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Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

Reciprocal aiming precision and central adaptations as a function of mechanical constraints

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ARTICLE INFO

Article history:

Received 23 May 2011

Received in revised form 16 August 2011

Accepted 19 August 2011

Keywords:

Movement accuracy

Attention

EMG

Motor control

ABSTRACT

The present study investigated the influence of mechanical constraints (load and movement velocity) on the movement accuracy during a reciprocal aiming precision task. Seven participants had to point rhythmically and alternatively (with flexion–extension of the right elbow) a cursor at two targets as accurately as possible. Two loads (applied to the limb effectors; 500 and 2500 g), two movement frequencies (1.25 and 1.75 Hz) as well as two target sizes (1 and 5 cm) were manipulated. Surface EMG activity of both *biceps brachii* and *triceps brachii* was recorded. Attentional demands, reflecting the central cost associated with the performance of aiming movements was assessed using a dual-task paradigm (via a probe reaction time task – RT). While the results demonstrated a significant degradation of pointing accuracy with mechanical loading (mean absolute error – AE for 500 g load: $0.32 \text{ mm} \pm 0.64$; mean AE for 2500 g load: $0.51 \pm 0.74 \text{ mm}$), no significant effect of movement frequency was found. For the two mechanical constraints, the mental effort to meet the task demands remained the same (mean RT_{-500g} : $370 \pm 123 \text{ ms}$; mean RT_{-2500g} : $395 \pm 119 \text{ ms}$). Electromyographic activity of both *biceps brachii* and *triceps brachii* muscles evidenced neural adaptations to changes in mechanical constraints. Put together, the present findings suggest that the cause of the observed loss of movement accuracy may probably result from more peripheral alterations such as an impairment of the afferent information processing.

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1. Introduction

Human skills to coordinate body segments in order to reach an object are common activities of daily living. The accuracy of such goal-directed movements is well known to dependent on the distance to be covered, the target width and the velocity of the performed action (Fitts, 1954). When movement amplitude and frequency are fixed, the movement precision is predicted to be altered with decreasing target size. The ability to control the impulsion of aiming movement, linked to muscle activation, is somewhat dependent upon muscle fatigue and/or the level of force to be exerted (Huang et al., 2006; Khan et al., 2006; Lorist et al., 2002).

Dealing with the influence of mechanical constraints on sensorimotor performance, Smits-Engelsman et al. (2006) observed that additional load applied to the limb effectors resulted in a degradation of the discrete tracking performance. However, the load had no effect on a cyclic tracking performance. The authors put forward the idea that the flexibility of higher central structures may be responsible (at least in part) for ensuring steady stable performance during

cyclic aiming movements. Likewise, Selen et al. (2007) suggested that the participants were willing to invest more attentional resources in order to make use of the predictability of the target motion in a tracking task.

The interplay between neuromuscular solicitations such as force requirements (i.e., conditions with additional load and/or higher movement frequency) and neural reorganisation (or task demands) already has been studied (Carson et al., 1999; Murian et al., 2008; Zijdwind et al., 2006). Overall, the literature suggests that the attentional demands associated with performing a motor task is dependent on the force to be exerted: the higher the exerted force, the larger the mental effort (“reflecting the amount of energy necessitated to perform and maintain a coordination pattern”, Temprado et al., 1999). More precisely, Murian et al. (2008) showed a significant increase in attentional cost during a loaded condition (while performing a bimanual coordination task), which was associated with a degradation of the motor performance. In these loading resistance conditions, the authors suggested that an increase of central drive necessary for the production of a higher force was linked to compensating activity of the higher motor centres (Murian et al. 2008).

Using surface electromyography (EMG), an indirect index of such neural adaptations may be given through investigation of muscle activation patterns. Such analyses already highlighted

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changes in the temporal muscle activation according to the level and/or type of load applied to the limb effectors. As an example, Mackey et al. (2002) observed alterations in muscle phase activity (duration, time of onset and offset of EMG bursts) as a function of both the level and the type of load (elastic, viscous, inertial) during rhythmic voluntary flexion and extension of the wrist. These results often have been interpreted as neural compensation to neuromuscular constraints/external load, evidencing the high flexibility of the Central Nervous System (CNS; Baldissera and Cavallari, 2001; Mackey et al., 2002). Facing the aforementioned literature, to our knowledge, what is currently missing is the mutual interaction between the sensorimotor performance, the neuromuscular constraints (load and movement velocity) and mental effort to-be-expended.

Dealing with the fundamental question of how the CNS specifies a set of muscle activation patterns while performing a motor task by minimizing a (energy) cost function (e.g., d'Avella et al., 2006; Kistemaker et al., 2010), the current design aims to evidence the role of attentional cost in respect to an optimal strategy for movement control when the mechanical constraints are altered. Thus, we tested the effect of load applied to the limb effectors and movement frequency on (a) the reciprocal aiming precision performed with elbow flexion–extension, (b) the patterns of muscle activity and (c) the associated attentional demands. We hypothesized a degradation of the pointing precision as a function of these mechanical constraints, accompanied with central adaptations at the level of the associated attentional demands and/or muscle activation coordination strategies.

2. Materials and methods

2.1. Participants

Seven right-handed volunteers (age: 22.1 ± 2.1 years) participated in this experiment. All subjects were informed of the possible risk and discomfort associated with the experimental procedures prior to giving their written consent to participate. They had normal or corrected to normal vision and had no previous experience with the experimental task. The experimental design of the study was approved by the local Ethical Committee and was carried out in accordance with the Declaration of Helsinki.

2.2. Task and apparatus

A schematic diagram of the experimental apparatus is shown in Fig. 1. The participant was standing in front of the experimental device and his arm was put down on a table so that the abduction of his shoulder joint was 90° (in the frontal plane) and only flexion–extension movements of the elbow joint in the transverse plan were possible. The participant had to point rhythmically and alternatively a small cursor (1 cm diameter sphere) at two targets as accurately as possible. The distance between the targets was constant (37 cm). To avoid the participation of shoulder muscles, the participant was instructed not to move his shoulder during the trial. The cursor was fixed to a rope and was moved by means of a small manipulandum with flexion–extension of the right elbow (the resulting movement amplitude was 90°). As guided via pulleys, the rope was tightened with a slight load (5% of the maximal isometric voluntary contraction in elbow flexion) applied to its extremity (see Fig. 1). Pilot experiments revealed that this load was enough to tighten the rope and sufficiently low to prevent additional muscle fatigue. In fact, the maximal voluntary contraction was not affected ($p = 0.46$; six subjects) by a 40-s task. The movement of the cursor was recorded with a linear position sensor connected to the rope (PT1 Scaimé®, Annemasse, France).

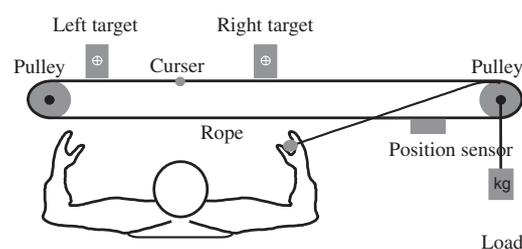


Fig. 1. Schematic representation of the experimental apparatus used during the pointing task. The participant had to point rhythmically and alternatively a small cursor (1 cm diameter sphere) at two targets as accurately as possible. The cursor was fixed to a rope and was moved by means of a small manipulandum with flexion–extension of the right elbow (the resulting movement amplitude was 90°).

2.3. Experimental design

Two loads (500 and 2500 g), two movement frequencies (1.25 and 1.75 Hz for each half cycle) as well as two target sizes (1 and 5 cm) were manipulated. The movement oscillation frequency was imposed by an auditory metronome. Using the dual-task paradigm (Abernethy, 1988), attentional demands associated with performing the pointing task were measured via a probe reaction time task (RT). For this latter, the participant had to respond as quickly as possible to 50-ms auditory signals by releasing a switch placed under his right heel. For each condition, the participant performed a double task trial of 40-s duration. The metronome was turned off after 10 s in order to avoid interference between RT signals and the metronome. Thus, only the remaining 30 s of the trials were analysed. The participants were instructed to “touch” the targets and had to keep their movement frequency closed to the frequency which was dictated previously by the metronome. As done in other works (e.g., Murian et al., 2008; Temprado et al., 1999), they also were instructed to give priority to the pointing task (i.e. to point the targets as accurately as possible) without giving up the secondary RT task. RT was defined as the time elapsed between the signal and the onset of releasing of the switch. During the last 30 s of a trial, the interval between RT signals varied randomly between 1 and 6 s (i.e., nine RTs per trial).

After a familiarization period, each participant completed the experimental session, composed of eight trials (2 load conditions \times 2 frequencies \times 2 target sizes), each separated by one minute of rest. The order of conditions across subjects was randomized.

2.4. Data recordings

Surface electromyographic (EMG) activity was recorded from the long head of the right *biceps brachii* and *triceps brachii* muscles. For each muscle, a pair of surface electrodes (In Vivo metrics – Ag–AgCl; 1 cm interelectrode distance) was attached to the skin. Prior to electrode application, the skin was shaved and cleaned with a mixture of alcohol and ether to minimize impedance. Each electrode was placed according to the recommendations of SENIAM (surface EMG for non-invasive assessment of muscles) (Hermens et al., 2000). *Biceps brachii* and *triceps brachii* are considered as the main synergist of elbow flexion and extension, respectively. The EMG signals were preamplified (gain = 600, bandwidth 6–400 Hz) and digitized at 1024 Hz with a 12-bit A/D converter (Myodata Compact, Electronique du Mazet, France). The RT response was sampled at 1000 Hz.

2.5. Data analysis and statistics

The time series of the cursor position were low-pass filtered with a dual pass second-order Butterworth filter, with a cutoff frequency of 10 Hz. A peak-picking algorithm was used to locate the

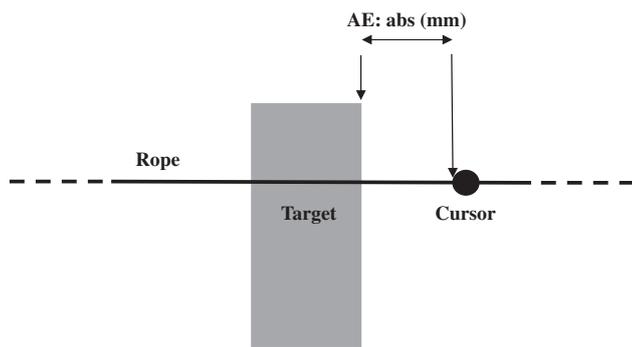


Fig. 2. Schematic representation for the calculation of the absolute error (AE). This calculation was identical at the right and left side of the target. As the cursor touched the target, there was no error.

reversal points of the movement. The mean movement frequency (Fq) was calculated for all trials and subjects. The position signal was calibrated according to the targets location in order to determine the absolute error (AE) of each reversal point. The mean AE of a trial was calculated as:

$$AE = \frac{\sum |x_{target} - x_{curser}|}{n_{reversal}} \quad (1)$$

where x_{target} and x_{curser} are the position of the target and the curser at a given reversal point and $n_{reversal}$ the number of reversal points in a given trial. For each reversal point, the greater the distance between the cursor and the target, the higher the AE and the lower the aiming precision (Fig. 2).

Each movement cycle was divided into 10 epochs of equal duration. The epochs 1–5 corresponded to the extension phase, and the epochs 6–10 represented the flexion phase of the elbow. The EMG root mean square (RMS) amplitude of the resulting data was calculated for each epoch. To display an overall pattern of muscle activity, the RMS values from all the cycles within a trial were averaged together. Then, 10 values were obtained with *biceps brachii* EMG RMS and *triceps brachii* EMG RMS. In order to determine the onset and offset of mean EMG burst activity for the *biceps brachii* (agonist muscle), a linear envelope first was computed through the application of a low-pass second-order Butterworth filter (dualpass, cut-off frequency = 3 Hz). Each cycle was interpolated to 100 points. Then, a mean linear envelope was obtained by averaging all cycles within a trial. EMG bursts of muscle activation were defined as the muscle activity between the period when the signal was above a threshold of 20% of the peak EMG value (onset) and the end of this phase (offset) (Dorel et al., 2009) (see Fig. 3). Hence, mean burst onset, offset, peak and duration (expressed in percent of movement cycle) were obtained for each trial.

For each trial, the minimum and the maximum RT as well as all RT values below 100 ms (considered as anticipated, see Abernethy, 1988) were removed from data set. Considering all participants and all conditions, only 3 of 504 RTs ($\approx 0.6\%$) were discarded because they were anticipated. The RTs within a trial then were averaged together. A decline in RT performance (i.e., higher RTs) is an assessment of increasing demands on central resources by the pointing task (Abernethy, 1988; Temprado et al., 1999).

All these dependent variables (AE, Fq, burst onset, offset, peak and duration, and RT) were submitted to a 2 (load) \times 2 (target size) \times 2 (frequency) ANOVA with repeated measures (Statistica[®]V6, Statsoft, Maison-Alfort, France). EMG RMS were submitted to a 2 (load) \times 2 (target size) \times 2 (frequency) \times 10 (epoch). LSD post hoc analyses were carried out when effects reached significance. A p -value below 0.05 was considered statistically significant.

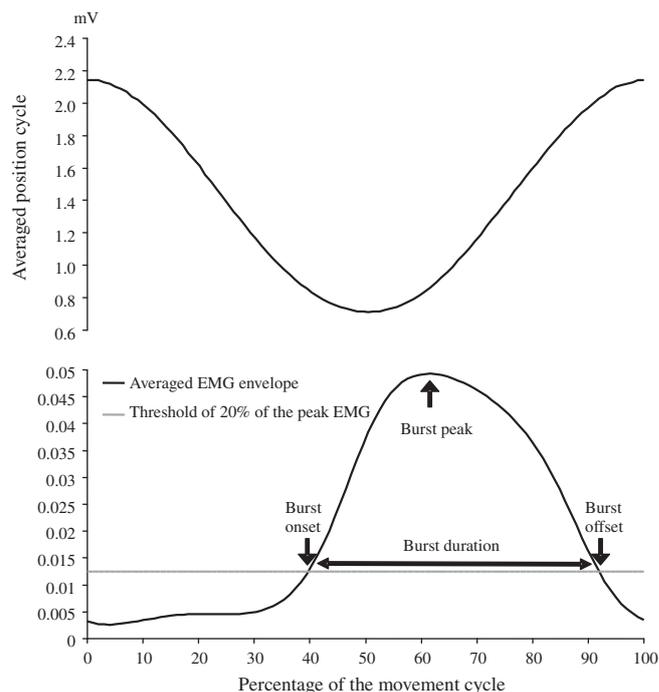


Fig. 3. Individual example of the determination of onset, offset, duration and peak of an EMG burst. EMG bursts of muscle activation were defined as the muscle activity between the period when the signal was above a threshold of 20% of the peak EMG value (onset) and the end of this phase (offset).

3. Results

Movement frequency

Neither main effect nor interaction concerning the load ($F(1,6) = 1.10, p = 0.33$) and the target size ($F(1,6) = 0.14, p = 0.72$) factors was found. The frequencies were highly closed, whatever the load condition (Fq1 – 500 g = 1.36 ± 0.09 Hz; Fq1 – 2500 g = 1.35 ± 0.07 Hz; Fq2 – 500 g = 1.79 ± 0.06 Hz; Fq2 – 2500 g = 1.79 ± 0.07 Hz).

Movement accuracy

The analysis showed the main effects of load ($F(1,6) = 6.71, p < 0.05$) and target size ($F(1,6) = 10.70, p < 0.05$). It also revealed a load \times target size interaction ($F(1,6) = 6.87, p < 0.05$). The effect of load was found only for 1 cm: the heavier the load, the higher the AE (Fig. 4).

EMG activity level

Concerning the *biceps brachii*, the analysis revealed a main effect of load ($F(1,6) = 38.41, p < 0.001$). The *biceps brachii* activity level was higher for the 2500 g condition compared to the 500 g load, whatever the epoch. A load \times epoch interaction ($F(9,54) = 16.39, p < 0.001$) (Fig. 5), a frequency \times epoch interaction ($F(9,54) = 32.49, p < 0.001$) as well as a load \times frequency \times epoch interaction ($F(9,54) = 9.41, p < 0.001$) were also highlighted by the analysis.

Regarding the *triceps brachii*, the ANOVA showed the main effects of load ($F(1,6) = 7.22, p < 0.05$) and frequency ($F(1,6) = 9.16, p < 0.05$). Overall, the *triceps brachii* was more activated for the 2500 g compared to the 500 g condition and for the highest compared to the lowest movement frequency. A load \times epoch ($F(9,54) = 10.65, p < 0.001$) and a load \times frequency \times epoch ($F(9,54) = 4.43, p < 0.001$) also have been revealed (Fig. 5).

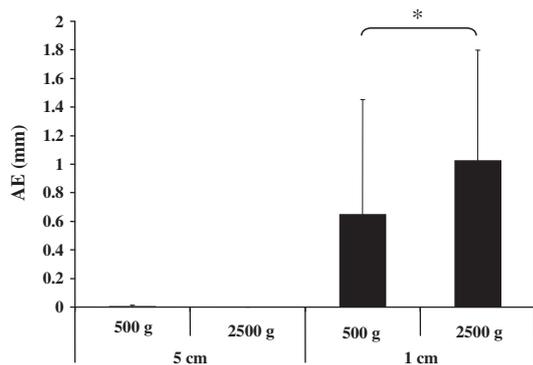


Fig. 4. Mean absolute error (AE) as a function of the load (500 and 2500 g) and the target size (1 and 5 cm). * Indicates a significant difference between the two loads ($p < 0.05$).

Burst analysis

Concerning the burst onset, offset, peak value and duration, the ANOVA revealed a main effect of movement frequency. Precisely, burst onset ($F(1,6) = 65.85, p = 0.001$), offset ($F(1,6) = 70.70, p = 0.001$) and peak value ($F(1,6) = 26.93, p = 0.01$) appeared earlier in the movement cycle for the highest frequency (1.75 Hz). Moreover, the burst duration was longer for the higher movement frequency condition ($F(1,6) = 29.89, p = 0.01$) (Table 1). A main effect

of load also was revealed: the burst offset ($F(1,6) = 8.71, p < 0.05$) and the peak value ($F(1,6) = 11.68, p < 0.05$) appeared earlier for the heaviest load condition (2500 g) (Table 1).

Attentional demands

Concerning the reaction time, the ANOVA revealed neither main effect nor interaction (effect of test for RT: $F(1,6) = 1.696, p = 0.234$). As an example, the RT values for the 500 g load condition (mean RT: 370 ± 123 ms) were similar to the RT observed for the 2500 g load condition (mean RT: 395 ± 119 ms).

4. Discussion

The present experiment first addressed the influence of muscle coordination impairment (induced by load applied to the limb effectors and movement frequency) on the performance in a reciprocal aiming task and on the concomitant attentional demands. Overall, the results are informative with respect to control strategies employed by the CNS when controlling limbs for movement accuracy. Contrary to our hypothesis, the way participants control the task requirements does not result in higher attentional load, whatever the mechanical constraints, as evidence of how the CNS could have regulated the cost function for optimizing the sensorimotor control when facing muscle activation alterations. Despite clear instructions (i.e., priority to the pointing task), subjects did not maintain the same level of accuracy performance

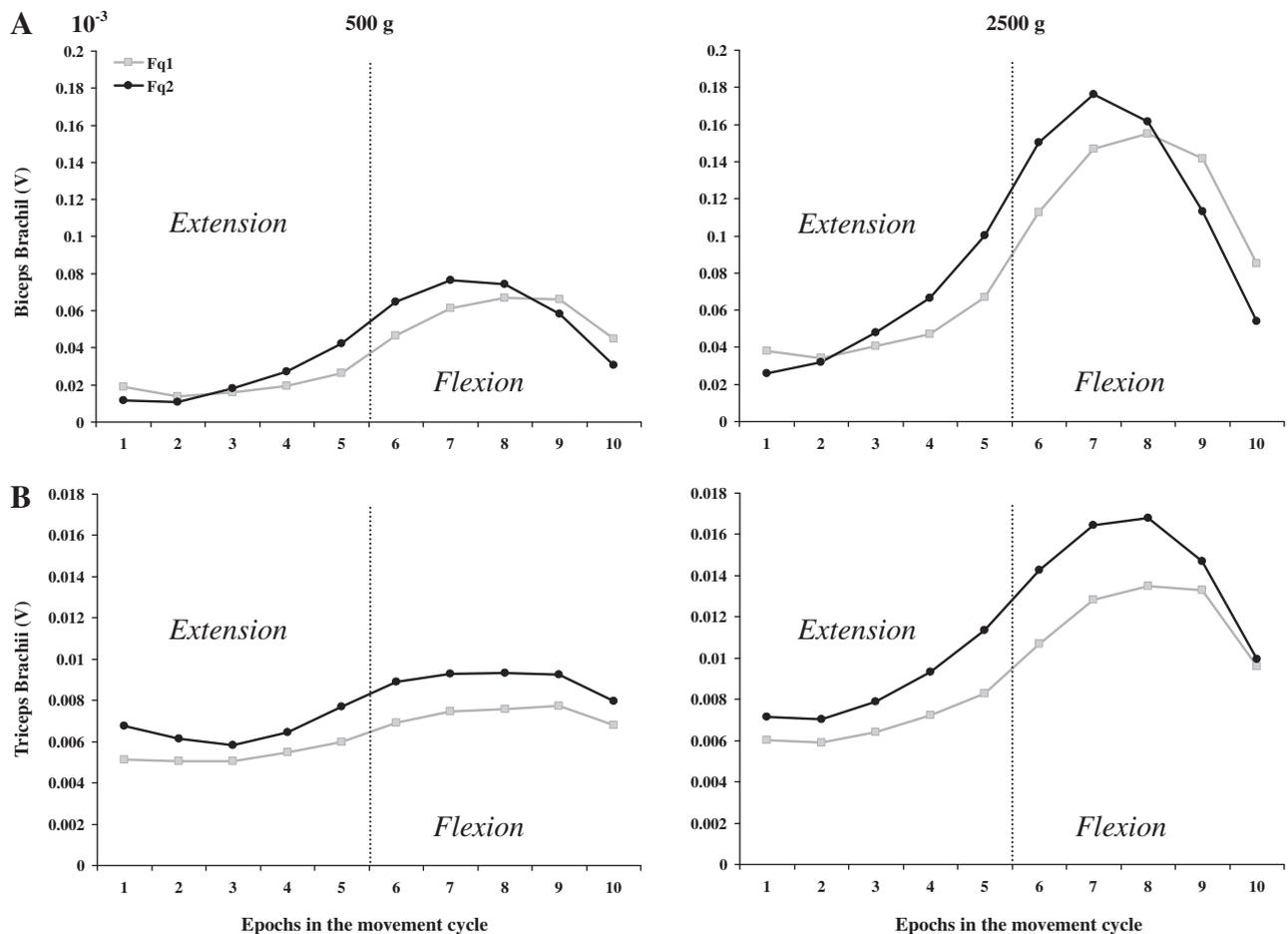


Fig. 5. The 10 mean root mean square (RMS) values from the *biceps brachii* EMG signal (A), and the *triceps brachii* signal (B) as a function of load (500 g: left panels and 2500 g: right panels) and frequency (● Fq2 and ■ Fq1), averaged between all participants (on an averaged movement cycle). * Indicates a significant difference between the two loads ($p < 0.05$).

Table 1
Changes in muscle activation timing induced by movement frequency or load.

Dependent variables (burst analysis)	Main effects of the movement frequency factor		Main effects of the load factor	
	1.25 Hz	1.75 Hz	500 g	2500 g
Onset	45.0 ± 5.14%	32.48 ± 6.80%*	39.44 ± 9.00%	38.06 ± 8.51%
Offset	98.38 ± 4.18%	94.22 ± 4.69%*	97.83 ± 4.68%	94.77 ± 4.67%*
Peak	75.14 ± 9.91%	66.15 ± 6.41%*	72.46 ± 10.30%	68.84 ± 8.27%*
Duration	53.35 ± 3.14%	61.74 ± 6.49%*	58.39 ± 7.26%	56.71 ± 5.86%

Onset burst, offset burst, burst duration and burst peak expressed in percent of movement cycle duration, as a function of frequency (1.25 and 1.75 Hz) and load (500 and 2500 g).

* Indicates a significant difference between the two conditions of main factors ($p < 0.05$).

during the highest loaded condition as reported previously by Murian et al. (2008). Even if participants adapted their control strategies because of mechanical constraints, they did not change the motor command generated by the CNS by regulating the higher-level cognitive inputs, in particular executive function (Royall et al., 2002).

The present results differ from those observed by Selen et al. (2007) who failed to reveal any influence of neuromuscular solicitations on tracking accuracy. The authors proposed that participants may have invested more attentional resources to meet the target location in the highest demanding experimental state. Because of the degradation of aiming performance with mechanical constraints, this present result raises the important issue of relationships between central resources investment and outcome conflicts in performance of a dual task composed of a reciprocal aiming task and a discrete RT task (Temprado et al., 2001). As a sign of outcome conflict, structures and/or mechanisms needed to perform the two tasks are likely to interfere, and arguably have a decisive influence in the capacity to discover/adopt the optimal control strategy for facing load (i.e. necessity to produce a higher muscle force).

It also is important to keep in mind that the subjects were instructed to hit ("touch") the targets and to follow the driven frequency as precisely as possible. This last instruction is a key point and quite different from the classical Fitts paradigm (we consequently decided to compute the absolute error because of the potential conflict between these two main instructions). With a given movement velocity, the movement accuracy was expected to be lower for the heaviest loading constraint. Firstly, it is worth noting that the manipulation of load did not have any effect on the actual movement frequency. Based on the original dual strategy hypothesis, speed-sensitive or speed-insensitive strategies (Corcos et al., 1989; Gottlieb et al., 1989), the control of rapid aiming movements is dependent upon whether participants can exert explicit control over their movement velocity. In this context, the decrease of movement accuracy as a function of load cannot be imputed to speed constraints. For the smallest target size, it was impossible for participants to reduce the movement speed in order to meet the accuracy demands. The results of this current experiment confirmed the influence of load on movement accuracy, especially when the target size is decreased.

Besides, interesting insights also should be provided from the neural compensation for peripheral load (Baldissera et al., 2000; Baldissera and Cavallari, 2001; Mackey et al., 2002). To this aim, another organisational principle that can play a role on the control of movement accuracy is the relationship between the load and central demands incurred by the CNS to perform rapid aiming movements (e.g., Ketelaars et al., 1999; Khan et al., 2006). In other words, from our results, we investigated whether neural effort for movement control is an important factor that influences how CNS arrives at kinematics and associated muscle activation patterns when mechanical constraints are altered (Dounskaia and Goble, 2011; Kistemaker et al., 2010). However, the alterations of accuracy movement caused by the load were not paralleled by a change

in mental effort as assessed through simple reaction times. This result is in accordance with the lack of effect on RT paralleled by a loss of movement stability caused by a change of the inertial load constraints in a bimanual rhythmic coordination task (Temprado et al., 2001). Unlike the proposition of Smits-Engelsman et al. (2006), our findings suggest that the origin of the load-induced accuracy degradation may be examined by investigating processes at a more peripheral level of the motor pathway, like a possible impairment of proprioceptive integration processes (Jaric et al., 1999; Taylor et al., 2002), at least as the movement speed is experimentally controlled and as the provided visual feedback facilitates the executive control (Khan et al., 2006; Selen et al., 2007).

Striking and interesting findings concern the EMG data. Our results evidenced neural adaptations to the different conditions, helpful to prepare, slow down and/or initiate the movement rapidly near the target. These results are supported by the EMG pattern analyses. Indeed, the *biceps brachii* is activated earlier and longer during the aiming movements at the highest frequency. Moreover, the present results revealed that the peak and the offset of the *biceps brachii* EMG activity appeared earlier in the movement cycle for the heaviest load condition. Overall, our findings are in accordance with the literature. Indeed, assuming a constant electromechanical delay of about 30–100 ms (Cavanagh and Komi, 1979), it would be expected that muscle activation occurs progressively earlier as movement velocity increases in order to develop force in the same part of the cycle/movement (Li and Baum, 2004). On the other hand, modifications in the temporal characteristics of the EMG signal (i.e., earlier bursts in the movement cycle) have been found under the effect of different load constraints. Such results should reveal neural adaptations to changes in neuromuscular constraints (Baldissera and Cavallari, 2001; Mackey et al., 2002). This hypothesis of neural control strategy is in accordance with recent findings demonstrating the consistency of muscle synergies, but with a slight timing adjustment of their activation regarding mechanical constraints during pedalling (Hug et al., 2011).

5. Concluding remarks

If our main results demonstrated a degradation of pointing accuracy with mechanical loading, the mental effort to meet the task demands remains the same as indicated by the reaction times. The cause of the observed loss of movement accuracy may probably result from more peripheral alterations such as an impairment of the afferent information processing. Indeed, proprioceptive integration such as neuromuscular spindle sensibility may depend upon adaptation to changes in stiffness due to changing task requirements (Gribble et al., 2003; Osu and Gomi, 1999; Taylor et al., 2002). Moreover, neural compensations for neuromuscular constraints may be interpreted as direct CNS adaptations to high muscle contributions (Selen et al., 2007; Smits-Engelsman et al., 2006). Such insight will give interesting data for understanding the influence of neuromuscular and cognitive constraints upon vol-

untary sensorimotor control in humans (Deschamps et al., 2004; Murian et al., 2007). It should be kept in mind that this pilot study involved only seven subjects, and that no specific assessment was made of when RT was measured. Indeed, the magnitude of RT would change depending on the distance of the probe signal relative to the target. It thus will be necessary to measure RT at different phases of the movement cycle. Considering these limitations, additional experiments with larger samples should be conducted to confirm this result.

Conflict of interest statement

None declared.

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