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Towards a Biomarker of Motor Adaptation Integration of Kinematic and Neural Factors

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***Abstract*—We propose an experimental protocol for the integrated study of motor adaptation during target-based movements. We investigated how motor adaptation affects both cerebral activity and motor performance during the preparation and execution of a pointing task, under different conditions of external perturbation. Electroencephalography (EEG) and movement analysis were simultaneously recorded from sixteen healthy subjects enrolled in the study. EEG signal was pre-processed by means of Independent Component Analysis and Empirical Mode Decomposition based Hilbert Huang Transform, in order to extract Event-Related Synchronization and Desynchronization parameters. Movement analysis provided several kinematic indexes, such as movement durations, average jerk and inter-quartile-ranges. Significant correlations between score, neural and kinematic parameters were found. Specifically, the duration of the going phase of movement was found to correlate with synchronization in the beta brain rhythm, in both the planning and executive phases of movement. Inter-Quartile Ranges and average jerk showed correlations with executive brain parameters and ERS/ERDcueBeta, respectively. Results indicate the presence of links between the primary motor cortex and the farthest ending point of the upper limb. In the present study we assessed significant relationship between neural and kinematic descriptors of motor adaptation, during a protocol requiring short-term learning, through the modulation of the external perturbations.**

***Index Terms*—Motor Adaptation, Event-related Synchronization/Desynchronization, Motor Kinematics.**

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I. Introduction

In the processes of learning skills, motor adaptation (MA) has been defined as the response of the neuro-musculo-skeletal (n-m-s) system, whereby previously known motor strategies are adjusted to match a new goal and/or a change in the environmental conditions [1]. MA involves the process of generalization, which is the ability of taking previously learned n-m-s paradigms out of their original contexts, to transfer them into the new one, and to modify them accordingly, in order to respond to the emerging needs [2].

The study of learning and adaptation has been generating increasing interest within the scientific community, and has been addressed from different points of view (e.g. neuroscience, biomechanics, psychology) with authors focusing on the multiple aspects of its manifestation, such as, for example: the contribution of the central nervous system (e.g. central planning and/or control) [3]-[4], the behavior of the peripheral neuromuscular system (e.g. muscle activation, motor performance) [5]-[6], or the consistency in the final outcome [1].

The adaptation of a well-established motor scheme to a new environment passes through an initial preparative planning and a subsequent ongoing correction of the motor strategy [7]-[8]. The initial planning stage involves multiple areas of the brain cortex, providing the neural encoding about *what to do* (fronto orbital cortex – Brodmann Area 11 and 12), *how to do it* (fronto lateral cortex – BA 44 and 45) and *when to do it* (supplementary area - BA 6) [9]. The electroencephalographic evidence of this process has been reported by independent studies, and mainly consists in a power decrease in the alpha frequency band (8-12 Hz), often referred to as Event Related desynchronization (ERD) [10]-[11]-[12].

Before the movement preparation ends, the executive part of the movement usually sets in. Correspondingly, ongoing correction takes place, largely relying on the visual and proprioceptive feedback [13]-[14]-[15]. Ongoing correction is explained as the generation of a differential error in the cerebellum, later processed by the same brain areas mentioned above, in a recursive manner [16]-[17]. Although the hodology (i.e. description of neural pathways) goes beyond the purpose of the present paper, we need to mention that the primary motor cortex (BA 4) also plays a primary role in the delivery of the motor command to the muscles, thus allowing a refined optimization of joint angles and the subtle tuning of torques

[3], [18]. Electrically, a correlate is observed in the beta frequency band (18-25 Hz), where some power decrease is found. Movement completion ends up in a sudden power increase in the beta band, also called event related synchronization (ERS) or *beta rebound* [12]. Through the repetition of the motor task, the cerebellar error is iteratively minimized, and thus the motor scheme is progressively perfected. The resulting effect is an improved behavioral outcome, also referred to as *adaptation*.

Although the electrical correlates of movement have been studied in previous works, the links between their modulation and the resulting effects on movement kinematics and dynamics still need exploration. Makienko et al. [19] reported adaptive changes in movement-related-potentials (MRPs) in a hand-fingers protocol, while Gentili et al. [20] and Del Percio et al. [21] reported decreased ERS and ERD with practice. Lastly, few studies dealt with the electrical effect in the brain due to the imposition of external forces to the arms [22]-[23]-[24]-[25]-[26].

Learning a new motor skill or adapting an old one involves the acquisition of a permanent ability in getting to the desired result with consistence, and regardless of the possible perturbation acting on the n-m-s system [1], [27]. Therefore, while increased variability in the final outcome is necessarily an index of poor ability, variability in motor performance cannot be read as such. Movement variability is inherently present even in very skilled individuals that repeat a well mastered movement (e.g. [1], [28]-[29]); therefore it is not, or not only, a manifestation of noise that hinders the final result [30]. The same result may in fact be achieved by choosing among different motor patterns that equally lead to a successful outcome [31]. Some authors have suggested that variance in motor performance can be a combination of noise and functional proprieties of the n-m-s system, and that it can represent a form of potential to adapt to perturbations [30], [32]-[33]. A few recent studies have supported this hypothesis by investigating the nature of movement variability both in clinics (e.g. [34]-[35]) and in sports (e.g. [29]). However, most of the meaning of performance variability and its relation with skills and skills generation/retention has still to be determined.

Furthermore, to our knowledge, literature lacks of integration between the different scales through which learning skills and motor adaptation can be observed. The only study on this topic has been conducted by Yuan and colleagues [36], who investigated the relationship between the kinematics of imagined and actual hand movement, i.e. the clenching speed, and the EEG activity in ten human subjects. Understanding the interaction and the possible relations between the cerebral activity, the transmission of central commands to the periphery, and, hence, motor performance and eventual results, is a fundamental step to a thorough knowledge of skills acquisition processes. This information may be beneficial for the study and the better knowledge of neuromuscular pathologies, for the selection of the most effective rehabilitative intervention, and for the improvement of training procedures (e.g. in sports). The implementation of a proper monitoring protocol may have application for all the aforementioned aspects.

Therefore, the aim of this study was: (i) to propose, implement and test an experimental protocol for the integrated study of motor adaptation during target-based movements; (ii) to assess how short-term motor adaptation affects both cerebral activity and motor performance during the preparation and execution of a pointing task, under different conditions of external perturbation; (iii) to look for possible relations between the different determinants of motor adaptation. In this study, we focused on the activity of the primary motor cortex and on the kinematics of the fingertip.

II. Materials and Methods

A. Subjects

Sixteen healthy subjects (13 males and 3 females) volunteered to take part in the study without receiving any reward for their participation. Their mean age was 24 years (SD 3.20, range: 19-30 years). The participants had no visual or musculo-skeletal impairment, nor any cardiovascular, neurological and neuropsychiatric diseases. None of them had knowledge of previous brain injury or parents affected by psychiatric pathologies. They were free from alcohol and drug dependency and also avoided taking coffee, medicine and alcohol before the experimental session. All the participants were strongly right-handed as assessed by the Edinburgh Handedness Inventory [37]. Anthropometric [38] and craniometrical measures [39] were taken.

The study was approved by the local Institutional Review Board and carried out at the Bioengineering Department of Politecnico di Milano. Participants were properly informed about aims of the research, testing procedures, personal data treatment and the possibility of withdrawal at any time. Written informed consent was obtained from each subject before taking part to the experiment.

B. Experimental protocol

The test consisted of 180 repetitive pointing movements, during which the participant was asked to touch a visual target as quickly and accurately as possible. The experiment was performed in a quiet room with dimmed light. The subject was seated on a comfortable chair. His/her position was constrained in order to limit the trunk, head and wrist movements. A 17” touch-screen monitor was placed on a table in front of him/her and used for the representation of the target space. The screen was placed at a distance from the acromion that corresponded to the 95% of the upper limb length (i.e. the distance between the acromion and the fingertip). This distance was arbitrarily chosen by the authors to allow the participants to reach the target space with their arm and without any appreciable involvement of other upper body joints.

Standardized instructions were given to each participant. During each trial (Fig. 1), the subject was asked to stare at a round cross-shaped cue sign first. This sign represented a preliminary warn of the target that would follow. This is the *planning* stage. The subject was instructed not to move at this stage, and to wait till a full circle (target) appeared in the same position. After the target appeared, the subject's goal was to touch the centre of the target as quickly and accurately as possible. This is the *execution* stage. A percentage score (0–100%), where 100% corresponded to the centre of the target and 0% to its border, was assigned to each trial by means of a computerized real-time routine. Bad results in terms of performance (e.g. missed target, double touches, …) or timing (e.g. anticipation of start, late touch, …) were all considered as a missed target and given a 0% score. The score represented the outcome variable. The interval between cue and target signs was randomized between 1.5 and 2.0 s in order to reduce movement anticipation. After movement execution, feedback information about the goodness of result was given in the form of a red dot indicating the point touched, together with the percentage score. After this stage, a new trial followed, after a time interval randomly chosen between 1 and 2 s.

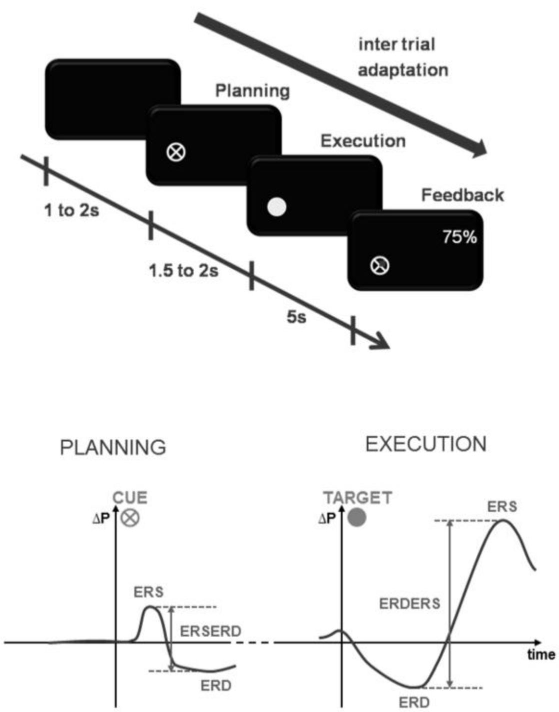

Fig. 1. TOP: The subsequent trial stages (left to right). The adaptation process is obtained through the repetition of planning, execution and feedback stimuli. BOTTOM: Neural parameters are schematized. Event-Related-Synchronization (ERS) and Event-Related-Desynchronization (ERD) are respectively an increase and decrease of power (ΔP), with respect to a baseline value. ERSERD and ERDERS measures are derived differentially from the previous ones. Thus, the reader should note that ERSERD and ERDERS measures are not independent, with respect to ERS and ERD.

The testing protocol consisted of 3 sessions (A, B and C) of 60 repetitions each, so that a total of 180 trials per subject were collected. A rest interval lasting 1 min and 40 s was respected between sessions. Trials 1–20, 21–40 and 41–60 of each session were considered for the study of early (E), intermediate (I) and late (L) adaptation. Four different target locations (T1–T4) were randomly repeated with the same probability during each session (15 reps) and sub-session (5 reps).

In order to make the experimental condition more controllable, the degrees of freedom of the trunk were limited, and possible compensatory movements prevented. The subject's trunk and head were wrapped around the back of the chair by Velcro straps and his/her wrist joint was locked by a rigid brace. Moreover, the subject could not use the fingertip to point the target but had to touch it with a rigid wand that substituted his/her forefinger. This allowed the exclusion of proprioceptive feedback contribution of the forefinger to brain processing. We reckon that these expedients did not negatively influence the final results, but rather made the task relatively new and more challenging. Each session was carried out under different environmental conditions, in which elastic bands with different elasticity coefficients were used to create external force acting on the wrist. The two ends of the band were tied on the wrist brace and on the base of the chair. Soft and hard bands [40] were employed in sessions B and C, while no band was used in session A. This was done to reiterate the adaptation request by increasing the change from a natural pointing movement.

C. *Data collection and processing*

Electroencephalography

A19-channel continuous EEG was recorded with a Sam32 (MICROMED, Mogliano Veneto, Italy) amplifier. Ag/AgCl electrodes were placed according to the standard international 10/20 system [39]. Two additional monopolar electrodes were placed over Cb1 and Cb2. A1 and A2 were used as reference with a midforehead placement of the ground electrode. Bipolar electrodes were used for the collection of eye movements (EOG) at the outer canthi and below the right eye. In addition, a bipolar derivation was acquired for the study of heart beat (electrocardiographic signal: ECG) and 3 bipolar derivations were used for recording the electromyographyc activity (EMG) of the biceps and triceps brachii and deltoid. The impedance of every electrode was below 5 kΩ. The A/D sampling rate was 1024 Hz. ECG and EMG were acquired during this study, but they are not included in the present analysis.

Electroencephalographic (EEG) data were digitally filtered offline using a band-pass finite-impulse response (FIR) filter (0.5–45 Hz) to remove noise and muscular artifacts. The filter was built as a combination of a low-pass (order 67) and a high pass FIR filter (order 6000) with linear phase. The necessity of steep transition bands and the high signal sampling rate motivated the high orders of the filters. Due to this choice, the filter initialization was not negligible, and the initial 7s of the signals were discarded. Then the signals were cleaned up from ocular artifacts through the independent component analysis (ICA) algorithm implemented in EEGlab toolbox [41], and then downsampled to 100 Hz. Cleaned EEG signal from the C3 electrode was selected for each subject, and then processed by means of an optimized version of the Empirical Mode Decomposition based on the Hilbert-Huang transformation method (EMD-HHT) [42], purposely adapted for the investigation of alpha and beta EEG rhythms. The method optimization is described in the following paragraphs of this section. Moreover, EEG power tracks obtained by means of the EMD-HHT method have been comparatively validated with traditional ERD/ERS technique proposed by Pfurtscheller et al. [10] in a previous work [43].

Cleaned EEG data were exported in MATLAB environment. Then, they were digitally filtered with a band pass FIR filter in two different frequency bands, alpha and beta, which were adaptively modified. The center frequency ω_c of the bandpass of the filter can range between 7 and 13 Hz for the alpha rhythm and between 13 and 25 Hz for the beta rhythm. After representation of a continuous wavelet decomposition of the signal in the whole frequency range, by means of EEGlab toolbox (starting from 3 Morlet wavelet cycles, and then increasing with frequency by a factor of 0.1), initialization of the center frequency ω_c was manually done, by choosing, in the alpha and on the beta ranges, respectively, the frequency with highest power value at zero time on the time-frequency wavelet representation. Then ω_c was adapted at every time step, according to the algorithm described below. The lower and higher cut-off frequencies of the adaptive passband were also adapted as ω_L= (ω_c–3) Hz and ω_H= (ω_c+3) Hz respectively.

Frequency updating algorithm

Signals were epoched into non-overlapping segments of 2.0 s duration (-0.5 to 1.5 s) relative to cue presentation and into periods of 7.5 s duration (-2.5 to 5.0 s) relative to target presentation. The baseline period was taken before the beginning of each trial (-3.5 to -2.5 s before the target onset that represent t= 0 s). The Empirical Mode Decomposition (EMD), was applied to alpha and beta EEG data as proposed in Huang et al. [42] in order to identify all extrema of the signal x(t); interpolations of minima (and of maxima) were performed, ending up with the envelopes $e_{\min}(t)$ and $e_{\max}(t)$. The mean *m(t)* was computed between the two envelopes, and the detail (also called Intrinsic Mode Function – IMF)

$$d(t) = x(t) - m(t) \tag{1}$$

was calculated. The procedure was repeated with a maximum of 5 iterations, until the resulting detail can be considered as zero-mean according to the stopping criterion. The stopping condition was that the ratio between the sum of the squared differences between two IMFs obtained at subsequent iterations and the sum of the samples obtained from the squared IMF of the previous iteration had to be lower or equal to 0.1. The window length of the EMD calculation was as short as 32 samples for the alpha rhythm and as 20 samples for beta. The Hilbert transform was then applied to the first IMF component.

The local instantaneous energy *IE*(*t*) was calculated for both alpha (α) and beta (β) bands:

$$IE_\alpha(t) = \int_{\omega_{\alpha L}}^{\omega_{\alpha H}} H_\alpha^2(\omega,t)\,d\omega \tag{2}$$

$$IE_\beta(t) = \int_{\omega_{\beta L}}^{\omega_{\beta H}} H_\beta^2(\omega,t)\,d\omega \tag{3}$$

where *H* is the "Hilbert transform", *ω* is the investigated frequency and *t* represents time.

Finally, ERS/ERD was estimated as follows, to highlight variations in the EEG frequency content with respect to the baseline period:

$$\Delta IE(t) = \Delta P_j(t) = \frac{\Delta P_j(t) - P_{Rj}}{P_{Rj}} \times 100 \tag{4}$$

Where $P_j(i)$ is the current power value in the j band, and P_{Rj} is the average power in the same band, calculated during the baseline condition [44]. If $\Delta P_j(i)$ is positive, ERS is detected at the maximum; if it is negative, ERD is found at the minimum, as represented in Fig. 2.

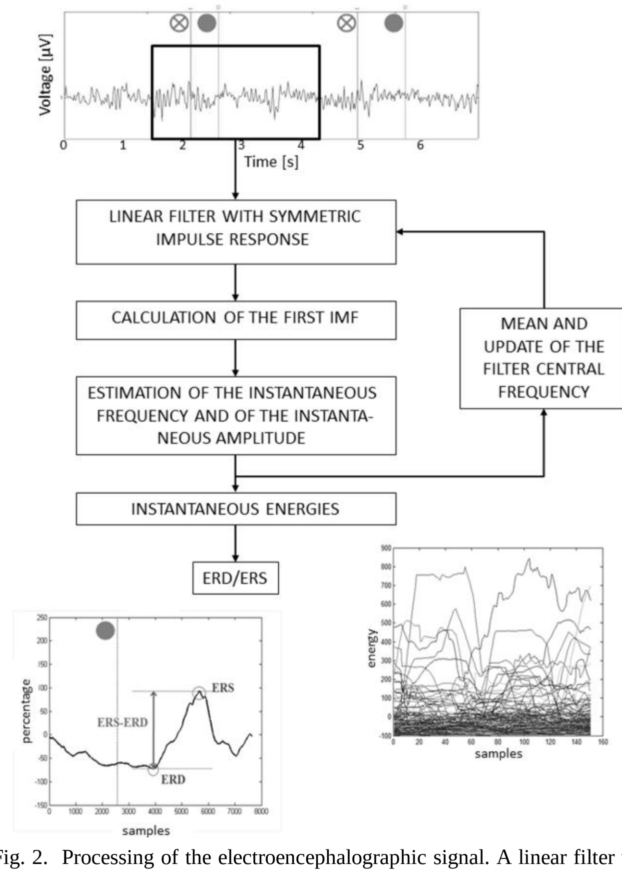

Fig. 2. Processing of the electroencephalographic signal. A linear filter with adaptive central frequency was applied to the data. EMD was run and IMFs were obtained. The first IMF was used for the estimation of the instantaneous frequency and amplitude. Last, instantaneous energies were calculated in each band, and ERD/ERS value was extracted.

In order to study the effects of task repetition, and thus adaptation, epochs were averaged separately for the different sessions (A, B, C) and sub-sessions (E, I, L). Before the averaging process, all the $\Delta P_j(i)$ epochs, related to the movement after target presentation were normalized to a total duration of 100 points independently from actual duration of the movement, with 0 being the presentation of the target stimulus and 100 the end of the movement on the target. For each of the two frequency ranges (Alpha and Beta), four independent neural parameters were extracted: (i) ERScue, reporting the maximum initial value of $\Delta P_j(i)$ following cue presentation, (ii) ERDcue, representing the minimum $\Delta P_j(i)$ following cue presentation, (iii) ERDtarget, containing the minimum $\Delta P_j(i)$ following target presentation, and (iv) ERStarget, providing the maximum value of rebound after target presentation. Then, two derived indexes were computed: (i) ERSERDcue, as the difference between ERScue and ERDcue, and (ii) ERDERStarget, as the difference between ERStarget and ERDtarget (Fig. 2, bottom panel).

Kinematics

Upper-limb movement analysis was carried out using a 6-camera optoelectronic system with passive markers (SMART-E, BTS, Milan, Italy) sampling at 50 Hz. Markers were attached to the subject's skin on selected body landmarks, according to the following marker set [38], [45]: on C7, on the sternum, acromion, elbow, ulnar and radial styloid processes, second metacarpal head and on the end of the wand that replaced the forefinger (fingertip). Additional markers were placed on the screen and on the targets during the initial calibration, to have the 3D coordinates of the target plane.3D coordinates of each marker and their derivatives were computed through SMART Analyzer software (BTS, Italy). However, only the kinematics of the fingertip has been taken into account in this study. The 3D coordinates of the marker were filtered using a low-pass second order zero-phase Butterworth filter (cut-off frequency = 10 Hz) [38]. Each movement was then divided into three main phases [38]: *going* phase (Δt_{go}, i.e. between movement onset and the time-point at which the finger-target distance drops below a distance threshold); *adjusting* phase (Δt_{adj}, i.e. between the end of going phase and the time-point at which the finger-target distance drops below a distance threshold); and *returning* phase (Δt_{ret}, i.e. between the end of adjusting phase and the return to the initial position).

Selected kinematic parameters were identified and calculated for each participant's trial [38], [45]: GMD (Going Movement Duration), time to target; TMD (Total Movement Duration), total trial time; PV (Peak of Velocity): maximum fingertip velocity during Δt_{go}; IC (Index of Curvature), ratio between the length of the 3D fingertip trajectory and the linear distance between its initial position and the final pointing position, representative of the movement smoothness during the ongoing phase [46]; dimensionless Average Jerk (AJ) [47,48]; $IQR_{avg\text{-}x|y|z}$ (Average Inter-Quartile Range along x, y or z direction), average spread of the bunch of trajectories that the fingertip draws during the subsequent repetition of the task [29].

D. Statistical Analysis

The effect of adaptation and/or environmental condition (independent variables) over a set of neural and kinematic variables (each one representing a dependent variable) was assessed through a 2-way repeated measure ANOVA. Sphericity of datasets was verified by applying Mauchly's test. Within-subjects effects were considered significant for $p<0.05$ and effect sizes (η^2) and observed power (OP) were also reported. Main effects were analysed by using pairwise comparisons with Bonferroni correction and assessed at $p<0.05$ significance level. Correlations between neural and kinematic variables, between kinematic variables and the outcome variable and between neural variables and the outcome variable were studied through Pearson correlation coefficients ($p<0.05$).

III. Results

All participants included in the study were able to complete the protocol without reporting fatigue or discomfort. An average of 30 minutes were necessary to prepare the subject and for setting up the equipment, while the execution of the 180 pointing tasks took about 40 minutes.

A. Behavioural results

Four possible errors, whose occurrence has been reported in Table 1, were observed during the execution of the pointing task: *omission*, i.e. movement not executed; *anticipation*, i.e. movement started before the appearance of the target; *postponement*, i.e. screen touched after the time limit; and *invalid*, i.e. movement dramatically different from the assigned task. No other type of error was observed.

Outcome scores (Table 2) put into evidence a significant main effect for phases ($p=0.001$, $\eta^2=0.360$, $OP=0.947$), with an increase in scores passing from the early to the intermediate ($p=0.035$) or late ($p=0.004$) phase. No significant changes were induced by the use of different resistance, or by interaction between the phase and condition factor.

TABLE I
Pointing Errors

Error Type	A			B			C		
	E	I	L	E	I	L	E	I	L
omission	0.063 (0.250)	0.063 (0.250)	0.188 (0.403)	0.313 (0.602)	0.000 (0.000)	0.063 (0.250)	0.125 (0.500)	0.000 (0.000)	0.063 (0.250)
anticipation	0.563 (0.727)	0.313 (0.602)	0.188 (0.403)	0.125 (0.500)	0.375 (0.500)	0.438 (0.814)	0.125 (0.342)	0.188 (0.403)	0.313 (0.793)
postponement	0.938 (1.289)	0.125 (0.342)	0.438 (0.814)	0.688 (1.401)	0.500 (1.265)	0.250 (0.577)	0.688 (1.250)	0.188 (0.403)	0.188 (0.403)
invalid	0.750 (0.931)	0.375 (0.619)	0.125 (0.342)	0.313 (0.602)	0.313 (0.602)	0.188 (0.403)	0.313 (0.602)	0.438 (0.727)	0.250 (0.447)
TOT	2.313 (1.722)	0.875 (0.992)	0.938 (1.197)	1.438 (1.368)	1.188 (1.667)	0.938 (1.029)	1.250 (1.639)	0.813 (1.074)	0.813 (0.882)

Errors committed for each condition (A, B and C) and for the three adaptation phases (E: early, I: intermediate; L: late). Data are expressed as mean values (standard deviation). TOT= any type of error (omission, anticipation, postponement and invalid).

TABLE II
PARAMETERS

	A			B			C		
	E	I	L	E	I	L	E	I	L
Behavioral parameter									
SCORE (%) *,§	0.55 (0.10)	0.59 (0.09)	0.59 (0.08)	0.57 (0.08)	0.59 (0.08)	0.62 (0.08)	0.56 (0.06)	0.58 (0.07)	0.58 (0.06)
Neural parameters									
ERDcueAlpha (%)	-22.54 (9.97)	-21.08 (16.52)	-24.07 (11.46)	-19.28 (16.47)	-21.15 (17.41)	-24.48 (20.66)	-23.97 (18.46)	-19.93 (15.65)	-24.25 (20.83)
ERSERDcueAlpha (%)	26.48 (13.15)	25.29 (19.19)	25.51 (18.81)	22.51 (17.95)	27.08 (12.78)	28.47 (21.10)	33.15 (19.97)	31.97 (31.23)	31.32 (21.84)
ERDcueBeta (%) ʔ	-23.92 (14.00)	-21.38 (19.92)	-20.42 (14.14)	-18.91 (18.07)	-15.69 (12.72)	-15.54 (14.89)	-24.93 (18.49)	-24.74 (14.96)	-29.70 (17.13)
ERSERDcueBeta (%) ʔ	26.91 (16.46)	21.63 (20.18)	22.93 (21.13)	22.95 (16.73)	18.11 (14.38)	20.88 (15.58)	33.27 (21.36)	28.53 (16.04)	34.34 (24.76)
ERDtargetBeta (%)	-45.98 (21.33)	-46.34 (23.23)	-42.31 (22.15)	-52.40 (23.93)	-47.89 (21.42)	-47.63 (24.96)	-53.00 (25.31)	-46.58 (26.48)	-48.35 (33.41)
ERStargetBeta (%) §,‡,◊	40.56 (47.45)	41.33 (38.70)	63.79 (48.82)	67.07 (61.97)	86.84 (63.05)	86.03 (69.15)	60.95 (40.38)	97.07 (80.76)	96.86 (62.51)
ERDERStargetBeta (%) ◊	86.54 (48.82)	87.66 (30.47)	106.10 (53.01)	119.47 (65.56)	134.73 (66.30)	133.66 (73.45)	113.94 (41.92)	143.65 (88.70)	145.21 (75.72)
Kinematic parameters									
GMD (s)	0.95 (0.10)	0.93 (0.09)	0.92 (0.11)	0.96 (0.11)	0.95 (0.12)	0.92 (0.11)	0.99 (0.13)	0.95 (0.12)	0.99 (0.10)
TMD (s)	2.70 (0.21)	2.59 (0.21)	2.61 (0.22)	2.66 (0.34)	2.67 (0.26)	2.65 (0.25)	2.75 (0.30)	2.67 (0.24)	2.69 (0.24)
PV (m/s)	0.85 (0.12)	0.86 (0.11)	0.86 (0.11)	0.87 (0.11)	0.86 (0.11)	0.84 (0.11)	0.81 (0.13)	0.84 (0.11)	0.82 (0.09)
IC	1.08 (0.14)	1.12 (0.21)	1.04 (0.02)	1.12 (0.17)	1.09 (0.12)	1.14 (0.18)	1.14 (0.12)	1.14 (0.09)	1.14 (0.13)
AJ §	59.86 (14.22)	66.08 (21.20	68.12 (20.22)	57.35 (12.43)	66.25 (20.34)	65.22 (15.76)	58.72 (15.68)	61.71 (20.64)	71.93 (18.40)
IQR_{avg-x} (m)	0.10 (0.04)	0.11 (0.05)	0.10 (0.04)	0.10 (0.04)	0.10 (0.04)	0.09 (0.04)	0.09 (0.04)	0.09 (0.04)	0.09 (0.04)
IQR_{avg-y} (m) §,†	0.19 (0.04)	0.18 (0.04)	0.17 (0.05)	0.17 (0.04)	0.17 (0.04)	0.16 (0.04)	0.17 (0.04)	0.16 (0.04)	0.15 (0.04)
IQR_{avg-z} (m) §	0.20 (0.02)	0.19 (0.02)	0.19 (0.03)	0.21 (0.02)	0.19 (0.02)	0.19 (0.03)	0.20 (0.02)	0.20 (0.02)	0.19 (0.02)

Parameters of the study group for each condition (A, B and C) and for the three phases (early, intermediate and late). Data are expressed as mean values (standard deviation).

Main effects for phase: *= p< 0.05, if compared E vs. I; §= p< 0.05, if compared E vs. L; †= p< 0.05, if compared I vs. L.

Main effects for condition: ◊= p< 0.05, if compared A vs. B; ‡= p<0.05; if compared A vs. C; ʔ = p< 0.05, if compared B vs. C

Abbreviations for electroencephalographic parameters: Event Related Synchronization (ERS), Event Related Desynchronization (ERD), difference between Event Related Synchronization and Desynchronization peaks (ERSERD or ERDERS, according to time progression). Values refers to the brain response aligned either to cue or target presentation.

Abbreviation for kinematic parameters: GMD: Going Movement Duration; TMD: Total Movement Duration; PV: Peak of velocity; IC: Index of Curvature; AJ: Average Jerk; IQR: Average Inter-Quartile Range.

A. Electroencephalography

No interaction effect between phase and condition factors was found for any of the neural parameters (Table 2). Two variables related to motor execution showed significant changes related to both phase and condition factors: ERStargetBeta (p=0.006, η^2=0.291, OP=0.856 for phase, p=0.002, η^2=0.341, OP=0.929) and ERDERStargetBeta (p=0.020, η^2=0.228, OP=0.719 for phase, p=0.001, η^2=0.352, OP=0.940 for condition). ERStargetBeta increased from the early to the intermediate (p=0.070) or late (p=0.033) phase, and from the condition without resistance to the ones with the alteration given by the elastic band (p=0.013 from A to B, p=0.007 from A to C). ERDERStargetBeta also increased from E to I-L and from A to B-C, with a trend that was significant for the A-B (p=0.005) and A-C coupling (p=0.007), and close to relevance for the E-L coupling (p=0.081). No significant differences were found between either the intermediate and late phase or the soft and hard resistance conditions.

Two different measures, ERDcueBeta and ERSERDcueBeta, manifested main effects with respect to the testing conditions. ERDcueBeta (p=0.008, η^2=0.273, OP=0.822) decreased from B to C (p=0.003). ERSERDcueBeta (p=0.023, η^2=0.223, OP=0.705) had an opposite behaviour and increased passing from a soft to a hard resistance (p=0.003).

B. Kinematics

No interaction effect between phase and condition factors was found for any of the kinematic parameters (Table 2). The duration of the movement did not change significantly in dependence of either phase or condition. IQR_{avg-y} ($p=0.000$, $\eta^2=0.437$, $OP=0.989$), IQR_{avg-z} ($p=0.001$, $\eta^2=0.360$, $OP=0.947$) and AJ ($p=0.010$, $\eta^2=0.265$, $OP=0.805$) reported a significant main effect for phase, with both the IQR indexes significantly decreasing from E to I and from I to L phase, and AJ significantly increasing from E to L. IQR_{avg-x} showed sensitivity to the change of condition ($p=0.018$, $\eta^2=0.234$, $OP=0.734$), getting lower from A to C, but with a p just above significance level ($p=0.080$).

C. Neuro-kinematic correlations

Correlations between kinematic and neural variables were significant ($p<0.05$) in 8 pairs of parameters: GMD correlated with all the four neural parameters listed (Table 3, second line), AJ showed correlation with ERSERDcueBeta, and IQR along the y and z axes showed some correlations with neural parameters as well (Table 3, last two lines). Correlations with the outcome score were significant for only 4 couplings (Table 3, first line and last column). The absolute value of the Pearson correlation coefficient was never greater than 0.26.

TABLE III
CORRELATIONS

	ERD cueBeta	ERSERD cueBeta	ERDERS targetBeta	ERS targetBeta	SCORE
SCORE	0.09	-0.08	0.12	0.19*	-
GMD	-0.19*	0.26*	-0.16*	-0.17*	-0.19*
TMD	0.06	0.02	0.02	-0.09	0.10
IC	-0.04	0.05	0.16	-0.01	-0.04
AJ	-0.13	0.21*	0.13	0.09	-0.03
IQRx	-0.05	-0.02	-0.08	-0.08	0.17*
IQRy	-0.05	0.03	-0.20*	-0.17*	0.12
IQRz	0.04	-0.17*	-0.01	0.05	0.19*

Pearson correlation coefficients rho for different measures (behavioral, neural and kinematic parameters). *= $p< 0.05$. For abbreviations see Table 2.

IV. Discussion

The adaptive processes underlying motor improvement have increasingly caught the interest of the research community in the last decade. Plenty of studies have been conducted with the aim of investigating either the role of the brain cortex or the modification of limbs kinematics in motor adaptation [6], [12]-[13]-[14]. Nevertheless, the correspondence between the neurophysiological and biomechanical modifications induced by adaptation is still largely unknown.

In this paper, we proposed a protocol for the integrated study of both neural and kinematic correlates underlying short-term motor adaptation, as well as their translation in terms of motor performance. Our interest was to highlight possible relations between the multiple determinants of motion, with the purpose of defining biomarkers for the detection and, possibly, quantify motor adaptation.

The protocol has proven to be effective: adaptation resulted to be statistically significant, with modification of several indexes across phases. Importantly, the duration of the protocol was well tolerated by all the participants enrolled in the study. None of the subjects reported nuisance and/or pain. The protocol allowed recording of a vast and valuable amount of data, a part of which has been examined and reported in the present paper.

As concerns the behavioral results, the analysis of errors showed the presence of significant main effect for phases. Specifically, an increase in scores emerged, passing from the early to the intermediate or late phase of the test. These results were representative of a progressive improvement in touching the centre of the target (i.e. getting the maximum score), and therefore in the accomplishment of the assigned task. Indeed, error signals play an important role to help the motor system smoothly correct movements; in so doing, adjustments in movement execution take over, thus allowing adaptation [49].

Modifications in brain response were found at both planning and execution level during motion. During the planning of the motor scheme, the index ERDcueBeta and the related index ERSERDcueBeta revealed significant diversity between neural responses to A and B conditions, thus suggesting the presence of some specificity of planning in the beta band with respect to the environmental characteristics; this result seems to be consistent with previous studies [18]. On the other hand, the executive part of movement seems to be the one mostly involved in modifications: the two related indexes ERStargetBeta and ERDERStargetBeta revealed changes dependent on phase, as some specificity was put into evidence for the different stages of movement refinement. The same two indexes also provided specificity with respect to conditions, revealing some modification between the “no band” naïve condition and the two modified environments requiring the coping with elastic bands. All considered, values reported in Table 2 seem to point out a prevailing influence of the elastic bands, overwhelming the short-term learning induced by phases: elastic bands seem to require some additional effort, which the protocol could only smooth but not dissolve, while the increase of ERStargetBeta and ERDERStargetBeta values through the single conditions (passing from E to I and L) could possibly indicate an increasing neural effort in refining the strategy, despite the improvement in behavioral performance. This interpretation would lead to the statement that neural control did not decrease through conditions, but rather supported behavioral improvement.

As concerns upper limb kinematics, results put into evidence a significant effect mainly for phases rather than for conditions. We found that the reproducibility of trajectory on the frontal plane (i.e. the plane where the screen was placed) was higher in the intermediate and late phases; this indeed corresponded to decreasing values of the IQR_{avg-y} and IQR_{avg-z} parameters. These data showed lower deviation in this plane after the early phase, and could be representative of motor adaptation. Interestingly, this is in line with the behavioral result emerging from the analysis of errors, which highlighted improving precision with the progression of phases. The average jerk also put into evidence increasing values passing from the early to the late adaptation phase: this change is representative of a decreasing smoothness during task execution. In presence of improvements in results subjects generally showed reduced regularity in the acceleration pattern during the same pointing task; this characteristic underpinned the conclusion that their movements had increased stop-start actions [47] due to the need of achieving a better performance with the subsequent repetitions. Therefore, the improvement in scores and outcome variability appeared to be obtained through an increased control over the movement rather than through a smoother motion that according to some authors reflects a mature and less-rigid mastery of the degrees of freedom of the neuro-motor system [28-33]. This may be due to the short time available for the subject to adapt. It may be hypothesized and verified in future studies that longer training processes may allow the participant to reach both improved performance and smoother movement patterns.

Moreover, whenever external conditions are modified, behavioral errors may arise from miscalibration of internal models, i.e. they may occur because the process of transforming the goal into motor commands rely on a not optimized, internal motor scheme. Execution errors then result in a generalization of brain internal models, and subsequently a change in motor commands [50]. In the present work, we repeatedly introduced modifications of the environmental conditions by means of two elastic bands of different resistances, starting from a naïve (“no band”) condition. Although the behavioral parameter (score) did not show significant main effect for condition, the latter could be observed at the neural (ERDcueBeta, ERSERDcueBeta, ERStargetBeta, ERDERStargetBeta) level, thus indicating some compensation phenomenon.

Correlations between behavioral, neural and kinematic parameters put into evidence some significant results, even though no strong rho values were found. The Pearson correlation coefficients, when significant, were anyway lower than 0.26. As the parameters used here are not intended to disentangle the elementary encoding underlying motor planning and execution, but rather to provide a synthetic representation of how motor control is accomplished by the brain cortex and the musculo-skeletal-system, these results are not surprising. Moreover, brain signals intrinsically contain a strong background activity, which is not expected to be related to the processes of learning and/or motor control. Interestingly, correlation was found between the outcome variable (i.e. score), and a kinematic index (i.e. IQR_{avg-z}): in this case, a higher precision in touching the centre of the target (score) seems to be connected to a lower deviation in movement execution (IQR_{avg-z}).

In this work, our attention was focused on the assessment of any link existing between the primary motor cortex and the fingertip, i.e. between the most central and non-invasively recordable structure of the central nervous system and the farthest ending point of the upper limb. The results described above encourage proceeding with further analyses, which will possibly take into account other intermediate landmarks. In fact, previous literature suggests that the elbow could be the leading joint in pointing task, subordinating all the others [51]-[52]. Additional investigation should also be conducted, including parameters other than those considered in this paper. Indeed, data acquired using the proposed experimental set-up can provide many additional descriptors of motion, including both conventional measures and innovative ones (e.g. non-linear dynamics measures, phasing relations, functional data analysis, …). Based on these observations, our results may represent a starting point to understand the interaction and the relations between the cerebral activity, the transmission of central commands to the periphery, and, hence, motor performance, so to improve the knowledge of processes which are at the base of skills acquisition.

This experimental set-up could find application also for clinics. In the evaluation of patients with neuromuscular pathologies, a better knowledge of the parameters related not only to biomechanics but also to the cerebral activation and the re-organization in a new situation after a specific rehabilitative treatment could allow a global assessment of the patients in terms both of motor output and of cerebral input, giving important indications in order to improve the rehabilitative options. Despite the potentialities of the protocol we have proposed herein, we need to mention its considerable duration, which for sure will have to come across some shortening and/or simplification before entering the clinics; pediatric adaptation will also need more engaging interface.

Lastly, we would like to point out possible limitations of this study, which was conceived as a first attempt toward a more refined and thorough analysis:(I) the limited number of participants, which may have influenced the strength of statistical findings; (II) the set of the neural and movement parameters chosen, which represented a subset of the many possible measure worth of observation and which will be integrated in the next analysis (e.g. upper-limb joint angles, connectivity between EEG channels, etc.); (III) the use of elastic bands, generating non-linearities, which could be substituted by some sort of controlled force field (i.e. manipulandum, etc…), with some advantage for modeling.

V. Conclusion

In the present study we investigated possible coupling between neural and kinematic descriptors of motor adaptation in healthy individuals. The assessment was performed through a protocol requiring short-term learning, and modulating the external perturbations, during a target-based task. From our analysis we identified significant parameters (i.e. behavioral, neural and kinematic parameters) able to describe motor adaptation. The proposed protocol has to be regarded as a preliminary, experimental implementation, which will possibly come across some simplification in the future, for application in more routinary and clinical settings.

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