Motor Variances in Arm Cranking as a Function of the Resistance

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Declarations

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Ethics approval and consent to participate

The study was approved by the Ethics Committee of the National Institute for Medical Rehabilitation (Budapest, Hungary) in accordance with the Declaration of Helsinki. All participants were informed about the purpose of the study. Written informed consent was obtained from all participants, and they participated voluntarily in the study.

Consent for publication

Not applicable

Competing interests. The authors declare that they have no competing interests.

Availability of data and materials. The datasets used and analyzed during the current study are available from the corresponding author.

Authors' contributions. MM and JL contributed to design the protocol. MM and LB collected and analyzed the data. LB developed software and visualization. MM, DP and NZ provided formal analysis, statistics. MM, DP and JL provided conceptualization, interpreted the results and contributed to writing of the draft. JL and DP contributed to editing.

Abstract

Background: Arm cycling on an ergometer is common in sports training and rehabilitation protocols, but has not been widely studied from an aspect of neural control. The hand movement is constrained along a circular path, and the user is working against a resistance, maintaining a cadence. Even if the desired hand trajectory is given, there is the flexibility to choose patterns of joint coordination and muscle activation, given the kinematic redundancy of the upper limb. With changing external load, motor noise and changing joint stiffness may affect the pose of the arm even though the endpoint trajectory is unchanged, unless a control mechanism maintains the same arm configuration in corresponding time points of the cycles. However, the effect of crank resistance on the variances of arm configuration and muscle activation has not been investigated, yet. Methods: Fifteen healthy participants performed arm cranking on an arm-cycle ergometer both unimanually and bimanually with a cadence of 60 rpm against three crank resistances. We investigated arm configuration variances and muscle activation variances. Arm configuration was given by inter-segmental joint angles, while muscle activation by surface EMGs of arm muscles. Applying multifactorial ANOVA we evaluated the effects of resistance conditions. Results: Arm configuration variance in the course of arm cranking was not affected by crank resistance, while muscle activation variance was proportional to the square of electromyographic muscle activity. Furthermore, the shape of the variance time profiles for both arm configuration and muscle activation was not affected by crank resistance independently on cranking being performed unimanually or bimanually. Conclusions: Contrary to the prevailing assumption that an increased motor noise would affect the variance of auxiliary movements, the influence of noise doesn't appear at the arm configuration level even when the system is redundant. Our

results suggest that neural control stabilizes arm configurations against altered external force in arm cranking. This may reflect the separation of kinematic- and force-control, via mechanisms that are compensating for dynamic non-linearities. Arm cranking may be suitable when the aim is to perform training under different load conditions, preserving stable and secure control of joint movements and muscle activations.

Keywords: load; arm configuration; muscle activity variance; kinematic control, force control

Introduction

Arm cycling on arm ergometers is often applied in sports training when the aim is to strengthen upper body muscles in neurologically intact people [1] or to assess muscle powers and evaluate performances [2] during sports activities. Arm cycling exercises are also included in medical rehabilitation protocols [3] to improve motor performance and motor control of individuals with spinal cord injury or stroke [4, 5]. Arm cycling exercises are also used in combination with functional electrical stimulation (FES) training of individuals with spinal cord injuries [6]. Despite a range of sport and rehabilitation applications [7], the literature on arm cycling movements is limited relative to that on lower limb cycling. However, the importance of arm cycling has recently been supported by several investigations. Arm cycling in males and females has been compared [8], and sex-related differences in peak and mean power were more pronounced in arm cycling than in leg cycling [2]. Other noteworthy studies include the physiological characteristics of eccentric arm cycling [9]; the influence of differences in arm cycling at various cadences on the modulation of supraspinal and spinal excitability, and the

influence of various arm cycling parameters (e.g., crank load) on interlimb reflex modulation (soleus H reflex) [10]. No significant differences were seen in the level of suppression of the H reflex at different crank loads. It was proposed, and supported by data, that neural coupling between the arms helps to increase movement symmetry and to ensure stable arm cycling [11]. It has been shown that arm cycling training improves strength, coordination of muscle activity during other types of motor tasks, such as walking, and neurological connectivity between the arms and the legs [12].

We investigate arm cranking from the aspect of motor variance. We asked the following question: "How are the variances of arm cycling (cranking) movements affected if the resistance of the crank is changed?" There is literature in robotics about control of manipulators where the movement of the end-effector is constrained, but the load on it is changed [13, 14]. This literature offers models for accomplishing the task. Our particular purpose was to analyze the physiological parameters of human subjects during a constrained motion (arm cranking) when an increasing load is applied (effect of crank resistance). The metrics we analyzed are the arm configuration variance (in joint space) and the muscle activation variance (in muscle space). These parameters can indirectly validate the type of control utilized for this complex task. The maintenance of the same arm configuration is not guaranteed as the resistance of the crank increases. It has been demonstrated that joint stiffness increases as the load at the end-effector increases [15]. Changes in joint stiffness at each joint can substantially change the pose of the arm even though the endpoint trajectory is unchanged. If the arm configuration variance is not affected by crank resistance, it ensures the separation of kinematic- and force-control [16, 17] where the kinematic task can be maintained safely when crank resistance is altered. Knowing the type of control strategy is important in training and rehabilitation protocols. It is not the aim of this paper to evaluate rehabilitation protocols, but to experimentally examine this potentially useful feature of arm

cranking movement and provide a validation of the already developed theories applied in robotic rehabilitation [18-20].

The research on motor variance of multi-joint and multi-muscle systems covers several motor tasks but regarding arm movements, most of them are unconstrained, reaching or pointing movements [21-23]. It has been reported that for object transporting arm movements, the joint configuration variance depends on the weight of the object held in the hand [24]. It is a remaining question and it is investigated to a smaller extent that how motor variances in joint space and muscle space are affected by external loads, in the case of constrained arm movements, when the end effector (hand) path in the workspace is constrained [25]. Here we extend these studies. When arm cycling on an ergometer, the hand path is constrained, and the variance in the endpoint trajectory is assumed to be very small if the hand moves on a fixed circle with constant angular velocity.

However, if the external load does not have an effect on the endpoint trajectory, it still may have an effect on the arm configuration variance. During arm cycling, when the endpoint trajectory is fixed, there is still the possibility for an infinity of change of the arm configuration that would result in the same endpoint trajectory. It is not trivial that the effect of the load does not appear at joint rotations. The underlining notion of motor variances famously reported by Bernstein [26], who observed that when the blacksmith wields the hammer, the hammer's trajectory is more reproducible than the arm configurations used to perform that movement. In our case, the endpoint variance is very small by definition, but it does not imply automatically that the joint configuration is reproducible during consecutive cycles.

In the present study, we investigate unimanual and bimanual arm cycling, focusing on motor variance at the joint and muscle levels.

Cycling was performed on an ergometer against three crank resistances. We hypothesized that variances in joint angular displacement are impervious to crank resistance but that variances in muscle activities (EMG) increases quadratically as the crank resistance increases. This behavior would underline that 1) there exists a mechanism for the concurrent control of motion and force where the two can be controlled separately, 2) the controller of the motion is linear, 3) there exists a predictive mechanism capable of compensating the dynamic non-linearities.

Methods

Participants

Fifteen right-handed, able-bodied participants (24 ± 4 years old) were recruited in the study who performed arm cranking movements on an arm cycle ergometer. The study was approved by the Ethics Committee of the National Institute for Medical Rehabilitation, Budapest, Hungary. Written informed consent was obtained from all participants, and they participated voluntarily in the study.

Experimental setup

Each participant was seated in a fixed chair in front of an arm cycle ergometer (MEYRA, Kalletal, Germany, (Fig. 1). The participant grasped the handle of the ergometer at the end of the crank, which was 10 cm long. The distance between the chair and the ergometer was set in such a manner that when the handle of the ergometer was at the most distant position with respect to the participant, the external angle of the elbow (the angle of the forearm with respect to the elongation of the long axis of the upper arm) was approximately 10-15 degrees. This corresponded approximately to the most extended elbow position. This angle was measured with

a protractor. The shoulders were strapped (with a chest strap) to the back of the chair to restrict the movement of the trunk. Note that because of the difference in participant size this configuration does not guarantee that each participant moves with the same angular displacement. For that case, the dimension of the crank would have need to change from participant to participant. Nevertheless, the subject dependent variation has been taken into account within our statistical analysis.

Insert Figure 1 near here.

Fig. 1 *Equipment and maker positions. A: arm cycle ergometer. B: schematic figure of the cycling participant. Black dots illustrate positions of markers placed on the body, on the handlebar of the ergometer and on the chair on which the participant was seated. Joint angles* α, β, γ *in the shoulder, elbow and wrist, respectively, were computed from marker coordinates.*

Ultrasound emitting markers as part of an ultrasonic movement analyzer system (ZEBRIS CMS HS, Isny, Germany) were placed on anatomical landmarks. In particular, we used markers of the following landmarks: acromion, distal end of the humerus, proximal end of the ulna, styloid process of the ulna, caput of metacarpal of the fifth digit. One marker was placed on the chair and one on the handlebar of the ergometer. The positions of the markers were recorded by three ultrasound-sensitive microphones, with a sampling frequency of 100 Hz.

The surface EMG activity was recorded by the EMG recording apparatus of the ZEBRIS system, from the right and left biceps (BI), triceps (TR), deltoideus anterior (DA), and deltoideus

posterior (DP) muscles, with a sampling frequency of 900 Hz. The skin was dry shaven and cleaned with 70% alcohol before placing pairs of NORAXON (Type 272) electrodes (interelectrode distance was 1.5 cm). The positions of the electrodes were based on the recommendations of the SENIAM project, "Recommendations for sensor locations on individual muscles" [27]. A reference electrode was placed at the elbow (over the olecranon).

Motor task

The participant was instructed to cycle with a cadence of 60 revolutions per minute (rpm), against three different crank resistances: low (1), moderate (2) and high (3). Cycling was performed bimanually and unimanually with the left or right arm. The resistance was quantified as the torque with which the crank resists rotation. In unimanual cycling, the low, moderate and high resistances were 1.16 Nm, 2.08 Nm, and 3.09 Nm, respectively. In bimanual cycling, they were 1.16 Nm, 3.09 Nm, and 6.14 Nm, respectively.

Cycling was performed by each participant unimanually (by the left and right arm) and bimanually under each of the three resistance conditions and using two different grasping forms (horizontal or vertical). In the horizontal grasping form, the palm was positioned horizontally, and the fingers were bent around the horizontal handle. In this paper we deal only with the horizontal grasping form. The order in which cycling conditions were chosen was random. In each condition, the participants cycled for 30 seconds. They had 1 minute of rest between conditions. A metronome was used to guide the participants keep a cadence of 60 rpm in each cycling condition.

Data processing and analysis

Recorded EMG signals were filtered using custom software in MATLAB (Mathworks, Natick, MA). Frequencies below 25 Hz and above 300 Hz were cut off (4th order Butterworth bandpass filter), as were frequencies from 49-51 Hz to eliminate the effect of the electrical power source (i.e. 50 Hz in Europe). After filtering, a root mean square (RMS) algorithm was applied to smooth filtered signals with a moving window of 0.088 ms (80 samples).

Recorded marker coordinates were filtered applying discrete cosine transformation (DCT) to eliminate artifacts (Shin et al. 2010). Here, DCT was used to transform the recorded kinematic signals from the time domain to the frequency domain. Then, we multiplied the results with a 3rd-order low-pass Butterworth gain function (cutoff frequency 10 Hz). Finally, inverse DCT was applied. The intersegmental angles at the shoulder, elbow and wrist were computed from filtered marker coordinates by trigonometric equations. Fig. 1 illustrates the joint angles: shoulder – α, elbow – β, wrist – γ. Arm configuration was defined by the inter-segmental angles $(\alpha,\ \beta,\ \gamma)$ thus resulting in a 3-dimensional joint space representation. Arm cranking is often represented as a planar movement in the sagittal plane, where the crank angular velocity is defined as a vector orthogonal to such a plane. We are aware that the movement is not completely planar, in the sense that there is a small ab-adduction angle at the shoulder and thus the elbow may deviate from the sagittal plane. However, the direction of the angular velocity of this rotation passes through the instantaneous center of rotation of the shoulder and the point of contact of the hand and crank. As described in publications illustrating the Uncontrolled Manifold [28], the variance of this degree of freedom does not influence the main task since the angular velocity vector of the crank and the angular velocity vector of the ab-adduction angle, are always orthogonal. We define the osculating plane ”Os(t)” as the plane orthogonal to the angular velocity around the elbow. Within this plane, we consider 3 degrees of freedom:

elevation of the shoulder ($\alpha(t)$), flexion-extension ($\beta(t)$) of the elbow, and flexion-extension of the wrist ($\gamma(t)$).

Variance calculations

Time courses of joint angles $\alpha(t), \beta(t), \gamma(t)$, and muscle activities (EMG amplitude) were segmented based on the number of cycles the subjects completed. Time normalization was applied to allow comparison of cycles. The time progression within each cycle was divided into 100 equally spaced time bins, and joint angles and EMG amplitudes were approximated with cubic spline interpolation at the beginning of the bins. The crank angle was defined as 0 when the crank was directed horizontally towards the participant (the handlebar was the closest to the participant). A complete cycle was defined by the crank angle, with each cycle starting at a crank angle of 0 and ending at a crank angle of 360, and this cycle was mapped to a time scale (1 to 100).

Then, angular variances (joint configuration variances) and muscle activity variances across cycles were computed at each percentage of cycle time.

Angular variance (joint configuration variance) per degrees of freedom:

$$V_{ang}(t) = \frac{\sum_{k=1}^{N} |\bar{a}(t) - a_k(t)|^2}{N * 3}$$

where $a(t) = [\alpha(t), \beta(t), \gamma(t)]$, $|\,.\,|$ denotes the vector norm (magnitude of the vector), and $t = 1, \ldots, 100$ (percentage of cycle time). The upper line denotes the mean across cycles, N is the number of cycles, and k is the serial number of a cycle.

Muscle activity variance per degrees of freedom:

$$V_{EMG}(t) = \frac{\sum_{k=1}^{N} |\bar{M}(t) - M_k(t)|^2}{N * 4}$$

where $M(t) = [\mathrm{BI(t)}, \mathrm{TR(t)}, \mathrm{DA(t)}, \mathrm{DP(t)}]$ and $t = 1, \ldots, 100$ (percentage of cycle time).

Variances were averaged across normalized time for each participant separately to characterize the variance by one number in each cycling condition for each participant.

Statistical methods

We calculated a multiple ways mixed factor analysis of variance (ANOVA) for both the variance norm of angles and Surface Electromyographic signals. In the analysis we considered 4 factors such as side=['left','right'], mode = ['Double','Single'], loading resistance = ['Level 1', 'Level 2', 'Level 3'], and the subject which should be considered as a random factor and therefore makes this a mixed model. We also included in the model both a pairwise and a three-way interaction between the factors. While we observed a large standard deviation for the whole population of angle variances, such a variable is the linear sum of the standard deviation due to each factor. Thus, by doing a multifactorial ANOVA we can pinpoint the size of the standard deviation for each factor and see which one makes us reject the null hypothesis. A post-hoc multicompare analysis based on Tukey's honestly significant difference criterion was also performed.

Results

Neither the factors nor their interaction with each other creates a significant difference for the arm configuration variance. On the other hand, we can observe a significant effect of both the

loading resistance (F=28.02, p<0.0001) and the mode (F=20.11, p=0.0005) for the variance of the EMG. Furthermore, there is a significant interaction between the subject and side (F=4.52, p=0.0282) and subject and mode, indicating that subjects perform the task with a statistically significant difference between the two sides, and between double-hand and single-hand cycling when compared to each other. This suggests that the subject is a confounding factor and must be considered as a random factor.

Kinematic variances

Crank resistance (CR) did not have a significant effect on angular variances (F=1.43, p=0.2573). Furthermore, the interaction between load and cycling mode was also not significant (F=0.28, p=0.7574) (Fig. 2 A_1 and A_2). Side (F=0.15, p=0.7062) and cycling mode (F=0.5, p=0.4894) did not have a significant effect on angular variances (Fig. 2B and 2C).

Insert Figure 2 here

Fig. 2 *Mean angular variances A_1) in low, moderate and high resistance conditions for bimanual cycling (mean across participants and sides); A_2) in low, moderate and high resistance conditions for unimanual cycling (mean across participants and sides). B) in bimanual and unimanual arm cycling (mean across participants, resistances and sides); C) in left and right arms (mean across partcipants, resistances and modes). Lines above bars denote standard errors of the mean (SEM).*

Angular variances (as functions of normalized time) were compared for low, moderate and high CRs (Fig. 3).

Insert Figure 3 here

Fig. 3 *Angular variance profiles. Time course of angular variance* $(V_{ang}(t))$ *in low, moderate and high crank resistances in bimanual cycling and unimanual cycling for the dominant (right) and non-dominant (left) arm. Continuous line: mean across particpants. Dotted line: Mean+SEM.*

Fig. 2 and Fig. 3 show that the magnitude of angular variances and the time profiles of angular variances were not significantly affected by crank resistance. This was observed in both bimanual and unimanual cycling.

Muscle activity variances

Higher crank resistance was associated with higher muscle activity variances (Fig. 4 A_1 and A_2) in all examined cycling conditions for both arms. This difference was significant when low and high crank resistance conditions were compared in either bimanual ($p<0.0001$) or unimanual cycling ($p<0.0001$) according to a post-hoc multicompare analysis based on Tukey's honestly significant difference criterion. This was also true when moderate and high RCs were compared in either bimanual ($p<0.00025$) or unimanual cycling ($p<0.0001$). Comparing bimanual and unimanual cranking, the muscle activity variance was higher for

unimanual than for bimnual cranking (Fig. 4 B). Comparision of muscle activity variances when cranking by the left and right arm did not show significant difference (Fig. 4 C).

Figure 4 here

Fig. 4 *Mean muscle activity variances A_1) in low, moderate, and high resistance conditions for bimanual cycling (across participants and sides); A_2) in low, moderate, and high resistance conditions for unimanual cycling (across participants and sides); B) in bimanual and unimanual arm cycling (mean across participants, resistances, and sides F=20.11, p=0.0005); C) in left and right arms (mean across participants, resistances and modes F=0.15, p=0.7062);*
Lines above bars denote standard errors of the mean.

In addition to comparing average muscle activity variances, muscle activity variance profiles were also compared among various cycling conditions. It was found that the shape of the variance profiles did not change for the specific arm, only its magnitude changed according to crank resistance. This finding is presented in Fig. 5.

Insert Figure 5 here

Fig. 5 *Muscle activity variance profiles. Time course of muscle activity variance* $(V_{EMG}(t))$ *in low, moderate and high crank resistances in bimanual cycling and unimanual cycling for the dominant (right) and non-dominant (left) arm. Continuous line: mean across participants. Dotted line: Mean+SEM.*

To represent quantitatively the similarity of the variance profiles, we computed correlation coefficients of the variance curves obtained in different resistance conditions. High correlation were observed when comparing variance curves, presented at Fig. 5, for different cranking conditions in unimanual cranking for both arms and bimanual cranking for the right arm. A weaker linear correlation was found between variance curves observed in bimanual cranking for the left arm (Table 1).

Table 1. Correlation coefficients of mean muscle activity variance time courses.

	Bimanual		Unimanual	
	Left arm	Right arm	Left arm	Right arm
low and moderate	0.51	0.77	0.95	0.96
low and high	0.35	0.77	0.89	0.93
moderate and high	0.90	0.92	0.98	0.93

Mean muscle activity variance time courses ($V_{EMG}(t)$) were correlated based on different resistance conditions (low, moderate, high).

Correlation coefficients between 0.40 and 0.59 were defined as 'moderate positive correlation', between 0.60 and 0.79 were defined as 'strong positive correlation', and between 0.80 and 1.00 were defined as 'very strong positive correlation'.

Naturally, if muscles are working against higher external resistance, the EMG amplitudes increase. On the other hand, the profile of the muscle activites does not necessarily need to remain the same, but we can reveal that it does within the same arm/condition. If the amplitude increases in such a manner that the signal with lower values is simply multiplied by a constant

365 c>1 then the variance will be multiplied by c^2. The result is not trivial because for this to happen,
366 the control variable needs to be linear [13], and the system to be controlled is highly nonlinear.
367 Indeed, the force of each muscle (and the activation signal that mediates it) is required to
368 accomplish 3 distinct tasks. These tasks are 1) providing the operational command for the hand
369 to follow the prescribed trajectory, 2) compensating non-inertial forces such as centrifugal and
370 Coriolis forces that are generated by the nonlinear dynamics as a result of the movement and, 3)
371 generating additional forces for matching the resistance. Thus, for the variance to change
372 quadratically between load conditions the controller must be able to decouple these components
373 to guarantee that the operational task remains the same and that the resistance force is matched.
374 We investigated how EMG amplitudes and muscle activity variances increased when crank
375 resistance increased. We found that the variances changed almost quadratically with respect to
376 the change in average muscle activities (EMG values). Fig. 6. presents that the average variance
377 of muscle activities (EMG signals) increases approximately at the same rate as the mean squared
378 EMG values when the crank resistance is increased.
379 For each participant, the average EMG values across time was computed for moderate and low
380 crank resistances separately. The average obtained for moderate resistance was divided by the
381 average that was obtained for low resistance. Thus, we get one ratio for each participant. The
382 squares of this ratios were averaged across partcipants and this average values are presented at
383 Fig 6 for different conditions (bimnaul/unimanual, left/arm right arm) separately. The same
384 method was used for the computation of square of ratios of EMG values obtained for high
385 resistance with respect to EMG values obtained for low resistance. The ratios of variances of
386 EMG magnitudes in moderate CR to variances of EMG magnitudes in low CR and the ratios of
387 variances of EMG magnitudes in high respect to low CR were also computed and presented. We

compared the ratio of variances and the square of ratio of muscle activities applying paired sample t-test, (p=0.05). There were no significant differences in any cycling conditions (Fig. 6.)

Insert Figure 6 here

Fig. 6 *The ratios of variances of EMG magnitudes compared to the squares of the ratios of EMG magnitudes. A) The ratios of variances of EMG magnitudes in moderate crank resistance respect to variances of EMG magnitudes in low crank resistance and the squares of the ratios of EMG magnitudes; B) The ratios of variances of EMG magnitudes in high CR respect to variances of EMG magnitudes in low CR and the squares of the ratios of EMG magnitudes. Ratios are presented for different combinations of conditions separately (left arm bimnaul, left arm unimanual, right arm bimnual, right arm unimanual). Mean and standard errors across participants are presented. The average variance of muscle activities (EMG) increases approximately at the same rate as the mean squared EMG values when the crank resistance is increased.*

Discussion

Arm cranking differs from the extensively studied reaching types of movements. It is a cyclic movement and it is a constrained movement. Considering cyclic arm movements, such as circle drawing, variances have been studied with regard to the endpoint trajectory and arm configuration [29-33]. When arm cranking on an ergometer, not only the fixed hand path is given that has to be tracked by the endpoint of a multijoint system, but there is also the need to produce an additional force. Equations for closed chain mechanisms, which show that both torque and angular

position must be controlled is presented in the appendix (based on the approach given in Yagiela et al. 2020). These formulas show that the variance of the angles must change unless another control mechanism is taking place.

Even if the desired hand path is given or it is fixed, there is still flexibility to choose patterns of joint coordination and muscle activation. However, our results suggest that neural control maintain the same arm configuration against altered external force in arm cranking.

Kinematic variances

In the cycling movement investigated here, each hand moved along a given path with a given velocity independently from the CR. In particular, the hand moved on a 2 dimensional path (circle). The variance in hand position was not affected by crank resistance by definition. It was unknown, however, whether arm configuration variance would be affected by crank resistance. The considered system is in fact redundant because the intersegmental angles in the shoulder, elbow and wrist are changing during the movement. The range of angular motion of the shoulder elbow and wrist was 42.75°±0.63; 68.87° ±0.49 and 23.35° ±1.16 (mean ±SEM) respectively. As the system is redundant there exist infinite mapping from joint space to operational space to accomplish the task. Our results found that variances in angular changes in joints space are not affected by crank resistance. This was found for both arms. This suggests that during arm cycling, central control ensures stable movement execution at kinematic level even if crank resistance is altered. The kinematic requirements of the task do not vary for altered crank resistance, what changes is the additional effort necessary to execute the movement. This aspect suggests that when a mapping is chosen between operational space and joint space, it is maintained as the resistance at the crank increases. Furthermore, it provides evidence for the existence of an independent control of force and position [16]. The central nervous system

435 (CNS) is able to handle two tasks separately. On one hand, it guarantees that the kinematic
436 trajectory and velocity is executed. On the other hand, it is able to regulate the force the hand
437 needs to apply without changing the kinematics, even though the kinematics and force generation
438 are highly coupled through the non-linear dynamics of the neuro-mechanical system. The CNS
439 thus parses muscular force for specific tasks, separately controlling the force necessary for the
440 kinematics and the additional force required for the increasing crank resistance.

441 Studies on bimanual circle drawing tasks found that movements of the non-dominant arm
442 was more variable than the movement of the dominant arm [34]. We did not find variance related
443 differences between the arms in our experiments on constrained arm cranking movements. This
444 may be explained by the fact that the hand path was fixed and the execution of the task did not
445 require high dexterity. Future work will require to study variability of movements of the two
446 arms in other constrained motor tasks and the relation of such variabilities to the dynamic-
447 dominance hypothesis that was developed and applied for targeted reaching movements [35].

448

449 *Muscle activity variances*

450 Cycling against a higher crank resistance requires increased muscle activity. It is a
451 general assumption that activation signals with higher amplitudes produce higher motor
452 variances due to signal-dependent noise. In the present study, we found that measured EMG
453 signals have higher variances in higher CRs. It is unknown whether larger variances, observed
454 when cranking was performed against higher CR, are a consequence only of higher signal
455 amplitudes or if other motor control factors also contribute. The magnitude of muscle activity
456 variances was significantly affected by crank resistance. However, the shape of the variance
457 curve did not depend on crank resistance (Fig. 5, Table 1).

Muscle activity variances during arm cranking increased with crank resistance. On the other hand, the resulting kinematic (angular) variances were unchanged. Our results support the idea that arm configuration variances while cycling on an ergometer are not affected by crank resistance and that during this motor task, neural control stabilizes arm configurations against altered external force. This conclusion held true for both arms. This suggests that the CNS is able to modulate separately a kinematic task and a force task. Impedance control has been proposed as a strategy for the execution of such combined tasks [16].

A proportional variation of the muscle activity profiles as the CR increases (and therefore a quadratic alteration of its variance) is possible only if the control system is linear. Given the nonlinearities of the dynamic system, such control can only occur if there is a prediction of the dynamic properties of the system and the CNS is capable of compensating the dynamic non-linearity.

Limitations

Considering the significant standard deviation of the variance in our measurements of the joints' angles, we can analyze different sources for this phenomenon. Errors could come from the instrumental setup; on the other hand, we have placed particular care on these aspects. Specifically, we have used a system that is able to measure the position of the limbs without direct contact and with submillimeter precision. Thus, we have avoided errors that can come from using systems like an encoder, where plays in the kinematic chain between hand and transducer via a transmission can affect the measurements. In our setup, measurements strictly depend on what the subject has performed and not from additional errors in the measurement chain. The precision of the ultrasound system we utilized is actually very high. Considering an

average length segment all about 300 millimeters with an error of identification at the tip of each segment equal to 1 millimeter, the average angular error due to the measurement system is about 0.2 degrees. It can be seen that the standard deviation of the angular variance is much larger than that. Therefore, we can see that the variance of the joint angular displacement does not depend on the measurement errors but is strictly depending on the task. Considering the Uncontrolled Manifold Theory, we can speculate that there are infinite poses that can guarantee proper tracking of the handle along the circular trajectory. Thus, the subject is free to choose among every possible solution without compromising the kinematics of the endpoint.

Useful insights for rehabilitation

An aspect for rehabilitation practice that the present study provides is to help to plan proper upper body exercises for people with paraplegia, whose lower limbs are paralyzed. It is essential to prevent further health problems, which would be the consequence of a physically inactive lifestyle of people with paraplegia. Arm-cycling on arm-cycle ergometer offers them an excellent exercise which helps to enhance physical capacity and maintain stable movement execution when employing increased crank resistances during the series of training sessions. As the arm configuration variance is not affected by crank resistance, this motor task may involve a stable movement execution and may be well used in rehabilitation and training protocols. Another example of a potential application is functional electrical stimulation (FES) driven arm cycling for people with tetraplegia, who are unable to move the arm crank voluntarily (Zhou et al. 2018). When spinal cord injured individuals are not able to generate active muscle forces voluntarily, FES controlled arm cycling is a useful exercise. The aforementioned practice helps to strengthen muscles by increasing crank resistance during the series of training sessions. If

504 muscle stimulation patterns are defined by observed muscle activity patterns of able-bodied
505 individuals, then when resistance is increased during FES driven cranking, the amplitude of the
506 stimulation has to be increased, and the stability of the control can be conserved. This may make
507 the FES control easily adaptable to increased crank resistance. In spite of the limitation that in
508 this study we investigated unimpaired participants, we feel that the results provide a starting
509 point and further studies may evaluate related training protocols for motor impaired individuals.
510

511 **Conclusions**

512 In summary, we investigated arm cranking movements performed by able-bodied
513 individuals on a cycle ergometer and addressed the question of how external load (crank
514 resistance) affects the variances of arm configuration and muscle activation. The arm
515 configuration variance was not affected by the crank resistance either in unimanual or bimanual
516 cranking. This aspect was surprising because even though the hand path and cadence were
517 constrained, a variability could be expected given that an increased resistance is associated to an
518 increased motor noise that could have affected the time profile of the arm configuration variance.
519 Muscle activation variances increased quadratically with respect to the change in average muscle
520 activities as the crank resistance increased, underlining a linear control system. This observed
521 kinematic, and muscle activity variances may reflect the separation of kinematic- and force-
522 control. While a single controller based on the equilibrium point hypothesis was proposed in
523 [20], more recent literature put forth the need for two separate controllers to compensate for
524 dynamical forces [18]. Our investigation suggests that the control scheme appears to allow a
525 stable control of the constrained movement while independently compensating for the additional
526 load and the effect of non-linear dynamics. Our experimental results are consistent with an

operational space control scheme that decouples the kinematics and the dynamics [13]. As suggested in [16], the modulation of force could be accomplished by proper modulation of stiffness, which would not change the pose of the arm as a function of the load (thus maintaining the arm configuration variance unaltered), but simply compensate for the additional crank resistance. Besides the importance of the relation of kinematic and force control in an arm movement task in which the hand path is constrained, these results may be relevant for planning rehabilitative training procedures. The results suggest that arm cranking can be performed in a comfortable, stable manner when external load alters.

Abbreviations

CR- Crank resistance, BI-biceps, TR-ticeps, DA- deltoidus anterior, DP deltoidus posterior

Acknowledgments

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Appendix

Equations for closed chain mechanisms, which shows that both torque and position must be controlled (based on the approach given in Yagiela et al. 2020):

Equations for closed multi-link chain mechanisms show that the variance of the angles must change (through a change in the transmission ratios) unless another control mechanism is taking place:

Let us assume that m_i is the mass of the ith link, I_i is its moment of inertia with respect to the center of gravity, $\dot{x}_i, \dot{y}_i$ are the translational velocities of the center of gravity with respect to the inertial frame and $\dot{\alpha}_\iota$ is the angular velocity of the link about its center of gravity. Furthermore, θ represent the angle of the crank.

We can define the generalized moment of inertia of the mechanism (crank + arm) with respect to the crank center of rotation as follows

$$I^*(\theta) = \sum_{i=1}^{n} \left(m_i \tau_{x_i}^2 + m_i \tau_{y_i}{}^2 + I_i \tau_{\alpha_i}{}^2 \right)$$

Where

$$\tau_{x_i} = \frac{dx_i}{d\theta} = \frac{\dot{x}_i}{\dot{\theta}}, \quad \tau_{y_i} = \frac{dy_i}{d\theta} = \frac{\dot{y}_\iota}{\dot{\theta}}, \qquad \tau_{\alpha_i} = \frac{d\alpha_i}{d\theta} = \frac{\dot{\alpha}_\iota}{\dot{\theta}}$$

are the transmission ratio of each link segment with respect to the crank angle θ. Notice that a change in variance of the links' degrees of freedom within a crank cycle is reflected in the change of transmission ratio if it is assumed that $\dot{\theta}$ is constant.

The dynamic equation of the mechanism is as follows:

$$I^*\ddot{\theta} + \frac{1}{2}\frac{dI^*}{d\theta}\dot{\theta}^2 = Q^*$$

Where Q^* is the torque at the crank.

Assuming a constant velocity of the crank $\dot{\theta} = const$ we obtain that $\ddot{\theta} = 0$ and thus:

$$\frac{1}{2}\frac{dI^*}{d\theta}\dot{\theta}^2 = Q^*$$

Assuming we are increasing the resistance of the crank by a factor k we obtain

$$k\left(\frac{1}{2}\frac{dI^*}{d\theta}\dot{\theta}^2\right) = kQ^*$$

If velocity of the crank is to remain constant, we have that

$$\left(\frac{k}{2}\frac{dI^*}{d\theta}\right)\dot{\theta}^2 = kQ^*$$

Therefore, we must have that the magnitude of the term $\frac{dI^*}{d\theta}$, representing the centrifugal and Coriolis dynamics components must increase k-fold. This implies a higher variance in the pose of the arm and, as a consequence, a possible higher variance of the joint angles with respect to the crank angle.

This can be further developed as we can calculate the derivative of the generalized moment of inertia as follows

$$\frac{1}{2}\frac{dI^*}{d\theta} = \sum_{i=1}^{n}\left(m_i\tau_{x_i}\frac{d\tau_{x_i}}{d\theta} + m_i\tau_{y_i}\frac{d\tau_{y_i}}{d\theta} + I_i\tau_{\alpha_i}\frac{d\tau_{\alpha_i}}{d\theta}\right)$$

And thus

$$\frac{1}{2\dot{\theta}}\sum_{i=1}^{n}\left(km_i\tau_{x_i}\frac{d\tau_{x_i}}{d\theta} + km_i\tau_{y_i}\frac{d\tau_{y_i}}{d\theta} + kI_i\tau_{\alpha_i}\frac{d\tau_{\alpha_i}}{d\theta}\right) = kQ^*$$

590

591 Since m_i and I_i are constants the terms $\left(\tau_{q_j}\frac{d\tau_{q_j}}{d\theta}\right)$, with q_j indicating a generic degree of

592 freedom, must all increase k-fold. These terms represent the product of the transmission ratios

593 for the generic degree of freedom q_j and its derivative with respect to . It is obvious that if the

594 transmission ratios do not change, we have that $\frac{d\tau_{q_i}}{d\theta} = 0$ and thus the result is absurd.

595 To allow for constant transmission ratio, and therefore to maintain the variance constant, there

596 needs to be an additional term in the equation that is able to control the torque without changing

597 the kinematic.

598

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696
697 **Figure Captions**
698
699 **Fig. 1**
700 Equipment and maker positions. A: arm cycle ergometer. B: schematic figure of the cycling
701 participant. Black dots illustrate positions of markers placed on the body, on the handlebar of the
702 ergometer and on the chair on which the participant was seated. Joint angles α, β, γ in the
703 shoulder, elbow and wrist, respectively, were computed from marker coordinates.

Fig. 2

Mean angular variances A_1) in low, moderate and high resistance conditions for bimanual cycling (mean across participants and sides); A_2) in low, moderate and high resistance conditions for unimanual cycling (mean across participants and sides). B) in bimanual and unimanual arm cycling (mean across participants, resistances and sides); C) in left and right arms (mean across partcipants, resistances and modes). Lines above bars denote standard errors of the mean (SEM).

Fig. 3

Angular variance profiles**.** Time course of angular variance ($V_{ang}(t)$) in low, moderate and high crank resistances in bimanual cycling and unimanual cycling for the dominant (right) and non-dominant (left) arm. Continuous line: mean across particpants. Dotted line: Mean+SEM

Fig. 4

Mean muscle activity variances A_1) in low, moderate, and high resistance conditions for bimanual cycling (across participants and sides); A_2) in low, moderate, and high resistance conditions for unimanual cycling (across participants and sides); B) in bimanual and unimanual arm cycling (mean across participants, resistances, and sides F=20.11, p=0.0005); C) in left and right arms (mean across partcipants, resistances and modes F=0.15, p=0.7062);

Lines above bars denote standard errors of the mean.

727 **Fig. 5**

728 Muscle activity variance profiles**.** Time course of muscle activity variance ($V_{EMG}(t)$) in low,
729 moderate and high crank resistances in bimanual cycling and unimanual cycling for the dominant
730 (right) and non-dominant (left) arm. Continuous line: mean across participants. Dotted line:
731 Mean+SEM

732

733 **Fig. 6**

734 The ratios of variances of EMG magnitudes compared to the squares of the ratios of EMG
735 magnitudes. A) The ratios of variances of EMG magnitudes in moderate crank resistance respect
736 to variances of EMG magnitudes in low crank resistance and the squares of the ratios of EMG
737 magnitudes; B) The ratios of variances of EMG magnitudes in high CR respect to variances of
738 EMG magnitudes in low CR and the squares of the ratios of EMG magnitudes. Ratios are
739 presented for different combinations of conditions separately (left arm bimnaul, left arm
740 unimanual, right arm bimnual, right arm unimanual). Mean and standard errors across
741 participants are presented. The average variance of muscle activities (EMG) increases
742 approximately at the same rate as the mean squared EMG values when the crank resistance is
743 increased.

Figures

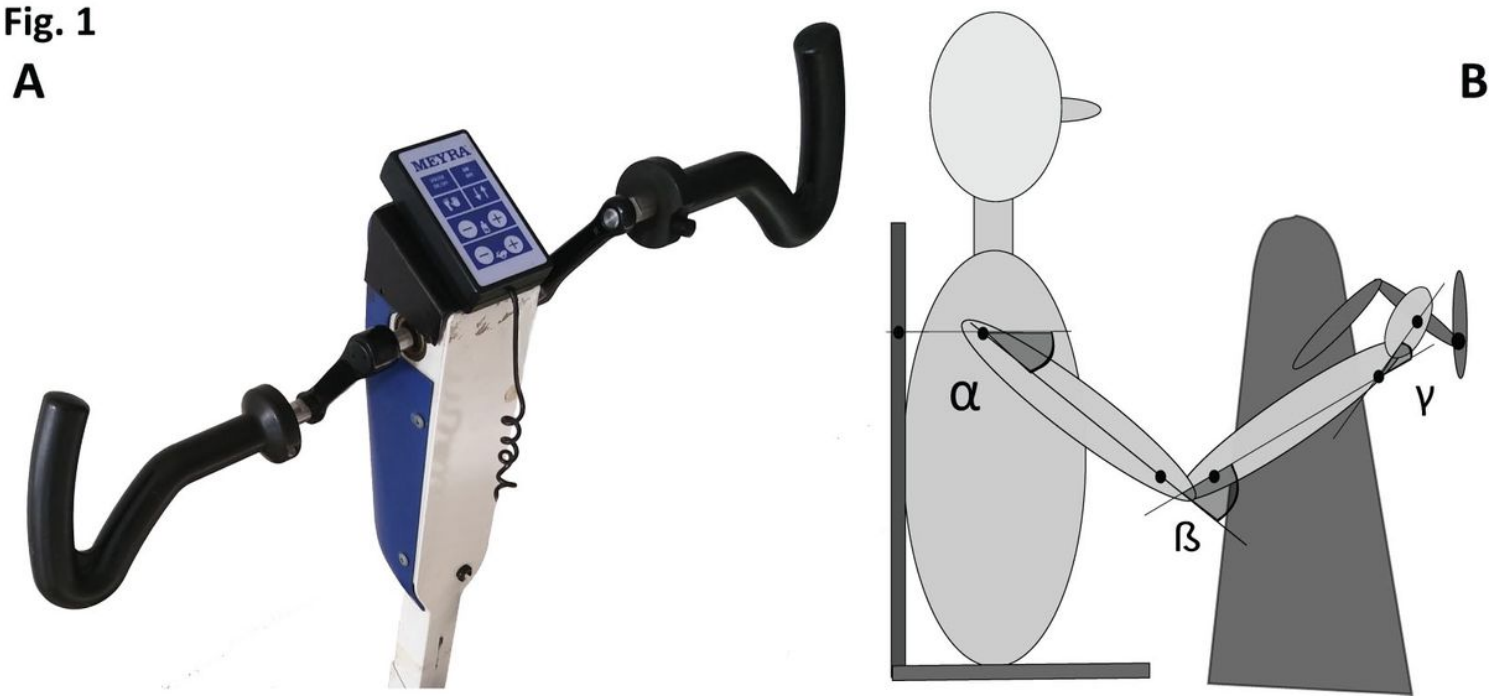

Figure 1

Equipment and maker positions. A: arm cycle ergometer. B: schematic figure of the cycling participant. Black dots illustrate positions of markers placed on the body, on the handlebar of the ergometer and on the chair on which the participant was seated. Joint angles □□□□□□□ in the shoulder, elbow and wrist, respectively, were computed from marker coordinates.

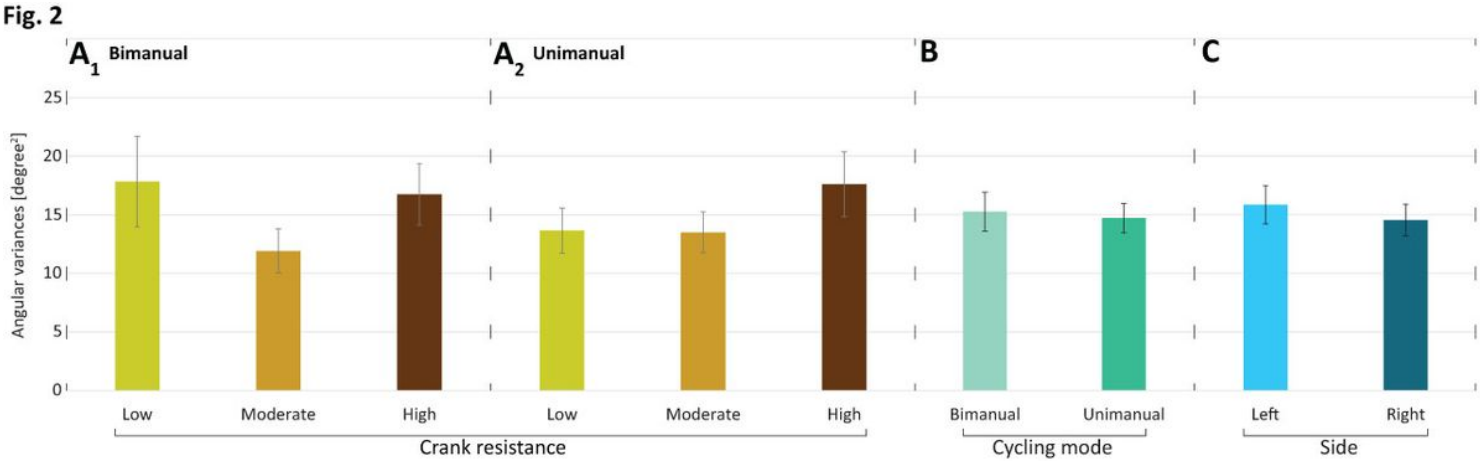

Figure 2

Mean angular variances A1) in low, moderate and high resistance conditions for bimanual cycling (mean across participants and sides); A2) in low, moderate and high resistance conditions for unimanual cycling (mean across participants and sides). B) in bimanual and unimanual arm cycling (mean across participants, resistances and sides); C) in left and right arms (mean across partcipants, resistances and modes). Lines above bars denote standard errors of the mean (SEM).

Fig. 3

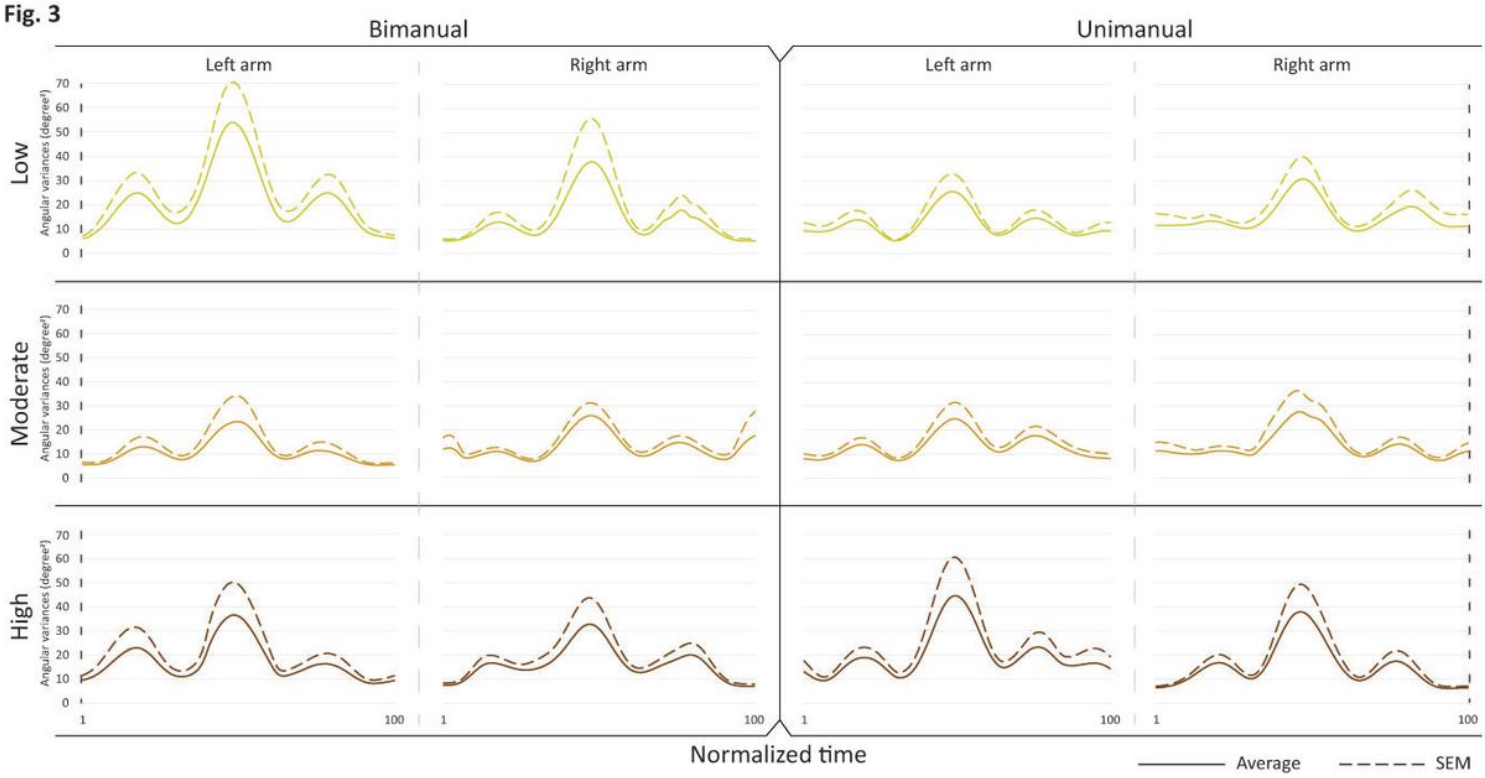

Figure 3

Angular variance profiles. Time course of angular variance (V_ang (t)) in low, moderate and high crank resistances in bimanual cycling and unimanual cycling for the dominant (right) and non-dominant (left) arm. Continuous line: mean across particpants. Dotted line: Mean+SEM

Fig. 4

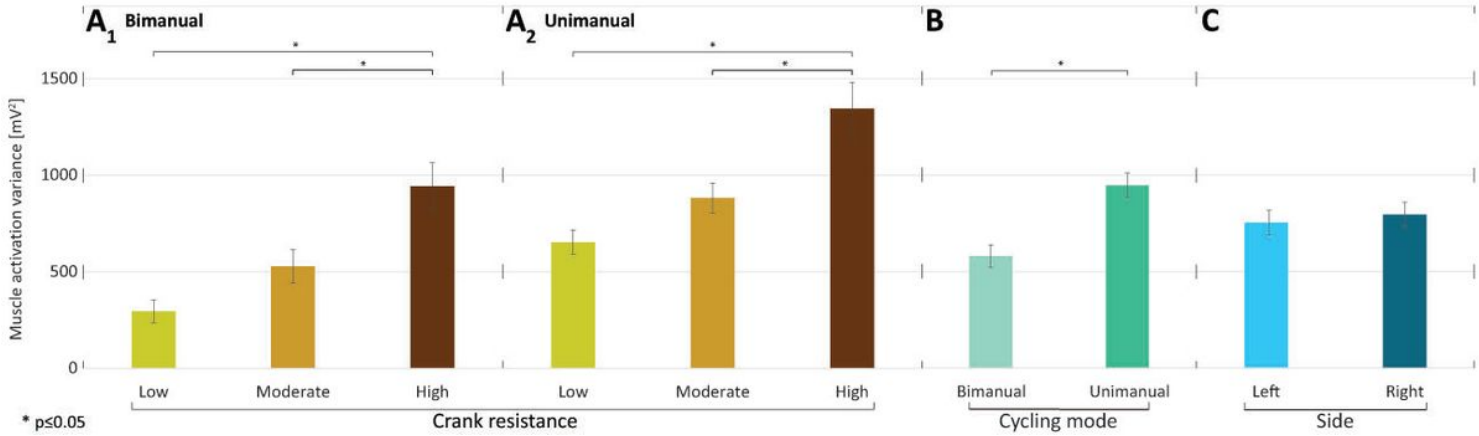

Figure 4

Mean muscle activity variances A1) in low, moderate, and high resistance conditions for bimanual cycling (across participants and sides); A2) in low, moderate, and high resistance conditions for unimanual cycling (across participants and sides); B) in bimanual and unimanual arm cycling (mean across participants, resistances, and sides F=20.11, p=0.0005); C) in left and right arms (mean across partcipants, resistances and modes F=0.15, p=0.7062); Lines above bars denote standard errors of the mean.

Fig. 5

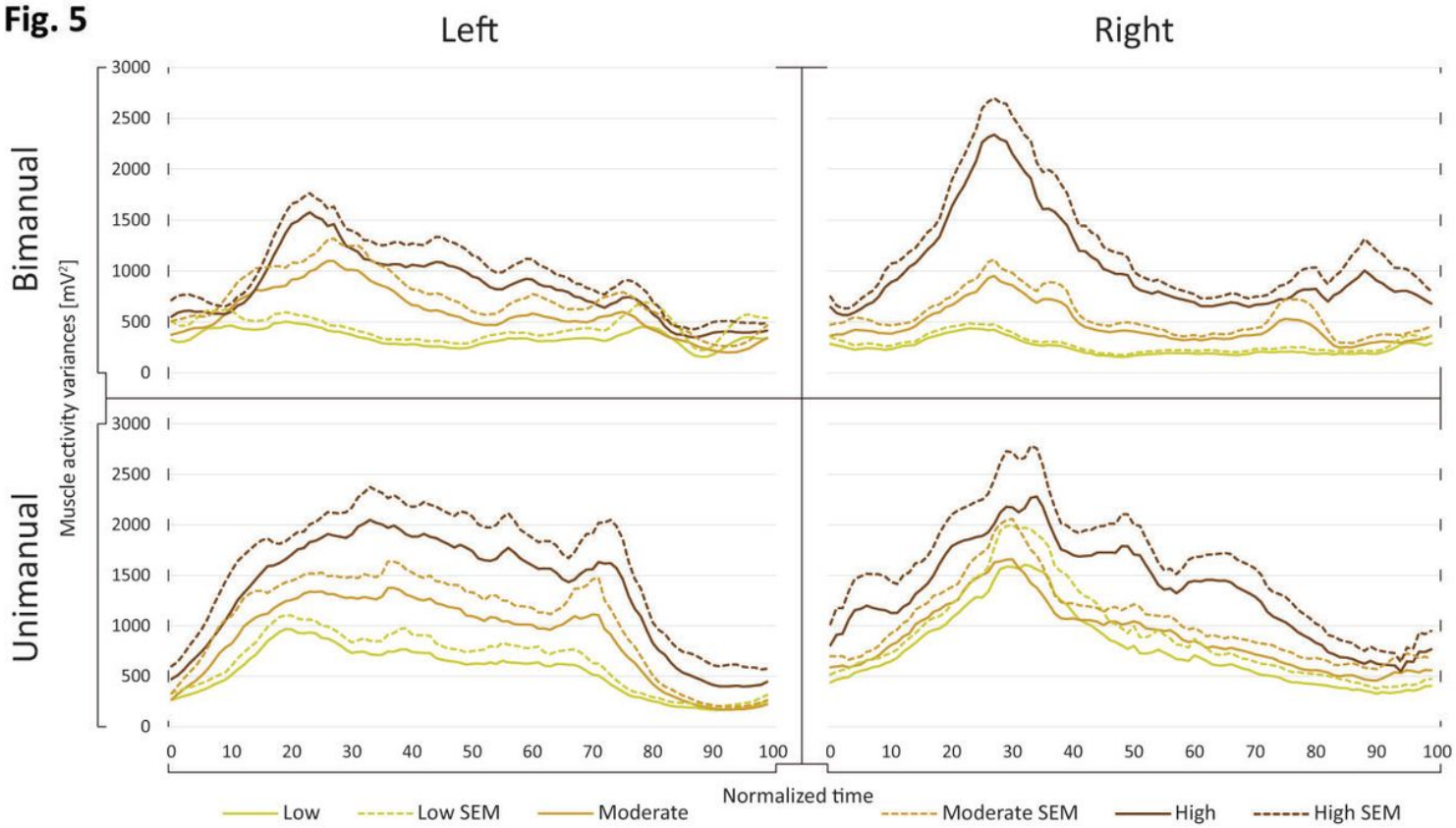

Figure 5

Muscle activity variance profiles. Time course of muscle activity variance (V_EMG (t)) in low, moderate and high crank resistances in bimanual cycling and unimanual cycling for the dominant (right) and non-dominant (left) arm. Continuous line: mean across participants. Dotted line: Mean+SEM

Fig. 6

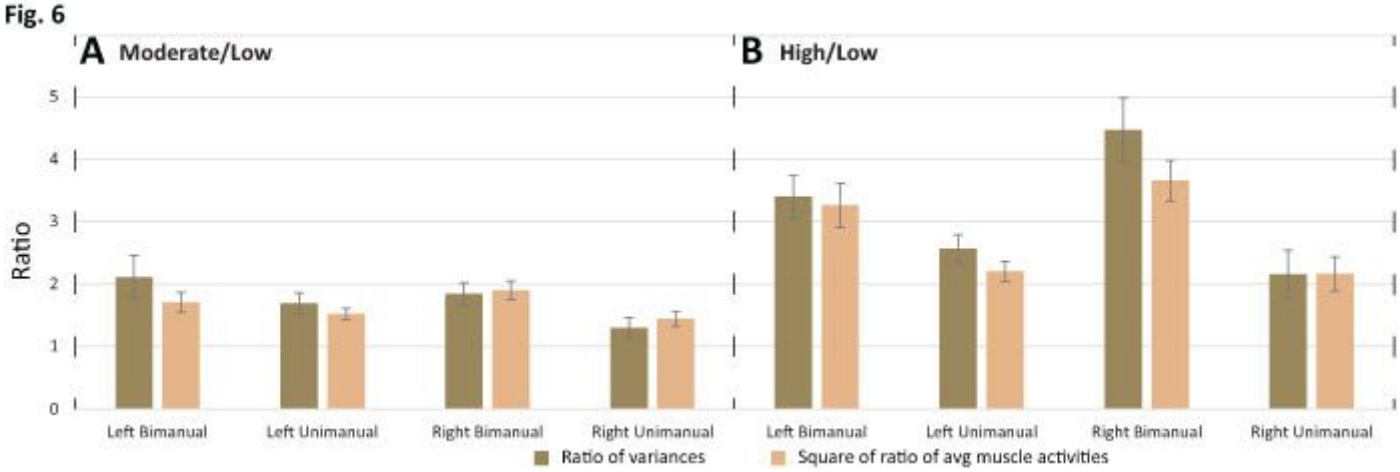

Figure 6

The ratios of variances of EMG magnitudes compared to the squares of the ratios of EMG magnitudes. A) The ratios of variances of EMG magnitudes in moderate crank resistance respect to variances of EMG magnitudes in low crank resistance and the squares of the ratios of EMG magnitudes; B) The ratios of variances of EMG magnitudes in high CR respect to variances of EMG magnitudes in low CR and the squares of the ratios of EMG magnitudes. Ratios are presented for different combinations of conditions separately (left arm bimnaul, left arm unimanual, right arm bimnual, right arm unimanual). Mean and standard errors across participants are presented. The average variance of muscle activities (EMG) increases approximately at the same rate as the mean squared EMG values when the crank resistance is increased.