國立交通大學

電機與控制工程學系

碩士論文

力回饋輔助對視覺運動協調之生理訊號變化

Physiological Signals Changes of Visuo-Motor Coordination under Force Feedback Assistance

研 究 生：董睿昕
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力回饋輔助對視覺運動理訊號變化

學生：董睿昕　　　　　　　　　　指導教授：林進燈 博士

國立交通大學電機與控制工程學系

中文摘要

本研究目的是探討力回饋輔助如何影響運動過程的效能表現、肌肉活動差異以及大腦的動態變化。受測者操控搖桿並進行軌跡跟蹤任務已完成手眼協調的任務。力回饋發生於當跟蹤軌跡誤差過大時，抵制偏移的力回饋輔助會正比於軌跡的偏差大小。在實驗一中，比較力輔助和無力輔助於運動行為能力的表現差異。在實驗二中，使用雙極電極以量測肌電訊號，以及六十四個電極的腦波帽量測腦電波訊號。並以肌電訊號能量差異、事件相關頻譜擾動(ERSP)之分析頻率在時間上變化差異了解力輔助帶來的影響。三個主要的力回饋產生的影響。第一，力輔助改善了跟蹤的運動效能特別在手臂的延展與收縮這兩個動作，但是當力輔助移除後運動效能也相對變差。第二，降低屈腕橈肌和肱二頭肌的肌肉使用力量。第三，降低α(8~12 Hz)和β(13~30 Hz)頻帶的腦電圖活動。因此，在力反饋輔助下的視覺運動軌跡跟蹤任務，減輕了調控時的肌肉負擔以及加速腦波學習的形成。這項研究提供了重要的觀點，對於力反饋輔助可能是一個工具來促進和加深人們在運動學習能力。

關鍵詞: 力回饋輔助、手眼協調、肌電訊號、腦電波、事件相關頻譜擾動

Physiological Signals Changes of Visuo-Motor Coordination under Force Feedback Assistance

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Abstract

The aim of this study was to investigate how short-term motor learning of force feedback assistance affected motor performance, muscle activity and brain dynamics. Subjects performed a trajectory tracking task of visuo-motor coordination with a joystick. Force feedback generated (maximal output 9N) by the joystick was proportional and resisted to the deviation of the tracking task. In the experiment 1 (n=30), motor behavior performance were compared between assist group and non-assist group. In the experiment 2 (n=9), electromyography (EMG) was recorded from 3 channels surface bipolar electrodes, electroencephalograph (EEG) was recorded from 64 scalp electrodes. EMG power and event-related spectral perturbation (ERSP) data were compared. There were three impacts of force feedback assistance. First, motor performance improved for the motion of flexion and extension, but no significant difference after removing the force feedback assistance. Second, lower muscle activity of flexor-carpi-radialis (FCR) and biceps brachii (BB) was performed. Third, there was a significant decreasing in alpha and beta power of EEG activity. Thus, force feedback assistance of visuo-motor trajectory tracking task was associated with decrease in EMG and attenuated alpha and beta power of EEG. This study provides important new light on the force feedback assistance may be a function as a tool to promote and deepen people's motor learning capabilities.

Keyword: Visuo-motor coordination, joystick, electromyography (EMG), electroencephalograph (EEG), alpha and beta power, event-related spectral perturbation (ERSP)

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感謝我的父母親及家人在這段求學、研究過程中的支持與鼓勵，有了你們的幫助，讓我能夠專心一致、持之以恆完成本篇論文。再者，感謝林進燈老師以及交通大學腦科學研究中心，提供豐富的教學資源、有系統的訓練課程、以及完善的實驗設備；另外加上腦科學中心成員的密集相互討論與交流，以及林君玲學姐的細心帶領，使得本篇論文漸趨完備。最後，感謝成功大學蕭富仁老師在認知神經科學的專業指導，交通大學楊谷洋老師提供搖桿程式設計與開發，以及楊秉祥老師給與在運動力學的相關資訊，使得本篇論文能夠以科學方法，透過量化與具體化的方式，以了解在運動過程中整體的行為效能、肌肉與腦部活化的差異。

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1. Introduction

1.1 Background

Movement, moving, physical activity are a fundamental component of human experiment that most of us take for granted. Human being spends a good deal of each day performing essential moment skills. Driving a car, riding a bike, eating, brushing teeth are among the more routine movement activities of everyday life. Novice picked these up by trial and error. It is considerable that longer to learn and practice in the movements are more complex and challenging. Consider, for example the actions of dentists and surgeons, pilots, athletes and performing artists. Many of their movements require years of practice, often under the watchful eyes of teachers, coaches, or other types of movement practitioners. When novice add to these tasks the abundance and variety of other skilled movements, it is important to know the concepts and the principle of motor learning and performance [1].

Motor learning refers to the relatively permanent gains in the motor skill capability associated with practice or experience [2]. The motor performance need to investigate in order to observe the change of motor learning. There are some fundamental differences between the concepts of motor performance and motor learning. Motor performance is always observable and influenced by many factors (e.g. attentional focus, fatigue). Motor learning, on the other hand, is an internal process or state that reflects a person’s current capability for producing a particular movement [1]. The best way for practitioners to assess motor learning is to observe people’s motor performance, noting the changes that occur systematically with additional practice. Only after considerable practice do people sometimes reach the final stage of learning where their performance is virtually automatic [3].

Movement in everyday life is often guided by external stimuli and multi-stage learning process [4] [5] [6]. It is a complex procedure for movement. Thus motor learning has been a subject of active research for over 50 years, and yet no deep understanding of mechanisms and methods had been found. However, research evidence suggests that systematic practice can increase a person’s level of motor learning [7] [8]. As early as the late 40’s [9], it was known that feedback played an important role in motor learning. Feedback is crucial to level of performance in the study of motor skill development. Ammons [10] gibes an overview of initial research done in the 1950’s, noting that “The more specific the knowledge of performance the more rapid the improvement and the higher the level of performance,” and that “the longer the delay in giving knowledge of performance, the less effect the given information has”. Bilodeau [11] states that knowledge of result is the “strongest, most important” variable determining performance and learning.

Different types of feedback also result in different motor performance when motor learning. However, in order to know what kinds of feedback can helpful for novice the character of feedback should be known. The effects of feedback can be classified as either “intrinsic” or “extrinsic” [12]. Intrinsic feedback relates to a person’s own sensory-perceptual information, and it helps to formulate a persons’ internal representation of the movement goal [13] [14]. There are three primary intrinsic communication channels for the novice to learn new movement or skills: auditory, visual, and tactile [15] [16] [17]. In general, novices can’t learn new skill with only one channel, and they will continually refine their motor skills achieving better and more consistent performance through multiple feedback. If novices practice incorrect motions during their motor learning, this practice could actually deteriorate users' skills, and cause injuries during the learning process.

Extrinsic feedback which usually comes from an outside source, is an alternative to intrinsic feedback. It has been categorized into either “knowledge of results” (KR) or “knowledge of performance” (KP) [18] [19] [20]. Magill [21] described that the KR is “externally presented information about the outcome of performing a skill”. The real-time KR feedback about one’s performance is the most important factor in learning new motor skills [22] [11] [23] [24] [25]. Feedback is very important to make correct and precise motions properly as early as possible in expert training of novices because the performance is seriously disrupted by lags of feedback of even [26].The time at which feedback is given is also extremely influential in human performance. Quickened feedback greatly enhances behavior and motor skill learning [27]. Conklin [28] also states that performance is seriously disrupted or made impossible by lags of less than 1.0 seconds. The best form of feedback is immediate instantaneous feedback, which allows the brain to connect synchronous actions to desired performances. The effective extrinsic feedback is based on three primary intrinsic communication channels. Thus KR feedback can take many different forms in motor learning applications, including verbal communication (i.e. knowledge of results) and visual and auditory signals [29] [30] [31] [32]. Although different in sensory modality, these types of feedback are completely indirect, meaning that the information they provide must be translated from a sensory coordinate system to the kinematic/proprioceptive coordinate system.

Millar and Al-Attar [33] found that performance on motor learning task could be improved when participants were provided with an external tactile reference frame during learning. The external cue aided tactile representation of spatial layout in a similar manner to external cues affecting the representation of spatial layout in visual memory (e.g. [34]), this phenomenon allowed for a more object-based representation of the tactile scene [35]. Lieberman and Breazeal [26] developed a system called *wearable vibrotactile feedback suit* to improve human’s motor learning, which could perform more rapid motor rehabilitation and postural retraining to combat repetitive strain injuries. In those studies, the external tactile reference frame is regard as the augmentative somatosensory channel, which utilized its kinesthetic, proprioceptive, and labyrinthine elements to give the users a greatly added view of their behavior. The external tactile reference frame could supplement the visual or auditory feedback, due to the tactile feedback presenting the most direct form of motor information.

It is well-known fact that motor performance skills rely on cognition and optimal muscle activation [36]. Electrophysiological signals can support us to study the effects of motor learning, such as electromyography (EMG) and electroencephalogram (EEG). EMG is used to study neuromuscular function, including identification of which muscles develop tension throughout a movement and which movements elicit more or less tension from a particular muscle or muscle group [37]. It is also used clinically to assess nerve conduction velocities and muscle response in conjunction with the diagnosis and tracking of pathological conditions of the neuromuscular system. Scientists also employ EMG techniques to study the ways in which individual motor unit respond to central nervous system command. A number of investigations has been shown that during sport or musical performance highly-skilled performer have a decrease EMG activity in comparison with non-skilled performer [38] [39].

A growing body of research has examined acute exercise effects on cognition, with results failing to provide consensus regarding the nature of this relationship. The EEG might be suited to the task of monitoring the changes in brain-state that occur when an individual performing a task comes to discover and adopt an effective strategy and to develop appropriate skills. Spectral features of the EEG in the theta and alpha and beta bands are sensitive to variations in attention and cognitive demands [40]. This suggests that EEG indices would be sensitive to practice-related changes in mental state and overall cognitive resource requirements. Furthermore, the topographical distribution of task-related modulation of the EEG might provide a gross measure of the region of cortex activated by task-specific neuro-cognitive strategies.

The evaluation of event-related brain spectral perturbation has provided additional insight into the underlying mechanisms involved in cognitive function beyond that of behavior measures. Preliminary information was obtained on how task practice affects spectral features of brain electrical activity [41]. Increases in performance accuracy and decrease in reaction times between the beginning and the ending portions of a testing session were accompanied by increased power in parietal alpha and frontal midline theta EEG spectral component. However, learning to perform new movements that are guided by external stimuli places high demand on the neural system. Different brain areas have to be activated to establish the cue-movement association.

1.2 Motivation

First, although literature has utilized the types of oscillation as the tactile reference frame, little information is available on the others types of external tactile reference frame. Second, most of the studies investigated the overall motor performance in a movement, however it was not sufficiently to realize the details of movement. Local aspect is discussed to tell the difference between different partitions of movement.

1.3 Aim of this thesis

In this study, subjects learned visually for tracking task and the external force feedback were activated immediately when subject deviated from the trajectory. The aim of this study was to know how was the impact of force feedback on motor performance, usage of muscle strength and brain dynamics oscillation. We designed two experiments to achievement the objectives. In a first experiment, motor behavior performance was discussed for force feedback assistance in different stages of learning. In a second experiment, in order to determine whether force feedback related physiology changes could be observed. Where the physiology indices were measured by Electromyography (EMG) and electroencephalogram (EEG), we hypothesized that the joystick with force feedback should minimize the deviation for motor skill learning in the tracking tasks during the sort-term learning. The impact of force feedback may be a function as a tool to accelerate and to deepen people’s motor learning capabilities.

2. Behavior Performance with Force Feedback Assistance

2.1 Methods

2.1.1 Subjects

There were 30 healthy right-handed volunteers (10 females and 20 males; age range: 18-27 years, mean age: 21.3 years) were paid to participate in this study. All subjects were without tremor disease detection by the Pullman Spiral System, shown in the Figure 2-1 (C). No history of neurological disease and with normal or corrected-to-normal vision from the answer of the appropriate questionnaires. For accurate evaluation of the performance, subjects were required not imbibe alcoholic or caffeinated drinks, or to participate in strenuous exercise 1 day prior to the experiments. The experiments were performed in the morning (9–12 AM) or afternoon (2–5 PM). Experiment protocol was approved by the Institutional Review Broad of Taipei Veterans General Hospital. Each subject was well instructed the procedures of experiment and none of them were aware of the experiment hypotheses. Subjects were sign the informed consent prior to the experiment.

2.1.2 Experimental Setup

All subjects had to do the spiral tests by holding the pen to along the spiral on the touch tablet (Pullman spiral acquisition and analysis: model 35910, Lafayette instruction company, Figure 2-1 (A)). Pullman Spiral Analysis is a noninvasive novel test that quantifies upper limb motor function based on kinematic and physiologic features derived from handwritten Archimedean spirals. The Pullman Spiral Analysis characterizes upper limb motor function with greater refinement, accuracy and objectivity than the clinical exam alone. Pullman Spiral Acquisition uses a digitizing tablet and writing pen connected to a computer to record position, force and time measurements from the handwritten spirals. Mathematical analyses quantify the kinematic parameters as well as other spiral features including shape, drawing speed, tightness of loops, and physiologic and essential tremor. Data were collect in the x-y plane and in the pressure axis, providing virtual "tri-axial" acquisition. Subjects were seated comfortable and instructed to start in the center of a 10 × 10 cm box on a 10 × 10 cm white paper with spiral curve. Subjects drawn both direction spiral tests, one is clockwise and the other is counter-clockwise, shown in the Figure 2-2.

The Pullman Spiral system was used to study the details of normal motor control, quantify normal and abnormal motor development, analyze movement disorders such as tremor disorders. Spiral analysis output consists of report of specific indices with visually striking graphical representations. If the subjects had the tremor disorders, then the spiral they did would look like as shown in Figure 2-1 (B). The importance of this measure lies in the detection and quantification of the largest power oscillation that would be clinically perceived as tremor. The red line represents a detection cutoff under 3 Hz. That is, the spiral analysis system assumes that these lower oscillations are caused by the drawing of the spiral loops themselves, rather than by an additional abnormality. The Figure 2-1 (C) shows the normal movement.

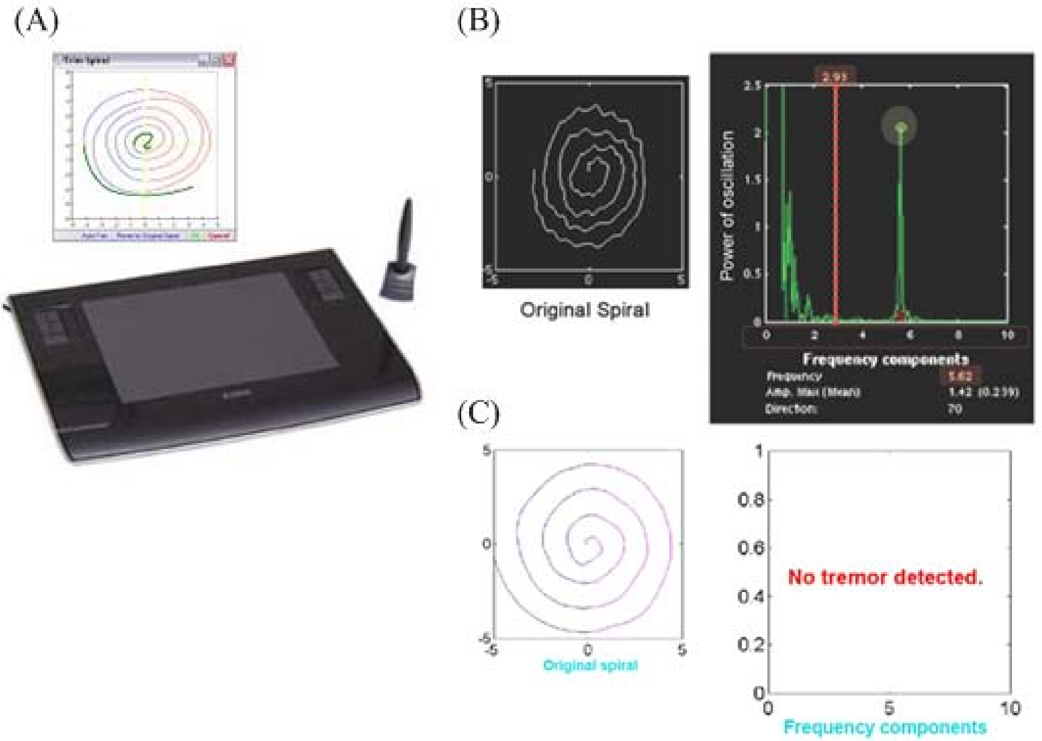


Figure 2-1: (A) Pullman spiral acquisition and analysis. (B) The example of spiral analysis for tremor movements and frequency component analysis. (C) The example of spiral analysis for normal movements

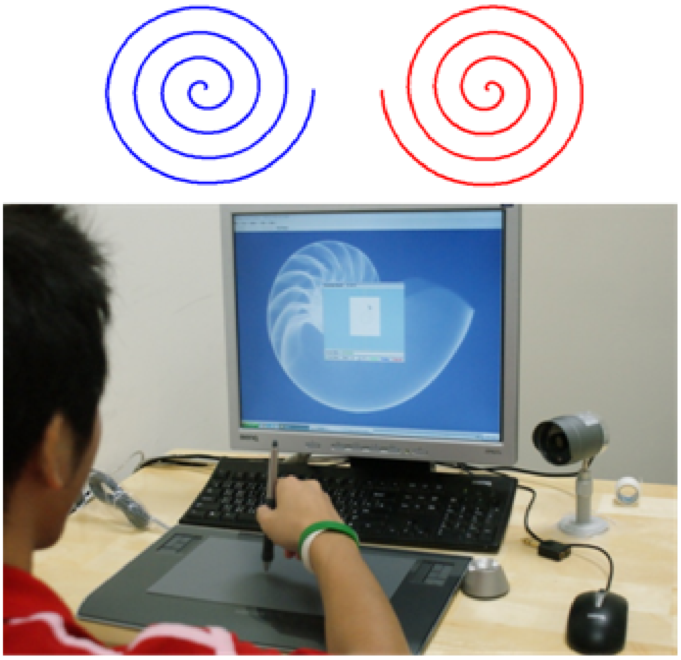


Figure 2-2: Clockwise and counter-clockwise spirals tests

The basic principle of our movement task was to increase tracking precision during the course of learning. In the motor tracking paradigm, all subjects seated in front of a 19” monitor (resolution of 1024×768 pixels) and learned to control the joystick (Immersion Impulse Stick, USA, Figure 2-4 (A)) to minimize the discrepancy between a changing foreground stimulus (the green cross with a diameter of 6 pixels controlled by the subject) and a constant speed background stimulus (the purple trajectory uncontrolled). The location of green cross applied to the location of joysticks and the purple trajectory located in the center of the black border, shown in the Figure 2-3. Experimental control was established using Windows DirectX and Immersion I-Force 2.0 Compatible in combination with MFC and OpenGL. The background of screen consisted of a square framed (660×660 pixels) by a black border with a dimension of 80 pixels presented in the centre of screen. There are four blocks in the vertical angle of the square. A shelter was utilized to avoid that subjects watched their hand and didn’t watch the screen when adjusting joystick (Figure 2-3). Furthermore, this study designed one mechanism to avoid the fatigue of the subjects’ hand (Figure 2-4 (B)).

The impulse stick is a robust joystick designed for rugged based entertainment and industrial control. In this device a local microprocessor loads and plays haptic effect. This microprocessor closes the haptic control loop, relieving your host computer of this burden. The force being played by the microprocessor can update approximately 100 times per second and directly projected to the screen, so that there was no discernible lag between the joystick and the visible cross. The Software Development Kit can be used as a rich, real-time graphical development and test tool for the device. All of the specifications of the impulse stick were showed in the Table 2-1.

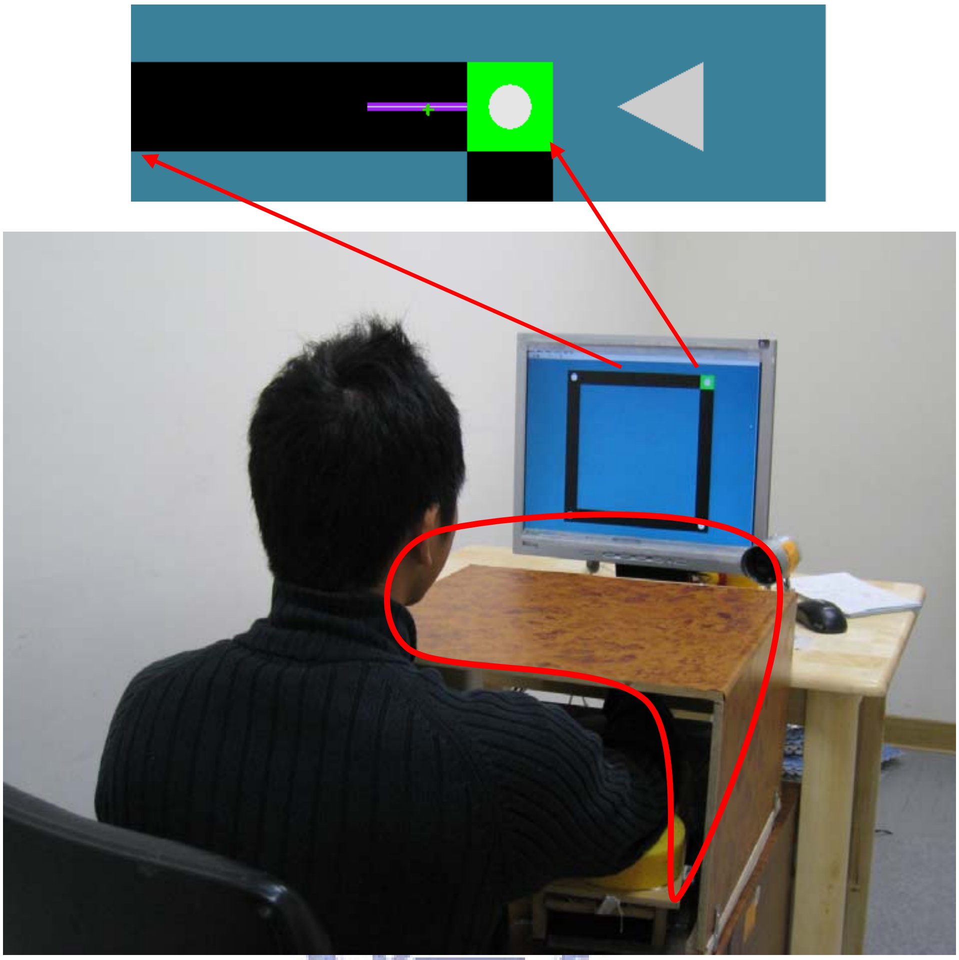


Figure 2-3: Experimental setup

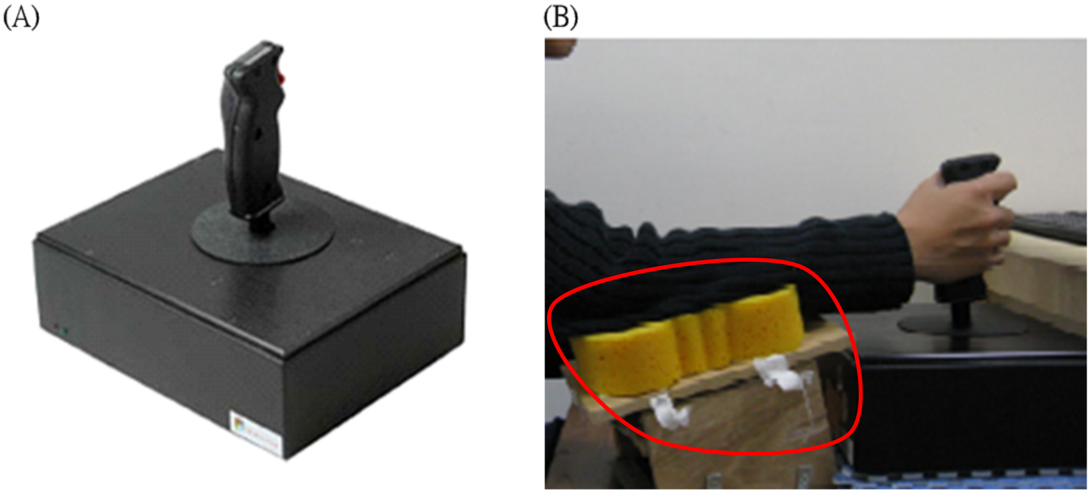


Figure 2-4: (A) Immersion Impulse Stick. (B) The mechanism to avoid the fatigue of the subjects' arm

Table 2-1: All of the specifications of the impulse stick

Specifications:	
Contruction Features:	
Heavy duty steel chassis	12”w x 9.5”d x 4” h
Belt transmission	4:1
Range of motion	40 degrees
Positional resolution	.01 degrees
Handle length	5.5”
Power	24VDC, 6.5A
Attaches 1” up from top of chassis	
Total length 6.5” from top of chassis	
Molded plastic handle with trigger and thumb buttons	
Host Compatibility:	
Communications	USB 1.1
Protocol	Windows DirectX, Immersion I-Force 2.0
Operating System	Windows XP, Windows 2000
Forces:	
Max force	14.5N (3.5lbf)
Continuous force	8.5N (1.9lbf)
Torque sensitivity	14.8 oz-in/A

This study proposed an augmented sensory force feedback theory to give subjects a real-time counterforce feedback and deepen the learning affect. The counterforce feedback acted immediately when subject deviated the trajectory more than 3 pixels in the experiment. And the counterforce was linearly increasing according to the deviation between the green cross (the movement of subjects) and the purple trajectory, shown in Figure 2-5.

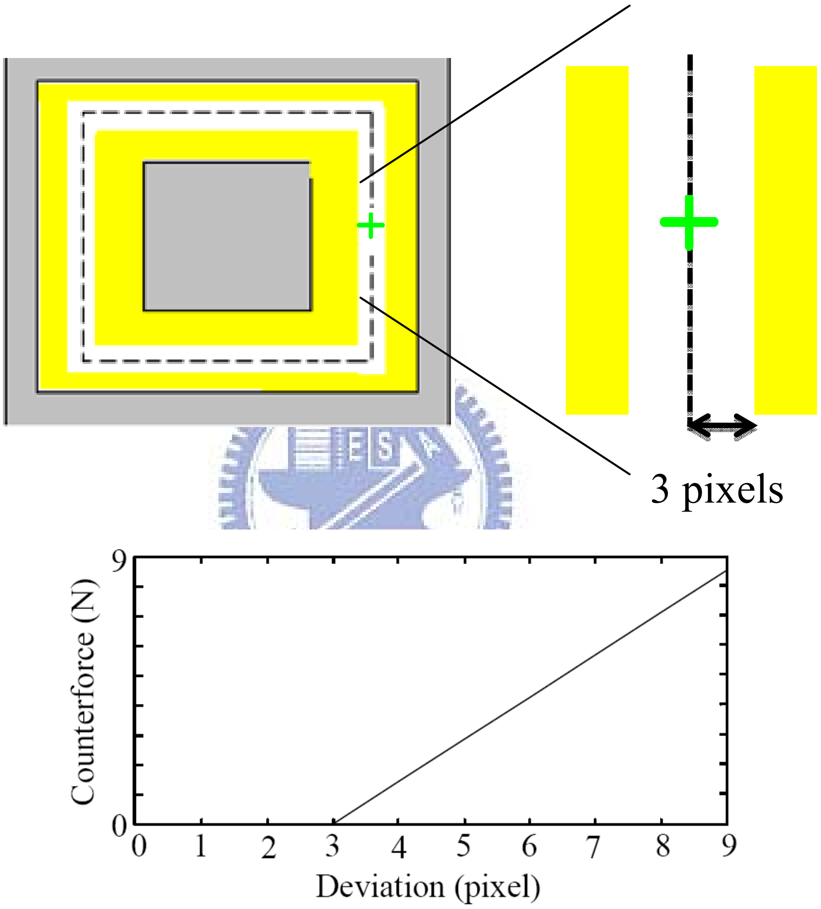


Figure 2-5: The condition of counterforce

2.1.3 Experimental Paradigm

In Figure 2-6, there were three stages in this experiment: the first, evaluation before the learning (earlier evaluation); the second, short-term learning of the trajectory tracking task; and the third, evaluation after the learning (later evaluation).

The first stage was the evaluation before the learning, which was utilized to evaluate the ability of the right hand. The third stage was the evaluation after the learning, which was the same as the first stage and utilized to evaluate whether the learning effect transform to another movement and the learning effect retain after removing the force feedback assistance.

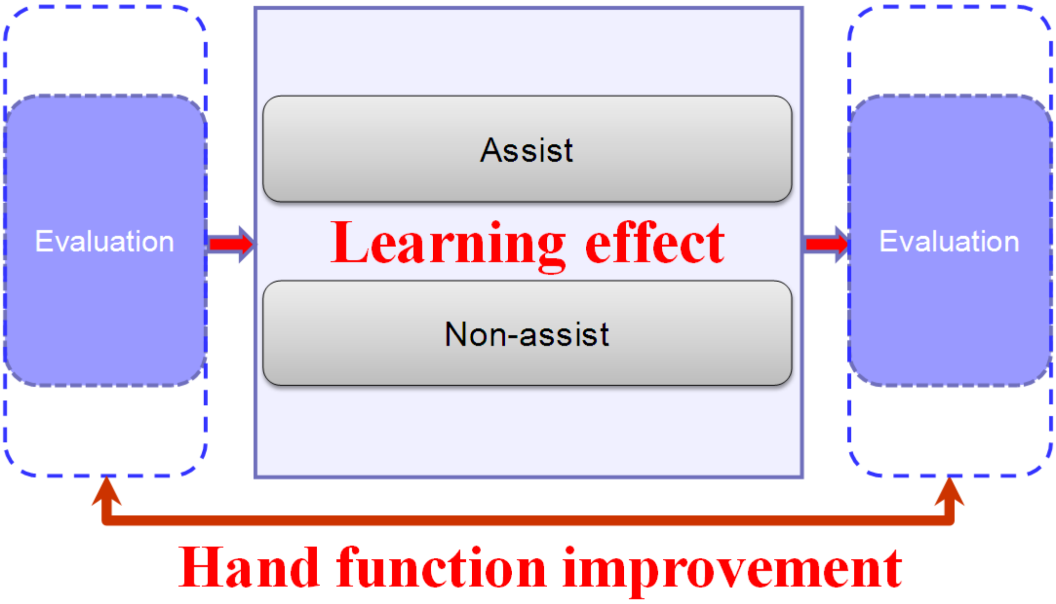


Figure 2-6: Three stages in the experiment 1.

Subjects were instructed to do tracking task in the earlier and later evaluation. Subjects had two trials to practice before the beginning of data collection in order to become familiar with the principle of the compensatory tracking task to handle the joystick. When subjects began tracking task, they were instructed to track the trajectory as precisely as possible by controlling the joystick.

There were four conditions per trial in the tracking task of evaluation. Four conditions were combined by two variables as the following. First, “direction” could be clockwise or counter-clockwise according to the direction of the triangle on the up-right side, shown in the Figure 2-7. Second, “disturbance” could be with or without disturbance force perpendicular to the trajectory direction and subjects should move back to the trajectory. The same strength of disturbance force would randomly occur once in all of four areas when tracking trajectory. There were four trials of each condition would be randomly assigned, totally sixteen trials in the evaluation.

All of the trials started with the up-right white circle and the triangle right to the up-right white circle indicated the direction of the trial, shown in the Figure 2-7. At the beginning of the black block presented for 3.2 seconds, subjects kept the green cross inside the white circle and waited. Following, the block transferred black into yellow for 0.8 seconds completing the cuing period and subjects had to ready for moving. Finally, when the yellow block became green block, subjects started to move the green cross out of the block and then the appearance of the trajectory displayed. Subjects commenced tracking the constant speed trajectory by controlling the joystick, shown in the Figure 2-8.

This trajectory located in the center of the black border and moved to the next white circle. Subjects had to instruct to continuously adjust the joystick as hard as necessary with their right hand to along the trajectory without the deviation for about 5.2 seconds. Whenever a deviation occurred, they immediately moved back to the trajectory. If the direction was counter-clockwise, then the green cross reached the up-left from the up-right. The color of block also was from black to the yellow first, and then the color of up-left is from yellow to the green. After leaving the up-left block the trajectory was presented and reached the down-left block. Thus all procedures of a trial were the same as the above in the evaluation.

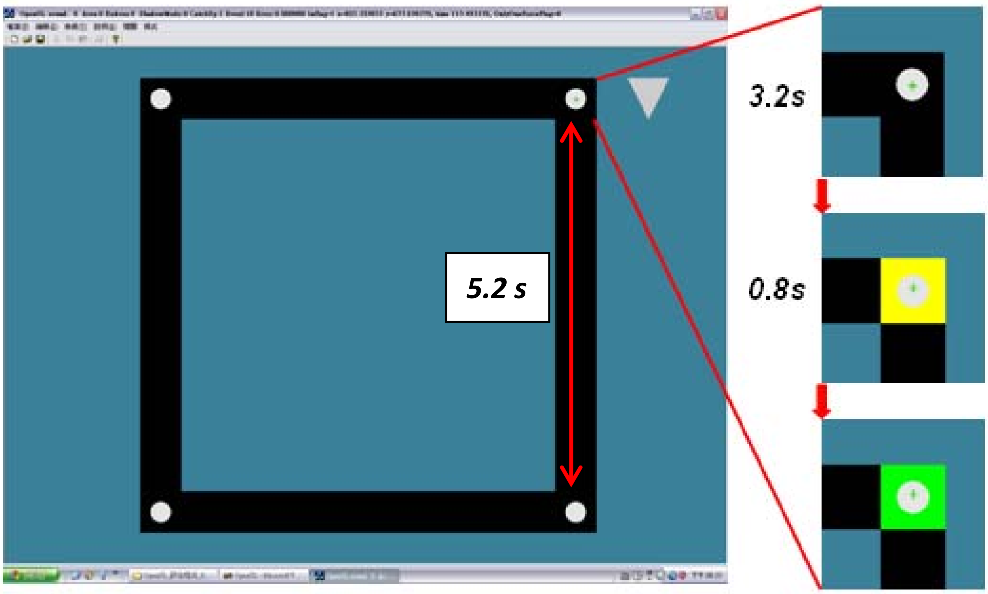


Figure 2-7: The procedure of a trial

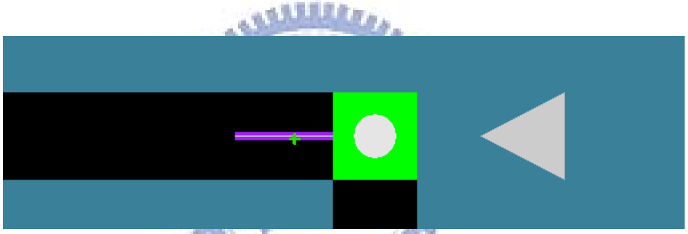


Figure 2-8: The appearance of the trajectory display

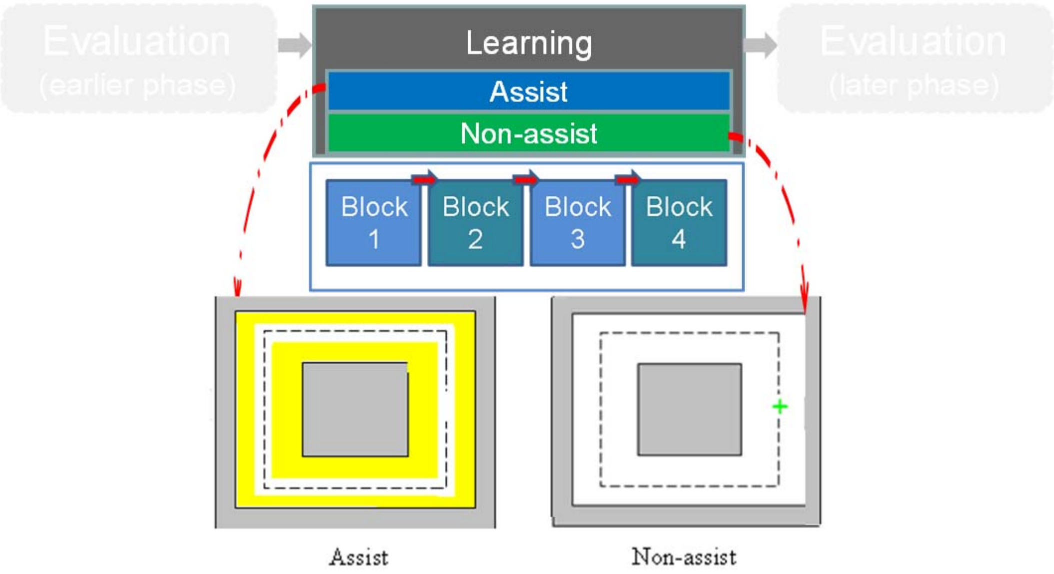


Figure 2-9: Assist group and non-assist group in the learning stage

In the second stage, two different experimental groups were investigated in the learning stage, shown in the Figure 2-9. Subjects were randomly divided into two groups and the number of females and males were equal in two groups, that is, there were 5 females and 10 males in each group. In order to understand the effects of the force feedback learning:

- *Assist group:* the assist group had the force feedback assistance to resist the excessive deviation which was made from the subject.
- *Non-assist group:* the non-assist group didn't have the assistance. Thus the non-assist group needed to correct the deviation by themselves.

In the second stage, subjects adjusted the joystick to track the trajectory only the counter-clockwise direction and without disturbance condition for about 6.2 seconds, slower than the movements (5.2 seconds) in the evaluation stage. A total of 40 trials for this experiment were performed in four blocks of each 10 trials. Subjects had to respond their physical fatigue strength for the right hand and their mental fatigue strength after the blocks. There were five scales (1-5) to represent the different fatigue strength (1 = no tired, 5= tired out). Subjects had short brakes of about 2~6 min between blocks according to fatigue strength level they responded. There was an interval of 30 min between the short-term learning and later evaluation; the whole experiment lasted approximately 1.5 hours.

2.2 Data Analyses

Because there were 4 areas (up, down, left and right sides of the square) in a trial, this study also tried to investigate if any difference effect of force feedback assistance in four areas. The movements were defined as extension, adduction, flexion and abduction for up, left, down and right sides of the four areas, respectively, shown in Figure 2-10. Thus the two aspects were discussed in the short-term learning and evaluation: *global* and *local*. Global aspect, tracking error was calculated by the mean value of the four areas, that is, the four areas had to group together in a trial. Local aspect, calculated the four areas individually.

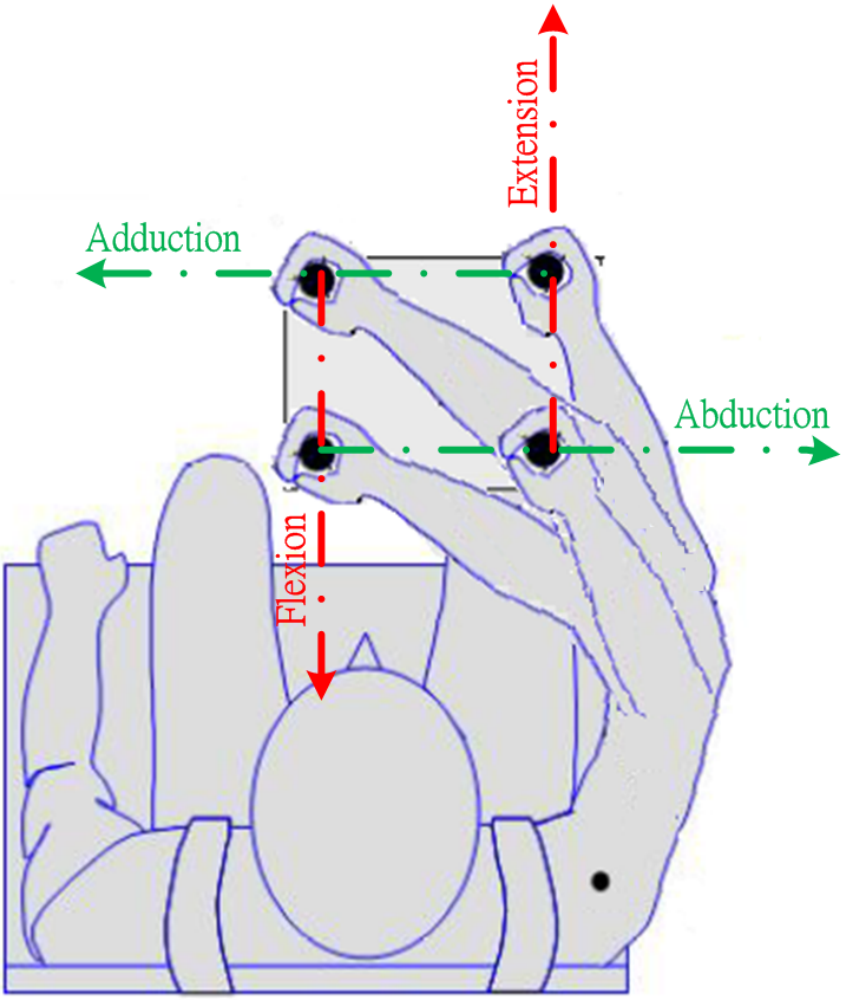


Figure 2-10: The movements were defined as extension, adduction, flexion and abduction.

This study defined two indices to evaluate the tracking performance of each movement: tracking error and variation.

Tracking Error

The joystick position (the center of green cross) and the required target position (the purple trajectory) were compared for each trial, which formed the sequence of one movement simultaneously, and calculate the mean absolute deviation in pixels for each trial, shown in the Figure 2-11.

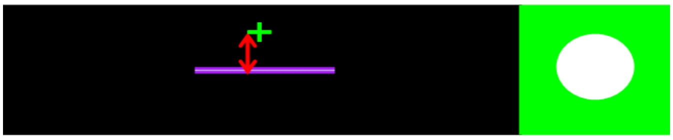


Figure 2-11: The tracking error was defined by the height of the red arrow

Variation

When the movements had a discrete deviation, the deviation was viewed as once variation and counted the number of the variation to this movement, shown in the Figure 2-12.

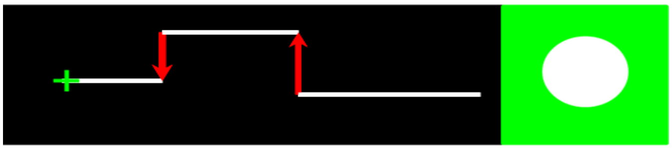


Figure 2-12: The variation was defined by the number of the red arrow

Curve fitting

In the learning stage, curve fitting was used to find the best fit to a series of tracking performance of the trials. Here two parameters were defined to investigate tracking performance: the initial performance and the learning rate. The larger number of tracking rate was the better learning effect for subjects. In order to obtain the initial tracking error and the learning rate parameters of the tracking performance, a nonlinear optimization algorithm such as *lsqcurvefit* implemented in MATLAB was used to fit the experimental data. The exponential function was used to fit the learning curve composed by the forty moments. The exponential function (1), according to the following, was used to calculate the values of the initial tracking error and the learning rate.

$$k_1 * \exp\{-k_2\} \tag{1}$$

where k_1 is the initial tracking error, k_2 is the learning rate.

Statistical analysis

All statistical analyses were conducted with SPSS 16.0 and MATLAB. To analyse the effect of force feedback assistance and task practice on performance of tacking error and variation in the learning stage, the factors of “group” (assist, non-assist) and “phase” (block1, 2, 3, 4) were be concerned, respectively. Tracking error values were statistically compared using mixed-design ANOVA by defining the “group” as between-subject factor and the “phase” as within-subject factor. The independent-samples *t* test was utilized to test the significant differences of two curve fitting parameters between the assist and the non-assist groups in the learning stage.

To compare the tracking performance (tracking error and variation) in the earlier evaluation (the first stage) with those in later evaluation (the third stage) to evaluate whether the learning effect retained after removing the resistance of force feedback by mixed-design ANOVA with the following factors: “group” (assist, non-assist), “phase” (earlier evaluation, later evaluation). The independent-samples *t* test was utilized to test the significant differences of the tracking error between the assist and the non-assist groups under the same evaluation stage of tracking test with disturbance force. The paired-samples *t* test was used to test between the earlier and the later evaluation of tracking test with disturbance force for both groups. The significant power of *p*-value was chosen by 0.05.

2.3 Experimental Results

Thirty subjects performed the three stages: *evaluation* before the learning, short-term *learning*, and *evaluation* after the learning. There are two groups in this study; 15 of the 30 subjects were the assist group and others were the non-assist group.

2.3.1 Learning Behavior Performance

Global aspect

In the Figure 2-13, the analysis of tracking performance (tracking error and variation) over all subjects revealed a significant difference between the assist and the non-assist group (ANOVA factor “group”, tracking error: $F(1,28) = 22.6$, $P < 0.001$; variation: $F(1,28) = 10.7$, $P = 0.003$). In the assist group, tracking performance was clearly better for all 4 blocks. Furthermore, a significant main effect for the factor “phase” (tracking error: $F(3,84) = 7.6$, $P < 0.001$; variation: $F(3,84) = 12.1$, $P < 0.001$) was registered. There was no significant interaction “groups × phase” for tracking error ($F(3,84) = 0.14$, $P = 0.938$), but significant interaction for variation. Non-assist group improved the tracking error and variation, assist group just improved the tracking error with learning. By visual of comparing the tracking performance of the assist group with the non-assist group a different pattern was obvious. The assist group in the first block was at a better tracking performance level compared to the non-assist group in the last block.

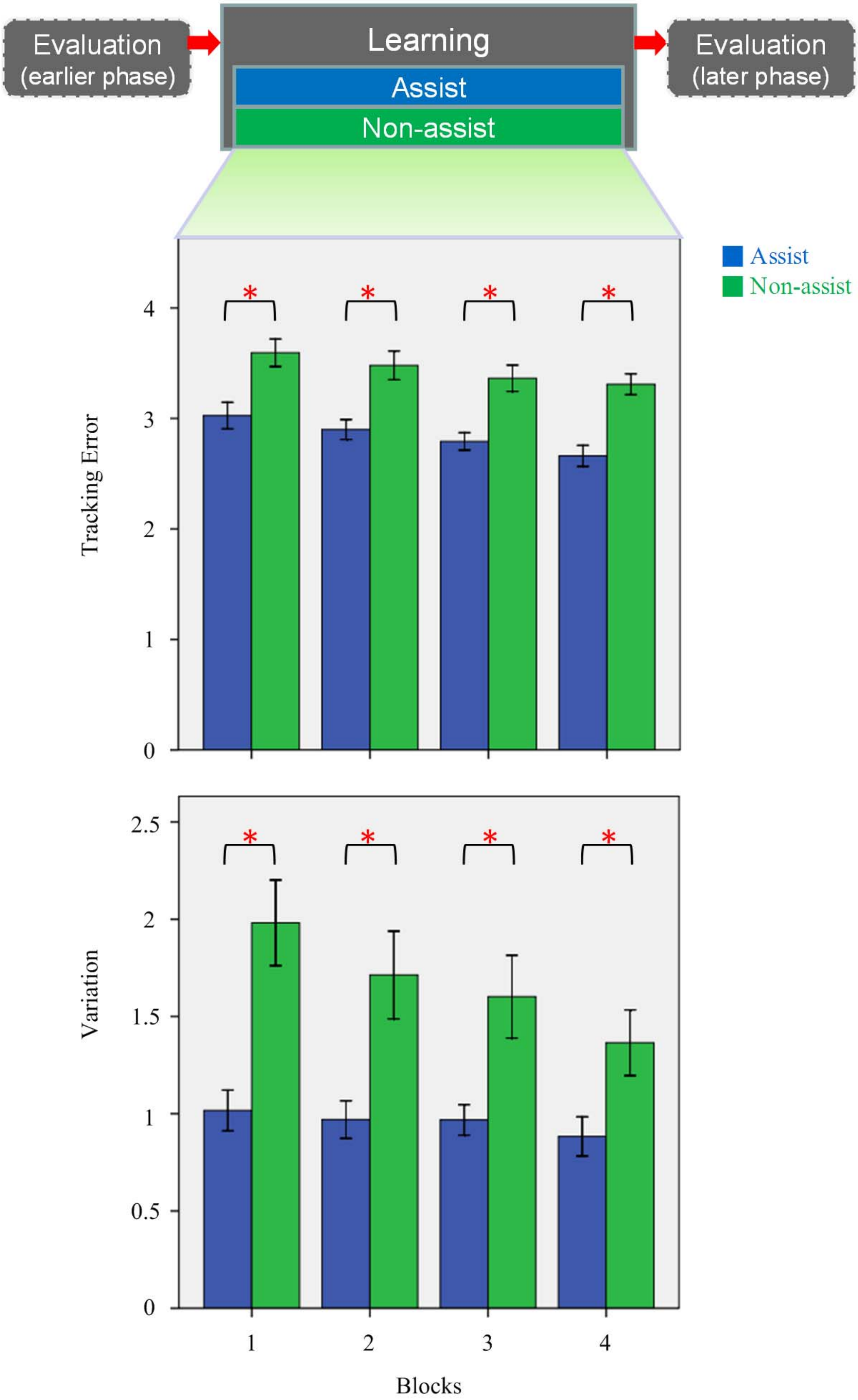


Figure 2-13: The analysis of tracking performance over all subjects in global aspect

Local aspect

Because there were four movements (adduction, flexion, abduction and extension), the effect of force feedback on the four movements was evaluated.

In the Figure 2-14 and Table 2-2, the analysis of tracking performance only significantly differed between the assist and the non-assist groups for some movements. Tracking error differed only for flexion and extension (flexion: $F(1,28) = 22,6$, $P < 0.001$, extension: $F(1,28) = 16.2$, $P < 0.001$), variation was for adduction, flexion and extension, (adduction: $F(1,28) = 9.5$, $P = 0.005$, flexion: $F(1,28) = 16.3$, $P < 0.001$, extension: $F(1,28) = 4.3$, $P < 0.047$). In the assist group, the tracking error was clearly better only for flexion (movement pulling inside) and extension (movement pushing outside), the variation was better for adduction, flexion and extension.

Tracking error had a significant main effect for the factor “phase” for abduction ($F(3,84) = 3.5$, $P = 0.018$) and extension ($F(3,84) = 4.3$, $P = 0.007$) was registered. There was no significant interaction “group × phase” for all the four movements. Both of the two groups improved the tracking error just for abduction and extension when learning. For variation, significant main effect for the factor “phase” for abduction ($F(3,84) = 5.0$, $P = 0.003$) was registered. There were significant interaction “group × phase” for abduction ($F(3,84) = 4.2$, $P = 0.008$). Only non-assist group improved the variation for abduction.

In the Figure 2-15, the tracking performance of all trials was compared between different movements for both assist group and non-assist group. The non-assist group had a significant smaller tracking error for adduction and abduction compared to flexion and extension, and smaller variation for abduction. The assist group had a significant smaller tracking error for abduction compared to adduction, flexion and extension, and smaller variation for flexion

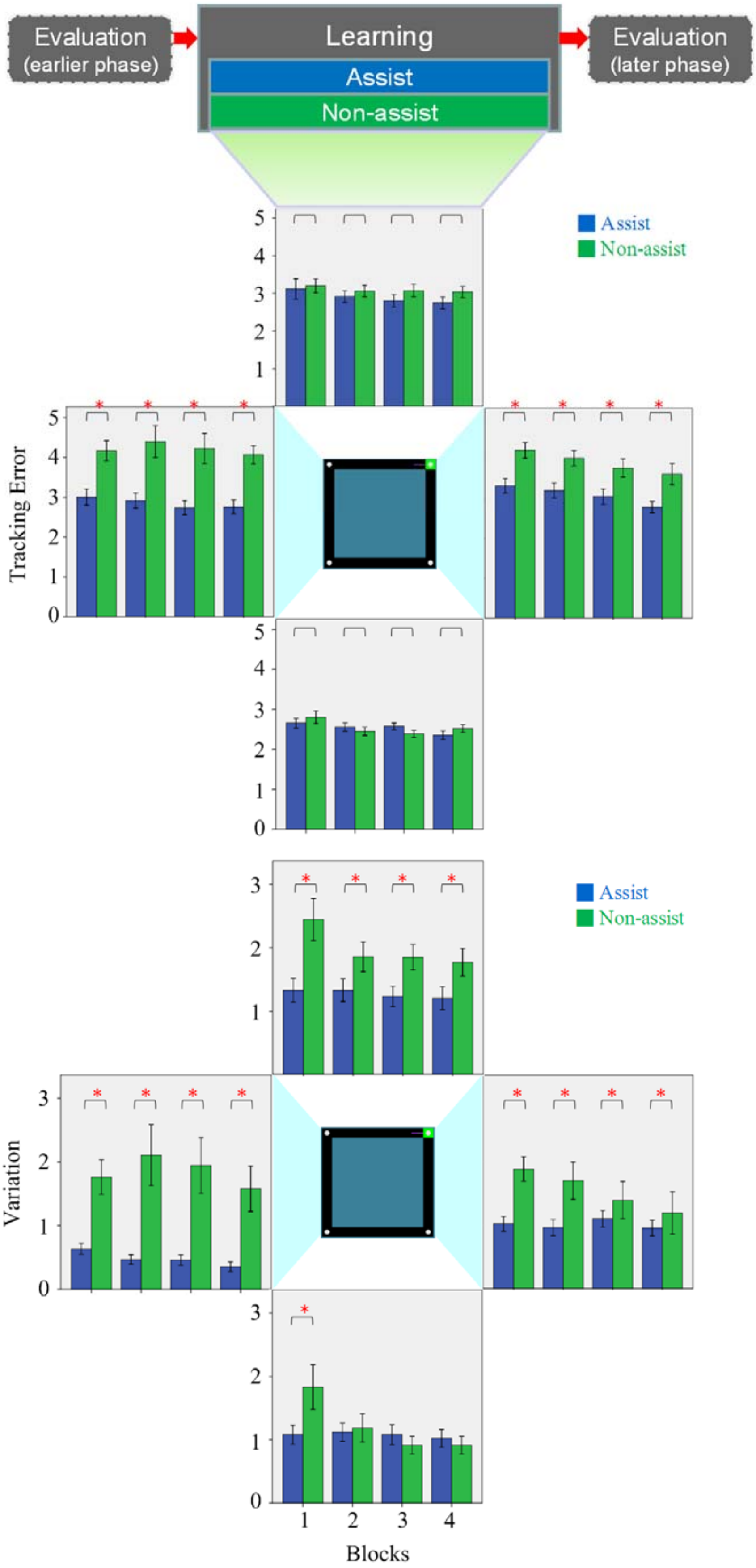


Figure 2-14: The analysis of tracking performance over all subjects in local aspect

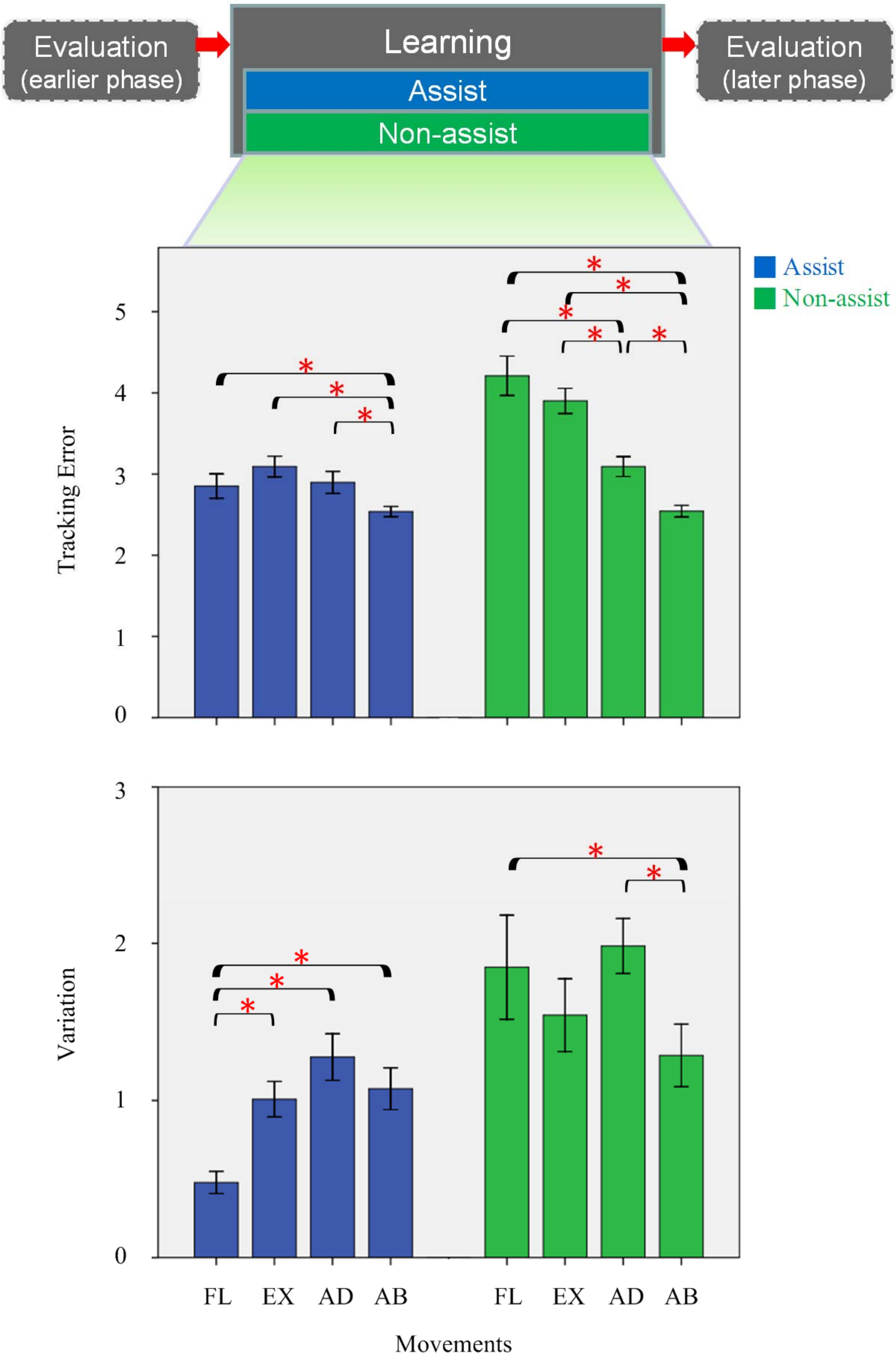


Figure 2-15: The tracking performance was compared between different movements.

Table 2-2: The statistical analysis of the tracking performance over all subjects

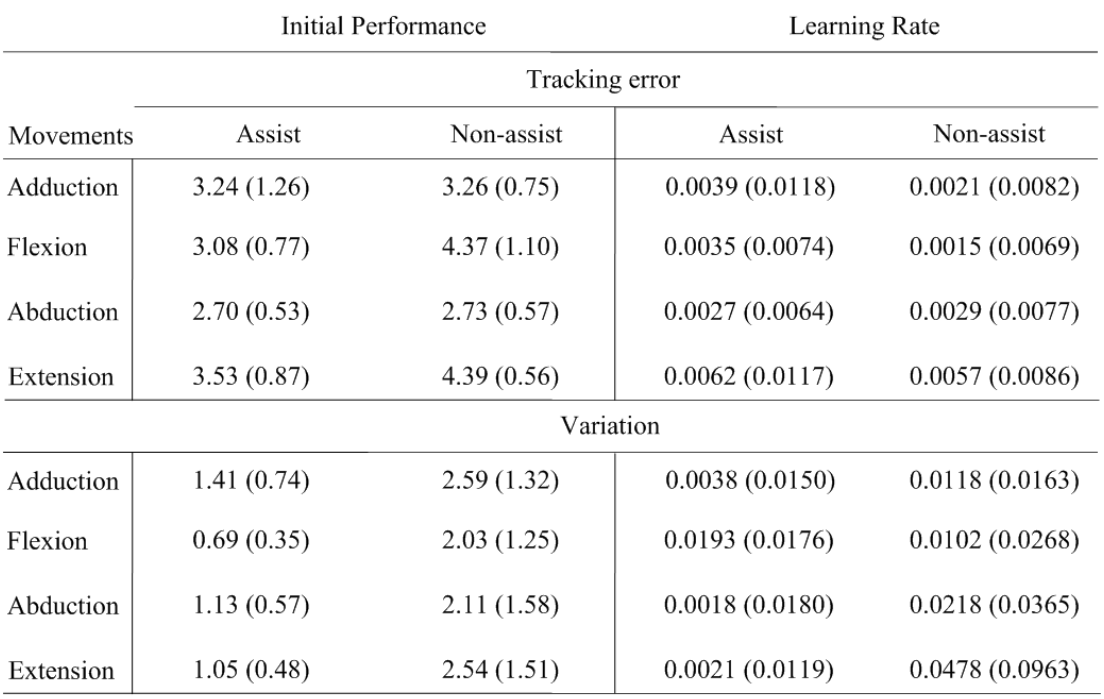
Mixed design ANOVA												
Tracking error												
Movements		*df*	*F*	*P*		*df*	*F*	*P*		*df*	*F*	*P*
Adduction	group	1	1.145	0.294	phase	3	1.329	0.267	group × phase	3	0.245	0.865
Flexion		1	22.644	0.000*		3	0.630	0.895		3	0.322	0.809
Abduction		1	0.003	0.957		3	3.523	0.018*		3	1.647	0.185
Extension		1	16.223	0.000*		3	4.298	0.007*		3	0.088	0.966
Variation												
Adduction	group	1	9.512	0.000*	phase	3	2.295	0.084	group × phase	3	1.347	0.265
Flexion		1	16.301	0.000*		3	1.200	0.315		3	0.881	0.454
Abduction		1	0.784	0.384		3	4.951	0.003*		3	4.216	0.008*
Extension		1	4.320	0.047*		3	2.492	0.066		3	2.429	0.071

Curve fitting

Because there were different effects between the four movements, we show only the results of the local aspect for tracking task in the learning stage. The curve fitting results shown on the Figure 2-16 was utilized the exponential function to fit the tracking performance cross all subjects. From this figure, there were larger gaps of tracking error between assist and non-assist groups for flexion and extension, it was the similar results mentioned before. Furthermore, curve fitting was performed to evaluate the initial performance index and learning rate index for each subject among two groups. Mean value and standard deviation of two indices are shown in the Table 2-3. The initial performance index revealed a significant difference between the assist and the non-assist group for some movements. Which were flexion and extension for tracking error (adduction, P = 0.455; flexion, P = 0.002; abduction, P = 0.804; extension, P = 0.016), adduction, flexion and extension for variation (adduction, P =

0.006; flexion, $P < 0.001$; abduction, $P = 0.135$; extension, $P < 0.001$). There was no significant difference for the learning rate index of tracking error (adduction, $P = 0.648$; flexion, $P = 0.836$; abduction, $P = 0.99$; extension, $P = 0.648$), and there was significant difference for the learning rate index of variation for flexion, abduction and extension (adduction, $P = 0.419$; flexion, $P = 0.014$; abduction, $P = 0.049$; extension, $P = 0.005$). The assist and non-assist groups had the same learning effect, but the force feedback assistance could improve the tracking performance effectively at the beginning of the task for flexion and extension. Because of the force feedback assistance, assist group had smaller initial performance index and learning rate index of variation.

Table 2-3: The analysis of curve fitting for tracking performance



	Initial Performance		Learning Rate	
	Tracking error			
Movements	Assist	Non-assist	Assist	Non-assist
Adduction	3.24 (1.26)	3.26 (0.75)	0.0039 (0.0118)	0.0021 (0.0082)
Flexion	3.08 (0.77)	4.37 (1.10)	0.0035 (0.0074)	0.0015 (0.0069)
Abduction	2.70 (0.53)	2.73 (0.57)	0.0027 (0.0064)	0.0029 (0.0077)
Extension	3.53 (0.87)	4.39 (0.56)	0.0062 (0.0117)	0.0057 (0.0086)
	Variation			
Adduction	1.41 (0.74)	2.59 (1.32)	0.0038 (0.0150)	0.0118 (0.0163)
Flexion	0.69 (0.35)	2.03 (1.25)	0.0193 (0.0176)	0.0102 (0.0268)
Abduction	1.13 (0.57)	2.11 (1.58)	0.0018 (0.0180)	0.0218 (0.0365)
Extension	1.05 (0.48)	2.54 (1.51)	0.0021 (0.0119)	0.0478 (0.0963)

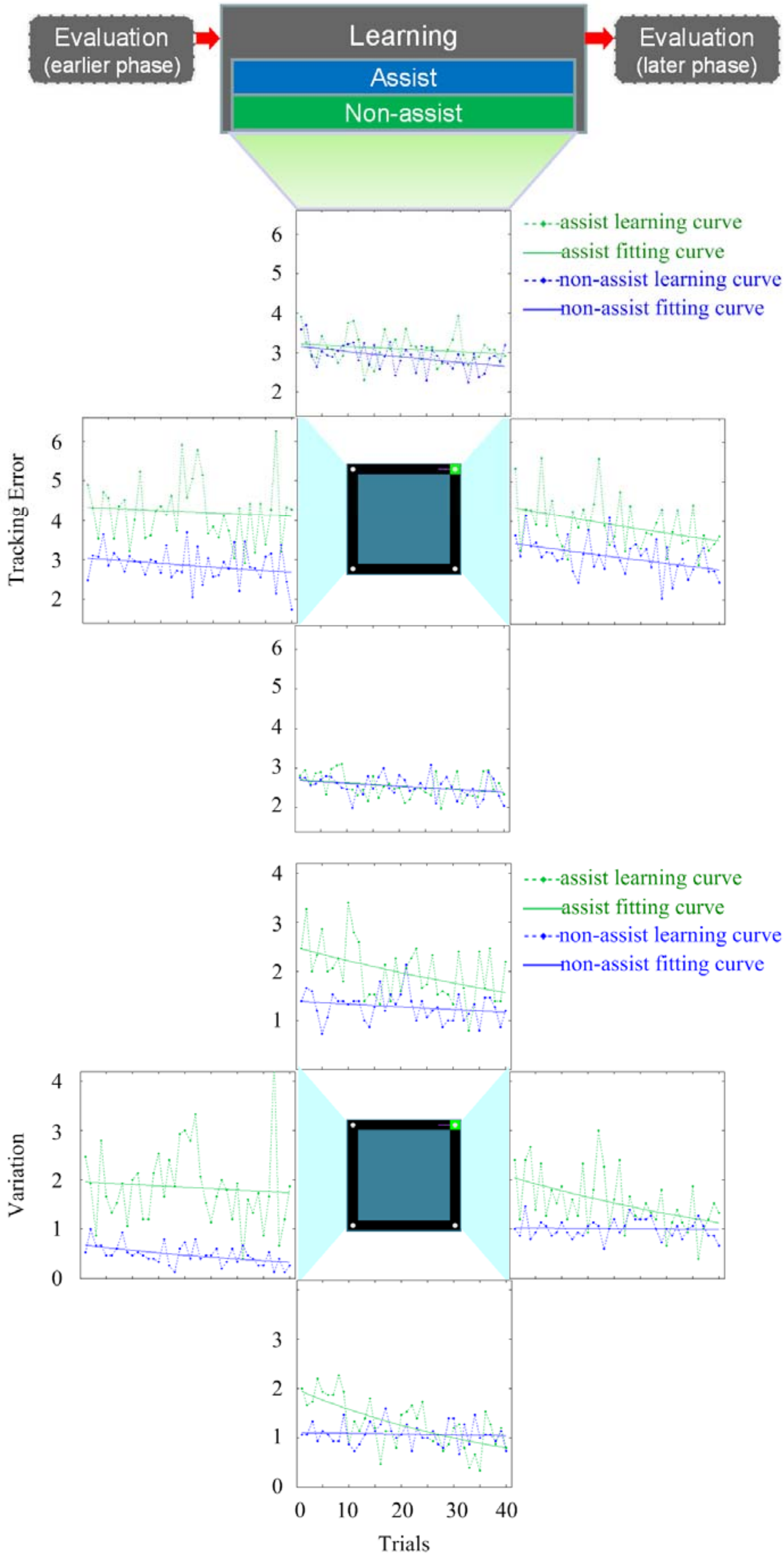


Figure 2-16: The curve fitted results of learning curve.

2.3.2 Evaluation Behavior Performance

There were four conditions in the tracking test of the evaluation. First, we showed the results of without/with disturbance force in counter-clockwise direction, this is the identical direction movement as the learning stage. Then we showed the results in clockwise direction.

Counter-clockwise direction: without disturbance force

Global aspect In the Figure 2-17, the tracking performance (tracking error and variation) didn't significantly differ between the assist and non-assist group in the evaluation (ANOVA factor "group", tracking error: $F(1,28) = 0.247$, $P = 0.623$; variation: $F(1,28) = 0.156$, $P = 0.696$). There were no significant difference of the tracking performance between two groups in the earlier evaluation, that is, we had the well-controlled groups in baseline tracing performance. In the later evaluation, the tracking performance didn't significantly differ between two groups after different learning situation. Therefore, better tracking performance would not remain after the removing of the force feedback assistance. Furthermore, a significant main effect for the factor "phase" (tracking error: $F(1,28) = 53.544$, $P < 0.001$; variation: $F(1,28) = 47.358$, $P < 0.001$) and no significant interaction "groups × phase" (tracking error: $F(1,28) = 0.548$, $P = 0.465$; variation: $F(1,28) = 2.313$, $P = 0.139$) was registered. Both of the two groups improved the tracking performance between the earlier and later learning. After learning the two groups both improve their tracking performance.

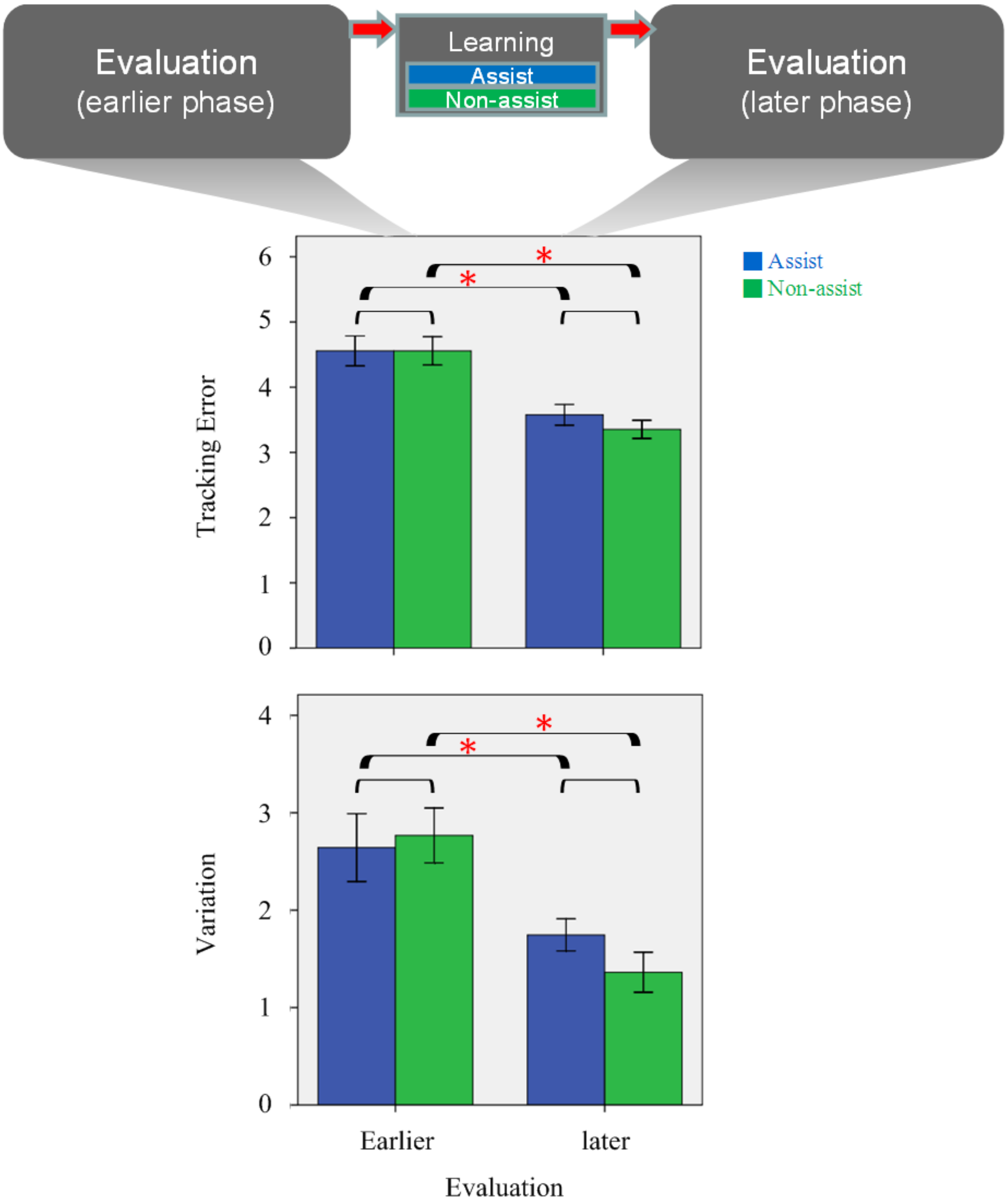


Figure 2-17: The analysis of tracking performance in the evaluation stage.

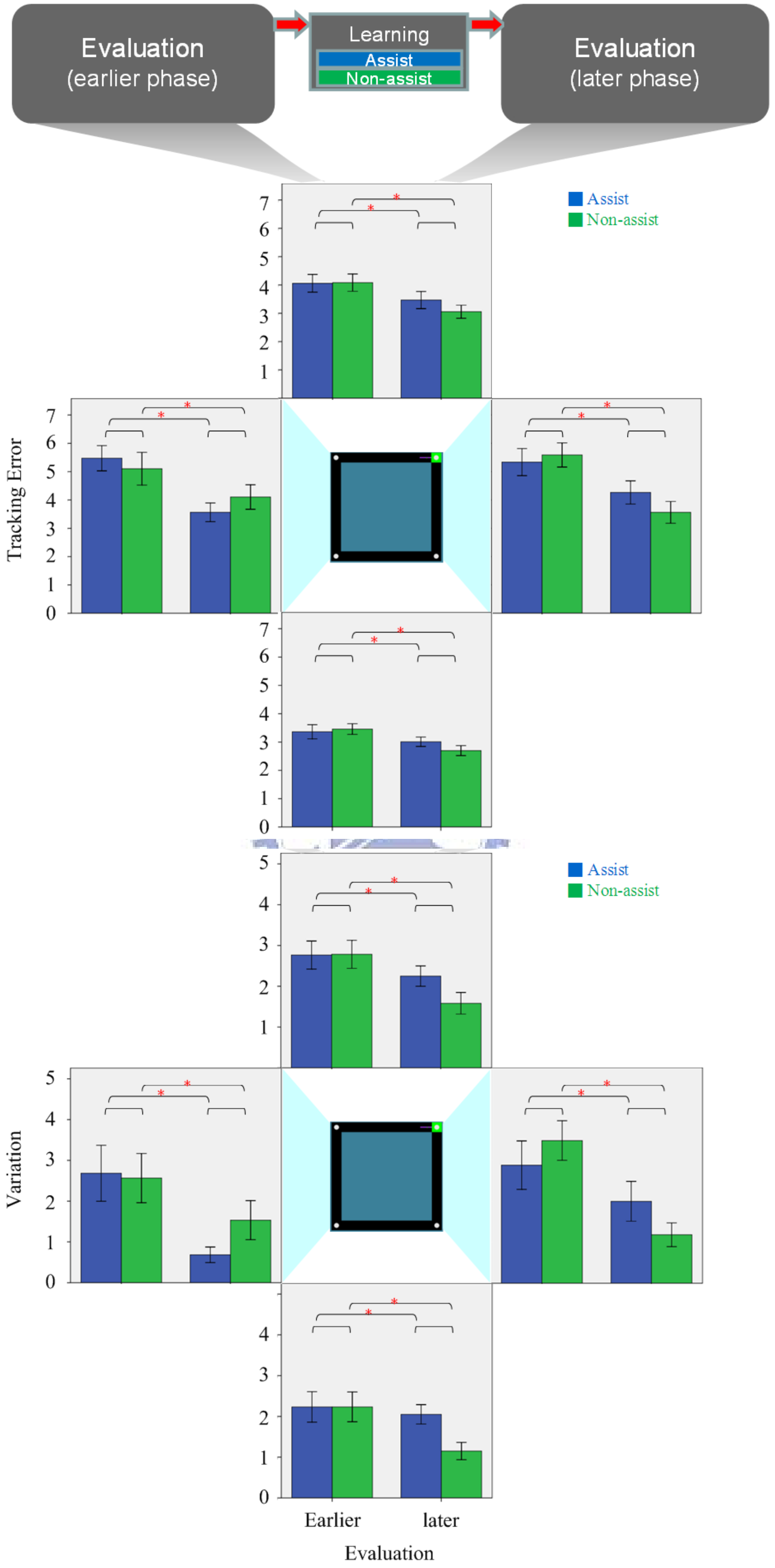


Figure 2-18: The analysis of tracking performance in the evaluation stage in local aspect

Local aspect In the Figure 2-18, the tracking performance didn't significantly differ between the assist and non-assist group in the evaluation of all four movements (ANOVA factor "group", tracking error: adduction, $F(1,28) = 0.295$, $P = 0.592$; flexion, $F(1,28) = 0.041$, $P = 0.841$; abduction, $F(1,28) = 0.267$, $P = 0.609$; extension, $F(1,28) = 0.225$, $P = 0.618$; variation: adduction, $F(1,28) = 0.867$, $P = 0.360$; flexion, $F(1,28) = 0.488$, $P = 0.490$; abduction, $F(1,28) = 1.506$, $P = 0.230$; extension, $F(1,28) = 0.041$, $P = 0.842$). Furthermore, a significant main effect for the factor "phase" (tracking error: adduction, $F(1,28) = 16.480$, $P < 0.001$; flexion, $F(1,28) = 8.831$, $P = 0.006$; abduction, $F(1,28) = 9.263$, $P = 0.005$; extension, $F(1,28) = 14.405$, $P = 0.001$; variation: adduction, $F(1,28) = 11.741$, $P = 0.002$; flexion, $F(1,28) = 8.361$, $P = 0.007$; abduction, $F(1,28) = 7.569$, $P = 0.010$; extension, $F(1,28) = 15.547$ $P < 0.001$) and no significant interaction "groups × phase" for all four movements were registered (tracking error: adduction, $F(1,28) = 1.183$, $P = 0.286$; flexion, $F(1,28) = 0.870$, $P = 0.359$; abduction, $F(1,28) = 1.244$, $P = 0.274$; extension, $F(1,28) = 1.371$, $P = 0.251$; variation: adduction, $F(1,28) = 1.860$, $P = 0.183$; flexion, $F(1,28) = 0.849$, $P = 0.365$; abduction, $F(1,28) = 3.821$, $P = 0.061$; extension, $F(1,28) = 3.079$, $P = 0.090$). After the learning of tracking task, both groups improved the tracking performance in each local movement. But there is no significant difference between assist and non-assist groups.

Counter-clockwise direction: with disturbance force

Local aspect In the Figure 2-19, for all four movements, there was no significant difference of the tracking error between two groups in the later evaluation (solid line). The force feedback assistance learning would not significant help the ability to control the disturbance force. Moreover, there was no significant difference of the tracking error between two evaluation stages for assist group (blue line) and non-assist group (green line). The learning effect of tracking task would not transform to the controllability of disturbance force. By visual of comparing the performance of four areas a different pattern was obvious. The tracking performance was better in the up and down sides than the left and right side. Although subjects encountered the same strength of disturbance force, it still had some different ability to handle the disturbance force. Once again different tracking performance existed in the local aspect.

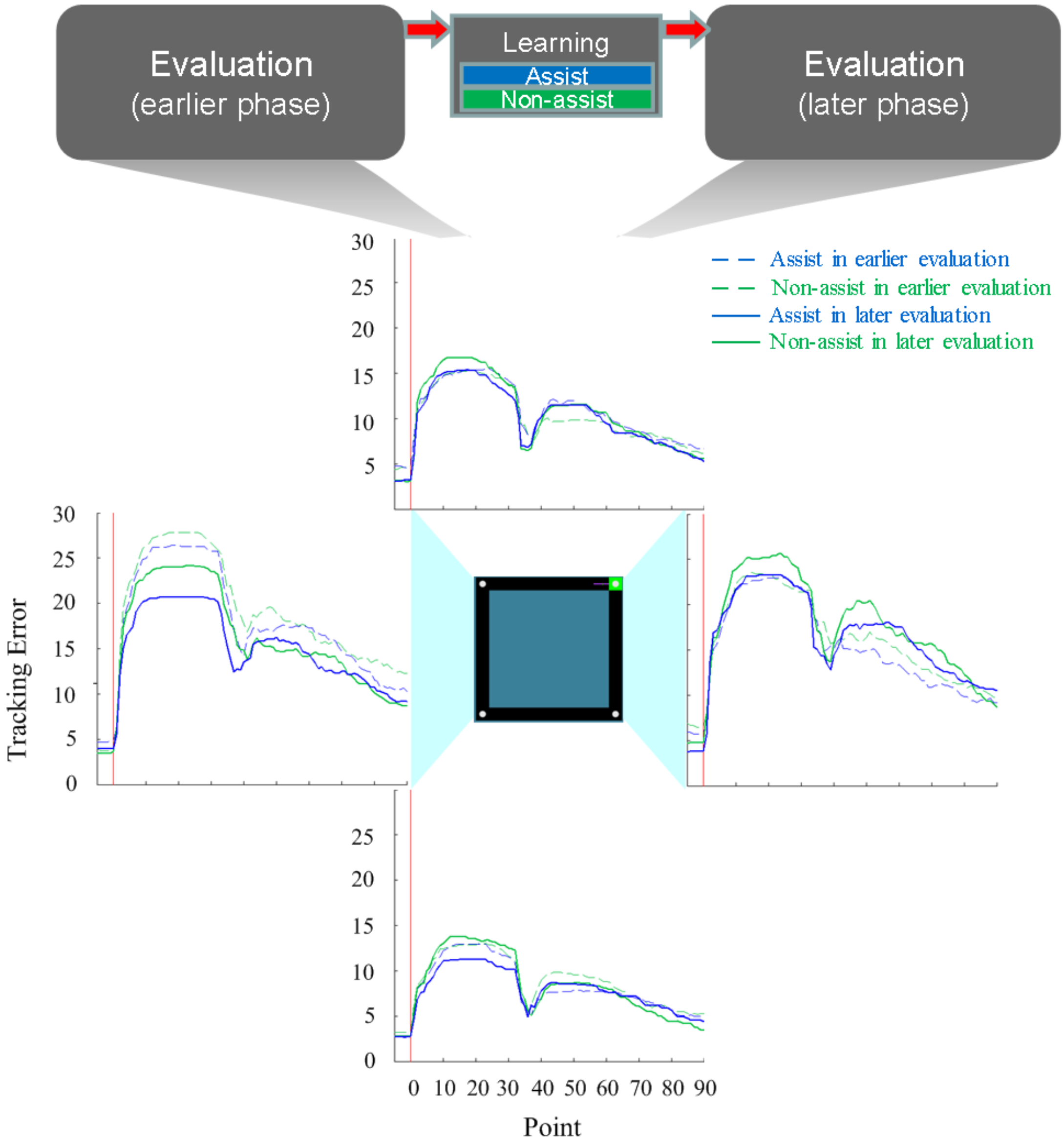


Figure 2-19: The tracking error with disturbance force for counter-clockwise direction.

Clockwise direction: without disturbance force

Local aspect The movements without disturbance force of clockwise direction in local aspect were comparable across groups and phases, in the Table 2-4. The tracking performance didn't significantly differ between the assist and non-assist group in the evaluation. Furthermore, a significant main effect for the factor "phase" only for flexion (*F*(1,28) = 9.057, *P* = 0.005) and no significant interaction "groups × phase" for all four areas were listed. Tracking performance of clockwise direction did not change significantly for adduction, abduction and extension after learning.

Table 2-4: The statistical analysis of the tracking performance for clockwise direction without disturbance force

Mixed design ANOVA												
Tracking error												
Movements		*df*	*F*	*P*		*df*	*F*	*P*		*df*	*F*	*P*
Adduction	group	1	1.891	0.294	phase	1	0.173	0.681	group × phase	1	1.602	0.216
Flexion		1	1.272	0.269		1	9.057	0.005*		1	1.454	0.238
Abduction		1	2.938	0.098		1	2.696	0.112		1	0.113	0.740
Extension		1	1.170	0.289		1	2.490	0.126		1	0.075	0.787
Variation												
Adduction	group	1	0.204	0.655	phase	1	0.575	0.455	group × phase	1	3.549	0.070
Flexion		1	0.822	0.372		1	15.003	0.001*		1	3.823	0.061
Abduction		1	2.501	0.125		1	0.153	0.699		1	0.220	0.643
Extension		1	0.160	0.692		1	3.311	0.080		1	0.246	0.624

Clockwise direction: with disturbance *force*

Local aspect In the Figure 2-20, the tracking error with disturbance force of clockwise direction had no significant difference between two groups in the evaluation. The tracking error with disturbance force of clockwise direction and counter-clockwise direction were contributed to the same results. The learning effect of the tracking task was not transferred to the handle the disturbance force.

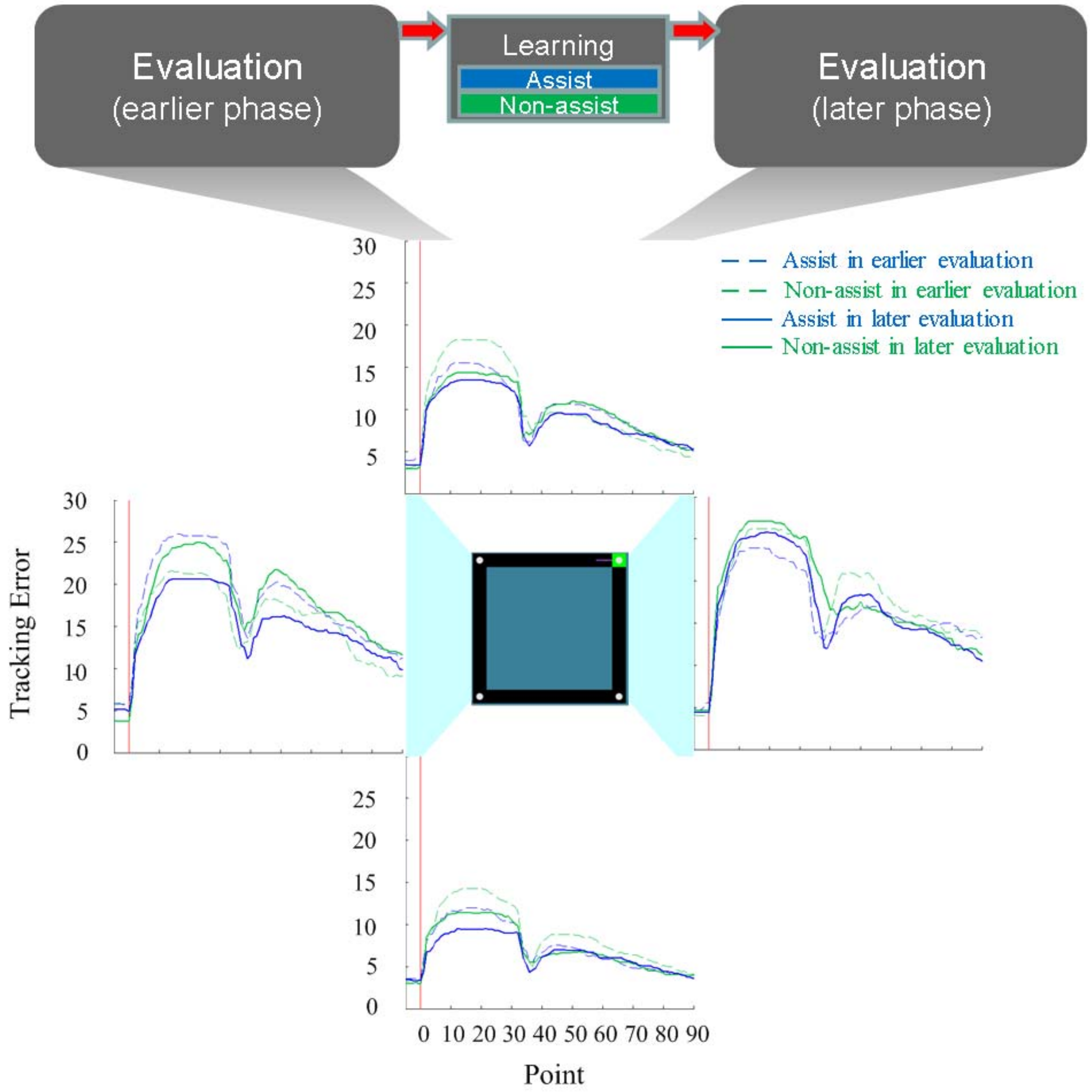


Figure 2-20: The tracking error with disturbance force for clockwise direction. The red lines were the onset of disturbance force. It revealed no significant difference between different groups and different evaluation stages.

2.4 Discussion

The movements of short-term learning depended on the dynamics of both arm and hand-held objects in our study. This study investigated learning effect in a tracking trajectory task, continuous force feedback assistance that afforded continuous monitoring and adjustment of motor output. We focused on two indices to discuss the tracking performance of force feedback assistance, namely, tracking error and variation in the short-term learning. The meaning of the tracking error and variation could be deemed tracking accuracy and tracking stability. First, the subjects of two groups didn’t significantly differ in the earlier evaluation, thus they had the same basic function of right-hand. Furthermore, the subjects of assist group answered that they could correct the deviation easily by the real-time force feedback after experiment.

In the learning stage, because there were different effects of force feedback assistance between four areas, we discussed tracking performance in local aspect. From the analysis of tracking error, assist group had better tracking accuracy for flexion and extension in comparison with non-assist group. Thus the non-assist group might consume most of their time in adjusting the accurate position for flexion and extension because subjects didn’t have the force feedback assistance. Thus incidence of under-correction or over-correction in the non-assist group was also greater than the assist group. The same results could be found from the analysis of curve fitting, initial tracking error indices were differ for flexion and extension. Subjects couldn’t track a trajectory so straight for flexion and extension [42]. In the phase condition, both groups improved the tracking accuracy significantly for abduction and extension with the increase in motor learning. There were similar learning effect at four movements whether or not force feedback assistance. This could be inferred as below.

For abduction, both groups improved the tracking accuracy with learning , since the operation angle at the elbow was approximately from 100° to 110° , the better mechanical advantages for the arm and biceps occurred between those angle at the elbow [43] [44]. For extension, although non-assist group improved tracking accuracy, it was still not better than assist group. The reason may be the movements for extension needed greater inertia to move the arm, and the greater inertia the harder it is to tracking trajectory. [45] [46]. For flexion, the effect of inertia influenced both groups more serious. Non-assist group could not learn how to adjust to promote tracking accuracy. It could be emphasized that force feedback assistance could be used, especially for improving the tracking error caused by greater inertia to move arm.

From the analysis of variation, assist group had better tracking stability for adduction, flexion and extension. Non-assist group had worse tracking stability at beginning of tracking task, but after the first block, they gained similar tracking stability as assist group with learning. The results of curve fitting analysis could further confirm that force feedback assistance could help effective enhance tracking stability. Especially for abduction, it had better tracking stability than the other movements and also could improved tracking stability with learning. Consequently, it could be responded to the results of tracking error that operation angle at the elbow was the better mechanical advantages for the arm to do the tracking task. However, for adduction, flexion and extension, there was no improvement for tracking stability. It may be the intrinsic instability of the hand still not improved and intrinsic instability never disappearance. [47] [42]. On the above discussion, we can know that force feedback assistance theory may be helpful to validate motor tracking task.

Yamamoto and Kitazawa [48] indicate that tactile signals were referred correctly to vertically aligned locations, which are reported to lead to normal judgments. In object manipulation, motor actions of the arm and hand, the motor system receives important sensory information from tactile receptors in the glabrous skin of the volar aspect of the hand [49] [50] [51]. In addition, muscles stabilizing the wrist must also be coordinated with muscles driving the arm when moving handheld loads. Moreover, recent research demonstrated that external focus is more effective than internal focus and examined importance of force feedback from the tool in the referral by manipulating the direction of force feedback in a virtual reality [52] [53] [54] . Computer controlled virtual environments, usually including a robotic manipulandum for force feedback, are often employed in this research as they allow experimenters to precisely control the parameters of their subject's mechanical and visual environment [55] [56].

This study also discussed whether the learning effect retains after removing the warning of force feedback by comparing the later evaluation with the earlier evaluation in the assist and non-assist groups. First of all, consider the same condition as the learning stage, that is counter-clockwise tracking task without disturbance force, the assist group improved the tracking error and variation after removing the force feedback in the later evaluation. The non-assist group was learning the motor skill by themselves, the tracking performances were also increasing at the later evaluation. Both groups improved the tracking performance similarly. There is no significant improvement for both groups doing the task with different conditions as the learning stage, such as tracking task at clockwise direction or with disturbance force. After the learning process only the same movement condition can be improved.

Compared the tracking performance at the end of learning stage and the later evaluation in the same condition of task, the tracking error and variation were worse in the later evaluation. That is after the removal of force feedback assistance, motor performance may not maintain as force feedback assistance. We assume that the subject doesn't control the joystick with force feedback stably in short-term learning. The subjects needed the force feedback to warn them of the deviation during the last block , even though the tracking error of assist group decreased in the short-term learning. Thus the tracking performance of assist group decreased after removing force feedback.

In the behavior experiment, the external force feedback assistance can reduce the tracking error and variation for motor skill learning especially for flexion and extension. And then we investigated the physiological signals changes under force feedback assistance. However, there was much difference of physiological activity between two groups to make objective comparison the force feedback effect. Therefore, we designed experiment 2 to compare the non-assist and assist condition in each subject to reduce the difference from subjects' effect. Since there were both difference for tracking error and variation for flexion and extension, we compared the physiological activity mainly at these sides. The physiological activity, such as oscillatory brain activity and muscle activity during the preparation, execution, and post-movement stage of a tracking task, were investigated.

3. Physiological Signals Activities with Force Feedback Assistance

3.1 Methods

3.1.1 Subjects

There were 9 healthy right-handed volunteers (all males; age range: 19-31 years, mean age: 22.3 years) were paid to participate in this study.

3.1.2 Experimental Environment

The experimental environment was at an anti-noise and soundproof room to reduce the noise to signal ratio. Noise Reduction and Noise Isolation Class according to the ASTM E 336-97 standard, the standard test method for measurement of airborne sound insulation in building, shown in the Figure 3-1.

3.1.3 Experimental Paradigm

In this experiment, there were totally 45 trials the same tracking task as in the expriment1. In the beginning, Subjects did the tracking task for five trials as the baseline without force feedback assistance. The following 40 trials were divided equally into four blocks. There were two different experimental conditions were investigated, one was with force feedback assistance and the other was not. Because the same conditions did not appear in two consecutive blocks, we had two kinds of order for experimental conditions to avoid the results from the specific order effect. Subjects were randomly assigned to one of the order (Figure 3-2). First two blocks were deemed non-assist 1 and assist 1, the others were deemed non-assist 2 and assist 2. Subjects responded three questions about the degree of difficulty tracking without deviation, physical fatigue strength and mental fatigue strength after each block.

Figure 3-1: The experimental environment was at an anti-noise and soundproof room

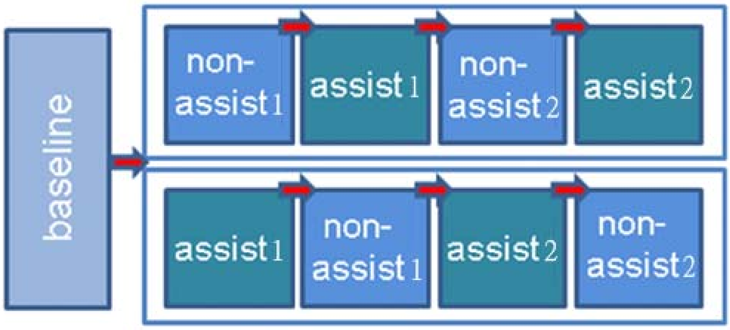


Figure 3-2: Experimental procedures and subjects were randomly assigned to one condition

3.1.4 EMG Data Acquisition

In the Figure 3-3, electromyography (EMG) were recorded by two 1.6-cm Ag/AgCl surface bipolar electrodes fixed about 3-5 cm apart and place on the flexor carpi radialis (FCR), extensor carpi radialis (ECR) and biceps brachii (BB) on the right hand. All of these were superficial muscles of the upper limb and were related the movements of the arm. The EMG signals were acquired with a 1000-Hz sampling rate, and notch filtered at 60-Hz.

We did the scrub to clean the skin above the muscles where electrodes were located, then we could get the better EMG signal. Electrodes and cables should be fixed by the bandage to reduce the noise from moving (Figure 3-4). To select robust sites for electrodes placement, the electrodes over these muscle sites were placed according to ask subjects to do some specific movement to test the clinical assumptions [57].

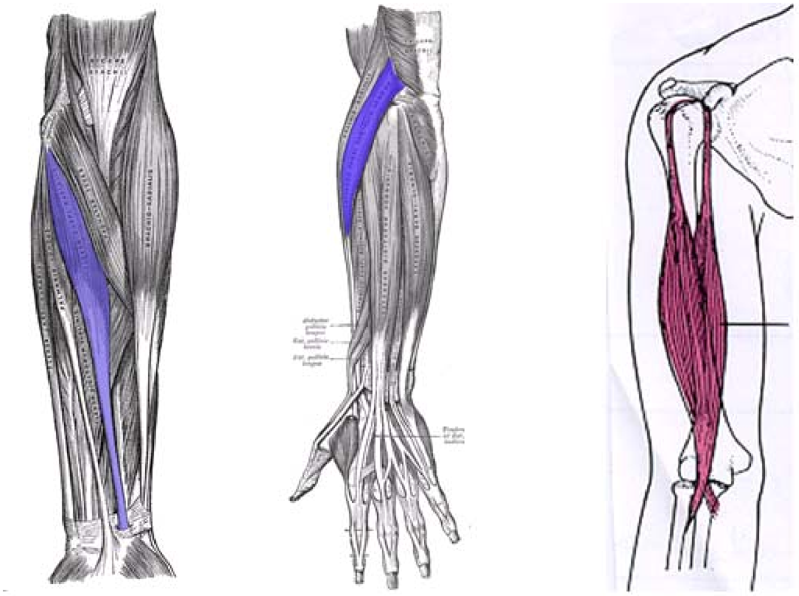


Figure 3-3: Three muscles for EMG signal measurement, those were flexor carpi radialis (FCR), extensor carpi radialis (ECR) and biceps brachii (BB) from left to right of the figure. Source: http://en.wikipedia.org/wiki/

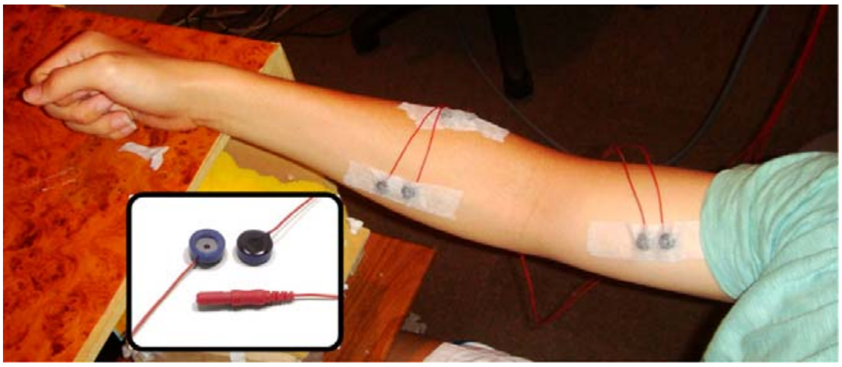


Figure 3-4: Three EMG channels were measured on the right hand of subjects.

3.1.5 EEG Data Acquisition

The electrical activity of neurons in the brain produces currents that reach the surface of the scalp. EEG provides a non-invasive method of recording the voltage differences of these scalp potentials.

EEG recordings were conducted using an electrode cap on which 64 Ag/AgCl electrodes (Figure 3-5(A)) were mounted. Subjects were with a movement-proof electrode cap for measuring the electrical activates of the brain and that was the electroencephalograph (EEG).All electrodes were placed in an elastic cap according to the international 10-20 system which was proposed by Jasper in 1958 (Figure 3-5(B)). The contact impedance between EEG electrodes and scalp was calibrated to be less than 10kΩ with conductive gel. EEG data were recorded and amplified with the Scan NuAmps2 Express system (Compumedics Ltd., VIC, Australia) with sampling rate at 1000 Hz and 32-bit quantization level, and notch filtered at 60-Hz.

At the beginning of the experiment, locations of the electrodes were digitized with the 3D digitizer (POLHEMUS 3 space Fastrak). Figure 3-6 shows the positions of the 3D digitized electrodes. Immediately, we confirmed the location of the electrodes and made the appropriate adjustments so that each electrode was located in the right position.

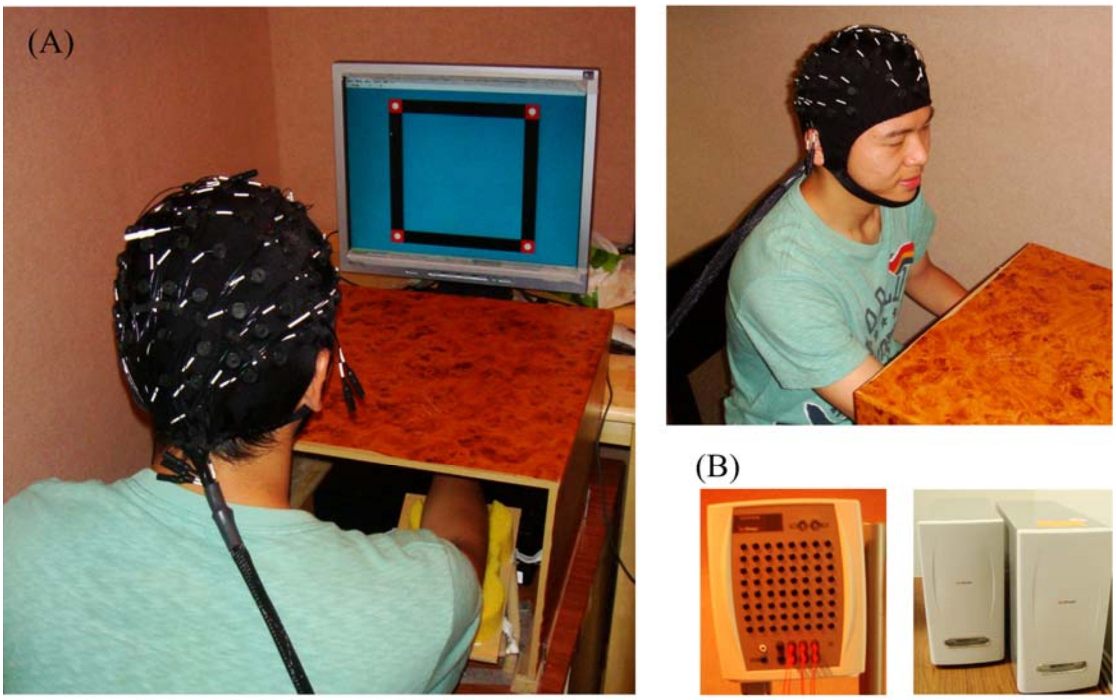


Figure 3-5: (A) EEG recordings were conducted using an electrode cap on which 64 Ag/AgCl electrodes (B) EEG data were recorded and amplified with the Scan NuAmps2 Express system

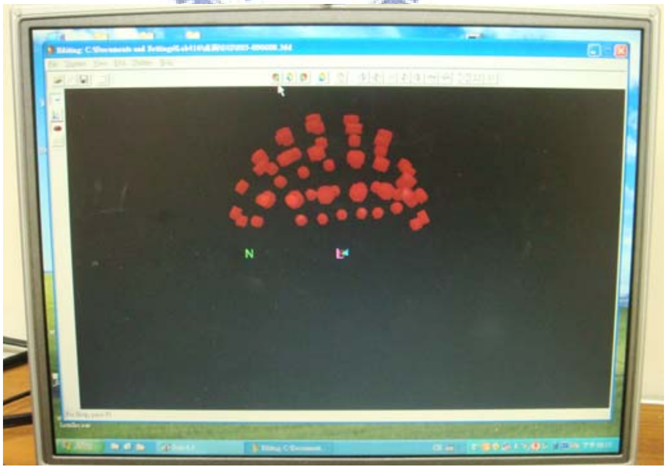


Figure 3-6: The picture shows the 3D Digitizer results.

3.2 Data Analyses

3.2.1 EMG Data Analyses

The EMG signals were first filtered by the band pass filter. A typical band pass filter was used to let all of the energy above 20 Hz through and then those close the gate at 300 Hz. The lower cutoff point eliminated much of the electrical noise associated with wire swayed and miscellaneous biological artifacts associated with slow-moving DC potential shifts. The upper cutoff point eliminated the tissue noise at the electrode site.

RMS approached the quantification of the EMG signal by squaring the data, summing the squares, dividing this sum by the number of observations, and finally taking the square root in a trial. Finally, averaged the RMS (2) values of trials in the “assist” condition and “non-assist” condition and compared the difference between two conditions.

$$\chi_{rms} = \sqrt{\frac{1}{n} \bullet \sum_{i=1}^{n} \chi_i^{2}} \qquad (2)$$

where χ_i for the measured values of EMG, n for the data points of EMG.

3.2.2 EEG Data Analyses

The flowchart of processing steps was shown as Figure 3-7. EEG data analysis was performed with EEGLAB 6.01, a freely available open source software toolbox (Swartz Center for Computational Neuroscience, La Jolla, CA; http://www.sccn.ucsd.edu/eeglab) running under MATLAB.

The EEG signals were first filtered with a low-pass and high-pass filter with the cut-off frequencies at 50 Hz and 2 Hz respectively to remove the line noise (60 Hz and its harmonic) and the DC drifting. Then EEG signal data were down samples to

250 Hz for data compression. The EEG filtered data were epoched into non-overlapping segments of 12 s duration (-3 to 9 s relative to cue presentation) in each block. That is the epochs were aligned to the onset of cue (yellow light). Since one trial of the tracking task were consisted by four different movements, the EEG response related to different areas (different movement strategies and muscle activity) were extracted and grouped separately. The Figure 3-8 shows the methods for extracting grouping epochs.

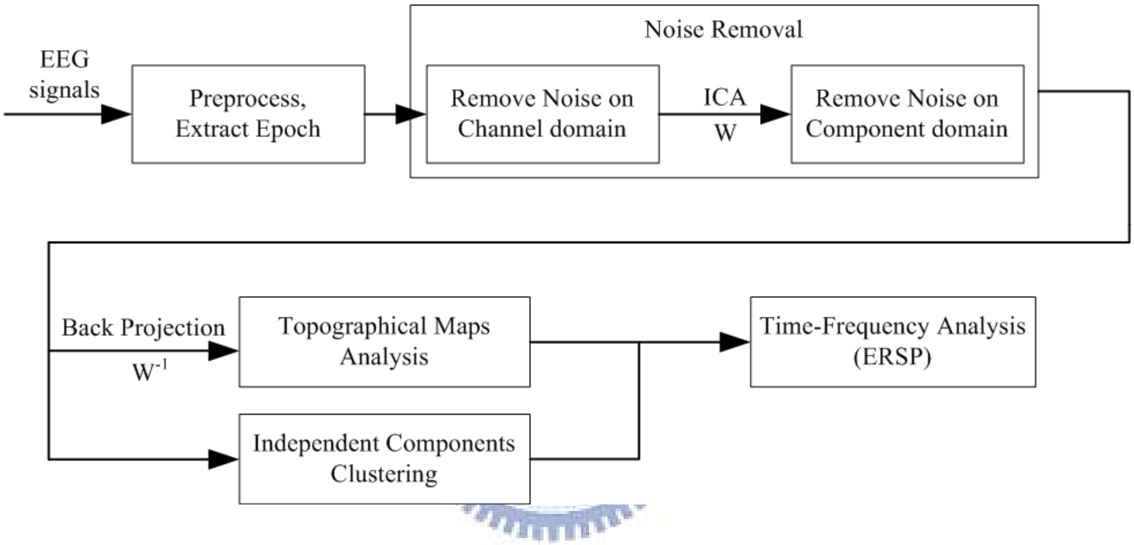


Figure 3-7: The flowchart of EEG data analysis.

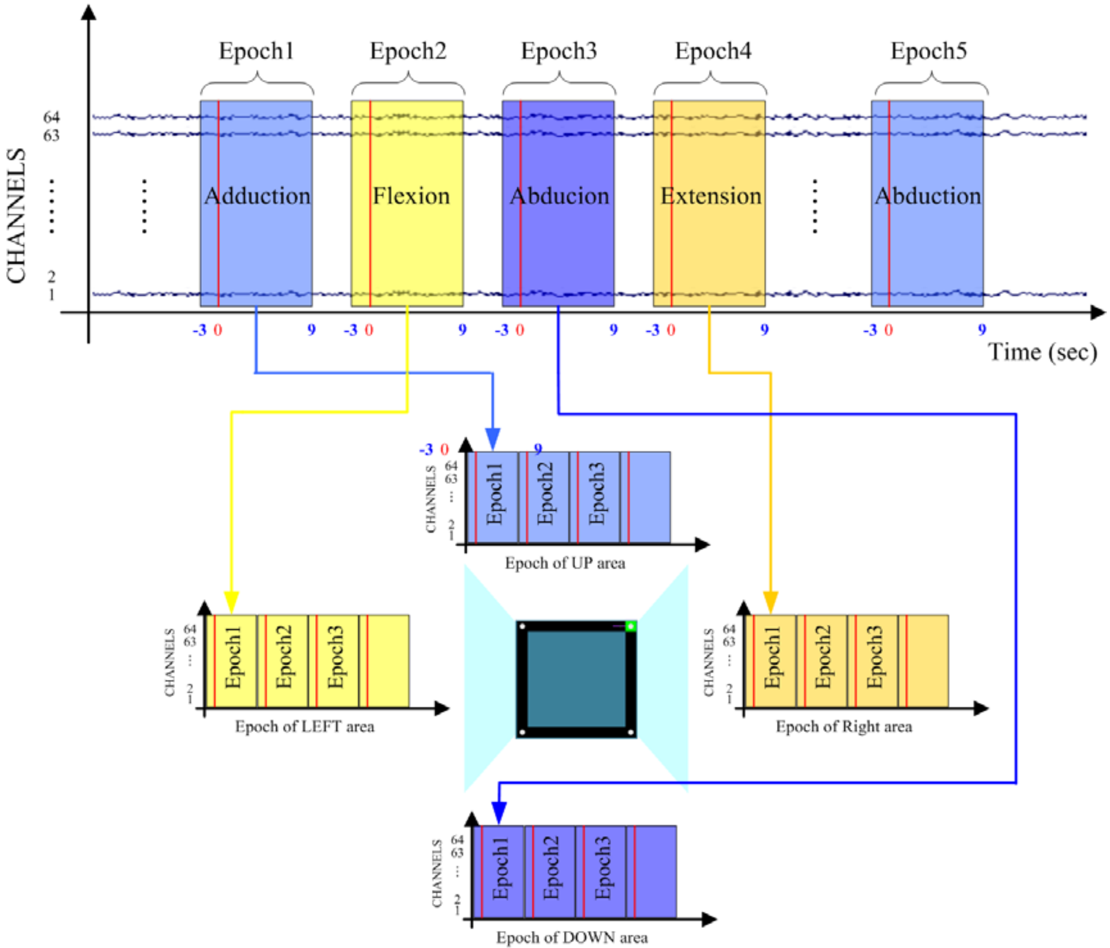


Figure 3-8: The illustration of epoch extraction.

Artifacts contaminated in the epoched EEG signals, those artifacts could be divided into physiologic and extra-physiologic activity. The physiologic artifacts (e.g. muscular activity and eye blinks) were generated from the body, which rose from other sources instead of the brain. Extra-physiologic artifacts were contaminated from the equipment or the environmental electrical noise. First, the epochs containing non-stereotyped artifacts (e.g., swallowing, cable movement, etc.) were rejected identified by visual inspection using the EEGLAB visualization tool and eliminated to enhance the signal to noise ratio. Figure 3-9 shows the example of the common artifacts contaminated in the EEG signals. Second, the EEG data were submitted to extended independent component analysis (ICA), in other words the EEG signals were transferred from channel domain to component domain. Independent components reflecting eye moment or other artifacts (e.g., temporal muscle and frontal muscle) were identified visually and moved the artifact components. The remainder components were back projected to the channel data domain [58] [59]. Back-projection was performed on the epoched data, from which all previously data containing non-stereotyped artifacts had been discarded. The methods of ICA were briefly described on the following paragraph 3.2.2.1.

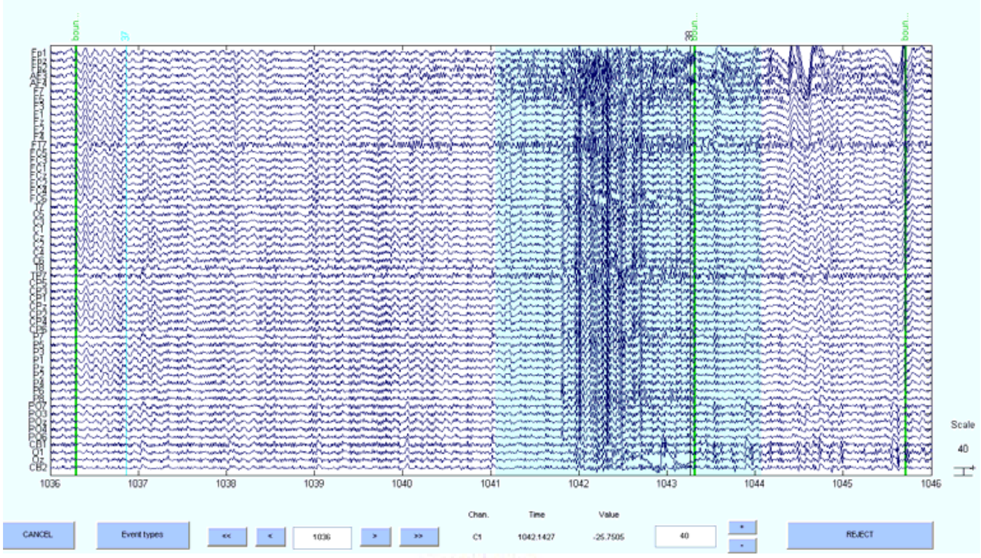


Figure 3-9: The example of common artifacts contaminated in the EEG signals, artifacts from cable moment

Topographical maps analysis was performed for all channels' data by plotting scalp map on certainly frequency band. Independent components clustering were performed for all components' data by k-means clustering, was described on the following paragraph 3.2.2.2. Finally, time-frequency analysis was performed for all channels and components by convolving the single-epoch data with Hanning-window sinusoidal wavelets [60]. For each epoch the mean baseline log power spectrum was subtracted from the spectral estimate to produce baseline period was set -2000 to -500 ms relative to the cue onset. Single-epoch spectral estimates were then averaged for the each subject. In order to study effect of force feedback assistance, epochs were averaged for the assist blocks as well as non-assist blocks, respectively. The principle of time-frequency analysis called Event Related Spectral Perturbation (ERSP) Analysis was described on the following paragraph 3.2.2.3.

3.2.2.1 Independent Component Analysis (ICA)

Each EEG channel activities is a time course of mixed voltage activities which collected from any point on scalp induces activity generated within a large brain area. Especially, large artifacts due to motion during exercise were contaminated with the brain source activity. Therefore the EEG source segregation, identification, and localization were very difficult. Nevertheless, ICA does offer great promise as a technique for artifact removal in the exercise research [61]. ICA is a higher order statistical method developed to extract individual signals (referred to as components) from mixtures of signals, based on the assumption that different physical processes (referred to as sources) will generate unrelated signals. One methodological caveat is that since the aim of the ICA is to separate underlying ‘source’ signals considered to be ‘statistically independent’ over time, it requires a relatively large amount of data in both length (EEG samples) as well as channels (the number of sources it yields is directly restricted by the number of recording electrodes used). Even on modern day computers, common ICA algorithms may take from minutes (for individual EEGs) up to hours (for very long EEG or if analysed en masse) to complete. In addition to the assumption of independence of source origins, common ICA algorithms (e.g. InfoMax, FastICA) operate under two further assumptions. Firstly that the underlying sources must exhibit non-Gaussianity, i.e. they must be non-normally distributed (other techniques do not require this assumption. Secondly the sources should be stationary, that is to say, they should each have a fixed location throughout the recording.

There were two conditions assumed for ICA computational processes: (a) the conduction of the EEG sensors is instantaneously and linearly such that the measured mixing signals are linear and the propagation delays are negligible. (b) The signal source of muscle activity, eye movements, and, cardiac signals are not time locked to the sources of EEG activities which reflected the synaptic activities of cortical neurons.

$$\mathbf{x}(t) = \mathbf{A}\mathbf{s}(t) \tag{3}$$

$$\boldsymbol{u}(t) = \boldsymbol{W}\,\boldsymbol{x}(t) \tag{4}$$

ICA is a statistical and computational technique for revealing hidden factors that underlie sets of random variables, measurements, or signals and it describes the source separation problem as the form of equation (3). Where $\mathbf{x}$ is the signal recorded form scalp and $\boldsymbol{A}$ is a linear transform called a mixing matrix. Blind sources s_i are statistically mutually independent. The ICA model estimates a linear mapping $\boldsymbol{W}$ such that the unmixed signals $\boldsymbol{u}(t)$ are statically independent as in equation (4). In ideal case, ICA components $\boldsymbol{u}$ is equal or approximate to activity sources $\boldsymbol{s}$. Figure 3-10 illustrates the concept of ICA application on EEG analysis.

Thus, the artifacts including the eye-movement (EOG), eye-blinking, heart-beating (EKG), muscle-movement (EMG), and line noises can be successfully separated from EEG activities and removed. Figure 3-11 shows the result of the scalp topographies of ICA weighting matrix $\boldsymbol{W}$ corresponding to each ICA component by projecting each component onto the surface of the scalp, which provided us to remove the artifact components, e.g., eye activity of components 7, 21 was projected mainly to frontal lobe, temporal muscle of components 27, 43 was projected mainly to temporal lobe. The activities from different brain sources, eye movement artifacts, muscles and channel noises were effectively separated into independent components 1 and 60.

Recent years have seen a remarkable proliferation of ICA related articles with successful applications described in reference to both artifacting [62] and EEG source modelling [63]. Evidence for the former includes artifact removal of muscle contamination, eye blinks and movement, noise, as well as cardioballistic phenomena. Figure 3-12 illustrates how ICA was used to extract an eye blink component from the raw EEG in data recorded from a sample of tracking task data.

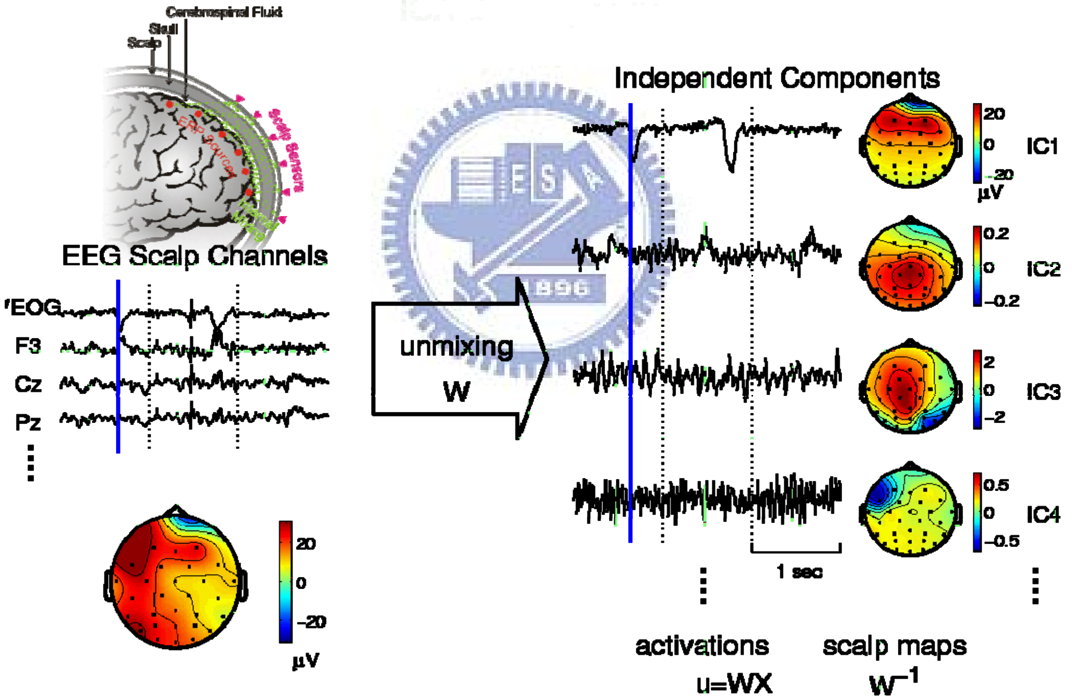


Figure 3-10: The illustration of ICA decomposed concept. EEG signals recorded from the scalp are mixed with multiple sources by a mixing matrix. ICA is a training phase to find a transformation matrix W to separates the mixed EEG signals into independent components which may have specific meanings, and Scalp maps is able to plot according to the weight of ***W*** [64]

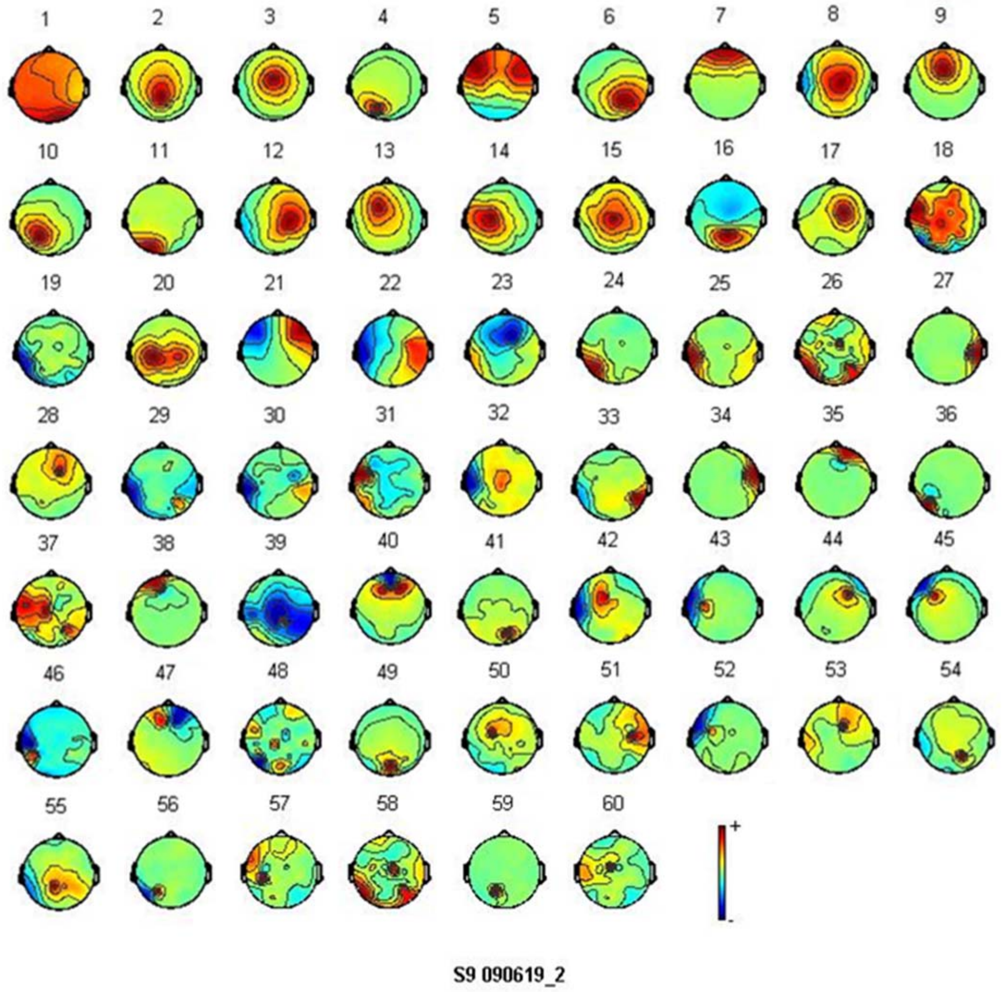


Figure 3-11: The typical example of scalp topography of ICA decomposition of one subject. 60 components are revealed in this example. The color bar is the amplitude of component signals

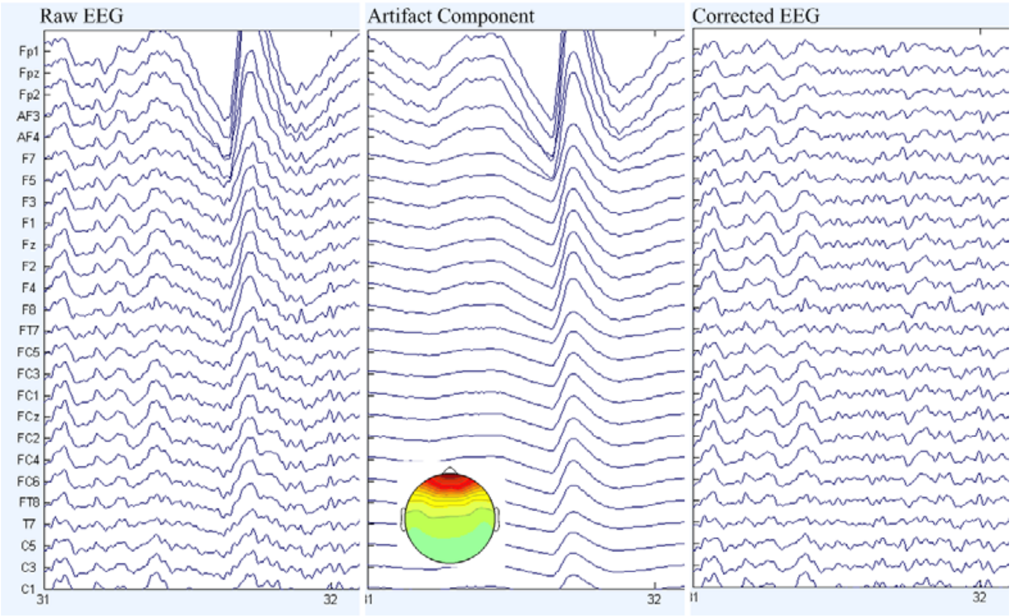


Figure 3-12: The example of artifact removal from the component 7, the eye blinking component.

3.2.2.2 Independent Component(IC) Clustering

To compare electrophysiological results across subjects, the usual practice of most researchers has been to identify scalp channels. Component clustering would be used to assess the consistency of ICA decompositions across subjects and sessions, and to evaluate the separate contributions of identified clusters of these data components to the recorded EEG dynamics.

ICA components from multiple subjects were clustered semi-automatically based on their gradient of scalp projections, EEG characteristic and equivalent dipole location. These criteria for each IC were measured and compressed into 10-dimensional feature vector by principal component analysis (PCA). Semi-auto K-means algorithm was used to group similar ICs together and the clustering result was finally adjusted according to the time-frequency property and equivalent dipole location of each IC. The grand mean scalp map and spectra property were then computed for each cluster to investigate the common characteristics brain activity in the task. The process of clustering was shown as Figure 3-13. Further analysis will base on the clustering result.

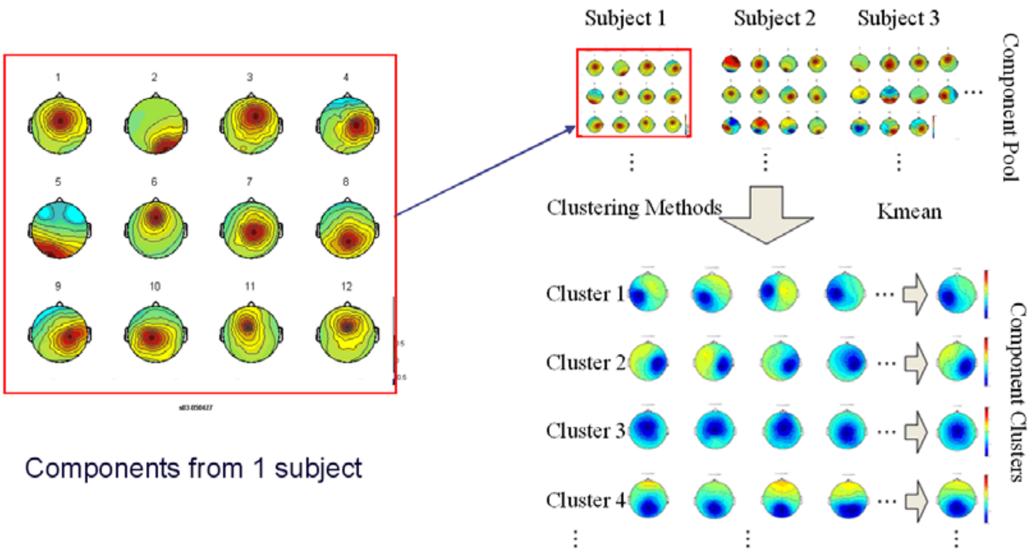


Figure 3-13: The flowchart of component clustering analysis using the K-means algorithm.

3.2.2.3 Event Related Spectral Perturbation (ERSP) Analysis

ERSP is a kind of time-frequency analysis of signals, which was first proposed by Makeig [65], can reveal those time-locked but not necessary phase-lock event related activities. ERSP analysis transforms time-course signal into spectral-temporal domain. The brain activity can be investigated in frequency domain, and we can choose the frequency band associated with tracking task to further analysis.

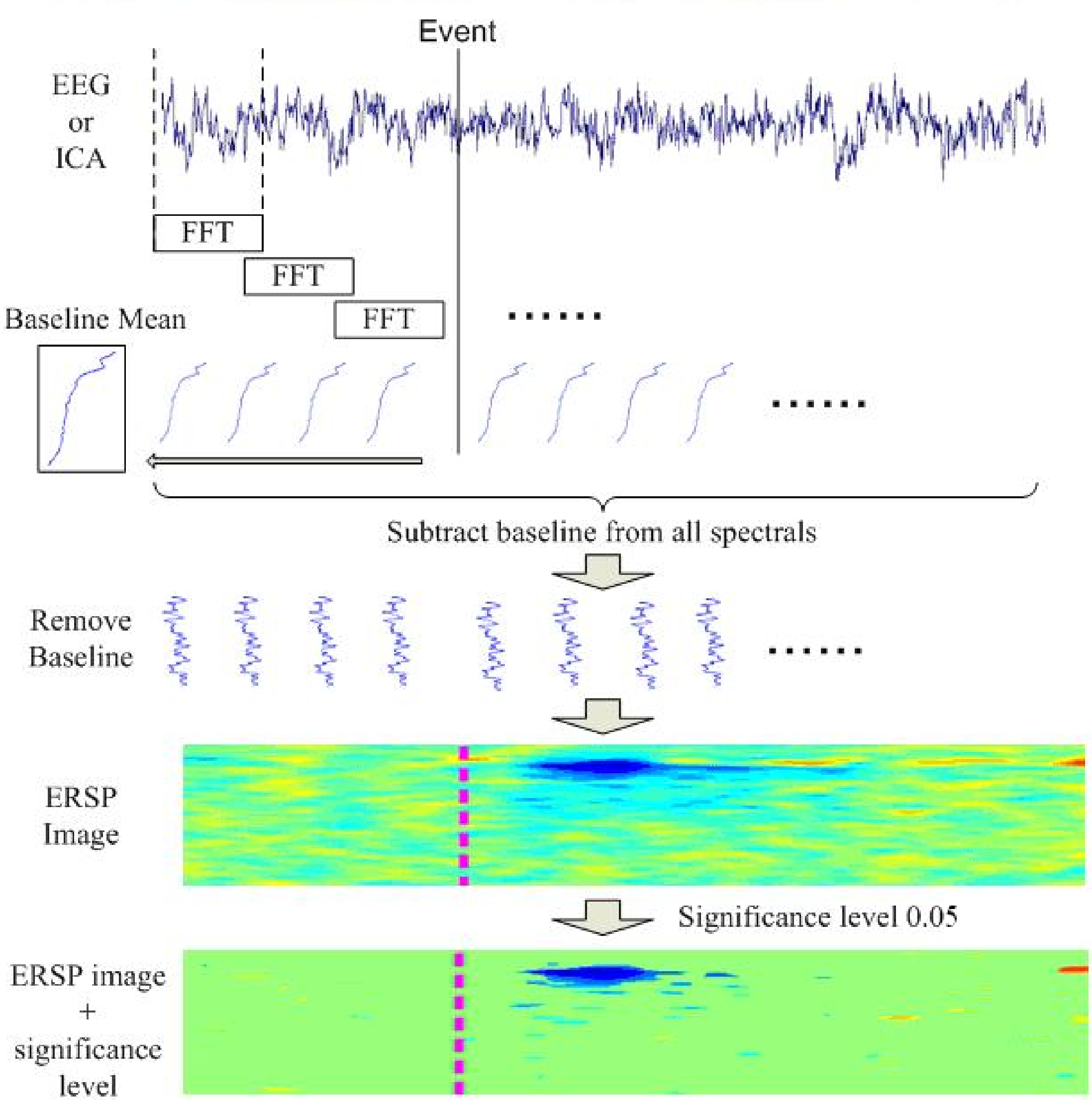


Figure 3-14: The picture shows the illustration of procedures in ERSP analysis. FFT is used a 1-s moving-window size. In the final step, non-significant parts of ERSP image are set to zero by the means of bootstrap [66].

The processing flow was shown in Figure 3-14.For each group of epochs, time series of EEG in each epoch k were transformed into time frequency matrix $F_k(f,t)$ using a 1-s moving-window fast Fourier transforms (FFTs). Log power spectra were estimated at 100 linearly-spaced frequencies from 0.5 Hz to 50 Hz, and then were normalized by subtracting the log mean power spectrum in the baseline (holding green cursor in the white circle) periods for each group of epochs. Time series of EEG in each trial were transformed into time-frequency matrix (200 x 100), with a frequency resolution about 0.5 Hz. Event-related spectral perturbation (ERSP) images, were obtained by averaging n time-frequency matrices from the same group using equation (5).

$$ERSP(f,t) = \frac{1}{n}\sum_{k=1}^{n} \left| F_k(f,t) \right|^2 \tag{5}$$

ERSP images were constructed to show only statistically significant ($p<0.05$) spectral perturbations (log power differences) from the mean power spectral baseline. Significance of deviations from power spectral baseline was assessed using a surrogate data permutation method. In the resulting ERSP images, non-significant time/frequency points were colored green. Through ERSP, we investigated the time-frequency information of brain activity in different force conditions during the tracking task.

3.3 Experimental Results

3.3.1 EMG Activities between Two Learning Conditions

In the Figure 3-15 (A), EMG power of single subject when tracking task were plotted and red line, blue line and green line were represented by cue, start and tracking, respectively. Subject performed with smaller EMG power for force feedback assistance condition. The following in the Figure 3-15 (B), EMG power for each subject (dash line) were normalized to non-assist condition as zero. Grand mean average EMG power was represented by solid red line. There was significant lower EMG power for force feedback assistance condition at FCR and BB (paired-t test, FCR: *P* =0.049; ECR: *P* = 0.353; BB: *P* =0.006). Therefore subjects could perform the tracking task with less muscle strength for force feedback assistance.

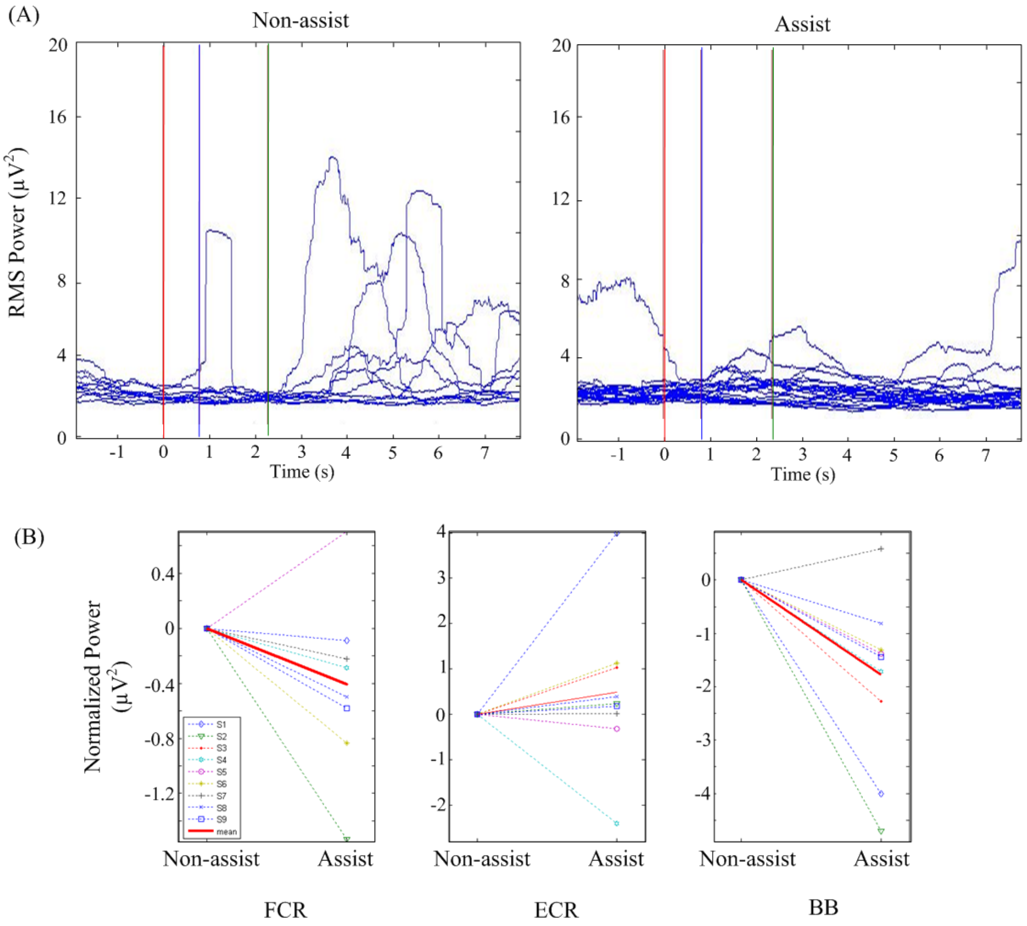


Figure 3- 15: EMG power for non-assist and assist condition.

3.3.2 Brain Dynamics between Two Learning Conditions

Topographical maps

In the Figure 3-16, grand mean average topographies (averaged across all scalp channels and subjects) for the theta (4~7 Hz), alpha (8~12 Hz) and beta (13~30 Hz) band for non-assist and assist conditions. For the theta band, cue onset a power increase was observed which was maximal over bilateral motor electrode sites. The following, start movement a power increase was shifted over centro-frontal sites and was shifted again over left motor when tracking trajectory. For the alpha band, tracking a power decrease was observed over all electrode sites and maximal over posterior-central sites. For the beta band, a power decrease for cue, start and tracking maximal over bilateral motor, central and widespread electrode sites respectively. From the above observation, the majority electrode sites of brain activities for the tracking task were located over frontal, central and left motor. Further ERSP analysis was performed over these areas.

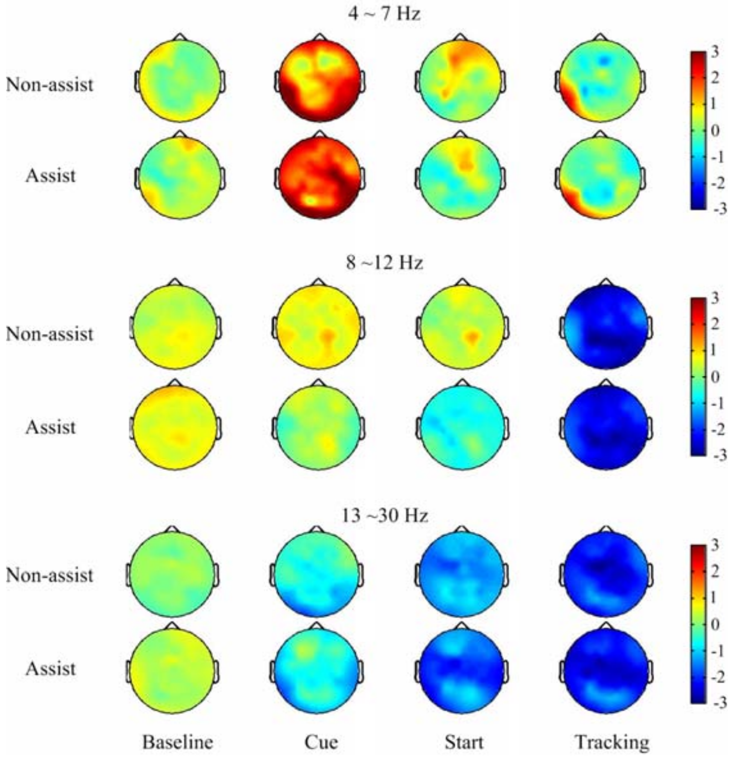


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Figure 3-16: Topographical maps analysis.

The frontal and central channels

In the Figure 3-17, the grand mean of ERSP images of the frontal and central channels across 7 subjects are shown for non-assist and assist conditions. The lower panels were the difference between non-assist and assist conditions for significant power 0.05 by paired-*t* sample test. There were significant difference between two condition for the alpha and beta band power while subjects starting movement. The alpha and beta band power attenuated while the tracking task under force feedback assistance.

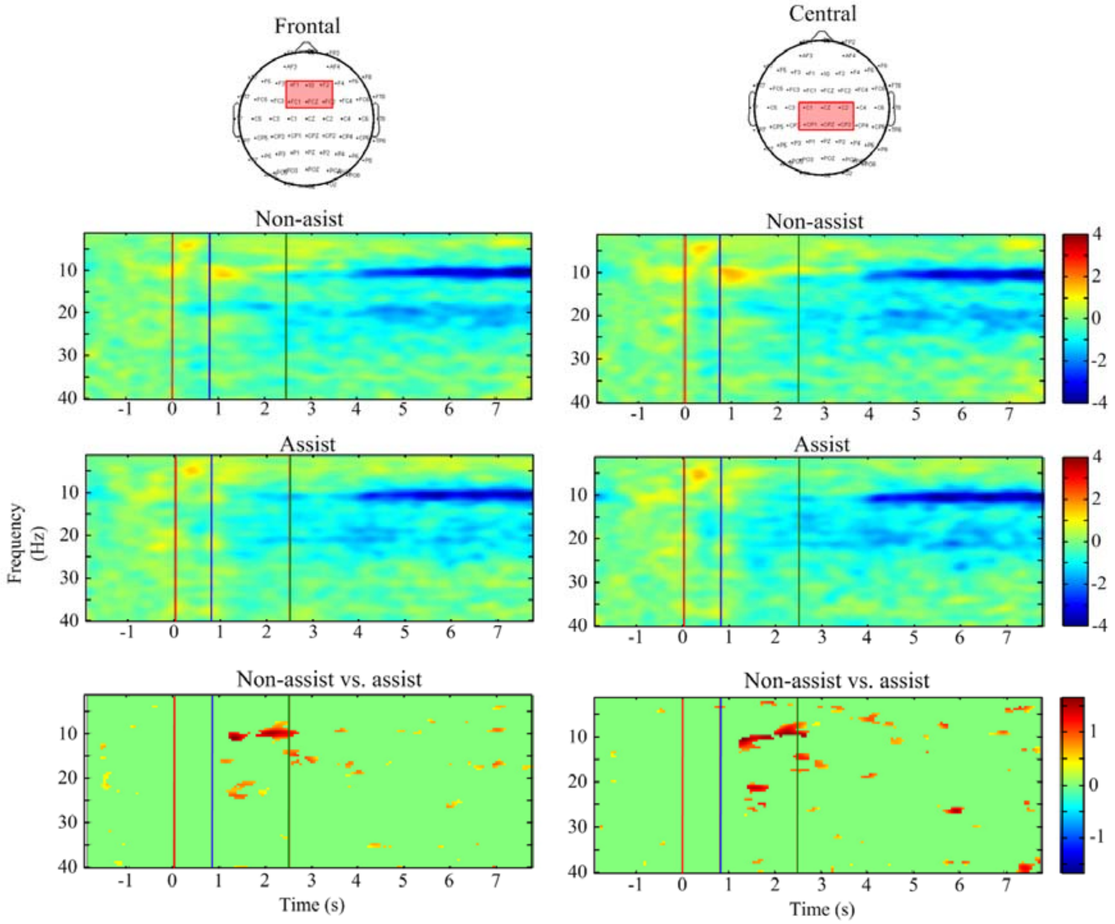

Figure 3-17: ERSP images of the frontal and central channels for non-assist, assist conditions and the difference.

The left motor channels

In the Figure 3-18, the ERSP images of the left motor channels are shown. There were significant difference between two conditions only for the alpha band power while subjects starting movement. The alpha band power also attenuated while the tracking task under force feedback assistance.

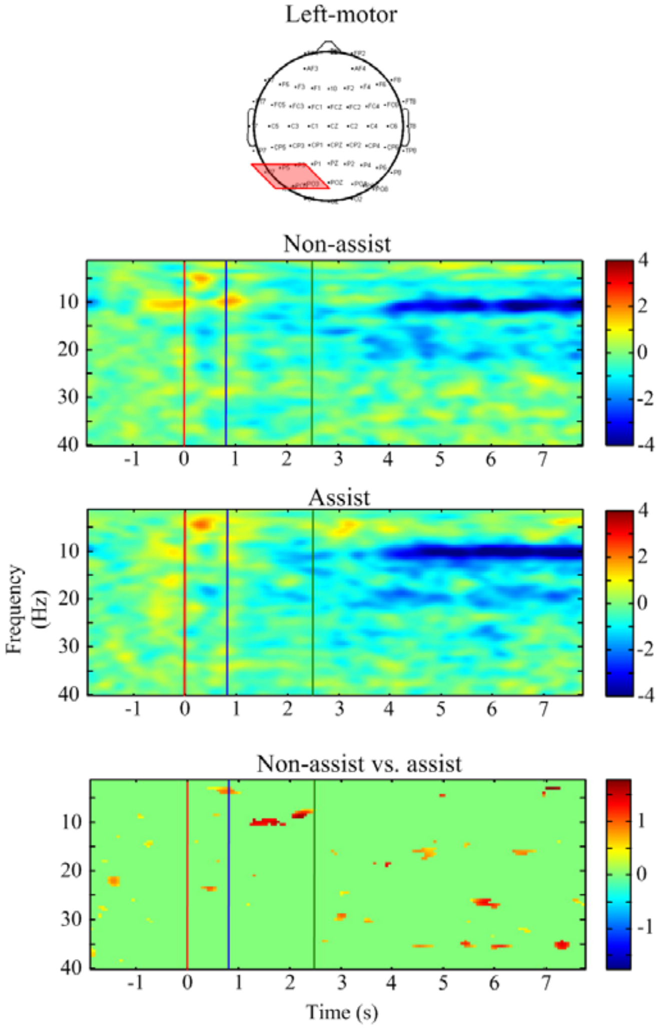

Figure 3-18: ERSP images of the left motor channels.

Independent components clustering and source localization

The results of the components clustering showed that the tracking trajectory at least involved several brain regions. In the Figure 3-19, the grand mean of the scalp map, the scalp maps from individual subjects and their equivalent dipole models for three major IC clusters (frontal, central and left motor clusters) across 7 subjects. For each clusters, all of the independent component were located at nearby areas. Based on these component clusters, the further ERSP images were investigated.

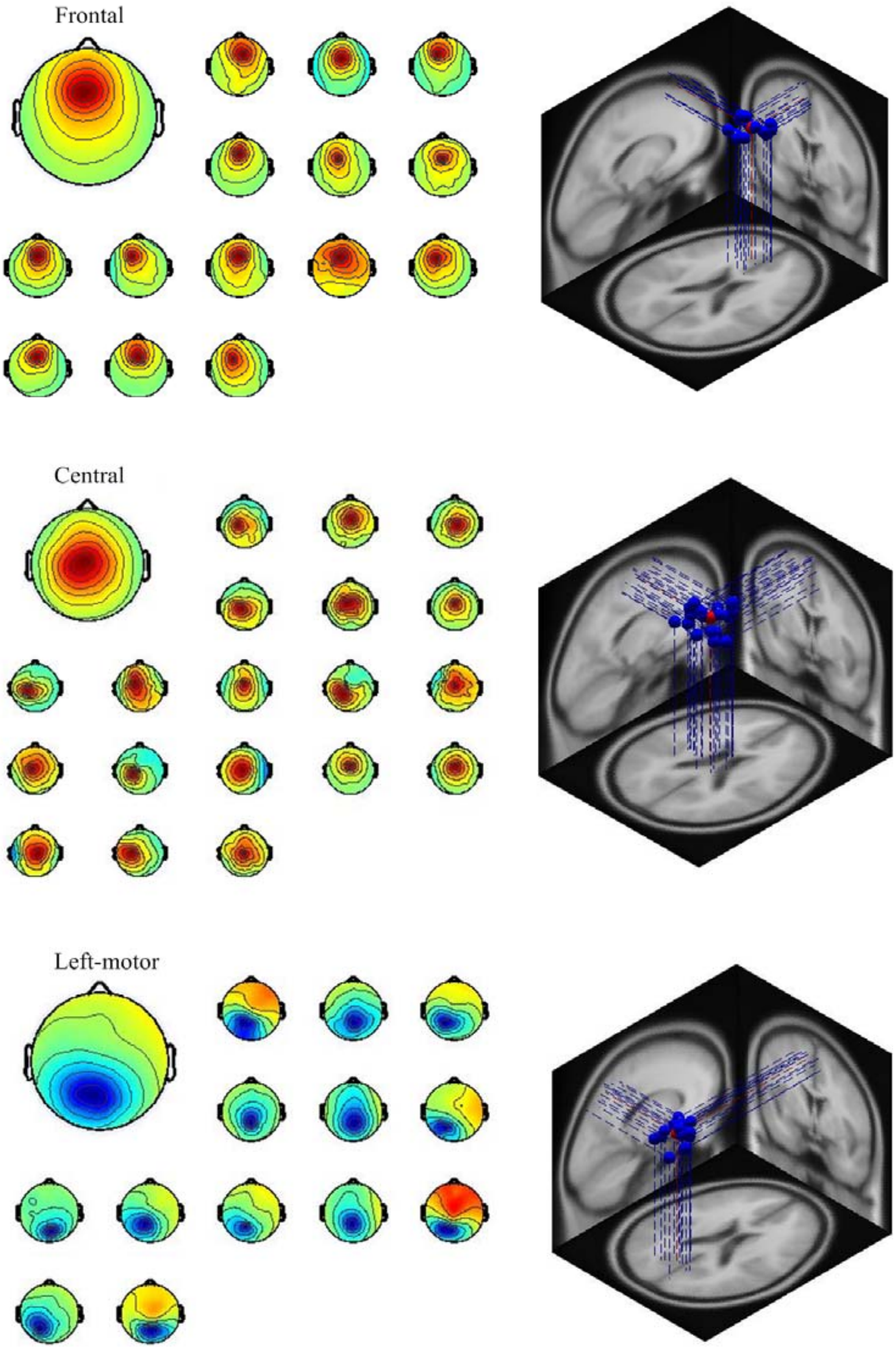

Figure 3-19: The analysis of independent components clustering and source localization

The frontal and central components

In the Figure 3-20, the ERSP images of the frontal and central components are shown. The results for non-assist versus assist condition were different in comparison with the frontal and central channels. There was no significant difference between two conditions over frontal component. The attenuation of the alpha and beta band power under force feedback assistance was presented mainly over central component when starting movement. Force feedback assistance would change the brain dynamic over central component.

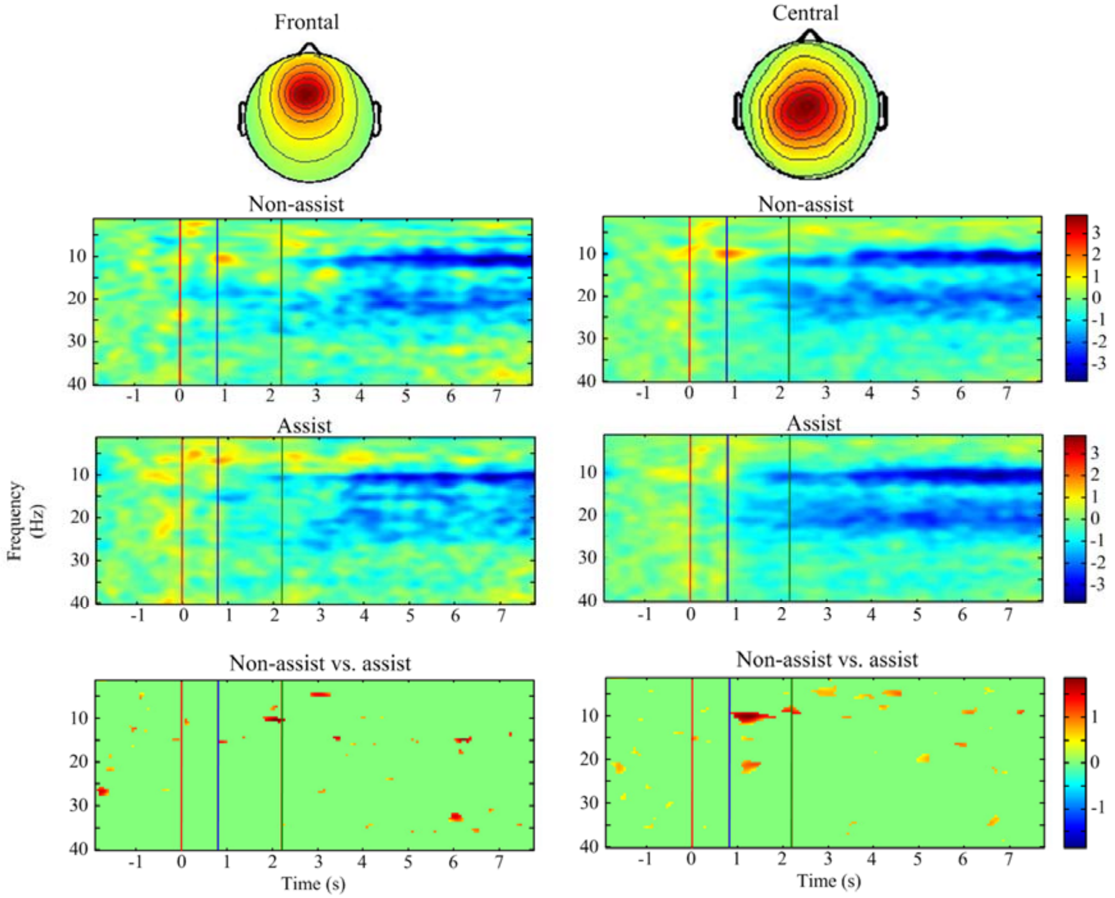

Figure 3-20: ERSP images of the frontal and central components

The left motor components

In the Figure 3-21, the changes of the neural activities related to process tracking trajectory over the motor IC cluster were characterized. There was no apparent difference between non-assist and assist conditions. Force feedback assistance would not influence the brain dynamic over left motor component, since the motor component of both groups had to handle the same task, that was tracking trajectory and minimize the tracking error.

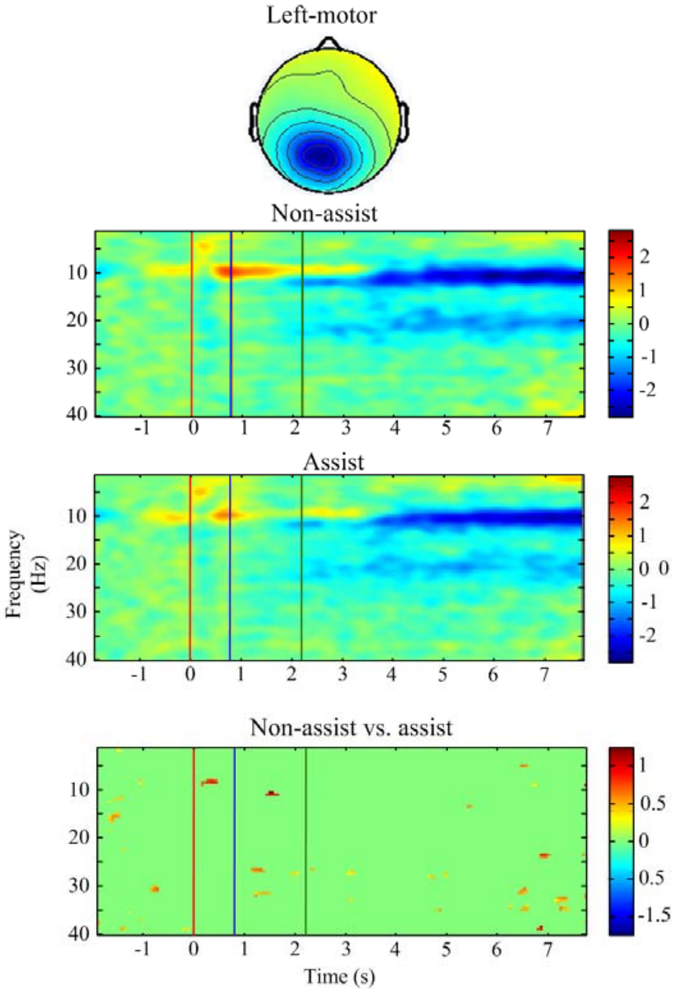

Figure 3-21: ERSP images of the left motor components

After the comparison with non-assist and assist condition for 3 clusters, the different learning blocks were investigated for non-assist 1 and non-assist 2 in comparison with assist 1. In the Figure 3-22, non-assist 1 vs. assist 1 was presented by the difference between non-assist 1 block and assist 1 block for significant power 0.05 by paired-*t* sample test, and non-assist 2 vs. assist 1 was the difference between non-assist 2 and assist 1. Results revealed that the brain dynamic of first learning block with force feedback assistance were similar with non-assist 2 block and different with non-assist 1 block. For the non-assist 1 vs. assist 1, the alpha and beta attenuation over the frontal cluster were found widespread when cuing, start and tracking. Over the central cluster, not only the alpha and beta attenuation when start but also the theta attenuation with force feedback assistance when tracking were found. There was no apparent difference over left motor cluster.

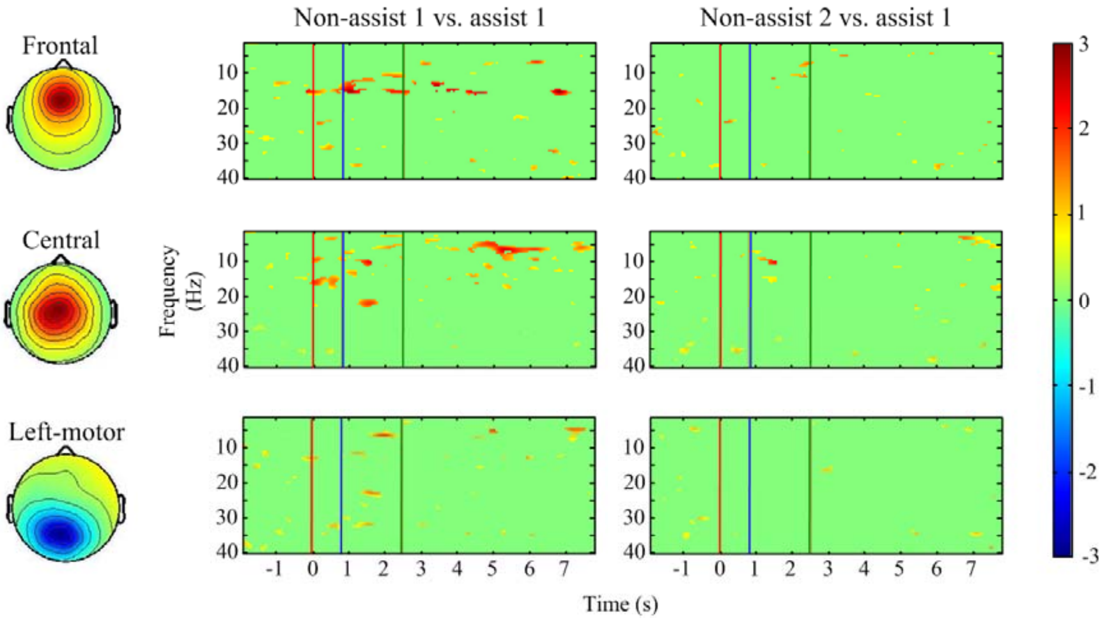

Figure 3-22: ERSP difference between different learning blocks and conditions.

3.3.3 Power Images between Two Learning Conditions

Form the above, the difference of brain dynamics between non-assist and assist conditions were mainly at alpha power of central component. In order to get the better understanding of brain dynamics at alpha power of central component, single trial power images would be investigate. First, for the relations between EMG power and EEG brain dynamic, force feedback assistance made subjects track trajectory with lower EMG power and attenuate the alpha power. Second, for the relations between learning trial and EEG brain dynamic, force feedback assistance made subjects attenuate the alpha power at the beginning of the trials, shown in the Figure 3-23.

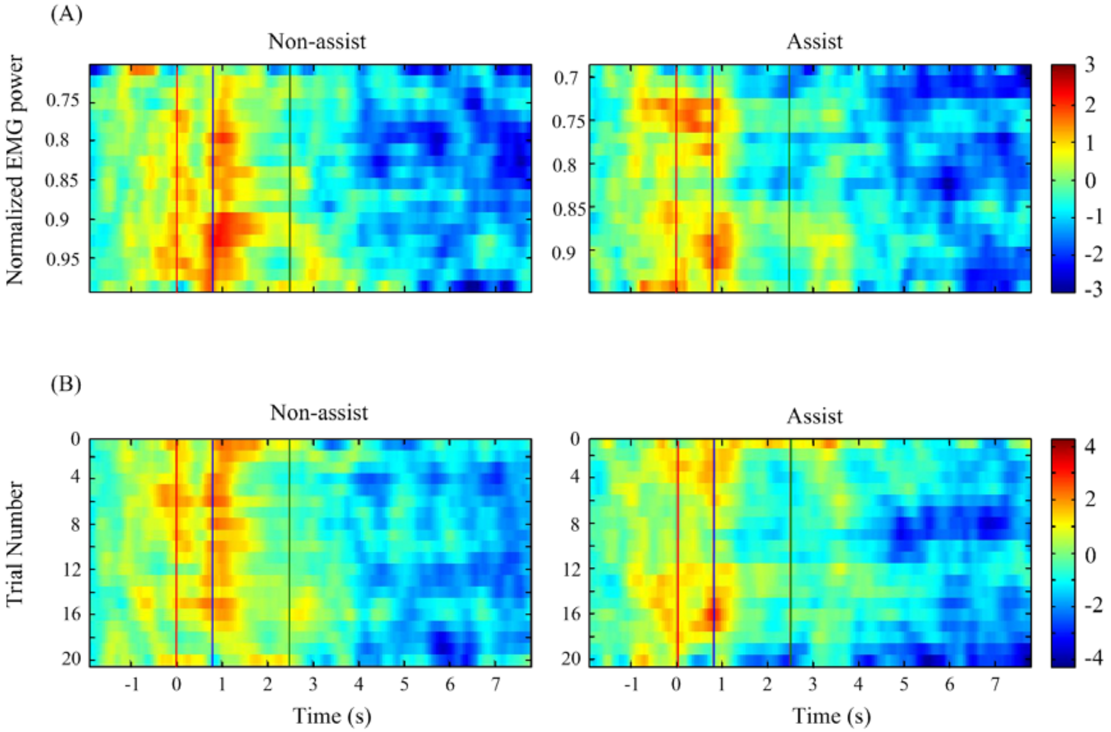

Figure 3-23: (A) Single trial power image sorted by EMG power.
(B) Single trial power image sorted by trial number.

3.4 Discussion

This study investigated muscle activity and oscillatory brain activity changes in a visually guided, continuous force feedback assistance that afforded continuous monitoring and adjustment of motor output. The results of topographical maps analysis showed the theta band power increased. Specifically, synchrony of theta oscillations has been suggested to be functionally related to temporal coordination of activity across distributed networks during mnemonic processes [67]. In contrast, the alpha and beta band power decreased maximal over central area. The central beta rhythm has a slightly more anterior focus than the central alpha rhythm, indicating different expression of these rhythmic brain activities in the motor and somatosensory cortex [68] [69]. Characteristic of rhythms within the alpha and beta band, occurring not only over the sensorimotor area but also over visual and auditory cortex, is their blockade (or ERD) when the corresponding area becomes activated. This might indicate a shift from an 'idling' to an 'active' state of cortical areas involved in a given task. Neurally, desynchronized alpha-band activity is believed to be a correlate of increased cellular excitability in interconnected thalamocortical systems [70]. Thus, it might be assumed that ERD reflects changes in local interactions between pyramidal neurones and interneurones controlling the frequency components of the EEG, while slow movement-related potentials would represent the response of cortical neurones to afferent signals [71].

The results of ERSP analysis of channel domain showed the alpha and beta band power attenuated under force feedback assistance. Since the alpha rhythm is typically attenuated by functional brain activity the above mentioned results suggest an inverse relationship between motor performance and cortical activation. This view is further supported by research applying imaging techniques such as positron emission tomography [72] or functional magnetic resonance imaging [73], whereas the fMRI studies suggest a gradual shift from cortical to subcortical brain regions as motor skills develop. Based on the well-established stage structure of sensorimotor skill acquisition, Hatfield and Hillman [74] suggested that the observed decrease in cortical activation as a function of skill level reflects a reduction of cognitive effort due to automaticity of information processing and this may minimize potential interference with visuomotor processes during task execution. The authors refer to this hypothesis as efficiency of psychomotor performance

The following, the results of the alpha and beta band power attenuated were dominated over central area. Zhuang [75] showed a decline in alpha band power that was maximal over the contralateral central region while subjects developed implicit knowledge. Hence, the mixed brain signals of channel domain were separated into different brain source activation. The further analysis which compared different learning block with force feedback assistance condition showed only the difference for non-assist 1 vs. assist 1. In the early learning block, force feedback assistance could attenuate the brain activity. The brain dynamic of force feedback assistance in the early learning block was similar to non-assistance condition in the later learning block. Hence, we could know that the brain activity also attenuated with the learning [59]. The force feedback assistance could accelerate the forming of brain function and reach to the later learning stage.

4. Conclusions

In this study, we investigated the differences of behavior performance and physiological activation under the force feedback assistance in the trajectory tracking task. The force feedback assistance provides an elegant and sufficient way to support subjects who performed a better motor performance. EMG power analysis revealed that subjects needed the lower muscle strength to perform the tracking task. Hence, subjects learned the movement more precisely and correct motor performance and muscle strength under force feedback assistance. Moreover, EMG signal can be used as the biofeedback loop to enhance training effects. Task-related EEG spectrum dynamics were analyzed using channel analysis independent component analysis, time-frequency and non-parametric static test. Our results supported the dissociation brain activity between the non-assist and assist conditions. Subjects who performed under force feedback assistance showed attenuated activation over frontal and central area when starting movement. The distinct brain power involved in the feedback learning may reflect that the different process to form the brain function for motor learning.

5. Future Works

The most exciting part of this study is that a proof of improvements in behavior and physiological activation. It merits further research to see how much those benefits can be stretched. It is possible for future experiments that studying the impact of force feedback strength and finding the best strength of force feedback assistance. There are many possible future tests and application worth mentioning.

5.1 Surgical Operation and Drafting

There are highly demands for accuracy and stability in the surgical operation and drafting. The doctor and artistic can learn the motion under force feedback assistance to improve the motor performance.

5.2 Rehabilitation

One of the original goals of this system was to provide a method of easy rehabilitation for those suffering from neurological trauma such as stroke. The tests performed show promise but no direct confirmation that such a system would benefit those recovering from such a trauma. Therefore, much needs to be done to confirm that this type of system in fact does aid those in that situation.

5.3 Motor Learning for the Blind

A still untested but likely hypothesis is that the force feedback assistance would greatly improve the blind community's ability to learn motor tasks. No official research has been performed as of yet, so this is as untapped element of research.

5.4 A Final Note

This study was to introduce a new method to improve people's live, to help them gain new abilities more effectively. The study shows great promise in this avenue. It is up to future work to utilize this new method, for everyone's benefit.

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Appendix: Experimental questionnaire

受測同意書

假如您健康狀況允許，且聽過實驗講解後，您本人是否同意參與此次實驗？

□同意　□不同意

填表者簽名：________________________

填表日期：_____ 年 _____ 月 ______日

受測者基本資料

第一部份：個人基本資料

下列各題，請就您個人背景逐一回答，於方格〝□〞內打〝V 〞或在橫線上填寫上正確資料。

1.1 姓名：____________________　1.2 性別：□ 男　□ 女
1.3 身高：_______ cm，體重：_______ kg
1.4 慣用手 :□ 左手　□ 右手
1.5 有無近視？□ 無近視 □ 有，左 _____度 右 _____度
1.6 有無老花眼？□ 無老花眼 □ 有，左 _____度 右 _____度
1.6 左右眼矯正後視力是否正常（1.0 以上）□ 是 □僅左眼正常 □僅右眼正常
1.5 教育程度 ： □ 國小　□ 國中　□ 高中/高職/五專 □ 大學 □ 研究所
1.6 出生年月日：_____年____月____日，年齡：______ 歲
1.7 聯絡電話：____________________
1.8 服務單位：_____________________
1.9 請簡述您工作內容：____
____

第二部份：過去或現在病史

於下列的問題的陳述中，按其選項選出最合適您認同的等級。謝謝！
（詢問以下問題僅是提供我們參考您是否適合作爲此次實驗對象）

Q1.您現有或曾有下列病症嗎？(可複選)

3.1 神經損傷 □ 無 □ 現有 □ 曾有 □ 治療中 □ 未治療 □ 已治療

如果有，發生於何處? ______________________________

3.2 關節炎 □ 無 □ 現有 □ 曾有 □ 治療中 □ 未治療 □ 已治療

如果有，發生於何關節? ______________________________

3.3 嚴重肌肉傷害 □ 無 □ 現有 □ 曾有 □ 治療中 □ 未治療 □ 已治療

如果有，發生於何處? ______________________________

第三部分生活型態評估

Q1.飲食篇

4.1-1 是否有飲茶習慣？ □是 □否 ________ cc 或 _______杯/一天

4.1-1 是否有喝咖啡習慣？ □是 □否 ________ cc 或 _______杯/一天

咖啡濃度爲 _________ 卡咖啡因/cc

Q2.飲酒、藥物習慣

4.2-1 您是否有飲酒習慣？（答是者續答下列各題） □是 □否

4.2-2 在上一個月，您大概有幾天有喝酒？__________________天

4.2-3 最近一次飲酒時間：_______ 年 _______ 月 ______ 日

4.2-4 你是否有使用藥物的習慣？□是 □否

Q3.運動習慣

4.3-1 您有沒有很規律的運動？ □是 □否

4.3-2 您每星期 ________ 次，每次運動時花 ________ 分鐘

4.3-3 常做哪些運動?

□ 跑步類(ex.慢跑、競走等) □ 太極拳 □ 氣功 □ 武術 □ 韻律操

□ 球類運動 □ 其它：______________________________

4.3-4 如果您有勾選跑步類的運動，您每次大約完成 ________ 公里的距離

4.3-5 寫出自己較喜歡的運動項目（至少三種）：

____

Q4.電腦或是遊戲篇

4.4-1 是否有常操作電腦或是玩遊戲（電腦或是電視）？ □是 □否

4.4-2 您每星期 ________ 天，每天操作時花 ________ 分鐘

4.4-3 您是否有常寫字或是練習書法？ □是 □否

您每星期 ________ 次，每次進行 ________ 分鐘

4.4-4 您是否常在家烹飪？ □是 □否

您每星期 ________ 次，每次進行 ________ 分鐘

實驗前填答：

1. 請問您前一晚的睡眠狀況
   □ 好　□ 中　□ 差
2. 本次實驗時間：___年___月___日___點___分　(請用 24 小時制)
3. 慣用手爲　□ 左手　□ 右手
4. 目前精神狀況 □ 好　□ 中　□ 差

受測者基本資料

姓名 ________________　年齡 _____　性別 ________

6. 您現在即將開始打膠並開始實驗，請寫下你目前的心情或心理狀態：

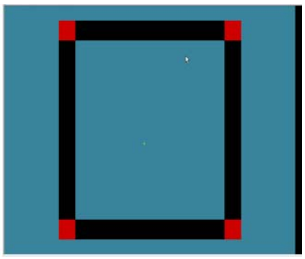
實驗中問答：

(以下問題，以程度 1～5 表示，程度 1 表示完全沒有，程度 5 表示非常嚴重)

1. 做完第一階段
   - ・Degree of difficulty(保持在中線)
   - ・Physical fatigue _____
   - ・Mental fatigue _____
2. 做完第二階段
   - ・Degree of difficulty
   - ・Physical fatigue _____
   - ・Mental fatigue _____
3. 做完第三階段
   - ・Degree of difficulty
   - ・Physical fatigue _____
   - ・Mental fatigue _____
4. 做完第四階段
   - ・Degree of difficulty
   - ・Physical fatigue _____
   - ・Mental fatigue _____
5. 是否有感受輔助力？ □ 有　□ 無

實驗後問答：

1. 實驗中時，覺得行走軌跡何處最難控制和最容易控制？（請在圖上標示出來）



2. 當有力輔助時，對自己的調控能力是否有幫助，並試著描述有何幫助？如無感受到力輔助就請您填寫無感受力輔助。

3. 在整個實驗下來，是否有感到自己的控制能力有所改變？如何改變？

4. 實驗的整個過程中，是否會感到手部會疲累？何處感到疲累？

5. 實驗過程中，是否會想要打瞌睡，感到疲累？

6. 操作儀器時，是否會感到不好操控？如有，請舉出。

7. 頭套是否會不舒服？在做實驗時，是否會感到不舒服？如有，請舉出哪裡會不舒服。

8. 眼睛是否會感到不舒服？何時？

9. 做實驗時，會感到環境太冷？或是太熱？

10. 請寫下您對本次實驗的建議與您的收穫：

11. 請問在訓練後是否認爲會幫助你日常生活的動作執行？有，請舉例說明。

◎ 謝謝您今天的參與，您的實驗數據將對腦波方面研究有重大貢獻～