行政院國家科學委員會補助專題研究計畫成果報告
期末報告

動作溢流對拔河選手手指生理震顫的影響

計 畫 類 別 ： 個別型計畫
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執 行 單 位 ：中山醫學大學物理治療學系

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中 華 民 國 103年02月10日

中文摘要： 動作溢流是一側肢體動作造成對側肢體非自主性的肌肉活化現象，也是交叉遷移訓練重要的生理基礎。動作溢流在肌肉疲勞發生時尤為明顯，並與個人所受的運動訓練與動作學習經驗有關。由於生理震顫較傳統肌電圖能更敏銳地反應疲勞特徵，更可以便利地評估個別手指的控制反應。本研究以手指生理震顫、肌電圖、與施力量測，探討在單側進行握力時，動作溢流造成對側手指生理震顫特徵的改變的現象。發現由於拔河選手的特定主要手部肌肉群需要特殊長期重量訓練，除了局部肌肉組織肥大之外，可能伴隨大腦塑性與傳出神經溢流的改變；拔河選手長時間的握力訓練將造成此拔河選手動作溢流量以及手指間活動自由度的改變，特別是進行需要高度專注力的活動。本計畫對拔河選手動作溢流的深入瞭解，不僅具學理上的參考價值，同時也有潛在運動傷害的應用價值，使患側肢體可利用健側活動在養傷期間獲得最大的肌力保留。

中文關鍵詞： 拔河、動作溢流、生理震顫、等長收縮

英文摘要： Contralateral motor overflow is an involuntary muscle activation associated with strenuous contralateral movement, subserving to a physiological basis of cross education in muscle strengthening. Motor overflow manifests with muscle fatigue, relating to previous motor training and movement experience. As the tug-of-war athletes receive special strengthening on hand muscles, we may well expect local tissue hypertrophy and adaptive change in brain connectivity that leads to efferent neuronal outflow. On account of superior sensitivity in detection of fatigue characteristics and inter-digit movement enslaving, the purpose of the project was to investigate the difference in motor overflow for the tug-of-war athletes using physiological tremors. We found contralateral motor overflow of the tug-of-war athletes contribute to different organizations of digit physiological tremors as compared to that of the healthy counterparts, especially when the motor task requiring high attentional focus is being conducted. In addition to advancing theoretical understanding of tugger’s motor overflow, the practical application of the finding is to realize preservation of muscle strength of the affected limb

using optimal contralateral activity.

英文關鍵詞： Tug-of-war, motor overflow, physiological tremor, isometric contraction

行政院國家科學委員會補助專題研究計畫成果報告

（□期中進度報告/■期末報告）

動作溢流對拔河選手手指生理震顫的影響

Contralateral motor overflow and restructuring of digit physiological tremors for the tug-of-war athletes

計畫類別：■個別型計畫　　□整合型計畫

計畫編號：NSC　101-2410-H-040 -017 –

執行期間： 101　年 8　月 1　日至 102　年 10　月 31　日

執行機構及系所：中山醫學大學 物理治療學系

計畫主持人： 陳怡靜 助理教授

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中　華　民　國　103　年　1　月 31　日

一、中英文摘要

動作溢流是一側肢體動作造成對側肢體非自主性的肌肉活化現象，也是交叉遷移訓練重要的生理基礎。動作溢流在肌肉疲勞發生時尤為明顯，並與個人所受的運動訓練與動作學習經驗有關。由於生理震顫較傳統肌電圖能更敏銳地反應疲勞特徵，更可以便利地評估個別手指的控制反應。本研究以手指生理震顫、肌電圖、與施力量測，探討在單側進行握力時，動作溢流造成對側手指生理震顫特徵的改變的現象。發現由於拔河選手的特定主要手部肌肉群需要特殊長期重量訓練，除了局部肌肉組織肥大之外，可能伴隨大腦塑性與傳出神經溢流的改變；拔河選手長時間的握力訓練將造成此拔河選手動作溢流量以及手指間活動自由度的改變，特別是進行需要高度專注力的活動。本計畫對拔河選手動作溢流的深入瞭解，不僅具學理上的參考價值，同時也有潛在運動傷害的應用價值，使患側肢體可利用健側活動在養傷期間獲得最大的肌力保留。

關鍵詞： 拔河、動作溢流、生理震顫、等長收縮

Contralateral motor overflow is an involuntary muscle activation associated with strenuous contralateral movement, subserving to a physiological basis of cross education in muscle strengthening. Motor overflow manifests with muscle fatigue, relating to previous motor training and movement experience. As the tug-of-war athletes receive special strengthening on hand muscles, we may well expect local tissue hypertrophy and adaptive change in brain connectivity that leads to efferent neuronal outflow. On account of superior sensitivity in detection of fatigue characteristics and inter-digit movement enslaving, the purpose of the project was to investigate the difference in motor overflow for the tug-of-war athletes using physiological tremors. We found contralateral motor overflow of the tug-of-war athletes contribute to different organizations of digit physiological tremors as compared to that of the healthy counterparts, especially when the motor task requiring high attentional focus is being conducted. In addition to advancing theoretical understanding of tugger's motor overflow, the practical application of the finding is to realize preservation of muscle strength of the affected limb using optimal contralateral activity.

Keywords: Tug-of-war, motor overflow, physiological tremor, isometric contraction

二、前言

動作溢流(motor overflow)是當一側肢體關節發生動作時，對側肢所產生同步非自主性的肌肉活化現象。動作溢流普遍發生於兒童、老年人與正常成人 (Parlow, 1990; Armatas, 1994; Mayston et al., 1999)，尤其是從事費力、困難的作業、動作疲勞、學習新運動技能時，動作溢流的產生最為明顯 (Lazarus 1992)；動作溢流過去經常被負向看待，被認為是造成動作障礙、對側關節靈活度降低的因素；但是近年來，在運動科學的研究發現動作溢流可以在肌肉重量訓練時，增加對側肢體肌肉活化的正面功能 (Shima et al., 2002; Seger & Thorstensson et al., 2005; Gabriel et al., 2006; Lee & Carroll 2007)；Shields 等人(1999)將年輕健康男性受試者，進行六週的右手握力耐力訓練，研究結果發現：即使是單側的重量訓練，研究發現對側的肌肉力量也會有明顯的增加，稱為肌力訓練的交叉遷移效應(cross education) (Hortobagyi et al.,

1997)。交叉遷移效應似乎受到肌肉收縮型式的影響，等速離心訓練組具有較高交叉遷移效果(Mahler, 1995)，同樣地，高速離心收縮訓練較低速離心收縮訓練有更明顯的交叉遷移效果(Brandenburg & Docherty, 2002; Farthing & Chilibeck, 2003)。由於拔河選手的特定主要手部肌肉群需要特殊長期重量訓練，除了局部肌肉組織肥大之外，可能伴隨大腦塑性與傳出神經溢流(efferent neuronal outflow)的改變。

三、研究目的

本研究計畫利用生理震顫、肌電圖、與施力量測，了解拔河選手的手部動作溢流與正常人的差異。計劃目的是探討「正常人與拔河選手在單側進行握力時，動作溢流造成對側手指生理震顫特徵的改變」。探討拔河選手長時間的握力訓練是否造成此拔河選手動作溢流量以及手指間活動自由度的改變，特別是進行需要高度專注力的活動時。

四、文獻探討

動作溢流發生的原因與胼胝體徑路(transcallosal pathway)活化有很大的關連性(Green, 1967; Zülch, 1969; Shimizu et al., 2002)；當單側大腦半球興奮時，傳遞至另一側的大腦半球的興奮量可藉由胼胝體徑路抑制機制(transcallosal inhibition)調整，當胼胝體徑路抑制高時，動作溢流少，對側大腦半球興奮相對低；反之，當胼胝體徑路抑制低時，動作溢流多，對側大腦半球興奮高。研究發現經常性重量訓練可降低胼胝體抑制，使兩側大腦半球運動區同步興奮，促進交叉遷移效果的發生。交叉遷移的概念目前已被應用在運動傷害的復健，當一側肢體因傷必須固定時，可以加強重量訓練在對側，使患側的肌力在養傷時期獲得最大的保留(Carroll et al., 2006; Gabriel et al., 2006)。

生理震顫是一種有規律性、但不完全自主的微小肢體活動，可用加速規 (accelerometer）或肌電圖來量測。近年來，生理震顫逐漸引起國內外運動控制學者的興趣，主要的原因是生理震顫訊號不僅容易測量，而且與運動表現有高度的相關性。例如：比較不同技術水準手槍射擊選手發現，有較好射擊表現的選手在瞄準期上肢生理震顫振幅明顯較小，且生理震顫的方向性也與射擊成績較差的選手不同(Tang et al., 2008)。同側肢體的活動會經由動作溢流的效應影響對側肢體生理震顫的結構與特徵；在正常年輕人利用非慣用手作最大程度用力的握拳動作時，對側慣用手的伸直手指生理震顫的振幅會增加，代表最大用力時動作溢流所產生的興奮作用，會擴展至對側慣用手；而且不同手指間的生理震顫交互訊息量(mutual information)會增高,使不同手指的生理震顫耦合程度變高,意謂動作溢流造成手指間動作自由度(degree of freedom in joint space)下降，手指活動的獨立性也受到影響。有趣的是主修鋼琴的大學音樂系學生，生理震顫的變化卻得到完全相反的趨勢；音樂系學生非慣用手單邊最大程度用力時，對側手指生理震顫不僅下降，而且指間生理震顫交互訊息量更加減少，可能是長期練習鋼琴使音樂系學生大腦塑性改變，利用未完全被確認的神經機制抑制動作溢流，使手部活動不受對側施力的干擾，提高兩手活動的自由度與獨立性(Chen et al., 2011)。

正式的國際拔河比賽起源自英國，團隊的拉力與個人肌力都是拔河比賽決勝的關鍵因素。拔河選手的主要肌肉群包括：肱二頭肌、胸大肌、三角肌、肱三頭肌、背闊肌、股四頭肌、前臂肌群等(邱定璿, 2007)；上述肌肉群都需要特別給予重量訓練，養成充足的肌力與耐力以應付比賽過程長時間的等長收縮(isometric contraction)(涂瑞洪,1997)。國內外的文獻均指出：拔河選手的握力明顯大於同年齡的一般人(涂惠芳，2005; Warrington et al., 2001; Smith & Krabak, 2002)。選手加強肌力訓練除了增加目標肌肉的肌蛋白形成與肥大肌纖維組織之外，

很可能也改變大腦傳出神經溢流(efferent neuronal outflow)、降低大腦運動區抑制性神經路徑的活動、並增加脊髓運動元的興奮性(Aagaard, 2003;Adkins et al., 2006)。

五、研究方法

(1) 受測者:

招募了16 位年滿18 歲沒有規律運動習慣的健康大學生(控制組) 與16 位年滿18 歲至少3年經驗的拔河選手(拔河選手組)參與實驗，受試者排除有神經肌肉的症狀或徵兆者。

(2) 實驗流程

在了解本研究計畫，同意參與本研究後，簽署受測者同意書。全部受測者於實驗開始時，先進行非慣用手最大握力(maximal voluntary contraction, MVC)測試，以最大力量抓握測力器3秒，決定最大握力。之後，受測者以非慣用手，依隨機順序執行放鬆不用力抓握(relax)、正弦抓握(sinusoidal isometric grip)、以及定量等長抓握(constant isometric grip) 等三種測試各三次；受測者舒適的坐在椅子上，且慣用手手腕和前臂固定於桌上的熱塑性副木上(圖１)。非慣用手則自然垂放身旁，以手部握住測力器。在定量等長抓握情況下，受測者將非慣用手手指彎曲施力握壓測力器產生力量輸出，穩定地施予75%最大抓握力量(75% MVC)，並維持20秒。正弦抓握情況下，受測者將非慣用手手指彎曲握住測力器，控制抓握力量在50%-100%最大抓握力量範圍50%-100% MVC)之間，以跟隨頻率0.6 Hz 之目標正弦波，收縮時間共計20秒；在放鬆不用力抓握情況下，非慣用手手指彎曲輕微扣住測力器而不施力，時間20 秒。本計劃採非慣用手的原因是預期為比慣用手更能產生較大的對側動作溢流。

在三種抓握測試情形下，受測者同時維持慣用手展開，即全部手指伸直指向前方並平行於地面，伸直之手指無任何支撐。五個加速規測量，固定在大拇指、食指、中指、無名指和小指的遠端指節的背面上，測量慣用手手指垂直方向的生理震顫。手指的震顫活動訊號十倍放大後被記錄。非慣用手與慣用手的伸指肌和屈指淺肌的肌肉活動分別被雙極表面的電極片所同時紀錄，伸指肌的電極片被置於手腕至手肘距離約四分之三處並可觸摸到肌肉之處，屈指淺肌的肌肉活動被一置於斜向、手肘內側下方大約四公分並可觸摸到肌肉之處的電極片所紀錄。全部的訊號被LabVIEW 平台上之custom program (LabVIEW v.8.5; National Instruments, Austin, TX)所呈現與記錄，抽樣頻率為1K 赫茲。

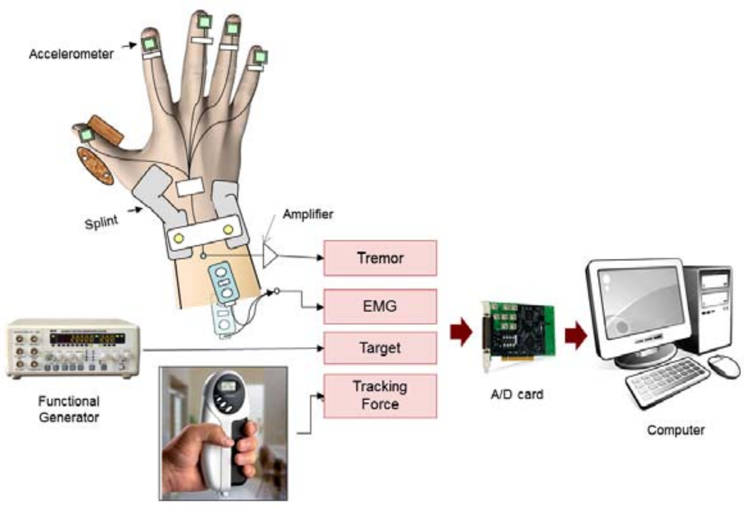


圖 1. 實驗整合系統示意圖，包含加速規(accelerometer)、肌電圖(EMG)、訊號產生器(functional generator)、目標視覺迴授訊號(target signal)、以及測力計系統。非慣用手施力作用於測力計(tracking force)，而同步量測慣用手手指的姿勢生理震顫。

(3) 訊號與資料處理

握力表現分析：在不同收縮情形下，測力系統所量測握力之平均值與均方根值，代表受試者努力程度。

肌電圖分析：主要收集訊號有執行維持姿勢慣用手、執行握力作業非慣用手之屈指肌、伸指肌，將四條肌肉的肌電圖訊號以帶通濾波器(截止頻帶：1, 400 Hz)濾波後，求取均方根值(root mean square, RMS)代表肌電訊號的強度，並經傅立業轉換後估計中位頻率(median frequency, MDF)，代表肌肉徵召運動單元(motor unit recruitment) 的策略。

生理震顫分析：計算均方根值代表生理震顫的強度；計算任何兩個手指生理震顫的交互訊息(mutual information)，交互訊息代表兩手指震顫時間序列的相同性質出現機率，交互訊息數值越高代表指間震顫有較高的相關性或受共同因素的影響較大，當動作溢流明顯，指間震顫有較高的交互訊息。

(4). 統計分析：

本研究主要關注慣用手的手指生理震顫、肌電圖各項參數是否受到非慣用手施力作用型式(定量等長收縮vs.正弦收縮)與族群間(拔河選手vs.控制組)而有不同。以二因子混合變異數分析(two way ANOVA)分析上述各項參數的顯著性，顯著水準p定為0.05，並視顯著性結果採適當之事後檢定。

六、結果與討論

拔河組共徵招16位業餘級拔河選手參與此研究(全為男性，年齡: 21.4 ±0.5歲; 體重: 81.9 ± 6.1公斤; 身高: 178.8 ± 4.6 公分) 。 這些選手來自大學的拔河隊，除了參與比賽之外每天進行拔河相關訓練約3-6小時，相關訓練資歷約3-7年。 控制組為16位沒有規律運動的大學生(全為男性，年齡: 21.3 ± 0.6歲; 體重: 63.6 ± 7.5公斤; 身高: 173.7 ± 3.2公分) 。所有測試者的慣用手皆為右手，且無任何神經肌肉系統問題的症狀。

握力表現分析

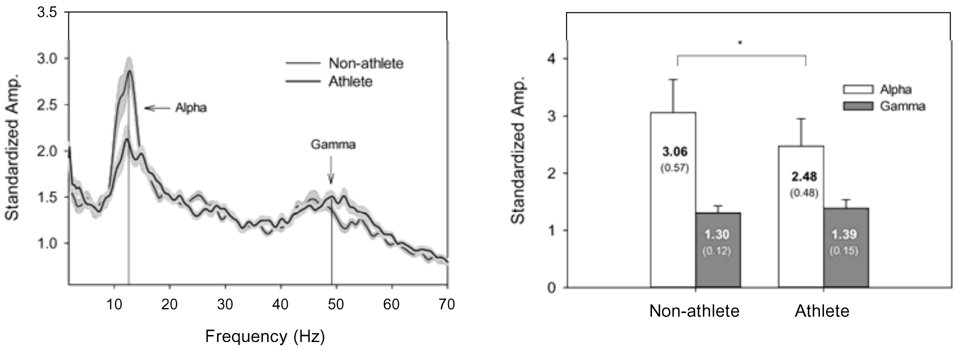
Independent t-test 的結果顯示在拔河組與控制組兩組間的最大握力值 (maximal gripping force) (t_{30} = -6.970, P < .001) 與最大握力與體重比值 (ratio of maximal gripping force to body mass) (t_{30} =-2.107, P = .044) 有顯著的不同. 拔河選手相較於健康大學生有較大的最大握力(拔河組:264.9 ± 42.6 N; 控制組: 170.3 ± 33.6 N) 與最大握力與體重比值(拔河組: 3.25 ± 0.59 N/Kg; 控制組: 2.73 ± 0.78 N/Kg) 。

表1為拔河組(tugger)與控制組(control)的在執行75%最大握力等常收縮時的握力表現與握力變異性分析。兩組之間在握力分析的振幅的均方根值的均方根值與握力變異性(即握力與目標握力之間的差異)的振幅的均方根值之間有統計上的差異，然而振幅的均方根值與握力變異性的振幅之間的比值在兩組間則沒有統計上的差異。此結果跟之前的研究相符(Warrington et al., 2001)，事實上，對於拔河選手在進行拔河比賽的過程中具有高強度的手抓握力是重要的。因為專業運動的特殊性，而造成手抓握力的增加也常見於手球(Del Percio eet al., 2010; Visnapuu and Jürimäe, 2007)、籃球(Visnapuu and Jürimäe, 2007)與柔道等選手中(Leyk et al., 2007)。

Amplitude variable	control	tugger	statistics
RMS_ME (N)	126.59 ± 19.00	194.75 ± 29.93††	$t30$ = -2.776, P = .009
RMS_FV (N)	3.06 ± 1.24	6.00 ± 2.60†††	$t30$ = -6.244, P < .001
RME/FV	51.62 ± 16.43	47.18 ± 17.55	$t30$ = -.487, P = .617

表1、 拔河組(tugger)與控制組(control) 左手在執行75%最大握力等常收縮時的握力表現與握力變異性分析。 (RMS_ME: root mean square of mean exertion; RMS_FV: root mean square of force fluctuation profile; RME/FV denotes amplitude ratio of the mean exertion to force variability)(††: Athlete > Non-athlete, *P* < .01; †††: Athlete > Non-athlete, *P* < .001)

肌電圖分析



(A) (B)

圖2、拔河組(athlete)與控制組(non-athlete) 左手在執行75%最大握力等常收縮時曲指淺肌 (flexor digitorum superficialis) 肌電圖的頻譜分析。(A) Pooled spectral profiles with the shaded area representing confidence interval of 1 standard error in the pooled standardized spectral profiles. (B) The means and standard deviations of standardized amplitude for 8-12 Hz and 35-50 Hz spectral peaks. (*: Non-athlete > Athlete, *P* < .05)

分析拔河組與控制組左手在執行75%最大握力等常收縮時曲指淺肌肌電圖的頻譜,發現頻譜中包含一個alpha spectral peak (8-12 Hz)與一個gamma spectral peak (40-60 Hz),其中gamma spectral peak的振幅較alpha spectral peak小一點。Independent t-test分析的結果顯示: EMG alpha oscillation有組間的差異(t_{30} = 2.367, *P* = .025),控制組有較大的alpha spectral peak,但是gamma spectral peak則沒有統計上組間的差異(t_{30} = -1.258, *P* = .218)。有關EMG頻譜中8-12 Hz的震盪被認為可能代表生理性震顫(physiological tremor) (Elble and Randall, 1976),也可以加速規測量(Hwang et al., 2009; Keogh et al., 2004)或EEG-rectified EMG 共譜(coherence spectra)分析測出(Safri et al., 2006)。此8-12 Hz的震盪可能是中樞對感覺動作皮質(sensorimotor cortex) 到脊髓動作神經元的共同驅動 (common central drive)促使運動單元活動同步化(Safri et al., 2006; Stegeman et al., 2010)。

生理震顫分析

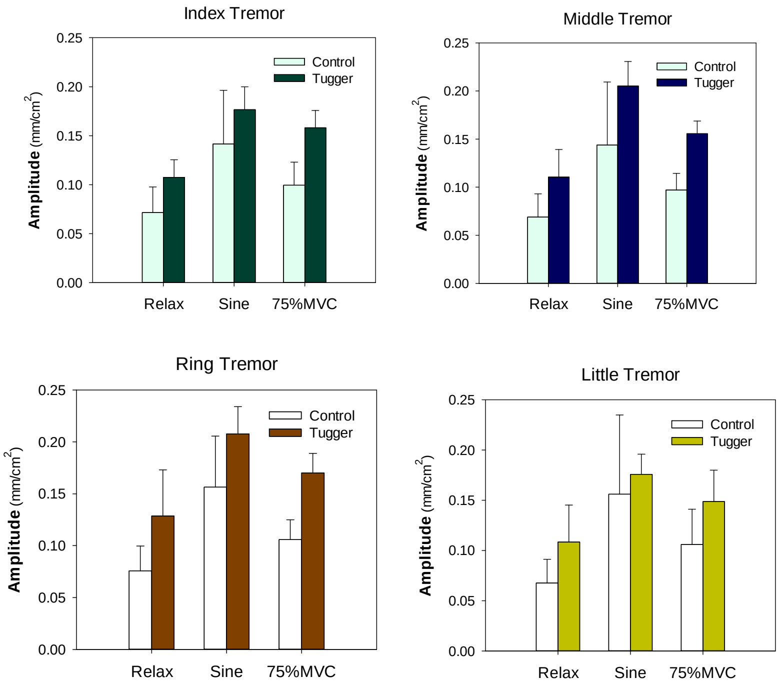


圖 3、慣用手手指在三種實驗情形（放鬆(relax)、正弦收縮(sine)、定量等長收縮(75%MVC))下的生理震顫均方根值與標準差。

拔河選手調節動作溢流的機制與正常人出現不同的趨勢。由圖 3 所示，在非慣用手執行正弦收縮與 75% MVC 定量等長收縮時，相較於放鬆的情形下的手指生理震顫，慣用手從食指到小指都出現較大生理震顫震幅的趨勢，而且正弦收縮似乎產生最大生理震顫以及動作溢流。同時資料也顯示生理震顫震幅有可觀的族群差異。拔河選手不論在何種實驗狀況下，動作溢流都引起的較大生理震顫。

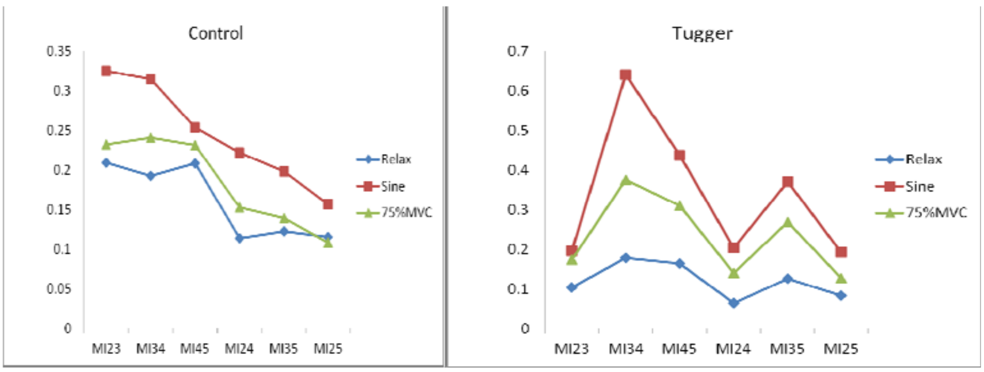


圖 4、慣用手手指在三種實驗情形(放鬆(relax)、正弦收縮(sine)、等量等長收縮(75%MVC))下的指間生理震顫交互訊息值。(MI23: tremor mutual information between the index and middle fingers, MI34: tremor mutual information between the middle and ring fingers, MI45: tremor mutual information between the ring and little fingers, MI35: tremor mutual information between the middle and little fingers, MI24: tremor mutual information between the index and ring fingers, MI25: tremor mutual information between the index and little fingers)

由圖 4 所示，在非慣用手執行正弦收縮與 75% MVC 定量等長收縮時，對側的慣用手呈現較大的指間生理震顫交互訊息值(mutual information, MI)，而且正弦收縮似乎產生最大指間生理震顫交互訊息值。資料呈現族群間的交互訊息值存在可觀的差異。拔河選手在握力實驗

狀況下都有較大的較高指間生理震顫的相關性，然而在放鬆的情形下，此趨勢並不明顯，代表拔河選手的動作溢流相對於正常受試者來說，可使手指間活動的相關性更為增加。

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八、附錄

說明: 本研究控制組有關握力表現分析已撰寫投稿，並獲接受發表。其他相關資料也將陸續撰寫成論文投稿。

發表論文:

Yi-Ching Chen, Yen-Ting Lin, Chien-Ting Huang, Chia-Li Shih, Zong-Ru Yang, Ing-Shiou Hwang (2013, Sep). Trajectory Adjustments Underlying Task-Specific Intermittent Force Behaviors and Muscular Rhythms. *PLoS ONE*, 8(9): e74273. **(SCI)**[**IF: 3.730**; Multidisciplinary science, Ranking: 7/56 (12.5%)] NSC 101-2410-H-040-017. (本人為第一作者)

Trajectory Adjustments Underlying Task-Specific Intermittent Force Behaviors and Muscular Rhythms

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Abstract

Force intermittency is one of the major causes of motor variability. Focusing on the dynamics of force intermittency, this study was undertaken to investigate how force trajectory is fine-tuned for static and dynamic force-tracking of a comparable physical load. Twenty-two healthy adults performed two unilateral resistance protocols (static force-tracking at 75% maximal effort and dynamic force-tracking in the range of 50%–100% maximal effort) using the left hand. The electromyographic activity and force profile of the designated hand were monitored. Gripping force was off-line decomposed into a primary movement spectrally identical to the target motion and a force intermittency profile containing numerous force pulses. The results showed that dynamic force-tracking exhibited greater intermittency amplitude and force pulse but a smaller amplitude ratio of primary movement to force intermittency than static force-tracking. Multi-scale entropy analysis revealed that force intermittency during dynamic force-tracking was more complex on a low time scale but more regular on a high time scale than that of static force-tracking. Together with task-dependent force intermittency properties, dynamic force-tracking exhibited a smaller 8–12 Hz muscular oscillation but a more potentiated muscular oscillation at 35–50 Hz than static force-tracking. In conclusion, force intermittency reflects differing trajectory controls for static and dynamic force-tracking. The target goal of dynamic tracking is achieved through trajectory adjustments that are more intricate and more frequent than those of static tracking, pertaining to differing organizations and functioning of muscular oscillations in the alpha and gamma bands.

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Introduction

Visuomotor tracking is a critical function of the motor system. However, intrinsic trajectory control is affected by variations in the state of the motor system [1,2], since motor responses are not strictly smooth. A larger size of force variability greatly drifts the force output away from an intended priori standard. The complexity of force variability is another dimension of force variability [3,4], typically indexed with entropy measures [3,5] to characterize the degree of fluctuation predictability over a force data stream [6,7]. The size and the complexity of force variability of a visuomotor task can be differently organized. For instance, tracking with visual feedback is more accurate and has a smaller size but a greater complexity of force variability than tracking without visual feedback [1,3,8]. An increase in force complexity is related to engagement of trajectory adjustments using on-line sensory inputs, rather than to task degradation [1,9]. One of the major sources of force or kinematic variability comes from sampled feedback processes of the visuomotor system [10,11] for enhancing the stability of the visuomotor system against long feedback delays [10,12]. However, sampled feedback brings about movement intermittency, as manifested with discrete blocks of pulse-like elements in movement trajectory. Movement intermittency becomes less evident in pursuit of a predictable target [13,14] or removal of visual feedback [10]. Both kinematic and force profiles exhibit intermittency, which is related to internal coding of the planned trajectory and error correction [14,15].

Exertion level is a key factor of force variability underlying progressive recruitment of fast-twitch motor units [16] and variations in code rating [17]. On account of an exertion-dependent increase in force variability [18,19], precise control of force is far more difficult at a higher force range than at a lower force range. Force stability at a higher force range presumably relies on task-dependent variations in code rating in that motor units are largely recruited [20]. As movement accuracy at large force output is insufficient for precision tasks, force scaling at a higher force range is often overlooked. Little attention has been paid to contrasting force variability properties between static and dynamic force-tracking at relatively high exertion levels. It is apparent that static and dynamic force-tracking challenge the visuomotor system to different extents, including visual information load [21], proprioceptive inputs [22], target constraints to produce the criterion force [23], and so on.

The present study sought to contrast the size and complexity of force intermittent behaviors between static and dynamic force-tracking at relatively high exertion levels of equivalent physical loads. Because of the different time and target constraints, we expected intermittent force behavior and the scaling property of individual force pulse for the two force-tracking tasks to be task-dependent. Another focus of this study was to explain the task-dependent intermittent force behavior with oscillatory activities in the working muscle. It was hypothesized that, in comparison to static force-tracking, dynamic force-tracking would lead to larger force intermittency and a smaller amplitude ratio of the a priori standard of intended pursuit relative to force intermittency, greater complexity and spectral dispersion of the force intermittency profile, and greater force pulse metrics with different statistical properties. In addition, muscular oscillations during static and dynamic force-tracking were differently organized with respect to tracking protocols. Our observations on force intermittency dynamics and muscular oscillations extend previous work to gain better insight into how force trajectories are planned to satisfy differing task needs.

Methods

Ethics Statement

The research project was approved by an authorized institutional human research review board (Chung Shan Medical University Hospital Institutional Review Board, CSMUH IRB), and all subjects signed informed consents before the experiment, conforming to the Declaration of Helsinki.

Subjects

Twenty-two male subjects (mean: 21.6±1.2 years) from a local community and a university participated in this study. All of the subjects were self-reported as being right-handed, and none of them had symptoms or signs of neuromuscular diseases.

Experiment Procedures

This study employed two unilateral resistance protocols of gripping, static and dynamic force-tracking. Each protocol consisted of three trials of 20 seconds, which were randomly completed by our participants with inter-trial periods of rest of at least 3 minutes. The subject sat on a chair with the left arm hanging naturally by the trunk and gripped a hand dynamometer (sensitivity: 0.01 N, bandwidth: DC–1 kHz, Model 9810P, Aikoh, Japan) connected to an analog amplifier (Model: PS-30A-1, Entran, UK). The force output and the target curve were displayed on a computer monitor to guide the force exertion of the force-tracking maneuver. Before the experiment, all subjects first performed 3 maximal voluntary contractions (MVC) of 3 seconds, separated by 3-minute pauses. The mean of the maximal force for the 3 MVCs was defined as the peak gripping force. During static force-tracking, the subjects needed to produce a constant force of 75% of peak gripping force with the aid of visual feedback. Dynamic force-tracking required the subjects to exert a load-varying isometric force to couple a 0.5 Hz sinusoidal target wave in the range of 50%–100% of peak gripping force. The target signal moved vertically in a range of 7.2° of visual angle (i.e., 3.6° above and 3.6° below the eye level on the screen), and visual feedback gain in terms of visual angle per MVC was identical for static and dynamic force-tracking. Muscle activity of the left flexor digitorum superficialis (FDS) was recorded by surface electromyography. A bipolar surface electrode unit (1.1 cm in diameter, gain = 365, CMRR = 102 dB, Imoed Inc., USA) was placed at an oblique angle approximately 4 cm above the wrist on the palpable muscle mass. All signals were sampled at 1 kHz by an analog-to-digital converter with 16-bit resolution (DAQ Card-6024E; National Instruments Inc., Austin, TX, USA), controlled by a custom program on a Labview platform (Labview v.8.5, National Instruments Inc., Austin, TX, USA).

Data Processing

The size and complexity of the force intermittency profile. Gripping force was down-sampled to 100 Hz in off-line analysis and then conditioned with a low-pass filter (cut-off frequency: 6 Hz) [24]. Mean gripping forces of an experimental trial for both force-tracking paradigms were determined. Then force output of the tracking tasks was dichotomized into two different force components, primary movement and force intermittency profile, akin to the algorithms proposed by Roitmen et al. (2004) and Selen et al. (2006) [25,26]. In brief, the primary movement was a smooth and deterministic force component of the force-tracking task, spectrally identical to the target rate. Also, the primary movement approximated target movement in amplitude. Therefore, the primary movement symbolizes the a priori standard of intended pursuit to couple the target signal. On the other hand, the force intermittency profile was a stochastic force component that contributed to force variability. The force intermittency profile was irregular, containing a number of individual force pulses (Fig. 1A). Recent studies have validated that force pulses are not noises, but part of an additive accuracy control to remedy tracking deviations from the target trajectory [26,27]. The dichotomy of gripping force was helpful to specify structural changes in the force intermittency profile (force variability) and to differentiate task effects on deterministic and stochastic force components for static and dynamic tracking. For the static force-tracking, the primary movement was a force level of 75% MVC. The force intermittency profile of static force-tracking could be obtained by removing the linear trend of the force time series (Fig. 1A, left). For the dynamic task, the primary movement was a 0.5 Hz sinusoidal wave with amplitude roughly in the range of 50%–100% MVC. The force intermittency profile of dynamic force-tracking was obtained by conditioning the force output with a zero-phasing notch filter that passes all frequencies except for a target rate at 0.5 Hz (Fig. 1A, right). The transfer function of the notch filter was $H(Z)=\mathrm{b}_0\frac{(1-e^{j\omega_0}z^{-1})(1-e^{-j\omega_0}z^{-1})}{(1-re^{j\omega_0}z^{-1})(1-re^{-j\omega_0}z^{-1})}$, $r=.9975$, $\omega_0=\pi/360$. Subtracting the force intermittency profile from the dynamic force output gave the sinusoidal component of the target rate in the gripping force, previously described as the primary movement for the dynamic task.

Root mean square (RMS) was applied to the primary movement and the force intermittency profile to calculate the amplitudes of the two force components. The RMS of the force intermittency profile symbolized the size of force variability. The amplitude ratio of the primary movement to force intermittency ($R_{PM/FI}$) was defined as the RMS of the primary movement divided by the RMS of the force intermittency profile. Spectral distribution of the force intermittency profile was estimated with the Welch method and a fast Fourier transform with a spectral resolution of 0.1 Hz. Mean frequency and spectral dispersion (spectral ranges between the 10th and 90th percentiles of the power spectra) were determined from the force intermittency spectral profile. The complexity of the force intermittency profile (i.e., the complexity of force variability) was quantified with multi-scale entropy (MSE) to reveal a sample entropy (SampEn) curve across different time scales (Appendix S1) [6,28]. Each time scale represented 10 ms for the sampling rate of 100 Hz. MSE areas under the time scales 1–25 (or 10–250 ms) and 26–60 (or 260–600 ms) were empirically determined to measure the complexity of the force intermittency profile on short and high time scales, respectively. The MSE area of the overall time scale of 1–60 was the sum of MSE areas under the time scales 1–25 and 26–60. A higher MSE area indicated a noisier structure with greater signal complexity.

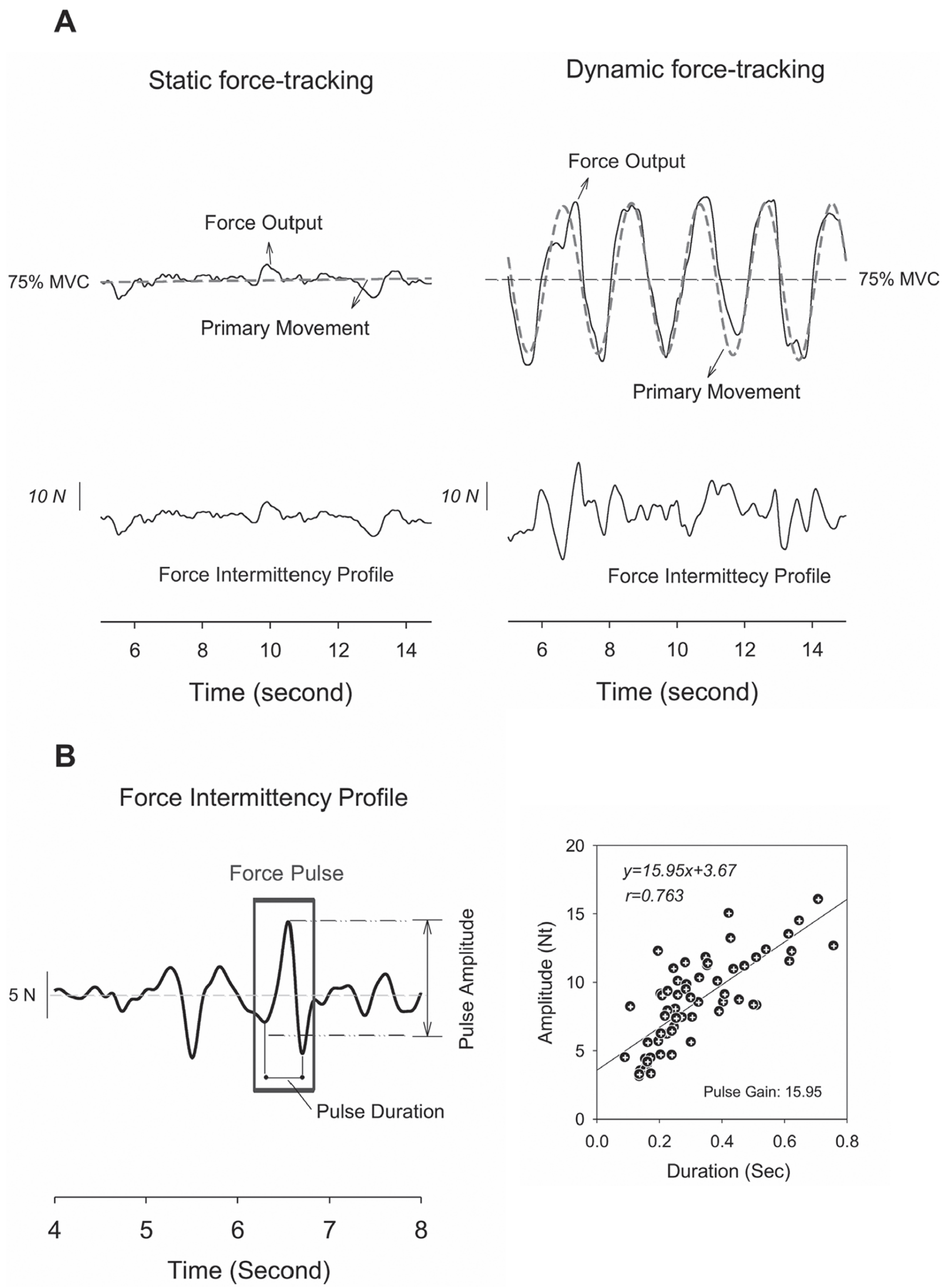

Figure 1. Illustrative examples of force intermittency profile, primary movement, and force pulse. (A) Feature extraction of force intermittency profile and primary movements from force outputs of static and dynamic force-tracking. (B) Representative force intermittency profile during dynamic and static tasks.
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Force pulse variables. Individual force pulses in a force intermittency profile were identified afterwards. Local peak in the force intermittency profile was defined as a force pulse, and a force intermittency profile contained many force pulses. Amplitude of each force pulse was the difference between a local maximum and the average value of the two nearest minima (Fig. 1B, left) [24,25]. The pulse duration was the time between two successive local minima in the force intermittency profile. For each subject, we characterized the pulse amplitude and duration of each pulse in a force intermittency profile during static and dynamic force-tracking and then calculated the probability distribution of pulse amplitude and pulse duration to get mean pulse amplitude and mean pulse duration. Linear regression between the pulse duration and pulse amplitude in a force intermittency profile provided a duration-amplitude regression slope, or pulse gain (Fig 1B, right) [24,25]. Force pulse gain of the three experiment trials during static and dynamic force-tracking was averaged across the subjects.

EMG variables. EMG of the FDS muscles were conditioned with band-pass filters (pass band for EMG: 1∼400 Hz). The amplitude of the EMG of the FDS muscles for the entire period of a trial was represented with RMS. The EMG data after band-pass filtering were rectified for spectral analysis [29]. Rectification of

Table 1. The contrast of amplitude variables of the primary movement and force intermittency between static and dynamic tracking.

Amplitude variable[1]	Static	Dynamic	*Statistics*
RMS_PM (N) [3]	126.23±4.97	124.19±4.36	
RMS_FI (N) [4]	3.49±0.32	5.68±0.26***	Λ =0.032, P=.000 [2]
$R_{PM/FI}$ [5]	48.01±3.49†††	22.52±0.51	

[1]Values were presented as mean ± se.
[2]Post-hoc for static force-tracking vs. dynamic force-tracking (***: Dynamic > Static, P<.001; †††: Static > Dynamic, P<.001).
[3]RMS_PM: root mean square of primary movement.
[4]RMS_FI: root mean square of force intermittency profile.
[5]$R_{PM/FI}$ denotes amplitude ratio of the primary movement to force intermittency.
doi:10.1371/journal.pone.0074273.t001

surface EMG is believed to enhance the spectral peaks that symbolize common oscillatory inputs or the mean firing rate of an active muscle [30,31,32]. The power spectra of the both un-rectified and rectified EMG signals were computed using Welch's method. A Hanning window with a window length of 1.6 seconds and an overlap of 0.4 seconds was used. Spectral resolution was 0.244 Hz. The spectral profile of rectified EMG of the three trials was averaged and then normalized with the mean spectral amplitude to reduce population variability. We obtained mean spectral peaks in the alpha (8–12 Hz), and gamma (35–50 Hz) bands from three tracking trials during static and dynamic force-tracking. All signal processing was completed using Matlab (Mathworks Inc., Natick, MA, USA).

Statistical Analysis. For each subject, all force and EMG variables of the three trials were averaged for the static and dynamic force-tracking tasks. A paired t-test was used to compare the mean gripping force between static and dynamic force-tracking. Hotelling's T^2 test was used to contrast the population means of force intermittency properties between static and dynamic force-tracking, including the amplitude parameter of force intermittency (RMS values of primary movement/force intermittency and $R_{PM/FI}$), spectral parameters of force intermittency (mean frequency and spectral dispersion), complexity of force intermittency (MSE areas in short, long, and overall time scales), scaling of force pulses (pulse amplitude, pulse duration, and pulse gain), and EMG variables (alpha peak and gamma peak, and RMS) of the FDS muscle. Post-hoc analysis was conducted for all Hotelling's T^2 tests with Bonferroni correction to determine the significance levels for multiple comparisons. For both tracking conditions, the correlation between the force amplitude variables (RMS_PM, RMS_FI, and $R_{PM/FI}$) and standardized amplitude of spectral peaks was examined with Pearson's correlation. Likewise, the correlation between the force intermittency complexity (MSE areas in low and high time scales) and standardized amplitude of spectral peaks was also examined with Pearson's correlation. The levels of significance for the determination of differences were 0.05. All statistical analyses were completed with the statistical package for Social Sciences (SPSS) for Windows v. 15.0 (SPSS Inc., USA).

Results

Basic Force Characteristics

The results of paired t statistics suggested an insignificant protocol effect on mean gripping force between dynamic force-tracking (128.79±6.05 N) and static force-tracking (130.89±6.11 N) (t_{21} = −1.471, P = 0.156), which validated that the physical work of the two loaded paradigms was very similar.

Force Intermittency Properties and Force Pulse Metrics

Table 1 contrasts the mean amplitudes for the primary movement (PM) and force intermittency (FI) profile between static and dynamic tracking. Hotelling's T^2 suggested a significant protocol effect on RMS values of the primary movement and force intermittency profile, as well as the amplitude ratio of $R_{PM/FI}$ (Wilks' Λ = 032, P<.001). Post-hoc analysis revealed that the RMS value of the force intermittency profile during dynamic force-tracking was greater than that during static force-tracking (P<.001), whereas the RMS value of the primary movement did not differ between the two force-tracking conditions (P = .301). Static force-tracking exhibited a greater $R_{PM/FI}$ (48.01±3.49) than did dynamic force-tracking (22.52±0.51) (P<.001). Figure 2A shows the power spectra of force intermittency between static and dynamic force-tracking for all subjects. The mean frequency and spectral dispersion of the force intermittency profile differed with force-tracking mode (Wilks' Λ = .035, P<.001), with greater mean frequency and spectral dispersion for dynamic force-tracking (P<.001) (Fig. 2B). Figure 3A shows the results of MSE analysis and pooled SampEn curves across different time scales for static and dynamic tracking. Dynamic force-tracking appeared to exhibit a larger SampEn in the low time scale 1–25 but a smaller SampEn in the high time scale 26–60 than those of static force-tracking. Hotelling's T^2 and post-hoc analysis were consistent with that observation (Wilks' Λ = .106, P<.001). Dynamic force-tracking had a larger MES area (59.7±0.3) under the time scale 1–25 than did static force-tracking (57.9±0.3) (P = .001), but an opposite trend was noted for the MES area under the time scale 26–60 (Dynamic: 68.8±0.5; Static: 75.6±0.5) (P<.001) (Fig. 3B). The MES area of the overall time scale 1–60 for dynamic tracking (128.6±0.6) was significantly lower than that for static force-tracking (133.6±0.7) (P<.001) (Fig. 3B), because of a more potent effect on the deceasing trend of the MES area in the high time scale.

The fundamental element in the force intermittency profile was the force pulse, the scaling parameters of which were examined between static and dynamic force-tracking (Table 2). Hotelling's T^2 statistics showed that the pulse variables differed with tracking protocol (Wilks' Λ = .135, P<.001). Post-hoc analysis suggested that the pulse amplitude of dynamic force-tracking (9.88±.53 N) was larger than that of static force-tracking (3.30±.35 N) (P<.001). Dynamic force-tracking exhibited a longer pulse duration (.448±.007 sec) than did static force-tracking (.378±.110 sec) (P<.001). The pulse gain (amplitude-duration regression slope) of dynamic force-tracking (26.59±1.39 N/sec) was significantly greater than that of static force-tracking (11.78±1.17 N/sec) (P<.001).

EMG Variables and Muscular Oscillations

Figure 4A contrasts the pooled spectral profiles of un-rectified/rectified EMG of the FDS muscle between static and dynamic force-tracking. Both EMG spectral profiles exhibited two prominent spectral peaks in 8–12 Hz and 35–50 Hz. Hotelling's T^2 statistics showed that EMG spectral variables varied with force-tracking protocol (Un-rectified EMG: Wilks' Λ = .722, P = .039; Rectified EMG: Wilks' Λ = .496, P = .003) (Fig. 4B). For rectified EMG, post-hoc analysis further revealed that static force-tracking (normalized spectral amplitude: 3.30±0.34) had a greater alpha spectral peak (8–12 Hz) than dynamic force-tracking (2.27±0.15) (P = .009). Conversely, the dynamic task (standardized amplitude:

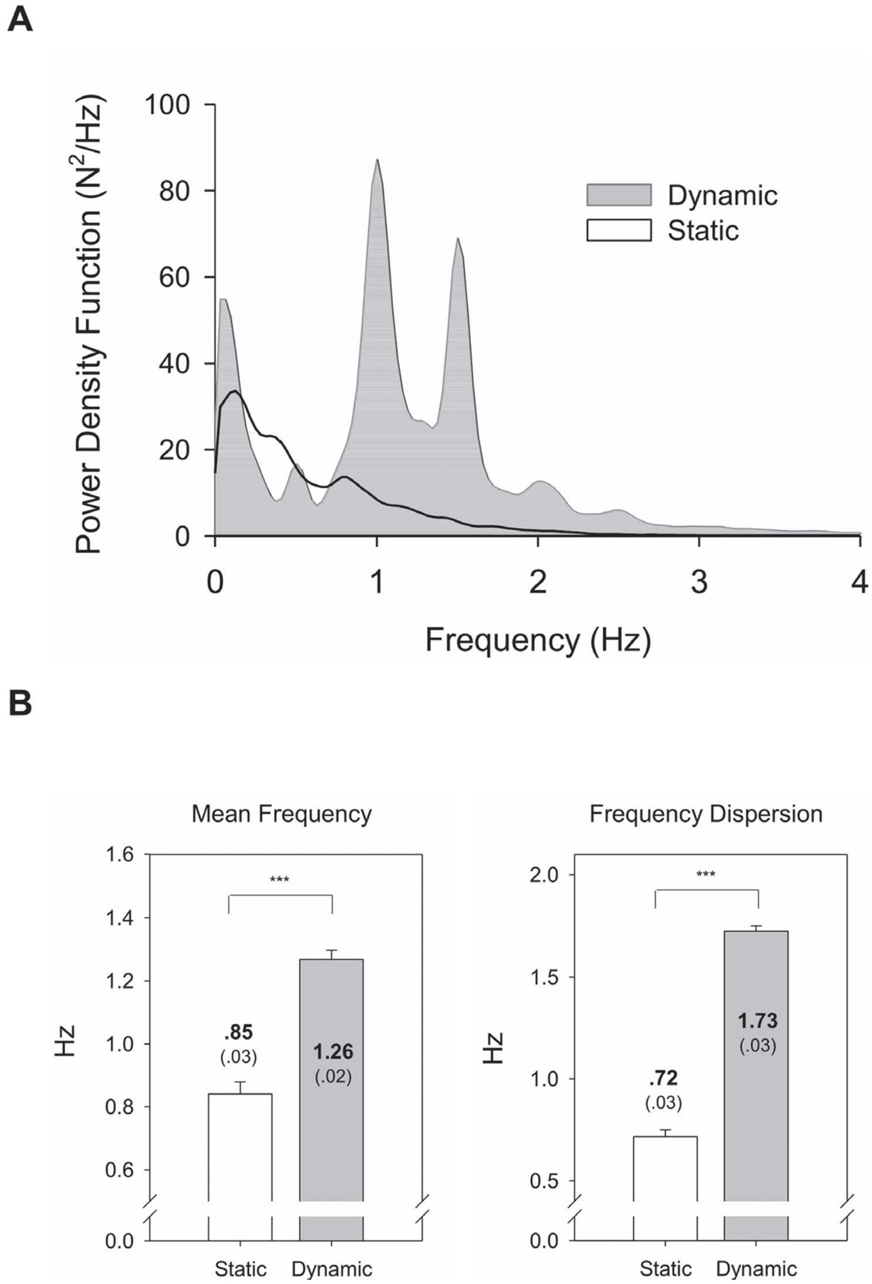

Figure 2. Contrast of spectral features of force intermittency profile between static and dynamic force-tracking. (A) Pooled spectral distributions of force intermittency profile during static and dynamic force-tracking, (B) population means of mean frequency and spectral dispersion for force intermittency profiles (Post-hoc test: ***: Dynamic > Static, $P<.001$).
doi:10.1371/journal.pone.0074273.g002

1.86±0.19) exhibited a larger gamma rhythm (35–50 Hz) than the static task (standardized amplitude: 1.42±0.07) ($P=.011$). Variations in standardized spectral peaks in the alpha and gamma bands between static and dynamic gripping for un-rectified EMG were similar to those of rectified EMG. Static gripping resulted in a greater alpha peak but a smaller gamma peak (alpha: 0.45±0.10; gamma: 2.20±0.13) than dynamic gripping (alpha: 0.24±0.03; gamma: 2.77±0.27) ($P<.05$). However, the EMG RMS of the FDS muscle was not significantly different between dynamic (0.065±0.005 mV) and static force-tracking (0.064±0.005 mV) ($P=.829$).

Table 3 shows relationships between the force intermittency variables and muscular oscillations for static and dynamic force-tracking. For static force-tracking, the standardized amplitude of the 8–12 Hz spectral peak was not significantly related to any force intermittency variables ($P>.05$). For dynamic force-tracking, the standardized amplitude of 35–50 Hz spectral peak was also correlated negatively and positively with force intermittency amplitude ($P<.05$) and the amplitude ratio of $R_{PM/FI}$ ($P<.001$), respectively. However, the standardized amplitude of the 8–12 Hz spectral peak was independent of any force variables ($P>.05$), though the muscular oscillation was significantly suppressed in comparison with that during static force-tracking.

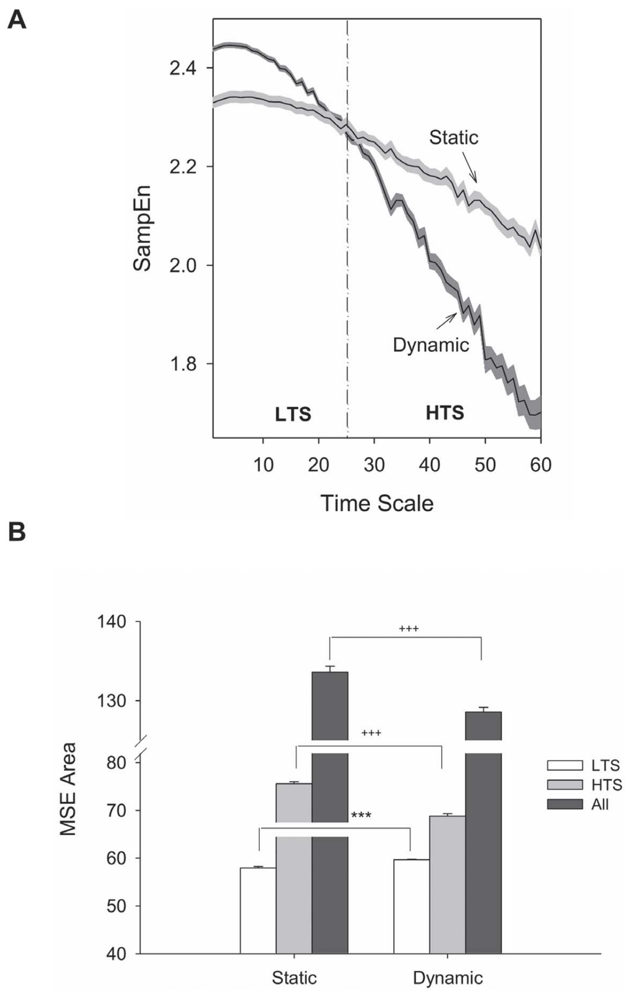

Figure 3. Contrasts of pooled complexity measures of force intermittency profile between static and dynamic force-tracking. (A) Sample entropy (SampEn) versus time scales, (B) Multi-scale entropy area (MSE area) for the low time scale of 1–25 (LTS), high time scale of 26–60 (HTS), and overall time scale of 1–60 (All). Each time scale represents 10 ms due to the sampling rate of 100 Hz. (Post-hoc test: ***: Dynamic > Static, $P \leqq .001$; †††: Static > Dynamic, $P < .001$).
doi:10.1371/journal.pone.0074273.g003

Discussion

The present study first revealed that the size and complexity of force intermittency as well as muscular oscillation were organized with the target goal of the force-tracking tasks. The dynamic force-tracking brought about a greater size of force intermittency with higher and wider spectral dispersion than did static force-tracking. In comparison with static tracking, dynamic tracking exhibited a greater complexity of force intermittency in the low time scale but, conversely, a greater regularity of force intermittency in the high time scale. Concurrent with task-dependent scaling of force intermittency, dynamic force-tracking exhibited a more potentiated 35–50 Hz muscular oscillation but a smaller 8–12 Hz muscular oscillation than did static force-tracking. In light of the force intermittency and muscular rhythm, there exist strategic differences in force regulation between dynamic and static force-tracking of a comparable load along with an underlying greater cognitive challenge for repetitive transient force changes during dynamic force-tracking.

Trajectory Optimization and Task-dependent Force Intermittency Properties

In this study, force output during a tracking maneuver was dichotomized into two force components, the smooth primary movement and the force intermittency profile. Contrary to a primary movement that signifies a priori standard preprogrammed in pursuit of a visual target [24,25,27], the force intermittency profile reflects an error-correction strategy in an attempt to remedy deviations during goal-directed movement. Under the framework of sampled movement control [10,11,12], force pulses in a force intermittency profile are centrally-scalable, superimposed onto the primary movement to tune a force trajectory [24,26,27]. Since dynamic tracking produced larger force intermittency and a smaller $R_{PM/FI}$ ratio than did static force-tracking (Table 1), dynamic force-tracking weighs more heavily on the error-correction process, entailing more intensive integration of proprioceptive and visual inputs than does static tracking [33]. Also, corrective adjustments to dynamic force-tracking were more frequent in order to generate motor commands in shorter time scales, on account of the higher number of high-frequency components with greater spectral dispersion in the force intermittency profile (Figs. 2A, 2B). Irrespective of static and dynamic tracking, force intermittency had a spectral range under 2 Hz, consistent with the Vaillancourt et al. (2002) [34], who reported a 0–2 Hz dominant frequency in force output during static continuous isometric contraction with low and high visual gains. Interestingly, force intermittency during dynamic force-tracking appeared to oscillate at harmonics of the target rates (primarily 1.0 Hz, 1.5 Hz, and 2 Hz). It is speculated that the subjects recurrently updated the trajectory control at particular rates, which have been noted to code kinematic properties of repetitive hand movement in the cortico-cerebello-cortical loop [35,36].

Although the complexity of force intermittency is typically characterized with approximate entropy [5,7,37] or uni-scale SampEn [23], this study adopted a new complexity measure with the use of multi-scale entropy (MSE). The methodological advantage of using MSE is that it allows assessment of SampEn across multiple time scales on the basis of multiple coarse-grained sequences and long-range temporal correlations, such that MSE accounts for time-dependent complexity and the presence of memory effects in physiological data [6,28]. In the low time scale 1–25, dynamic force-tracking exhibited a greater force intermittency complexity (larger

Table 2. The contrast of force pulse variables between static and dynamic tracking.

Force pulse variable[1]	Static	Dynamic	*Statistics*
Mean Amplitude (N)	3.30±.35	9.88±.53***	
Mean Duration (Sec)	.378±.011	.448±.007***	Λ = 0.135, P = .000 [2]
Pulse Gain[3] (N/Sec)	11.78±1.17	26.59±1.39***	

[1]Values were presented as mean ± se.
[2]Post-hoc for static force-tracking vs. dynamic force-tracking (***: Dynamic > Static, $P < .001$).
[3]Pulse gain also denotes amplitude-duration slope of force pulse.
doi:10.1371/journal.pone.0074273.t002

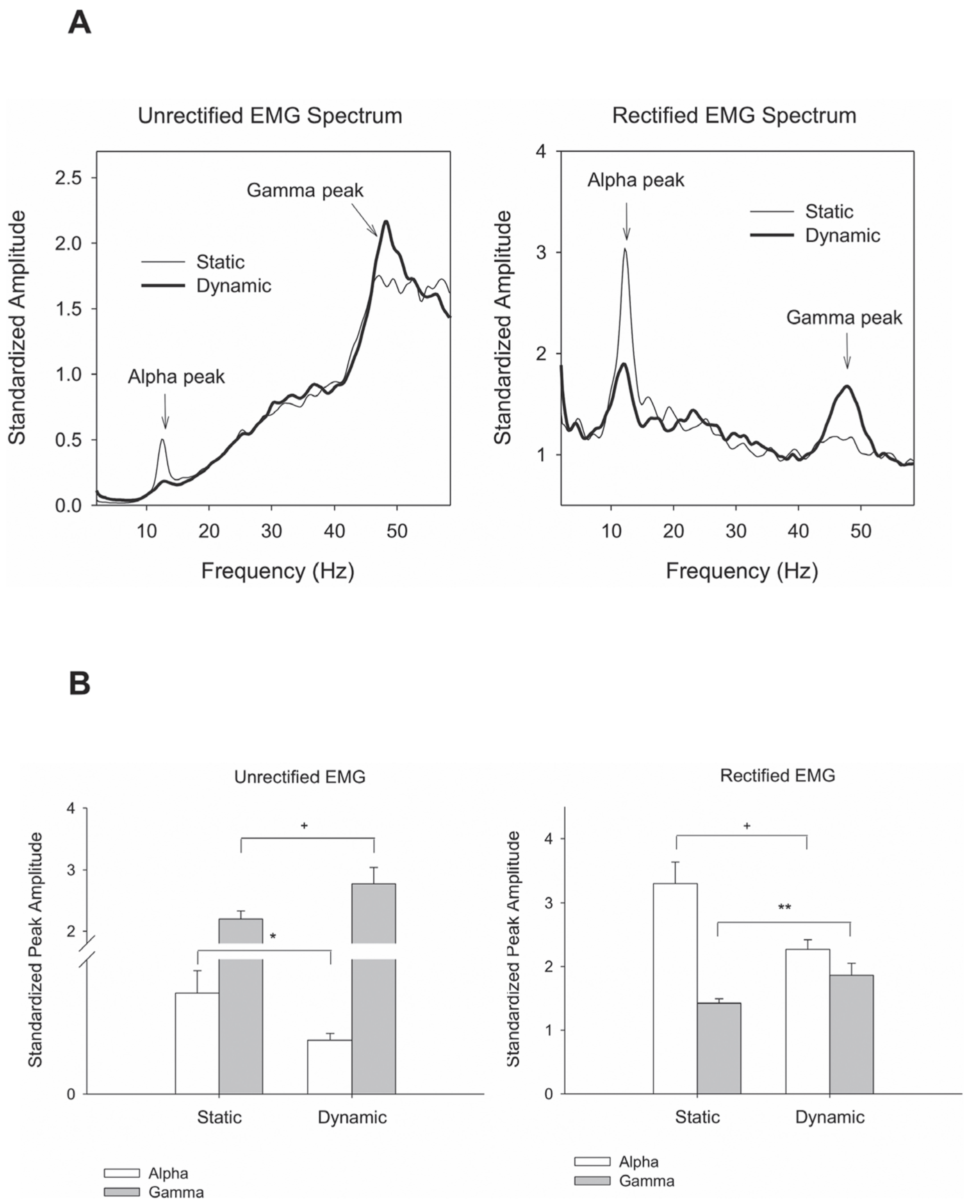

Figure 4. Contrasts of spectral features of the EMG between static and dynamic force-tracking. (A) Pooled spectral profiles of un-rectified and rectified EMG, (B) The means and standard errors of standardized amplitude for 8–12 Hz and 35–50 Hz spectral peaks. (Post-hoc test: *: Dynamic > Static, $P<.05$; ††: Static > Dynamic, $P<.01$; †: Static > Dynamic, $P<.05$).
doi:10.1371/journal.pone.0074273.g004

MSE area) than did static force-tracking (Figs. 3A, 3B), physically in accordance with the wider spectral spreads in high frequency of the force intermittency profile. Dynamic force-tracking in the shorter time scale was more informative, probably because the force tracking system adapted the required force output to multiple changing sensory inputs from the periphery to remedy tracking deviations in a short interval [33]. However, force intermittency of dynamic force-tracking in the high time scale 26–60 were conversely more regular (smaller MSE area) than those of static force-tracking (Figs. 3A, 3B). Since the target cycle of 0.5 Hz for dynamic force-tracking was 500 ms, it was very likely that the force intermittency data in the former half of the target cycle shared some stochastic properties with the latter half of the target cycle. Therefore, the force intermittency sequence after the course-gaining process with a window length exceeding half of the target cycle (time scale = 25) presented memory effects with higher possibility of predictability (lower SampEn curve) than did the force intermittency sequence in the static condition. This scenario suggests that fine-tuning of force trajectory during dynamic tracking was rhythmically encoded in every half a target cycle. The trajectory corrective mode for time-to-valley force and time-to-peak force during dynamic force-tracking could be analogous. Because the effect of SampEn in the high time scale on complexity measures overpowered that in the low time scale, the overall MSE area of dynamic force-tracking was still lower than that of static force-tracking (Fig. 3B). This observation on overall MSE area can explain a more regular movement trajectory for

Table 3. Pearson's correlation coefficients between force intermittency characteristics and muscular oscillations.

	Static	Dynamic	
(n = 22)	Alpha	Alpha	Gamma
$RMS_{\_PM}$[1]	r = −.336, P = .126	r = −.289, P = .192	r = −.223, P = .319
$RMS_{\_FI}$[2]	r = −.365, P = .095	r = −.208, P = .354	r = −.426, P = .048[*] [7]
$R_{PM/FI}$[3]	r = .382, P = .079	r = −.381, P = .081	r = .654, P = .001[**] [7]
$MSE_{\_LTS}$[4]	r = .052, P = .189	r = .282, P = .204	r = .295, P = .183
$MSE_{\_HTS}$[5]	r = .118, P = .602	r = −.104, P = .645	r = .057, P = .800
$MSE_{\_All}$[6]	r = .088, P = .698	r = −.031, P = .892	r = .089, P = .695

[1]$RMS_{\_PM}$ represents root mean square of primary movement.
[2]$RMS_{\_FI}$ represents root mean square of force intermittency profile.
[3]$R_{PM/FI}$ represents amplitude ration of primary movement relative to force intermittency profile.
[4]$MSE_{\_LTS}$ represents multi-scale entropy area of low time scale 1–25.
[5]$MSE_{\_HTS}$ represents multi-scale entropy area of high time scale 26–60.
[6]$MSE_{\_All}$ represents multi-scale entropy area of overall time scale 1–60.
[7]The shaded area indicates a significant level of correlation coefficient. (*: $P < .05$; **: $P < .005$).
doi:10.1371/journal.pone.0074273.t003

tracking a periodically-moving target [5,23]. Like force intermittency properties, force pulse metrics were differently organized with target accuracy constraints. Dynamic tracking exhibited greater pulse amplitude and pulse duration than did static tracking (Table 2). In addition, to keep in line with a rhythmic target movement, the central nervous system had to multiply pulse gain (or scaling amplitude-duration slope) during dynamic tracking (Table 2). Therefore, the dynamic target goal was accomplished by additive accuracy control that preferentially increased the gain of spatial scaling of force pulse more than the gain of temporal scaling of force pulse. A similar change in scaling amplitude-duration slope of kinematic submovement was reported, when tracking speed progressively increased during circular manual tracking [24,25].

Oscillatory Muscular Activity and Task-dependent Trajectory Adjustments

The variations in force intermittency property for the static and dynamic force-tracking pertained to differing organization of muscular oscillations at 8–12 Hz and 35–50 Hz in the FDS muscle (Fig. 4A). Research has shown that muscular oscillations in the EMG spectral peaks are related to grouped motor unit firing rates, especially enhanced EMG rectification that suppresses EMG spectral features related to the motor unit action potential shape (higher-frequency components) [31,38]. Although we did not directly measure the EEG-EMG piper rhythm (EMG-EEG coherence), it is likely that the muscular oscillations at 8–12 Hz and 35–50 Hz were physiological tremor [39,40] and the gamma band of the EMG piper rhythm [1,41], respectively. They could be the peripheral parts of EEG-EMG piper rhythm serving to regulate motor unit firing during force tracking maneuvers. For dynamic force-tracking, the most noteworthy finding was the potentiation of the low gamma band of the EMG piper rhythm (Fig. 4B). In fact, oscillatory muscle activity in 35–50 Hz is in line with converging evidence that the gamma band in corticomuscular coherence presents during phasic movement [1,42] and repetitive isotonic contraction [43]. The occurrence of gamma synchrony is thought to be of functional relevance when a motor task entails temporal modulation in movement patterns with global alertness to integrate sensory-motor information [1,42,44]. Our observation adds to this hypothesis by showing a significant negative correlation between 35–50 Hz muscular oscillation and force intermittency amplitude (Table 3). Hence, we may well argue that the gamma EMG piper rhythm is specified for fine-tuning force trajectory during dynamic tracking. The more gamma EMG piper rhythm associates with the lesser corrective attempts and the greater priori standard of tracking maneuver relative to force intermittency ($R_{PM/FI}$). In addition, we noted a significant suppression of alpha muscular oscillation during dynamic force-tracking, as compared with that of static tracking (Fig. 4B). Iyer et al. [45] also reported a roughly 12 Hz motor unit discharge during static and quasi-sinusoidal isometric contraction at the same mean force level. However, what is still not completely clear is the role of 8–12 Hz muscular oscillatory activity in the shift of tracking mode in this study.

In this study, muscular rhythm was assessed with spectral peaks of surface EMG. EMG rectification is a prevailing approach prior to calculating corticomuscular coherence for maximizing information about the grouped firing rate frequencies of active motor units [30–32]. However, some researchers argue against the appropriateness of the pre-processing procedure, as rectified EMG does not necessarily enhance the peak detection of corticomuscular coherence and may produce inconsistent coherence spectra in some cases [46,47]. Regarding this methodological controversy, we also validated our observations with spectral analysis using raw EMG. Two prominent spectral peaks were consistently noted in the spectral profiles of raw and rectified EMG, with similar parametric changes in standardized peak amplitude with respect contraction mode. In addition, we did not observe a significant EMG oscillation in the beta range (13–21 Hz) in either profile during static force-gripping, though previous studies have shown that the beta EMG-EEG piper rhythm is critical to maintaining force stability during sustained isometric contraction [1,29,42]. Physically, an evident EMG-EEG coherence in the beta band just represents a relatively high degree of an in-phase oscillation at 13–21 Hz for both EEG and EMG signals; however, it does not mean a prominent beta oscillation as compared to other spectral ingredients in the EMG signal. Despite this fact, future work is still needed to find cortical control over the task-specific scaling of force intermittency force-tracking of different patterns, on account of the functional interactions between cortical and spinal oscillatory networks.

Conclusions

In light of characteristic differences in the primary movement and force intermittency, we noted that neuro-mechanic control of force trajectory for static and dynamic force-tracking at a relatively high exertion level was task-dependent. Dynamic force-tracking exhibits a greater amount of force intermittency, with higher spectral components and greater complexity in the low time scale than that of static force-tracking. The target goal of dynamic force-tracking is achieved through frequent and vast trajectory adjustments, underlying intricate short-term and similar error-correction processes over half a target cycle. Unlike during static force-tracking, alpha muscular oscillation is markedly suppressed during dynamic force-tracking. The emergence of gamma muscular oscillation during dynamic force-tracking is likely to be responsible for the scaling of force intermittency and force trajectory adjustments. At a relatively high exertion level, modulations of muscular oscillation and force intermittency properties agree with the theoretical postulation that internal force coding to stabilize movement trajectory differs vastly with target constraints.

Supporting Information

Appendix S1 Calculation of MSE Area.
(DOCX)

Author Contributions

Conceived and designed the experiments: YCC ISH. Performed the experiments: YTL CLS. Analyzed the data: YCC YTL. Contributed reagents/materials/analysis tools: CTH ZRY. Wrote the paper: YCC ISH.

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國科會補助計畫衍生研發成果推廣資料表

日期:2014/01/30

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<tr><td>計畫主持人: 陳怡靜</td></tr>
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101 年度專題研究計畫研究成果彙整表

計畫主持人：陳怡靜			計畫編號：101-2410-H-040-017-				
計畫名稱：動作溢流對拔河選手手指生理震顫的影響							
成果項目			量化			單位	備註（質化說明：如數個計畫共同成果、成果列為該期刊之封面故事...等）
			實際已達成數（被接受或已發表）	預期總達成數(含實際已達成數)	本計畫實際貢獻百分比		
國內	論文著作	期刊論文	0	0	0%	篇	
		研究報告/技術報告	0	0	0%		
		研討會論文	0	0	0%		
		專書	0	0	0%		
	專利	申請中件數	0	0	0%	件	
		已獲得件數	0	0	0%		
	技術移轉	件數	0	0	0%	件	
		權利金	0	0	0%	千元	
	參與計畫人力（本國籍）	碩士生	0	0	0%	人次	
		博士生	0	0	0%		
		博士後研究員	0	0	0%		
		專任助理	0	0	0%		
國外	論文著作	期刊論文	1	1	100%	篇	
		研究報告/技術報告	0	0	0%		
		研討會論文	0	0	0%		
		專書	0	0	0%	章/本	
	專利	申請中件數	0	0	0%	件	
		已獲得件數	0	0	0%		
	技術移轉	件數	0	0	0%	件	
		權利金	0	0	0%	千元	
	參與計畫人力（外國籍）	碩士生	1	1	10%	人次	
		博士生	1	1	10%		
		博士後研究員	0	0	0%		
		專任助理	0	0	0%		

其他成果 (無法以量化表達之成果如辦理學術活動、獲得獎項、重要國際合作、研究成果國際影響力及其他協助產業技術發展之具體效益事項等，請以文字敘述填列。)	無

	成果項目	量化	名稱或內容性質簡述
科教處計畫加填項目	測驗工具(含質性與量性)	0	
	課程/模組	0	
	電腦及網路系統或工具	0	
	教材	0	
	舉辦之活動/競賽	0	
	研討會/工作坊	0	
	電子報、網站	0	
	計畫成果推廣之參與（閱聽）人數	0	

國科會補助專題研究計畫成果報告自評表

請就研究內容與原計畫相符程度、達成預期目標情況、研究成果之學術或應用價值（簡要敘述成果所代表之意義、價值、影響或進一步發展之可能性）、是否適合在學術期刊發表或申請專利、主要發現或其他有關價值等，作一綜合評估。

1. 請就研究內容與原計畫相符程度、達成預期目標情況作一綜合評估

■達成目標

□未達成目標（請說明，以 100 字為限）

□實驗失敗

□因故實驗中斷

□其他原因

說明：

2. 研究成果在學術期刊發表或申請專利等情形：

論文：■已發表 □未發表之文稿 □撰寫中 □無

專利：□已獲得 □申請中 ■無

技轉：□已技轉 □洽談中 ■無

其他：（以 100 字為限）

3. 請依學術成就、技術創新、社會影響等方面，評估研究成果之學術或應用價值（簡要敘述成果所代表之意義、價值、影響或進一步發展之可能性）（以 500 字為限）

學術成就：已達成有 1 篇 SCI 全文(full article)形式期刊發表，為 SCI Multidisciplinary science,領域排名 12.5%之期刊。本計劃主要在於延續過去動作控制的研究經驗，瞭解一般人與拔河選手之間因特別化的肌力訓練可能造成動作控制機制不同的影響。技術創新:利用加速規測量動作控制而造成的生理震顫。

社會影響:一般人與拔河選手之間的動作控制機制的瞭解是運動科學的新知識，可做為訓練學理參考。