classified as category 4 (recreational) or 5 (athlete noncyclist)

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cyclists according to Ansley and Cangle's classification (category 1 would be professional cyclists riding more than 30000 km per year and category 6 would be sedentary noncyclists).¹² Each participant was informed of the experimental procedure and signed an informed consent form before the study. The experimental design of the study was conducted in accordance with the Declaration of Helsinki and approved by the University of Toulouse ethical committee. Participants were asked to avoid high-intensity or exhaustive exercise at least 72 hours before the laboratory trials.

Protocol

The cycling tests were performed using an electromagnetically braked Excalibur (LODE, Groningen, Netherlands). The rate of power output variation imposed by the ergometer was set to 1000 W·s⁻¹. Participants first performed a cycling test to determine their spontaneous sit-to-stand transition power. This methodology was used in analogy with the walk-run transition to provide different percentages of power output according to a common biomechanical reference for the participants. In this test, phases of 40 seconds at a crank power of 50 W were alternated with phases of 20 seconds with an increment in power. In these 20-second phases, the starting crank power was 200 W, and was incremented by 25 W at each step. This protocol including intervals was chosen after preliminary trials showed that some participants may reach exhaustion before the sit-to-stand transition when using different protocols of crank power increased without recovery. The participants were instructed before the test to maintain a pedaling cadence of 90 revolutions per minute (rpm) provided by a visual feedback. During the test, the participants were encouraged to maintain 90 rpm as soon as their cadence was under or above 90 ± 5 rpm. The sit-to-stand transition

power was considered as the crank power at which participants rose from the saddle during at least 10 seconds when they could not maintain 90 rpm in a seated position.

After a 5-minute rest period, participants performed 12 randomized trials in either a seated or standing position at crank power corresponding to 20%, 40%, 60%, 80%, 100%, or 120% of their sit-to-stand transition power. Each pedaling trial recording lasted 10 seconds and was started as soon as the participant was able to steadily pedal at the target crank power at 90 rpm. The pedaling cadence was standardized for all the participants because of its possible influence on the upper limb kinetics. Three minutes of passive rest were allowed between each trial.

Standardized Positioning

To limit bike positioning effects, standardized settings were established. The seat tube angle was set to 73° and the crank length was 0.17 m. The pedal cleat was positioned under the first metatarsal bone.¹³ The saddle height was set to obtain a 150° knee angle during full leg extension. The handlebar was flat and positioned to standardize drop and reach lengths according to torso and arm lengths.¹⁴ The positioning of the 2 hands on the handlebar was left up to the participant (handlebar width: 0.7 m, sweep angle: 0°).

Kinematics

Kinematic data from 18 passive markers on the upper limbs were recorded by 12 infrared cameras (VICON, Oxford, UK) at 200 Hz (Figure 1). The anatomical landmarks were selected according to de Leva.¹⁵ The wrist and elbow joint centers were defined as the midpoints between the processus styloideus ulnae and radii,

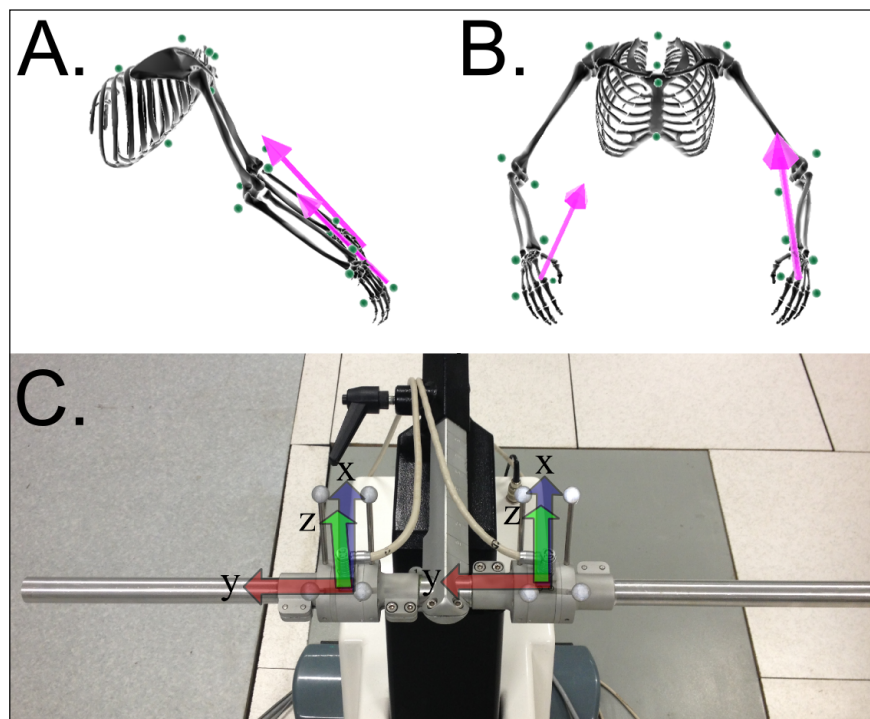


Figure 1 — (A) Sagittal view of the marker set used to reconstruct the upper limb kinematics. (B) Frontal view. The arrows represent the reaction forces recorded by the handlebar sensors and the dots represent the passive markers used to model the bone movements and orientations. (Figures made with Opensim 3.2 software.³⁴) (C) Handlebar with 6-component sensors and the marker set used to locate the sensor origins and orientations; x, y, and z represent the anteroposterior, mediolateral, and vertical axis, respectively.

and between the lateral and medial epicondyles, respectively. The shoulder joint center was located using the SCoRE method.¹⁶ For this method, a specific procedure previously suggested to locate the hip joint center was applied.¹⁷ In a preliminary recording, the participants were asked to repeat shoulder flexion-extension, abduction-adduction, and circumduction. The SCoRE method then allowed the localization of the shoulder joint centers relative to a coordinate system attached to the thorax. Coordinate systems for each of the 3 segments were defined as recommended by the International Society of Biomechanics (ISB).¹⁸ To provide upper limb loadings according to crank angle, 2 additional markers were added on the anterior and posterior part of a strip embedded externally to the pedals to determine the position of each pedal axis.

Handlebar Sensors

The 3D force and moment vectors applied to the handlebar were recorded from 2 tubular sensors (SENSIX, Poitiers, France) positioned on each side of the handlebar (Figure 1C). These dynamometers had a maximum 1% error on each direction (combining linearity and hysteresis errors), and a maximum 1.5% error on the 6 component combination. All measures were given at the sensor reference point, which was localized by 4 reflective markers in the global reference frame. These markers also allowed the localization of the sensor orientation and were fixed to the sensors with a built-in interface.

Inverse Dynamic and Data Analysis

A Newton-Euler recursive algorithm was implemented to perform the bottom-up inverse dynamics protocol used to calculate the 3D joint moments, successively at the wrist, elbow, and shoulder joints (details of the equations in Winter¹⁹). Handlebar reaction forces and moments vectors were translated from the sensor reference point to the wrist joint center using Varignon's formula. Body segments' inertial parameters were derived from the scaling equations given in de Leva.¹⁵ Handlebar loadings and upper limb kinematics were synchronized using Nexus 1.7.1 system (VICON, Oxford, UK) and marker position data were filtered using a fourth-order, zero-phase-shift, low-pass Butterworth with an 8-Hz cutoff frequency.²⁰

To obtain the joint loadings in function of crank angle, data from a total of 15 pedaling cycles were averaged for each value of crank angle obtained during the pedaling trials of the second part of the study (from 2% to 120% of the sit-to-stand transition power in a random order). Moments applied to the upper limb were the net joint moments at the wrist, elbow, and shoulder, and the external forces were the handlebar reaction force and the joint force acting on the upper arm at the shoulder. Joint moment vectors were expressed according to the segment coordinate systems described in the Kinematics section earlier. External forces and moments developed mechanical energy if they produced or absorbed power during the movement.²¹ Joint powers were calculated at each joint as the product of the joint moment and joint angular velocity, the power transferred across the shoulder joint was computed as the dot product between the shoulder reaction force and its linear velocity, and handlebar power was considered to be zero given the fixed cycling ergometer.² The instantaneous power equation specific to the upper limbs in cycling was therefore:

$$P_t = \sum_1^3 (P_j) + \mathbf{F}_{sf} \cdot \mathbf{V}_{sf} = \frac{dE_t}{dt}, \quad (1)$$

where P_t represents the total upper limb power, P_j represents the joint powers, \mathbf{F}_{sf} represents the reaction force at the shoulder, \mathbf{V}_{sf}

represents the shoulder linear velocity, and $\frac{dE}{dt}$ represents the change in mechanical energy of the upper limb. The upper limb network, $W_{UL(NET)}$, was defined as the mean over 15 complete crank revolutions of the time-integral of the total upper limb power as:

$$W_{UL(NET)} = \int_0^{1 \text{ cycle}} [P_t(t)] dt = \int_0^{1 \text{ cycle}} \sum_{i=1}^6 [\mathbf{M}_i(t) \cdot \omega_i(t)] dt + \int_0^{1 \text{ cycle}} \sum_{i=1}^2 [\mathbf{F}_{sf}(t) \cdot \mathbf{V}_{sf}(t)] dt \quad (2)$$

where M_i and ω_i were the i^{th} joint moment and joint angular velocity, respectively. These calculations were performed in the sagittal plane to allow direct comparisons with the power produced at the crank. To compare the total work of the upper limbs to the total work done at the crank, $W_{UL(NET)}$ was divided by the time-integral of the crank power over a crank revolution (W_{crank}) to obtain $W_{UL(RELATIVE)}$.^{10,11,20}

$$W_{UL(RELATIVE)} = \frac{W_{UL(NET)}}{W_{crank}} \times 100. \quad (3)$$

The absolute work done by the upper limbs ($W_{UL(ABSOLUTE)}$) during 1 crank revolution was computed as the sum of the absolute value of the work done by the 2 arms and the shoulder transfers, and considers both eccentric and concentric work as positive^{19,22} as:

$$W_{UL(ABSOLUTE)} = \int_0^{1 \text{ cycle}} \sum_{i=1}^8 |P_i(t)| dt = \int_0^{1 \text{ cycle}} \sum_{i=1}^6 |\mathbf{M}_i(t) \cdot \omega_i(t)| dt + \int_0^{1 \text{ cycle}} \sum_{i=1}^2 |\mathbf{F}_{sf}(t) \cdot \mathbf{V}_{sf}(t)| dt \quad (4)$$

The entire data processing was performed using custom-made codes written in Scilab 5.4.0 (SCILAB, Scilab Enterprises, Versailles, France).

Data Reduction and Statistics

Results are reported as mean \pm standard deviation. Moments, powers minima and maxima, and works were extracted at each joint for each pedaling condition and these variables were divided by the participant's body mass. Two-way repeated measure ANOVAs (position = seated and standing \times crank power = 20%, 40%, 60%, 80%, 100%, and 120% of the sit-to-stand transition power) to test the effect of the position and of the crank power on the upper limb moment, power, and work magnitudes were performed after checking for data normality using Shapiro-Wilk tests and for variance homogeneity using Levene tests. Post hoc analyses were performed using Bonferroni's method. All statistical analyses were performed using the STATISTICA software (STATSOFT, Maisons-Alfort, France). An alpha value of .05 was defined as the level of statistical significance.

Results

The 6 power outputs corresponding from 20% to 120% of the sit-to-stand transition power were of 112 ± 19 W (1.6 ± 0.3 W·kg⁻¹) for 20%; 225 ± 38 W (3.2 ± 0.6 W·kg⁻¹) for 40%; 337 ± 56 W (4.7 ± 0.9 W·kg⁻¹) for 60%; 450 ± 75 W (6.3 ± 1.2 W·kg⁻¹) for 80%; 562 ± 94 W (7.9 ± 1.5 W·kg⁻¹) for 100%; and 675 ± 113 W (9.5 ± 1.8 W·kg⁻¹) for 120%. Because the participants could select their mediolateral hand position on the handlebar, which could have affected upper limb kinetics, their hand position across pedaling conditions has been checked. The pedaling condition had no effect on hand position (Table 1).

A significant effect of crank power was observed in 16 out of 18 moments analyzed in the seated position and in 14 out of 18

Table 1 Hand position across pedaling conditions

	20%	40%	60%	80%	100%	120%
Seated						
Absolute	0.254 ± 0.031	0.256 ± 0.028	0.252 ± 0.028	0.253 ± 0.024	0.250 ± 0.027	0.251 ± 0.028
Relative	0 ± 0	0.002 ± 0.017	-0.002 ± 0.016	-0.001 ± 0.018	-0.003 ± 0.016	-0.003 ± 0.014
Standing						
Absolute	0.259 ± 0.021	0.258 ± 0.024	0.264 ± 0.023	0.259 ± 0.021	0.257 ± 0.026	0.262 ± 0.023
Relative	0.005 ± 0.026	0.004 ± 0.022	0.010 ± 0.022	0.005 ± 0.024	0.003 ± 0.023	0.008 ± 0.022

Note. Mean ± *SD* position of the center of the left hand on the mediolateral axis. “Absolute” represented the horizontal distance in m between the hand center (defined as the midpoint between the wrist joint center and the midpoint between the 2 metacarpal heads) and the handlebar center. “Relative” represented the difference in m between the hand center in the seated 20% condition and the hand center in the assessed condition. No significant difference was observed between pedaling conditions.

in the standing position, showing that joint moments increased in magnitude with increasing crank power (Figure 2–4). A significant effect of position was observed, with most of the moments higher in magnitude in standing position in comparison with the seated position.

Normalized moments developed about the wrist, elbow, and shoulder joints across the pedaling cycle at each percentage of the sit-to-stand transition power in the frontal, sagittal, and transverse planes are presented in Figures 2, 3, and 4, respectively.

Maximum negative power in the seated position increased for all joints with increasing crank power (Figure 5), whereas maximum positive power increased only at the shoulder joint level. Both negative and positive power was higher in standing position compared with seated position. In standing position, crank power affected shoulder and elbow maximum negative powers, whereas only shoulder maximum positive power increased with increasing crank power.

Net joint work ($W_{UL(NET)}$) presented no significant effect of crank power in the seated position (Figure 6A). Conversely, $W_{UL(NET)}$ significantly increased in the standing position at 100% of the sit-to-stand transition power in comparison with 20%. $W_{UL(NET)}$ at 120% of the sit-to-stand transition power was significantly higher than all the other crank power conditions. $W_{UL(NET)}$ was significantly higher in the standing position compared with the seated position at 120% of the sit-to-stand transition power.

The total work done by the upper limbs expressed in proportion to the total work generated at the crank ($W_{UL(RELATIVE)}$) ranged from 0.5% to 2% in the seated position and from 1.5% to 2.5% in the standing position (Figure 6B). No significant effect of crank power on upper limbs’ total work was observed either in the seated or in the standing position.

The amount of work in absolute value generated by the upper limbs ($W_{UL(ABSOLUTE)}$) increased with crank power in both seated and standing positions, with $W_{UL(ABSOLUTE)}$ at 120% of the sit-to-stand transition power different from all other crank powers (Figure 6C). $W_{UL(ABSOLUTE)}$ was higher in the standing position than in the seated position at all crank powers.

Discussion

The aim of the current study was to assess, for the first time, upper limb joint loadings in cycling, and to test if these variables were affected by crank powers and pedaling positions. Results showed that increase in crank power led to an increase in magnitude of the upper limb moments in both the seated and standing positions. The crank power value determined in this study was higher than the one

of a previous study, reporting 419 ± 30 W as the power above which standing position should be chosen to maximize performance.²³ Because Hansen and Waldeland²³ compared times to exhaustion during cycling at different percentage of maximal oxygen uptakes, methodological differences in the criterion defining the optimal sit-to-stand transition power may explain this result.

The increase in upper limb joint moments with increasing power suggests new biomechanical constraints in these conditions. The common point between the variables affected by the crank power was that they were associated with handlebar pulling, whereas the variables not affected were mostly associated with handlebar pushing. Previous studies confirm this evolution of handlebar actions with increasing power.^{3–5} However, why handlebar pulling magnitudes increase with increasing crank power, but not handlebar pushing, is still unclear. A first hypothesis could be that cyclists activate their upper limb muscles to better activate their lower limb muscles.^{7,8} This could explain the increase in handlebar pulling, but not the stability in handlebar pushing. Another explanation could be found in the main goal of the cycling task: producing crank power. When this constraint increases, it is necessary for a given cadence to produce higher pedal force and thus more vertical force. At one point, these vertical pedal forces may counterbalance the body weight. This finding is consistent with the observation of a reduction of the saddle vertical force with increasing crank power.^{5,24} The observed increase in handlebar pulling may be used to avoid body elevation when the pedal vertical forces exceed the body weight. Therefore, the increase in upper limb joint moments may allow the performance by trunk stabilization. Given the fact that experts develop higher crank power outputs²⁵ and that they have been observed to push more and pull less on the pedal than less-trained cyclists,²⁶ pedaling in a seated position for this population may imply higher levels of upper limb loadings, because of the necessary higher handlebar pulling to remain seated. Studies assessing maximal upper limb moments in isolated joint testing reported higher magnitudes than the one observed in our study.^{27,28} This comparison suggests that, in cycling, upper limb joint moments may not reach the maximum magnitude possible. However, it cannot be ruled out that the upper limb muscles may face important demands in force given the specific context of pedaling and the likely nonoptimal position for each joint to produce moments, and because of the decrease in joint torque production associated with the increase in joint angular velocity.²⁹ In this sense, previous findings showed that myoelectrical activity of the forearm muscles is close to maximal during maximum power pedaling.⁶ To improve performance by creating high pedal forces with limited trunk movements, and to avoid the standing position to decrease aerodynamic drag,³⁰ one

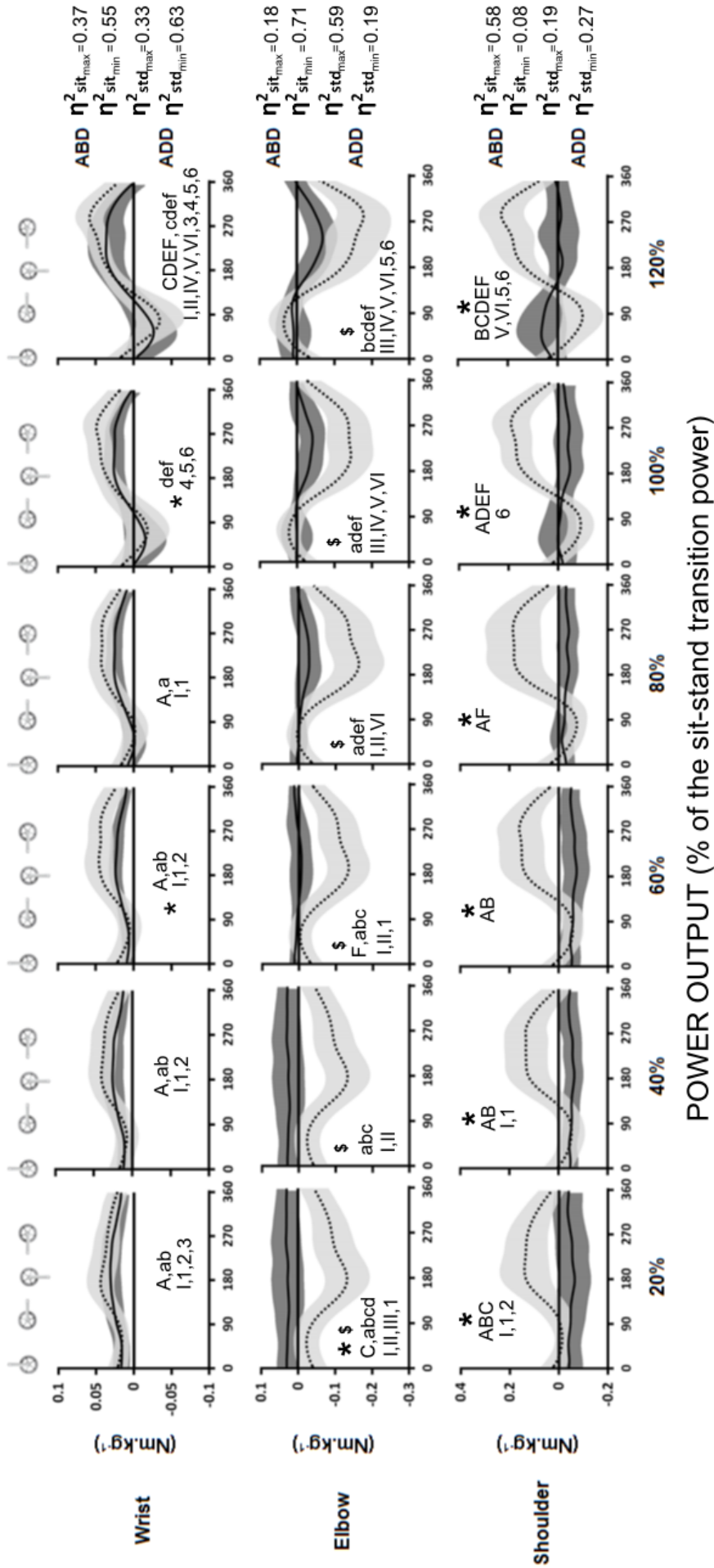


Figure 2 — Frontal plane moments in the joint coordinate systems for the left upper limb at the wrist, elbow, and shoulder in function of the right crank angle. Mean seated cycling (solid black line) and standing cycling (dotted black line) are represented \pm one standard deviation (dark gray shading for seated/light gray shading for standing). Data are normalized by body mass over 15 pedaling cycles at 90 revolutions per minute (rpm). ABD = abduction; ADD = adduction. * Significant difference between maximum seated and standing position. \$ Significant difference between minimum seated and standing position. ABCDEF indicate a significant difference compared with 120%, 100%, 80%, 60%, 40%, and 20% of the power output corresponding to the sit-to-stand transition for the maximum magnitude in the seated position; abcdef indicate a significant difference compared with the minimum magnitude in seated position; I,II,III,IV,V,VI indicate a significant difference compared with maximum magnitude in the standing position; and 1,2,3,4,5,6 indicate a significant difference compared with minimum magnitude in the standing position. η^2_{sit} = eta-squared in the seated position. η^2_{std} = eta-squared in the standing position.

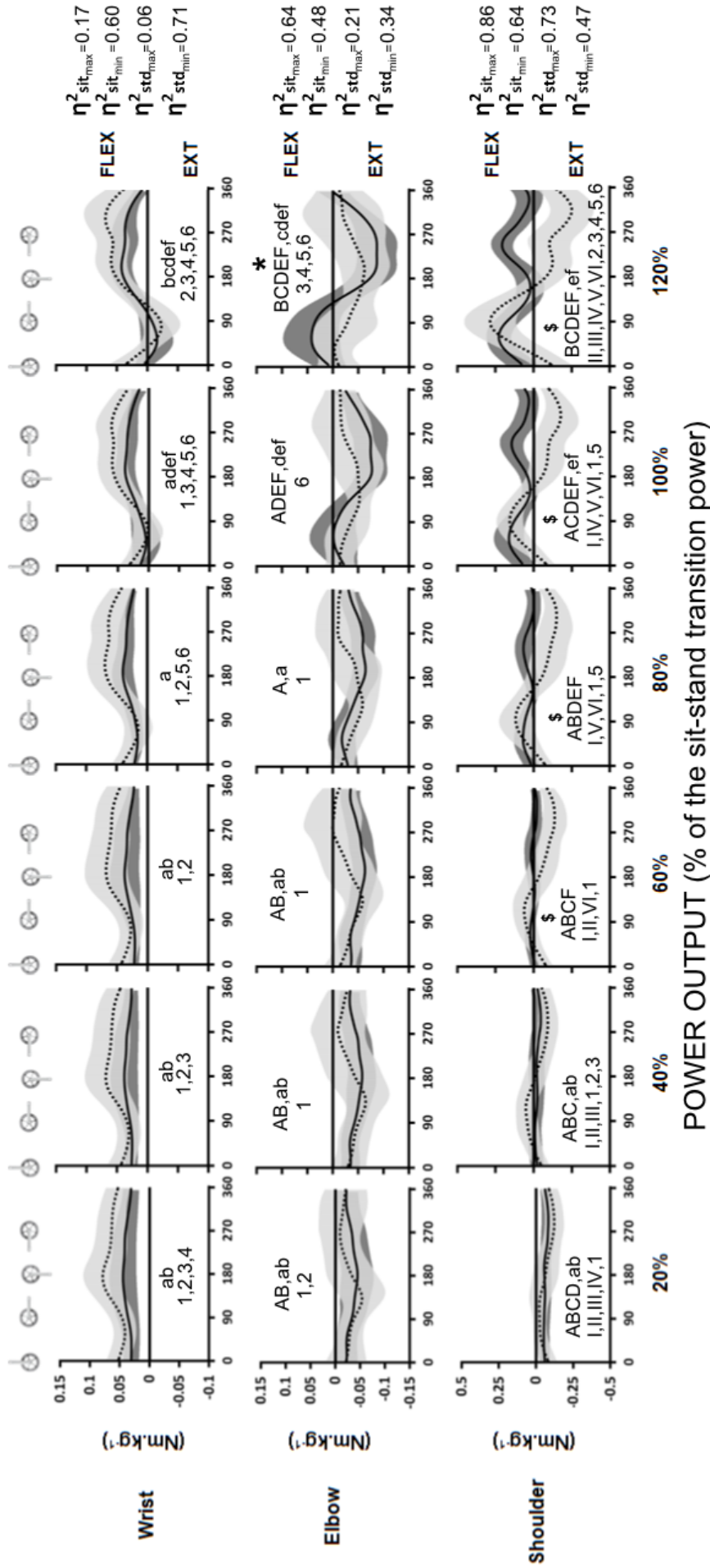


Figure 3 — Sagittal plane moments in the joint coordinate systems for the left upper limb at the wrist, elbow, and shoulder in function of the right crank angle. Mean seated cycling (solid black line) and standing cycling (dotted black line) are represented \pm one standard deviation (dark gray shading for seated/light gray shading for standing). Data normalized by body mass over 15 pedaling cycles at 90 revolutions per minute (rpm). FLEX = flexion; EXT = extension. * Significant difference between maximum seated and standing position. \$ Significant difference between minimum seated and standing position. ABCDEF indicate a significant difference compared with 120%, 100%, 80%, 60%, 40%, and 20% of the power output corresponding to the sit-to-stand transition for the maximum magnitude in the seated position; abcdef indicate a significant difference compared with the minimum magnitude in seated position; I,II,III,IV,V,VI indicate a significant difference compared with maximum magnitude in the standing position; and 1,2,3,4,5,6 indicate a significant difference compared with minimum magnitude in the standing position. η^2_{sit} = eta-squared in the seated position. η^2_{std} = eta-squared in the standing position.

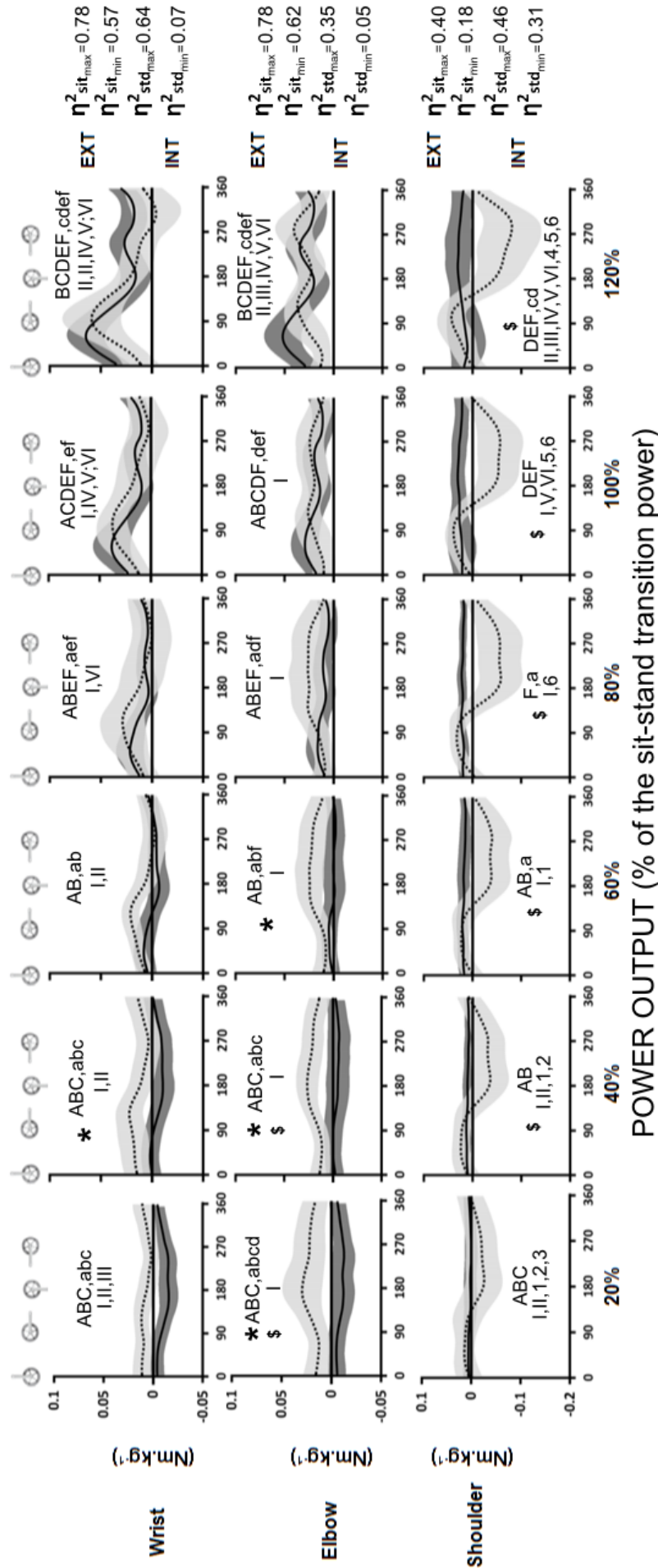


Figure 4 — Transverse plane moments in the joint coordinate systems for the left upper limb at the wrist, elbow, and shoulder in function of the right crank angle. Mean seated cycling (solid black line) and standing cycling (dotted black line) are represented \pm one standard deviation (dark gray shading for seated/light gray shading for standing). Data normalized by body mass over 15 pedaling cycles at 90 revolutions per minute (rpm). EXT = external rotation; INT = internal rotation. * Significant difference between maximum seated and standing position. \$ Significant difference between minimum seated and standing position. ABCDEF indicate a significant difference compared with 120%, 100%, 80%, 60%, 40%, and 20% of the power output corresponding to the sit-to-stand transition for the maximum magnitude in the seated position; abcdef indicate a significant difference compared with the minimum magnitude in the standing position. η^2_{sit} indicate a significant difference compared with maximum magnitude in the standing position; and 1,2,3,4,5,6 indicate a significant difference compared with minimum magnitude in the standing position. η^2_{sit} = eta-squared in the seated position. η^2_{std} = eta-squared in the standing position.

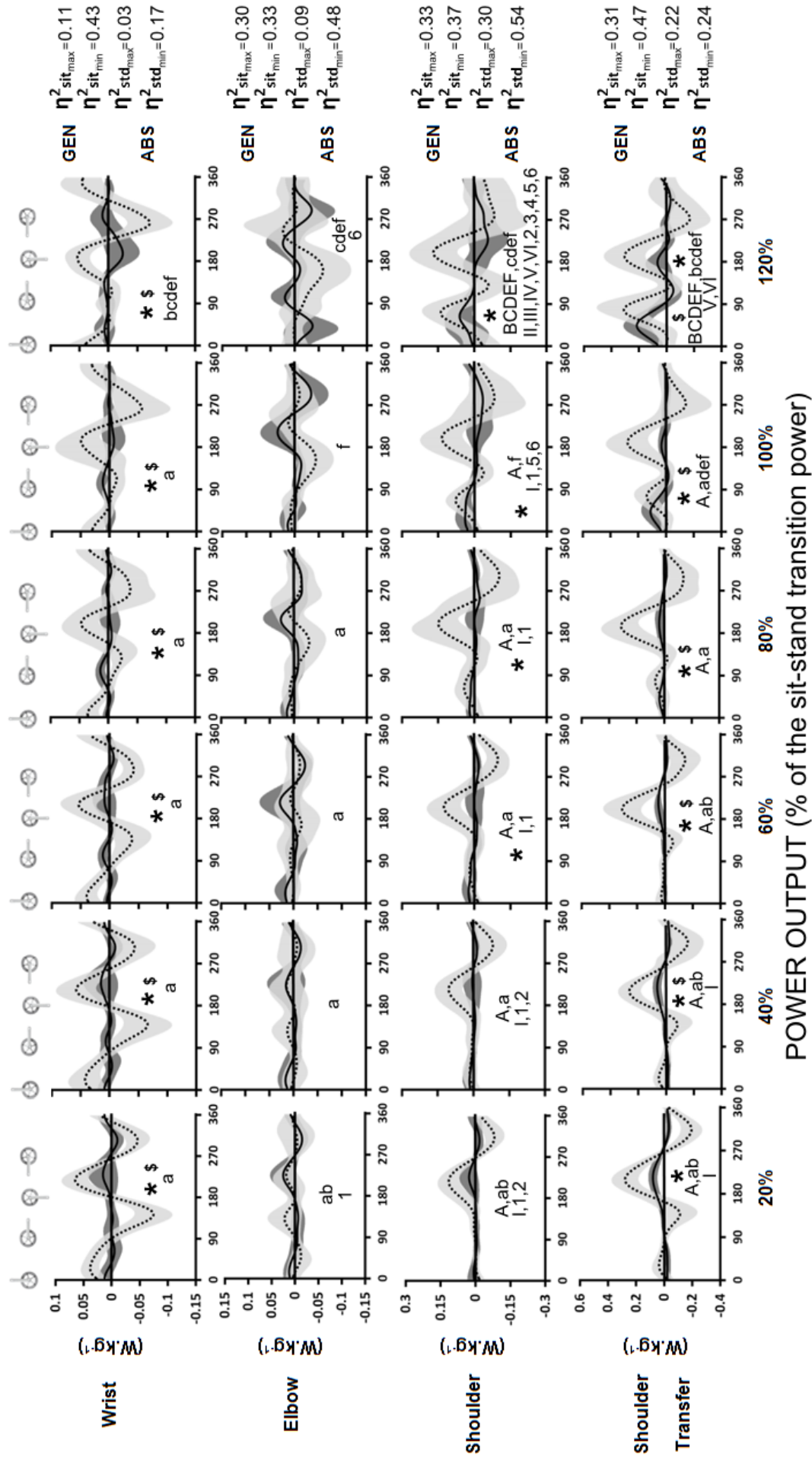


Figure 5 — Powers in the sagittal laboratory plane for the left upper limb at the wrist, elbow, and shoulder in function of the right crank angle. Mean seated cycling (solid black line) and standing cycling (dotted black line) are represented \pm one standard deviation (dark gray shading for seated/light gray shading for standing). Data normalized by body mass over 15 pedaling cycles at 90 revolutions per minute (rpm). GEN = power generation; ABS = power absorption. * Significant difference between maximum seated and standing position. \$ Significant difference between minimum seated and standing position. ABCDEF indicate a significant difference compared with 120%, 100%, 80%, 60%, 40%, and 20% of the power output corresponding to the sit-to-stand transition for the maximum magnitude in the seated position; abcdef indicate a significant difference compared with the minimum magnitude in the standing position; I,II,III,IV,V,VI indicate a significant difference compared with maximum magnitude in the standing position; and 1,2,3,4,5,6 indicate a significant difference compared with minimum magnitude in the seated position. η^2_{sit} = eta-squared in the seated position.

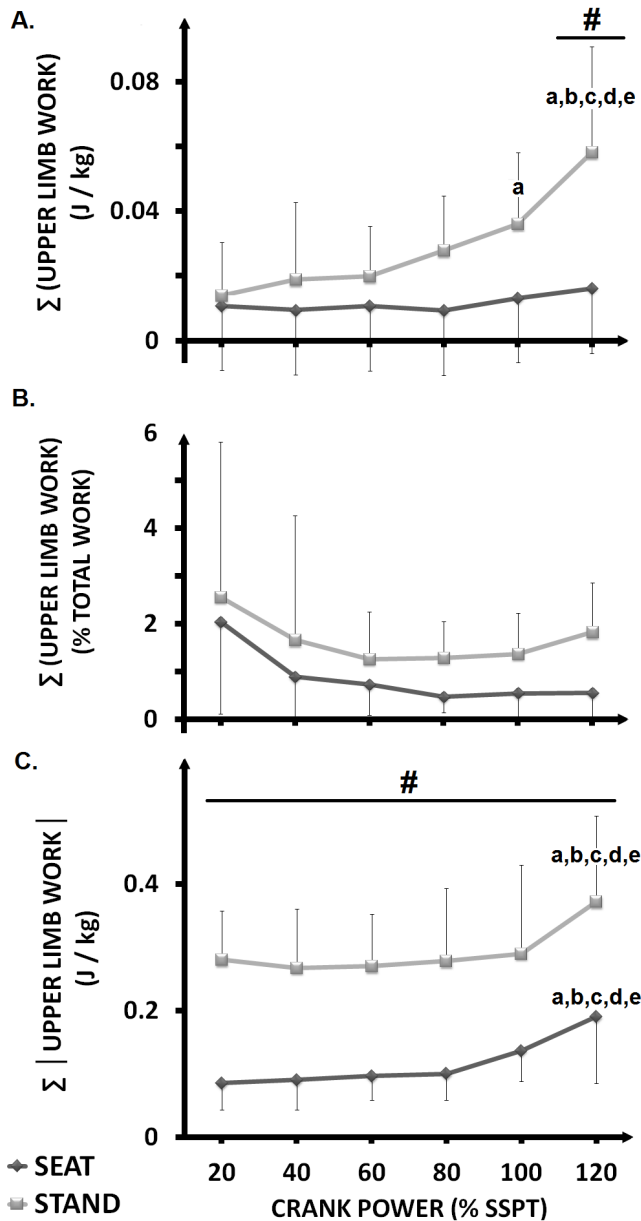


Figure 6 — Work done by the upper limbs. One dot represents the sum of the two wrists, elbows, shoulders, and shoulder transfer work for the given position (seated = dark gray squares; standing = gray squares) across crank power outputs. (A) $W_{UL(NET)}$. (B) $W_{UL(RELATIVE)}$. (C) $W_{UL(ABSOLUTE)}$. Power output from 20% to 120% of the sit-to-stand transition power correspond to 76 ± 13 J (1.1 ± 0.2 J·kg⁻¹), 151 ± 26 J (2.1 ± 0.4 J·kg⁻¹), 227 ± 39 J (3.2 ± 0.6 J·kg⁻¹), 302 ± 52 J (4.2 ± 0.8 J·kg⁻¹), 378 ± 65 J (5.3 ± 1 J·kg⁻¹), and 453 ± 78 J (6.3 ± 1.2 J·kg⁻¹) works generated at the cranks by pedaling cycle, respectively. # Significant difference compared with seated cycling ($P < .05$). a,b,c,d,e Significant difference compared with 20%, 40%, 60%, 80%, and 100% of the sit-to-stand transition power, respectively.

implication of these results may be the use of strength conditioning programs involving the upper limbs and focusing on the muscular actions associated with handlebar pulling.

The mechanical energy cost of the upper limbs was represented in our study by measurement of $W_{UL(ABSOLUTE)}$, which cumulates values of work associated with both concentric and eccentric contractions. The increased $W_{UL(ABSOLUTE)}$ could be associated with an

increase in metabolic cost with increasing crank power. This result is in line with one study which demonstrated a significant reduction in metabolic cost when the torso was stabilized to decrease upper limb actions in cycling.³¹ The study by McDaniel and colleagues suggested that upper limbs represented about 2% of the pedaling energy cost. However, their study was limited in the crank power they could impose on their subjects (maximum 250 W). From the increase in joint moments and powers observed in our results for power outputs > 250 W, it could be expected that the absolute value of the metabolic cost due to the upper limbs would be higher for higher power outputs. Previous studies showed increasing magnitudes of hip transfer power in the standing position⁹ and with increasing crank power from 3% of the pedal power at 250 W to 8% at about 1000 W.^{10,11} However, the relative upper limb work generation, $W_{UL(RELATIVE)}$, remained low in both positions and lower than the reported data on hip power transfer. That could mean that other parts of the body such as the back and/or the contralateral leg could participate in the power transmitted through the hip joint force, and that the upper limb direct contribution to the crank power remains low. Further investigations using inverse dynamics with instrumented pedals and seats and lower limb kinematic reconstruction are needed to confirm that the upper limbs could only be partially responsible for the so-called hip transfer power in the literature. These considerations related to moments and works created by the upper limbs suggest that future models should consider both upper and lower limbs when modeling cycling biomechanics and energy expenditures, especially when the focus is on high crank power, elite athletes, and/or on the standing position.

It is important to note that pedaling on a cycling ergometer and cycling in the field are 2 different things.³² Even if the former is a common practice for rehabilitation and training, it does not allow lateral bicycle oscillations, which are supposed to interfere with the pedaling technique. Thus, upper limb kinetics may presumably be modified in the field, particularly in the standing position in which roll angles up to 24° have been observed and may increase the upper limb moments specifically in the frontal plane.³³ Other factors such as slope or vibrations may have effects on the upper limb kinetics, in addition to other handlebar systems like road or time-trial, with other drop and reach settings. To test these effects, sensors recording full-body kinematics, and three-dimensional loads applied on each of the bicycle's supports are necessary to provide inverse dynamics analyses of cycling in field conditions. These methods often used to provide insights into the biomechanical causes of the observed movement are limited in accuracy by the combination of kinetic, kinematic, and anthropometric errors, however these sources of error should not change the conclusion about both position and crank power having an effect on upper limb kinetics. Further studies are necessary to fully understand which parameters are leading a cyclist to spontaneously transit from seated to the standing position.

In summary, this study measured three-dimensional moments, powers, and works done at the upper limb joints for various position and crank power conditions. We conclude that upper limb kinetics are affected by both crank power and the pedaling position. By providing a first reference of upper limb kinetic measurements, and by showing that these variables are affected by both the power output and the cycling position, these results have implications for clinicians, athletes, coaches, and sport scientists aiming to enhance performance and prevent injuries in cycling.

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