Effects of Functional Electrical Stimulation on Trunk Musculature during Wheelchair Propulsion

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Submitted to the Graduate Faculty of

School of Health and Rehabilitation in partial fulfillment

of the requirements for the degree of

Ph.D. of Rehabilitation Science and Technology

University of Pittsburgh

2005

UNIVERSITY OF PITTSBURGH

SCHOOL OF HEALTH AND REHABILITATION SCIENCE

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During wheelchair propulsion, in order to apply power to the pushrim effectively, shoulder and trunk stabilization is needed to control arm movements and the consequent transfer of power from the limbs through to the pushrim. Available trunk control may be one of the most important force-generating mechanisms during wheelchair propulsion, particularly when an individual is fatigued or propelling through a difficult or demanding situation. Consequentially it is a worthwhile pursuit to further understand and study the process of trunk muscle recruitment during propulsion and the effects of reduced trunk control on propulsion biomechanics. In the first of three studies contained in this dissertation is, trunk muscle recruitment patterns using surface electromyography (sEMG) electrodes during wheelchair propulsion under different speed conditions. The results of this first study provided insight into the functional role of specific trunk muscles during propulsion.

In the second study, a biomechanical analysis was utilized to examine the effect of functional electrical stimulation (FES) on trunk musculature during five minutes of wheelchair propulsion. The findings revealed that a trunk FES device could help an individual to generate more propulsion power and increase gross mechanical efficiency during wheelchair propulsion. Consequentially, with these improvements in propulsion biomechanics, a user with a trunk FES device may be able to more easily negotiate demanding propulsion tasks, ultimately improving quality of life.

The third study investigated the influence of surface electrical stimulation of trunk musculature on shoulder muscle recruitment patterns during wheelchair propulsion. The results showed that trunk FES may help individuals to generate wheelchair propulsion power without placing additional demands on shoulder musculature. With trunk stimulation, the functional role of the shoulders may shift from stabilizers to a prime movers contributing more directly to propulsion.

In the future, improvements can be made in delivering FES to specifically targeted muscle groups to more accurately simulate volitional trunk control. With advanced programming, a FES device could be better synchronized with the propulsion cycle to avoid continuous stimulation of the trunk which can be uncomfortable or fatiguing. It would be ideal to provide stimulation during pre-push and early push phase of the propulsion cycle when trunk stability has been shown to be most critical. Individuals could potentially benefit from using FES more during challenging situations or tasks of short duration, such as pushing up a ramp or across thick carpeting.

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PREFACE

I would like to foremost thank my doctoral advisor, Dr. Alicia M, Koontz for her academic and research guidance over the last few years. Also, with my deepest gratitude, I would like to thank Dr. Cooper and Dr. Boninger for providing me with the opportunity to be a part of a highly prestigious and worldwide respected research group. In addition, I would like to thank Dr. Triolo and Dr. Fitzgerald for their professional advice and support. Furthermore, I would like to thank Dr. McCrory, Dr. Holm, Dr. Ding, and Dr. Guo for their guidance throughout my dissertation study. I would also like to thank Annmarie Kelleher and Rosemarie Cooper for their great assistance in IRB modification over past few years. Moreover, I have to thank Jennifer Mercer, Ian Rice, Megan Yarnall, Brad Impink, Eliana Chaves, and Fabrisia Woolard for assisting with subject recruitment and technical issues related to this project. I would deeply thank all staff and graduate students at HERL for their great help and supports in so many ways in past few years.

Very special thanks go to my parents, and my brother and sister for their support although you live thousands miles a way. I also like to express great appreciation to my previous colleagues at Taipei City Psychiatric Center for inspiring me throughout past four years. Thanks to all of you.

DEFINITIONS AND ABBREVIATIONS

sEMG: Surface electromyography

FES: Functional electrical stimulation

SCI: Spinal cord injury

MWUs: Manual wheelchair users

MVC: maximum voluntary contraction

RA: rectus abdominis

EO: external oblique

IO: internal oblique

LT: longissimus thoracis

IL: iliocostalis lumborum

MU: multifidus

ASIS: anterior superior iliac spine

PSIS: posterior superior iliac spine

ANOVA: Analysis of variance

HIGH: the stimulation level which was adjusted for the 50% of difference between the self-reported maximal tolerable amplitude and the minimal stimulation amplitude causing muscle contraction.

LOW: the stimulation level which was adjusted for the 25% of difference between the self-reported maximal tolerable amplitude and the minimal stimulation amplitude causing muscle contraction.

OFF: the stimulation level which no stimulation was given.

GME: gross mechanical efficiency

PM: pectoralis major

AD: anterior deltoid

MD: middle deltoid

PD: posterior deltoid

BB: biceps brachii

TB: triceps brachii

1. INTRODUCTION

This dissertation examined the effects of functional electrical stimulation on trunk musculature during wheelchair propulsion through a series of three inter-related studies. This dissertation consists of seven chapters as follows: Chapter 1 presents background literature and a general outline of the overall research designs of the work. Specific aims of the dissertation are stated at the end of each paragraph. Chapter 2 describes surface electromyographic (sEMG) activity of the trunk muscles in unimpaired individuals during wheelchair propulsion at various speeds. Chapter 3 describes the effects of functional electrical stimulation on trunk musculature during wheelchair propulsion in twelve individuals with spinal cord injury (SCI). As part of this study, propulsion kinetics, kinematics and physiological energy expenditure were assessed during a five-minute propulsion trial. Chapter 4 consists of a sEMG analysis of shoulder muscle activity during wheelchair propulsion with varying levels of trunk functional electrical stimulation. The EMG activity of six shoulder muscles was recorded from eleven individuals with SCI during a five-minute propulsion trial. Finally, Chapter 5 states the conclusions and limitations of this study.

1.1. Background

This dissertation work builds upon a collaborative study (VA Rehab R&D B3043-C) between the University of Pittsburgh's Human Engineering Research Laboratories (HERL) and the Cleveland Functional Electrical Stimulation (FES) Center in Ohio to 1) determine the phasing and intensity of abdominal and back muscle activity during wheelchair propulsion of unimpaired individuals, 2) quantify the effects of the surface FES induced trunk stability on wheelchair propulsion biomechanics of individuals with SCI, and 3) examine changes in shoulder muscle activity during wheelchair prolusion with functional electrical stimulation applied to the abdominal and back muscles. An overview of the three studies is described below (Figure 1):

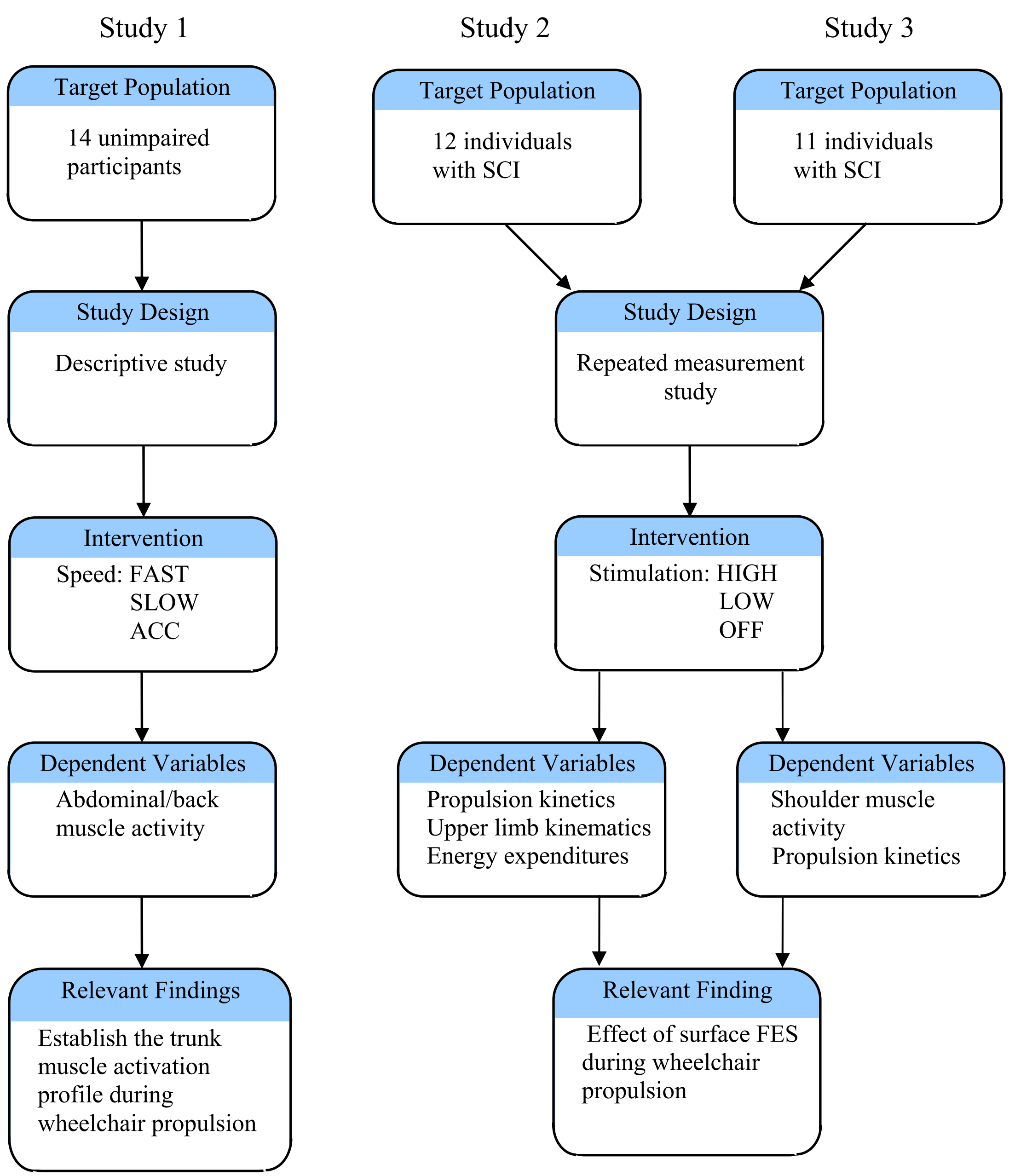

Figure 1 Chart of the research design

1.2. Trunk muscle activity during wheelchair propulsion

Trunk instability due to the absence or impairment of abdominal and back muscle control usually leads to a “C”-shaped kyphotic posture with flattened lumbar spine, and posterior pelvic tilt among individuals with SCI (Hobson and Tooms 1992). This functional sitting posture allows individuals with SCI to shift the trunk center of gravity back and secure it within their base of support without losing balance in a wheelchair. However, this passive kyphotic sitting posture can cause back pain, rotator cuff injury, and painful chronic health problems (Sinnott, Milburn and McNaughton 2000; Rintala, Loubser, Castro, Hart and Fuhrer 1998; Curtis, Drysdale, Lanza, Kolber, Vitolo and West 1999; Samuelsson, Tropp and Gerdle 2004).

Impaired trunk control also could compromise trunk stability during wheelchair propulsion, thereby affecting propulsion performance and limiting the ability of individuals with SCI to overcome fatigue during wheelchair propulsion. (Dallmeijer, van der Woude, Veeger and Hollander 1998; Schantz, Bjorkman, Sandberg and Andersson 1999; Newsam, Rao, Mulroy, Gronley, Bontrager and Perry 1999; Rodgers, Gayle, Figoni, Kobayashi, Lieh and Glaser 1994; Rodgers, Keyser, Gardner, Russell and Gorman 2000). One review article (Vanlandewijck, Theisen and Daly 2001) indicated that there are few biomechanical studies specifically addressing the functional role of the trunk during wheelchair propulsion. Trunk control ability may be one of the most important force-generating mechanisms during fatigue, and other situations of increased demand or difficulty during wheelchair propulsion. Information regarding trunk muscle recruitment during the propulsion cycle may be valuable in understanding the effect of reduced trunk control on propulsion biomechanics.

Therefore, the purpose of the first study was to establish and describe trunk muscle recruitment patterns using surface electromyography (sEMG) from unimpaired individuals during wheelchair propulsion under three different propulsion speed conditions: 0.9 m/sec (SLOW), 1.8 m/sec (HIGH) and acceleration (ACC). Surface electromyographic activity of six trunk muscles (rectus abdominis , external oblique, internal oblique, longissimus thoracis, iliocostalis lumborum, and multifidus) was collected using a MyoSystem 1200 (Noraxon U.S.A. Inc., Scottsdale, AZ). The results of the first study provide insight into the functional role of specific trunk muscles during the push and recovery phases of propulsion. Based on this trunk muscle activation profile, stimulation patterns could be programmed into FES units to reproduce a similar muscle activity for trunk stability of individuals with SCI during wheelchair propulsion.

1.3. Biomechanical analysis of functional electrical stimulation on trunk musculature during wheelchair propulsion

Functional electrical stimulation (FES) is a technique that artificially generates neural activity in order to overcome lost functions of paralyzed muscle or sensory impairments. FES has been used extensively in individuals with SCI, primarily to induce ambulation, standing, hand grip and cycling exercise (Jaeger, Yarkony and Smith 1989; Yarkony, Jaeger, Roth, Kralj and Quintern 1990; Triolo, Bevelheimer, Eisenhower and Wormser 1995; Wieler, Stein, Ladouceur, Whittaker, Smith, Naaman, Barbeau, Bugaresti and Aimone 1999; Kukke and Triolo 2004). Studies of FES-assisted walking have demonstrated improvements in functional mobility, but with slow ambulation speed and high cost of energy expenditure (Moynahan, Mullin, Cohn, Burns, Halden, Triolo and Betz 1996; Shimada, Sato, Abe, Kagaya, Ebata, Oba and Sato 1996; Kobetic, Triolo and Marsolais 1997; Klose, Jacobs, Broton, Guest, Needham-Shropshire, Lebwohl, Nash and Green 1997; Brissot, Gallien, Le Bot, Beaubras, Laisne, Beillot and

Dassonville 2000; Jacobs and Mahoney 2002). Because of its modest performance of low ambulation speed with high energy cost, FES-assisted walking is not used extensitively in the community or at home. Most users of FES systems use them for exercise, standing, and short distance walking, but still rely on a wheelchair as their primary means of mobility during daily activities (Kobetic, Triolo, Uhlir, Bieri, Wibowo, Polando, Marsolais, Davis and Ferguson 1999; Moynahan et al. 1996). It may be possible to integrate a user's existing FES system or use a surface FES device to induce trunk stability for wheelchair use. Providing trunk stability by artificially stimulating the abdominal and back muscles may create a better base of support of the shoulder girdle complex, thereby helping a user to generate larger propulsion forces and power.

The purpose of the second study was to examine whether a surface FES system, applied to the abdominal and back muscles of manual wheelchair users (MWUs), could help to stabilize the trunk, thus improving propulsion technique and efficiency without a significant increase of energy expenditure. It was hypothesized that using stimulation on a MWU's trunk musculature would 1) produce a significant increase in propulsion forces, torques, mechanical effective forces, and power production, 2) cause a greater range of motion at shoulder, elbow and wrist joint along with increased trunk flexion, and 3) generate no significant increase of energy expenditure. Twelve participants with SCI received three stimulation levels (HIGH, LOW and OFF), randomly applied to their abdominal and back muscle groups with a surface functional electrical stimulation (FES) device, during a five-minute propulsion trial at a constant speed of 1.34 m/sec. Propulsion kinetics, kinematics, and metabolic variables during a propulsion trial were measured using the SMART$^{\text{Wheels TM}}$ (Three Rivers Holdings, Inc., Mesa, AZ), a three-dimensional OPTOTRAK motion analysis system (Northern Digital Inc., Ontario, Canada) and a

SensorMedics 2900 Metabolic Cart (SensorMedics, Yorba Linda, CA) respectively. Effects of stimulation level amongst the above variables over three time intervals within a five-minute trial were examined using a two-way repeated measures ANOVA. The results of the second study indicate potential benefits of trunk FES that will help manual wheelchair users improve propulsion efficiency.

1.4. Shoulder muscle activity during wheelchair propulsion

Manual wheelchair propulsion requires a large amount of static work from proximal shoulder muscle synergy and cocontraction to stabilize and adjust the shoulder girdle complex with respect to the trunk, for gripping and applying force to the hand rim during the push phase of propulsion (van der Helm and Veeger 1996; van der Woude, Dallmeijer, Janssen and Veeger 2001; van der Woude, Veeger, Dallmeijer, Janssen and Rozendaal 2001; Vanlandewijck et al. 2001). Prolonged manual wheelchair use can lead to pain and repetitive strain injury (RSI) in the upper extremities (Subbarao, Klopfstein and Turpin 1995; Nichols, Norman and Ennis 1979; Pentland and Twomey 1991; Dalyan, Cardenas and Gerard 1999; Boninger, Towers, Cooper, Dicianno and Munin 2001). In fact, the shoulder is the most commonly reported site of musculoskeletal injury in MWUs. The high prevalence rate of shoulder pain is likely related to overuse of the arms during transfers and wheelchair propulsion. Lack of trunk stability due to the absence or impairment of abdominal and back muscle control has been associated with post-injury shoulder pathology (Sinnott et al. 2000).

Individuals with paralysis of the lower extremities due to SCI rely on their upper limbs to push their wheelchairs for mobility. The power required during wheelchair propulsion originates from the musculature of the upper limb and shoulder, and results in moderate to high intensity and duration of shoulder muscle electromyographic activity (Harburn and Spaulding 1986; Masse, Lamontagne and O'Riain 1992; Mulroy, Gronley, Newsam and Perry 1996; Schantz et al. 1999; Mulroy, Farrokhi, Newsam and Perry 2004). It has been suggested that lack of trunk stability, leading to a less erect posture and poor support of the shoulder girdle complex, may limit production of maximal upper limb strength that is required to push a wheelchair. Studies have investigated a variety of devices for improving trunk stability during propulsion and sitting posture, such as a rigid backrest, inclination of seat frame angles, or artificially stimulating paralyzed trunk muscles (Parent, Dansereau, Lacoste and Aissaoui 2000; Samuelsson, Tropp, Nylander and Gerdle 2004; Yang, Koontz, Triolo, Mercer and Boninger 2005). However, few studies have examined the effects of these devices on shoulder muscle intensity and duration during wheelchair propulsion.

Therefore, the purpose of the third study was to investigate shoulder muscle activation and duration of activity in response to varying intensity levels of trunk stimulation during wheelchair propulsion. It was hypothesized that surface electrical stimulation applied to abdominal and back muscles in individuals with SCI would provide a better base of support for the shoulder girdle complex thereby reducing the intensity and duration of shoulder muscle activity to achieve the same propulsion demand. Eleven manual wheelchair users with SCI participated in three repeated five-minute wheelchair propulsion trials at a constant speed of 1.34 m/sec. During each propulsion trial, one of three stimulation levels (HIGH, LOW and OFF) was applied in random order to the participant’s abdominal and back muscle groups with a surface functional electrical stimulation (FES) device. The surface electromyographic (sEMG) activity of six shoulder muscles and their corresponding propulsion kinetics were recorded using a TELEMYO 2400T (Noraxon U.S.A. Inc., Scottsdale, AZ) and SMART$^{\text{Wheels TM}}$ (Three Rivers Holdings, Inc., Mesa, AZ). Effects of stimulation level on the sEMG and kinetic variables over three time intervals within a five-minute trial were examined using a two-way repeated measures ANOVA. The results of the third study provide insight into the potential application of surface electrical stimulation of trunk muscles as a preventative mechanism to minimize the risk of shoulder pain and injury in long-term wheelchair users.

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2. SURFACE ELECTROMYOGRPHY ACTIVITY OF TRUNK MUSCLES DURING WHEELCHAIR PROPULSION

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ACKNOWLEDGEMENTS

This research was supported by the U.S. Department of Veterans Affairs, Rehabilitation Research and Development (B3043-C), and American Society of Biomechanics Graduate Student Grant-In-Aid (2004).

2.1. ABSTRACT

Objective: To describe trunk muscle recruitment patterns using surface electromyography (sEMG) from unimpaired individuals during wheelchair propulsion under various propulsion speed conditions.

Methods: Fourteen unimpaired subjects participated in the study after providing informed consent. Subjects propelled a test wheelchair under three different speed conditions: 0.9 m/sec (SLOW), 1.8 m/sec (HIGH) and acceleration from rest to their maximum speed (ACC). Lower back/abdominal sEMG, and upper body movement were recorded for each speed condition. Based on the hand movement during propulsion, the propulsion cycle was further divided into five stages in order to better define the roles of the trunk muscles.

Results: Both abdominal and back muscle groups revealed significantly higher activation at early push and pre-push stages when compared to the three other three stages of the propulsion phase. With increasing propulsion speed, trunk muscles showed increased activation ($p<0.0001$). Back muscle activity was significantly larger than abdominal muscle activity across the three speed conditions ($p<0.0001$), with lower back muscles predominating.

Conclusions: The present study identified trunk muscle recruitment patterns at three different propulsion speeds. Abdominal and back muscle groups cocontracted at late recovery phase and early push phase to provide sufficient trunk stability to meet the demands of propulsion. Customizing the wheelchair (e.g., using a rigid backrest or inclining the seat) or artificially stimulating paralyzed trunk muscles may increase trunk stability, hence improving propulsion performance.

Keywords: muscle recruitment; spinal cord injury; propulsion cycle

2.2. INTRODUCTION

Trunk instability due to the absence or impairment of abdominal and back muscle control usually leads to a “C”-shaped kyphotic posture with flattened lumbar spine, and posterior pelvic tilt among individuals with spinal cord injury (SCI) (Hobson and Tooms 1992). This functional sitting posture allows SCI individuals to shift the trunk center of gravity back and secure it within their base of support without losing balance in a wheelchair. However, this passive kyphotic sitting posture can cause back pain, rotator cuff injury, and painful chronic health problems (Sinnott, Milburn and McNaughton 2000; Rintala, Loubser, Castro, Hart and Fuhrer 1998; Curtis, Drysdale, Lanza, Kolber, Vitolo and West 1999; Samuelsson, Tropp and Gerdle 2004). Furthermore, impaired trunk control could compromise trunk stability during wheelchair propulsion and influence propulsion performance. Dallmeijer et al. (1998) found that individuals with tetraplegia placed their hands in a more backward position on the pushrim at the start of the push phase as compared with individuals with paraplegia. The difference between these two groups in the start angle during propulsion was believed to be related to reduced trunk stability in the tetraplegia group (Dallmeijer, van der Woude, Veeger and Hollander 1998). Thus, pushing with the hands in more backward position appears to be a compensating strategy for individuals without trunk control to secure their balance and achieve sufficient trunk stability.

Schantz et al. (1999) compared the patterns of body movement between individuals with paraplegia and tetraplegia during wheelchair propulsion at three different speeds. They discovered that participants with paraplegia had more trunk flexion at the start of push while accelerating the wheelchair in comparison to participants with tetraplegia. The greater volitional control of the trunk and arm muscles allowed participants with paraplegia to have longer push phases, thereby increasing their propulsion speed (Schantz, Bjorkman, Sandberg and Andersson

1999). Newsam et al. (1999) assessed upper extremity motion during wheelchair propulsion among persons with different levels of spinal cord injury. They reported that participants with high cervical lesions yielded greater range of trunk motion during propulsion. They suggested that augmented stabilization of the trunk may help participants who lose voluntary control of trunk musculature to maintain a consistent propulsion stroke patterns (Newsam, Rao, Mulroy, Gronley, Bontrager and Perry 1999).

Power et al. (1994) compared shoulder isometric strength of individuals with tetraplegia and paraplegia. They found that shoulder strength, which is responsible for providing the primary propulsion force, of the tetraplegia group was significantly lower than for the paraplegia group. They believed that lack of trunk stability, which resulted in less erect posture and poor support of the shoulder girdle complex, limited production of maximal strength. (Powers, Newsam, Gronley, Fontaine and Perry 1994). People with paraplegia have additional upper extremity muscle function and more trunk and shoulder muscle stability compared to people with tetraplegia. These differences likely influence propulsion efficiency.

Impaired trunk control also limits the ability of individuals with SCI to overcome fatigue during wheelchair propulsion. Rodgers et al. (1994) investigated the influence of fatigue on trunk movement during wheelchair propulsion. They reported a significant increase of trunk forward lean with fatigue (Rodgers, Gayle, Figoni, Kobayashi, Lieh and Glaser 1994). This increase in forward lean might aid the application of force to the pushrim and enable the transfer of propulsion power from the trunk and upper extremity to the pushrim (Sanderson and Sommer 1985). Rodgers et al. (2000) also found that subjects with increased trunk flexion during propulsion, which was accentuated with fatigue, had greater shoulder flexion and elbow extension when compared to subjects with a more erect posture. The trunk flexion and upper arm movement patterns appeared to be a compensatory strategy to generate a propulsion moment during muscle fatigue (Rodgers, Keyser, Gardner, Russell and Gorman 2000). Based on these previous findings, it is unlikely that individuals without volitional trunk control will be able to adopt a trunk flexion propulsion style, thereby restricting their ability to generate effective propulsion moments.

One review article (Vanlandewijck, Theisen and Daly 2001) indicated that there are few biomechanical studies specifically addressing the functional role of the trunk during wheelchair propulsion. Trunk control ability may be one of the most important force-generating mechanisms during fatigue, and other situations of increased demand or difficulty with propulsion. Information regarding trunk muscle recruitment during the propulsion cycle may be valuable in understanding the effect of reduced trunk control on propulsion biomechanics. Therefore, the purpose of this study was to establish and describe trunk muscle recruitment patterns using surface electromyography (sEMG) electrodes from unimpaired individuals during wheelchair propulsion under typical propulsion speeds and conditions.

2.3. METHODS

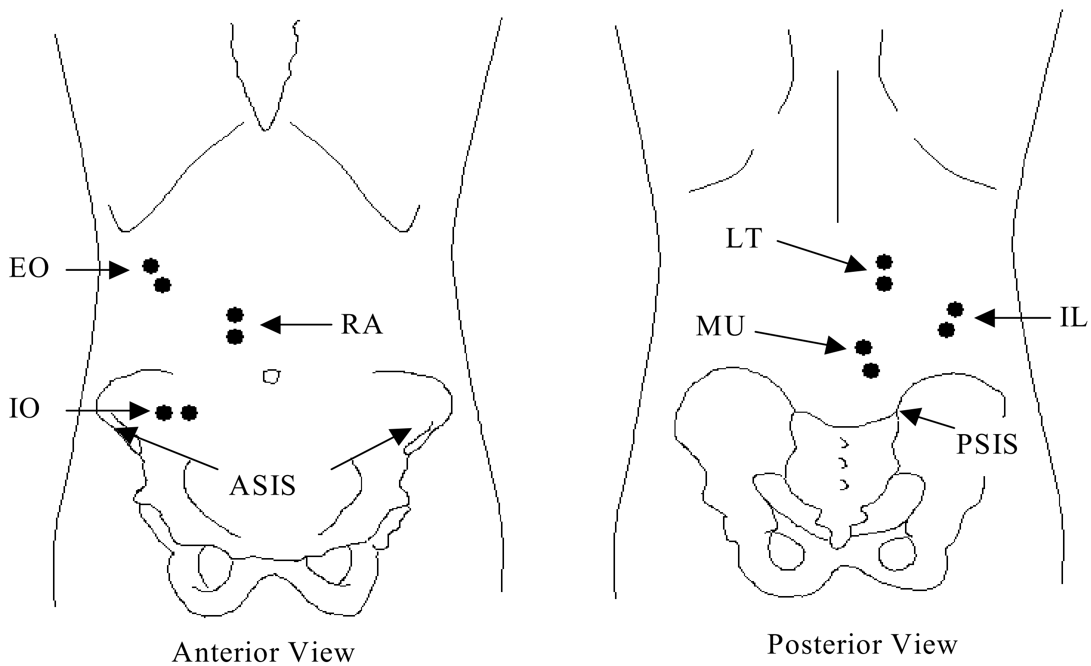
2.3.1. Subjects:

Fourteen unimpaired subjects (11 male and 3 female, mean age 24.7$\pm$3.6 years old, mean weight 69.3$\pm$14.3 kg and mean height 173$\pm$7 cm) provided informed consent in accordance with the procedures approved by the Institutional Review Board of the Veterans Affairs Pittsburgh

HealthCare System prior to participation in the study. None of these subjects reported any previous history of upper extremity pain or low back disorders that would impair propulsion.

2.3.2. Electromyography

Surface electromyographic (sEMG) activity of abdominal and back muscles was recorded with bipolar surface electrodes over three abdominal muscles (rectus abdominis — RA, external oblique — EO, internal oblique — IO), and three back muscles (longissimus thoracis — LT, iliocostalis lumborum — IL, multifidus — MU). Electrode placement was verified with isolated manual muscle tests (Figure 2).



RA = rectus abdominis; **EO** = external oblique; **IO** = internal oblique
LT = longissimus thoracis; **IL**= iliocostalis lumborum; **MU** = multifidus

Figure 2 Electrode placements for abdominal and back muscles

The electrodes for the RA were placed one centimeter above the umbilicus and two centimeters lateral to the midline. For the EO, the electrodes were placed just below the rib cage and along a line connecting the most inferior point of the costal margin and the contralateral pubic tubercle. For the IO, electrodes were placed one cm medial to the anterior superior iliac spine (ASIS) and beneath a line jointing both ASISs (Ng, Kippers and Richardson 1998). The electrodes of the LT were placed over the muscle belly at T12 level and along a line connecting the most superior point of the posterior axillary fold and the S2 spinous process. For the IL, the electrodes were placed at the L2 level and aligned parallel to the line between the posterior superior iliac spine (PSIS) and the lateral border of the muscle at the 12^{th} rib. For the MU the electrodes were placed at the L5 level and aligned parallel to the line between the PSIS and the L1-2 interspinous space (De Foa, Forrest and Biedermann 1989). The ground electrode was attached to the sternal notch. Prior to electrode attachment, the skin surface was shaved, slightly abraded and cleaned with alcohol to ensure low inter-electrode impedance.

To determine background noise level inherent to the acquisition system, ten seconds of EMG data were collected with the subject laying supine on a mat table at rest. In addition, ten seconds of maximum voluntary contraction (MVC) EMG data during maximal effort manual muscle tests were recorded. When subjects were lying on the mat table, the following standard manual muscle testing positions was performed to assess the maximum effort of each muscle:

- RA: trunk forward flexion against resistance with hips and knee flexed while lying supine (Kendall, McCreary and Provance 1993)
- EO and IO: oblique trunk flexion and rotation against resistance with hips and knee flexed while lying supine (Kendall et al. 1993);

- LT, IL and MU: trunk extension against resistance while lying prone with the legs stabilized (Kendall et al. 1993).

2.3.3. Kinematic marker positions

Infrared-emitting diode (IRED) markers were placed on the subject's upper body (acromion process, lateral epicondyle, and the head of the third metacarpal), and hip (greater trochanter) to record motion of the upper limbs and trunk in a global reference frame via a three-dimensional motion analysis system (Northern Digital Inc., Ontario, Canada). The subjects' trunk angles during the propulsion trials were assessed by calculating the angle between a reference line in the resting position while sitting on the test wheelchair and the same reference line during the propulsion trials. This reference line was drawn between the acromion process and the greater trochanter in the sagittal plane (Vanlandewijck et al. 2001). The motion system was synchronized with the SMART^{Wheel} and EMG system to record kinematics of the upper body, propulsion forces and muscle activity during propulsion.

2.3.4. Experimental Protocol.

A test wheelchair (Quickie R2 ultralight wheelchair, seat height 48 cm, and seat width 38 cm) was fitted bilaterally with SMART^{Wheels} (Three Rivers Holdings, LLC., Mesa, AZ), three-dimensional force and torque sensing wheels. Subjects were asked to push the test wheelchair, secured to a dynamometer with a four-point tie down system, to become familiar with the test setup before testing began. To investigate trunk muscle recruitment patterns during propulsion, subjects were instructed to push the test wheelchair without leaning their backs against the backrest. Subjects were asked to propel the test wheelchair at a two steady-state speeds: SLOW

(0.9m/s) and FAST (1.8 m/s) for 20 seconds respectively. Also, the subjects completed one acceleration trial (ACC) that involved a quick acceleration to their fastest possible propulsion speed, and maintaining the speed for a six second period. Real-time propulsion speed was displayed on a 17-inch computer screen placed in front of the subjects during all trials.

2.3.5. Data analysis

EMG signals were collected with a MyoSystem 1200 (Noraxon U.S.A. Inc., Scottsdale, AZ) using a bandwidth of 150 to 500 Hz. The data were then sampled and digitized on a computer at a rate of 1000 Hz. Afterward, the data were full wave rectified and smoothed with 4^{th} order Butterworth low-pass filter (10 Hz cut-off). EMG signals during propulsion were normalized as %MVC for each muscle. Significant EMG activity was defined as activity with an intensity of at least 5% MVC and for longer than 5% of the entire propulsion cycle (PC) (Mulroy, Gronley, Newsam and Perry 1996; Mulroy, Farrokhi, Newsam and Perry 2004). In order to compare muscle activity across subjects for the various speeds, the PC time was normalized to 100% for each subject. Additionally, the time spent in the push or recovery phase was expressed as a percentage of the entire PC. Data for each subject were then normalized to the group mean percentage of PC for push and recovery phases respectively. The push phase was further divided into two stages: early push and late push (Figure 3). The transition from early to late push was defined as the point when the hand passed the top-center position of the pushrim (Newsam et al. 1999). Recovery phase was separated into three smaller stages of follow-through, hand return, and pre-push according to maximal anterior and posterior hand position during the recovery phase (Newsam et al. 1999).

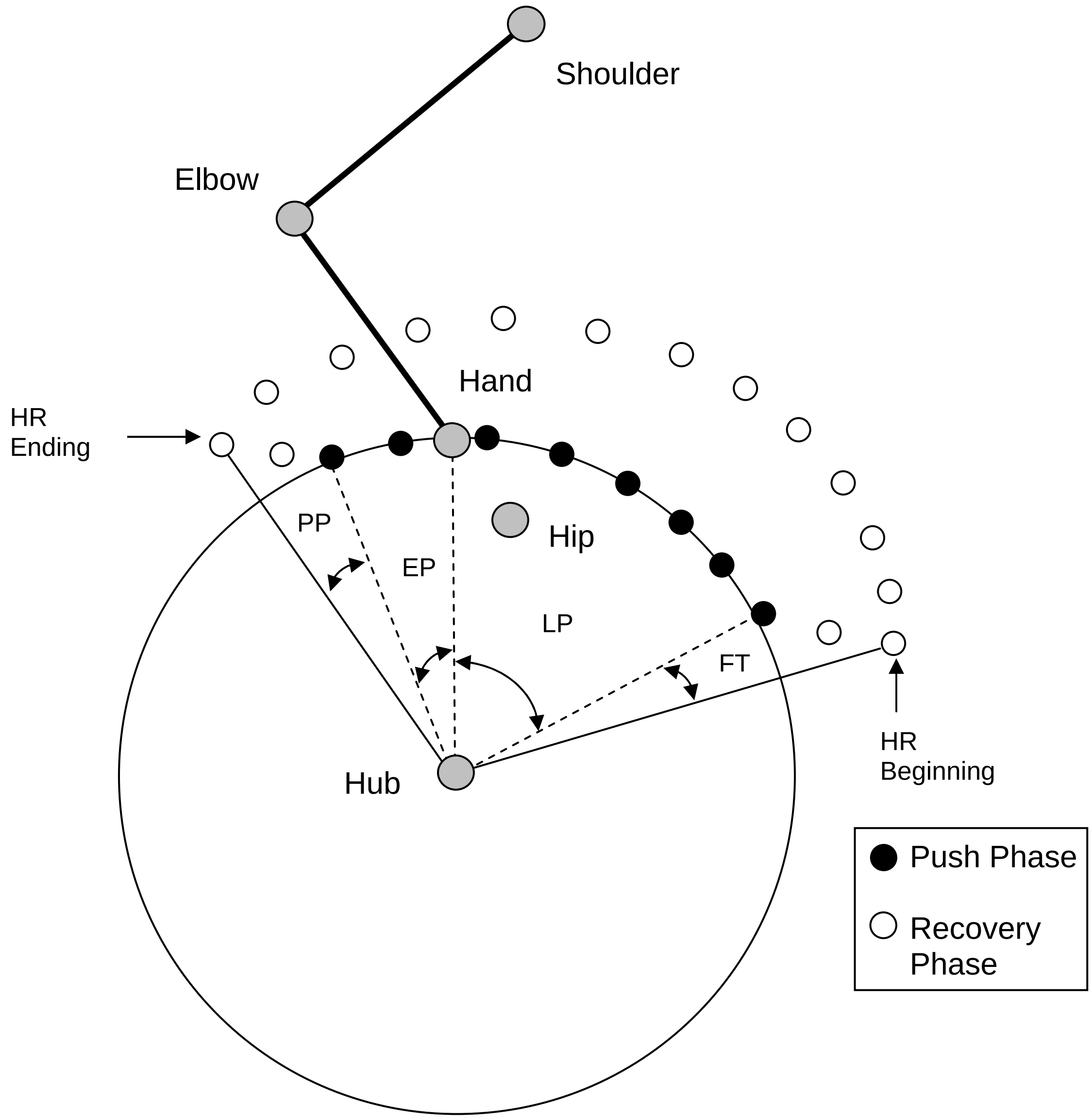


Figure 3 The motion marker placement and their relative wheelchair propulsion stages.
PP = pre-push; **EP** = early push; **LP** = late push;
FT = follow-through; **HR** = hand return.

The start and end of the push/recovery phase was determined by visual inspection of the presence/absence of forces and torques as detected by the SMART^{WheelsTM}. SMART^{WheelsTM} data were collected at 240 Hz and filtered with an 8th order Butterworth low-pass filter, zero lag and 30 Hz cut-off frequency (Cooper, Robertson, VanSickle, Boninger and Shimada 1997). Afterwards, the kinetic and EMG data were linearly interpolated for synchronization with the kinematic data collected at a rate of 60 Hz. In order to obtain a representative muscle activation profile at SLOW and FAST speed conditions, EMG data from ten consecutive strokes were averaged together to provide a single EMG profile of muscle activity during a complete propulsion cycle.

The ACC trial was divided further into a start-up phase, the initiation of wheelchair motion, and a steady-state phase when a constant propulsion speed was maintained. Data during the start-up phase can most likely represent a majority of daily wheelchair use. In order to discriminate between start-up and steady-state phase, the mean push phase time of the subject group from the first six strokes of the ACC trial was analyzed by one-way repeated measures ANOVA with a Bonferroni post-hoc test ($\alpha = 0.05$). The results showed that the first two strokes had a significantly longer push phases than the later four strokes ($p<0.05$). These first two strokes were then considered start-up strokes for all subsequent analyses, and data from these two strokes were averaged together to provide a representative value for start-up propulsion.

2.3.6. Statistics

Some trunk muscle EMG data during the push and recovery phases showed a skewed distribution with a certain period of inactivity. Therefore, the median intensity for each muscle EMG value during each stage and the three different propulsion speed conditions was reported as a descriptor rather than mean value of each muscle EMG activity. Afterward, the representative median values of each muscle EMG data were screened for normality of distribution with the Wilk-Shapiro W statistic ($\alpha=0.05$). The preliminary results indicated non-normal distribution of each muscle EMG data ($p<0.05$). Hence, Friedman two-way (propulsion speeds $\times$ stages) nonparametric repeated-measures analysis of variance (ANOVA) analysis with Post-hoc analysis using Wilcoxon signed ranks test was used to examine differences in the trunk muscle activation across the different stages and speed conditions during propulsion. The level of statistical significance was adjusted using Bonferroni corrections for multiple comparisons between five stages ($p = \alpha/4 = 0.0125$, where $\alpha = 0.05$) and three speed condition ($p = \alpha/2 = 0.025$, where $\alpha = 0.05$). All statistical analyses were performed using the SAS System for Windows 9.0 and SPSS 11.0 for Windows software package.

2.4. RESULTS

2.4.1. Push phase EMG Activity

During the SLOW condition, muscles with dominant activity during early push and late push stage included three back muscles (LT, IL, MU) and one abdominal muscle (EO) (Figure 4). Abdominal muscles (RA, IO) showed less activity. The IO was only active during early push phase (0%-7% of the PC). The IL was active in the middle of push phase (4%-27% of the PC), and the RA remained inactivate. The MU and EO both remained active throughout the entire push phase (0-50%). While subjects pushed their wheelchairs at the FAST and ACC conditions, abdominal muscles (RA, IO, EO) increased their EMG intensity level as did the back muscles (LT, IL, MU) (Figure 5 & 6). The IO, EO, LT, IL, and MU were all active throughout the entire push phase. The RA was inactive during the FAST condition, but contracted in the beginning of push phase (0%-45% of the PC) during the ACC condition. The median EMG intensity of the MU (17.2 %MVC) displayed the highest activity of all six muscles for all speed conditions and the RA (7.6 % MVC) showed the least activity during the push phase. Overall, the intensity of the back muscles across three speed conditions was significantly higher than abdominal muscle intensity during the push phase ($p<0.01$).

Figure 4 Group average trunk muscle activation patterns for the SLOW condition

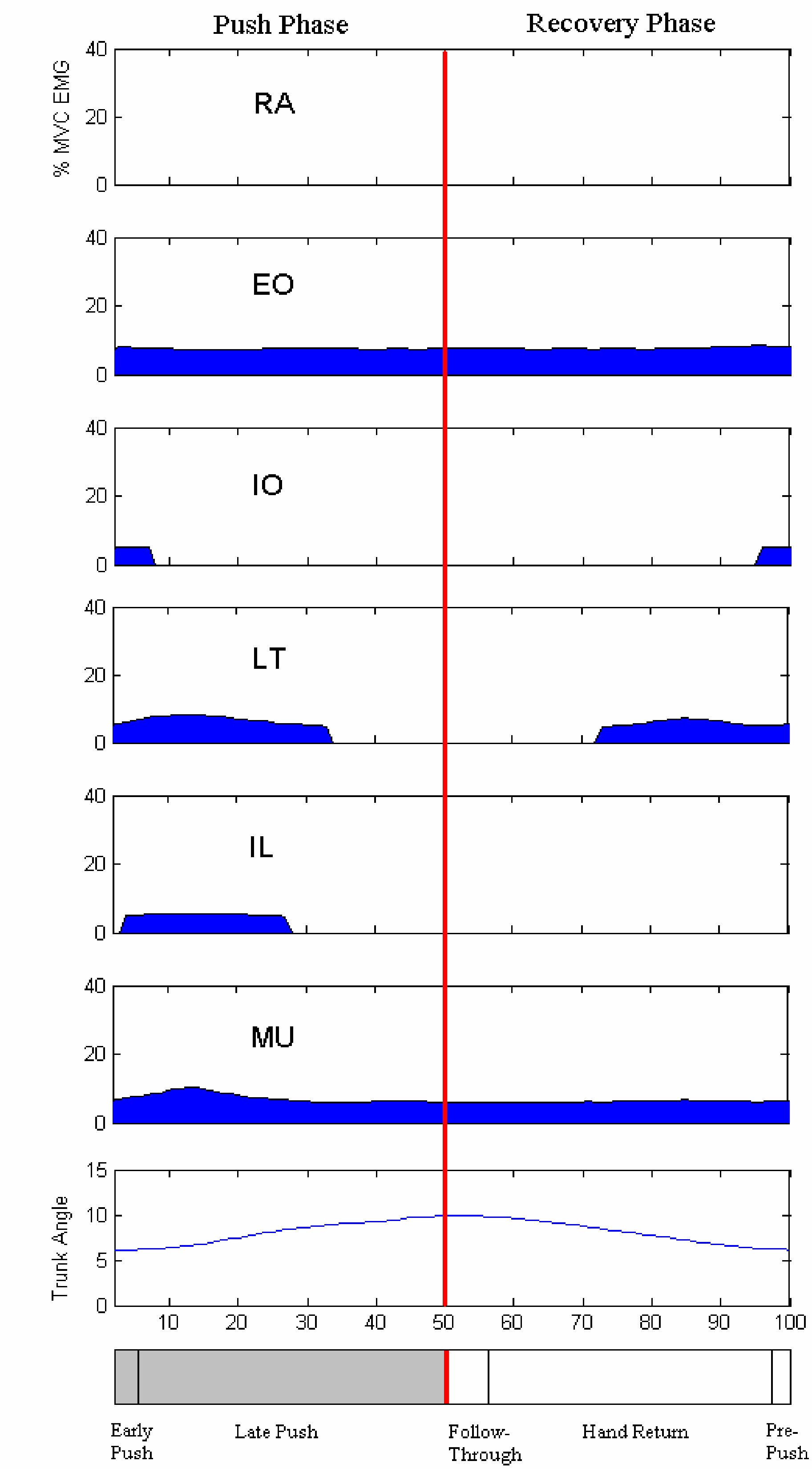


Figure 5 Group average trunk muscle activation patterns for the FAST condition

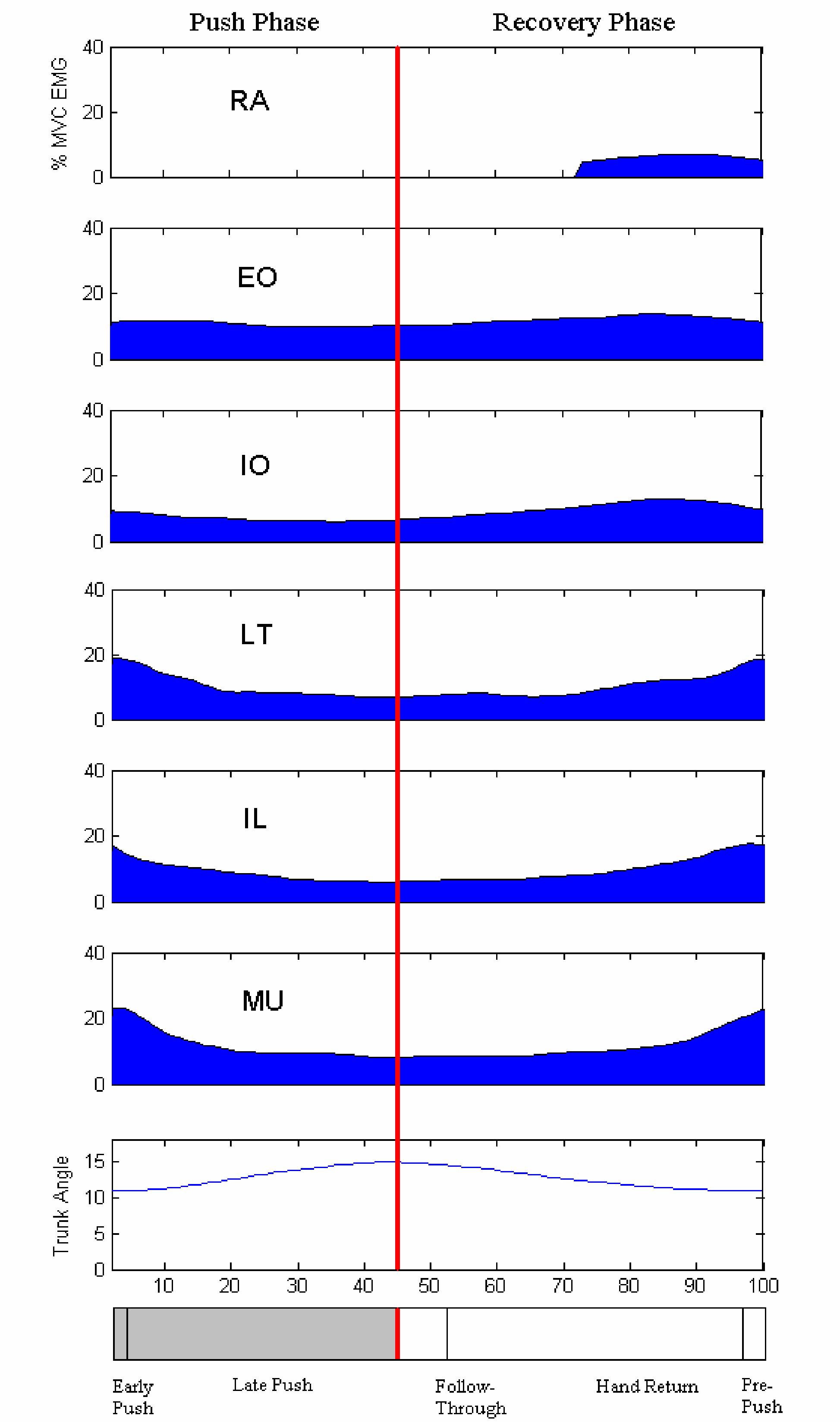
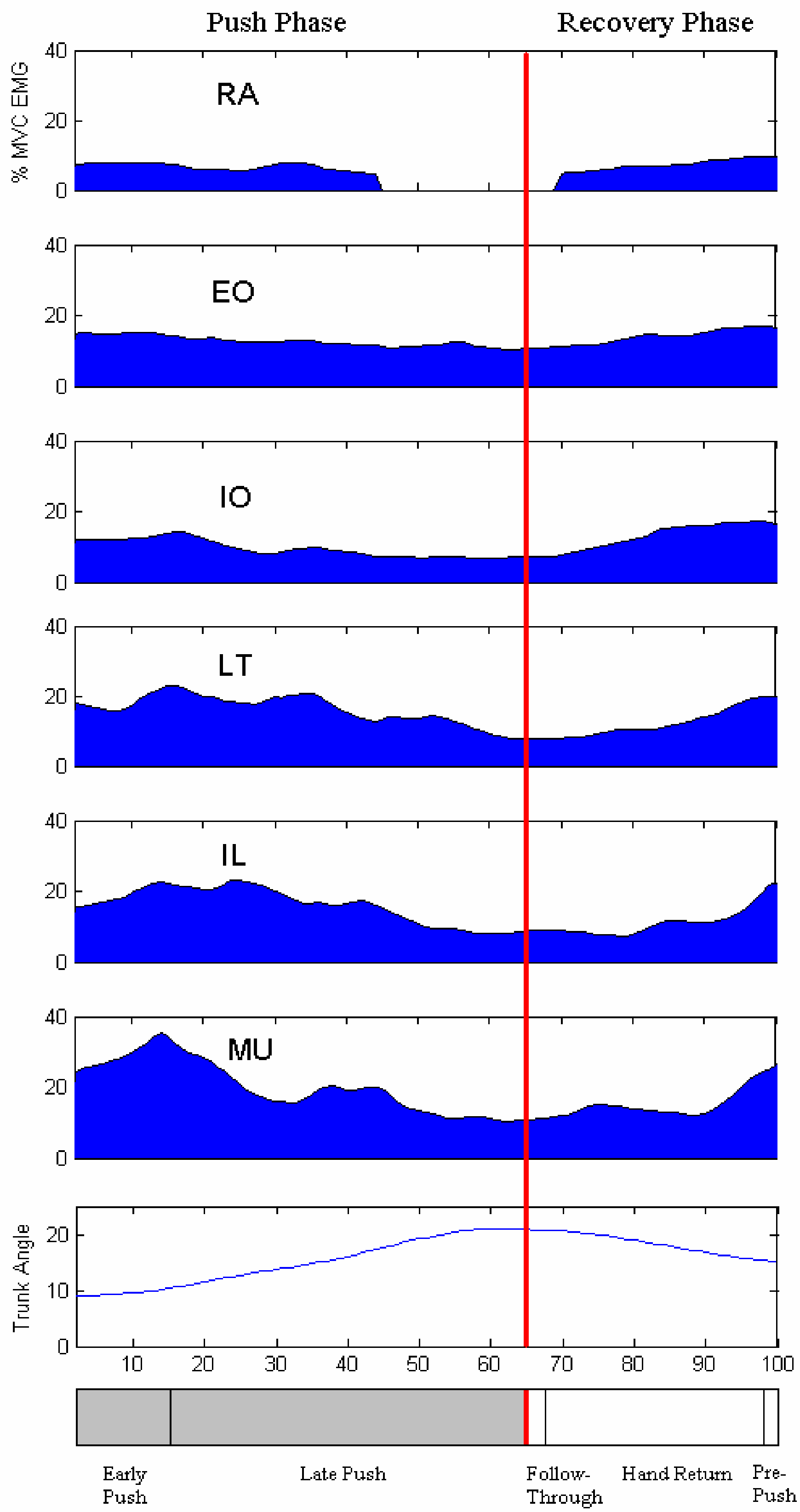


Figure 6 Group average trunk muscle activation patterns for the ACC condition



2.4.2. Recovery Phase EMG Activity

During the SLOW condition, the dominant muscles of the recovery phase included the same muscles that were active during the push phase (LT, MU and EO). The EO and MU muscles remained active throughout the entire recovery phase. The LT did not show EMG activity until the late recovery phase (73%-100% of PC). The RA and IL remained inactive during the recovery phase (Figure 4). During the FAST and ACC conditions, the intensity of abdominal muscles and back muscles activity increased (Figure 5 & 6). The IO, EO, LT, IL, and MU all showed activity during the entire recovery phase for the fast speed and acceleration conditions. The RA appeared active in the middle of recovery phase and remained active until the next PC (73%-100% of the PC) at the FAST and the ACC conditions (70%-100% of the PC), respectively. Similar to the push phase, MU showed the highest activity of all six muscles for all speed conditions (14.6% MVC). The overall activity of the back muscle groups across all three speed conditions was significantly larger than abdominal muscle activity during recovery phase ($p<0.01$).

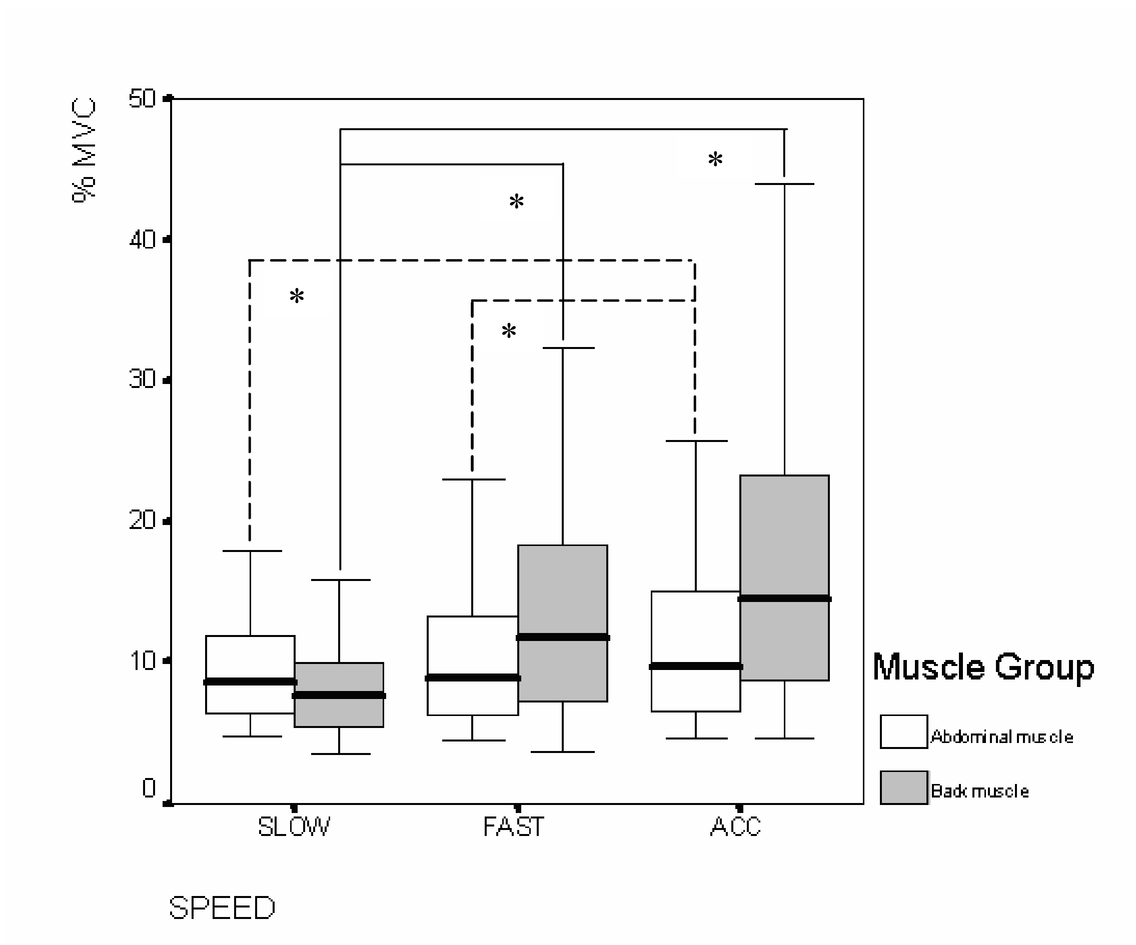
2.4.3. Trunk motion during propulsion

Subjects exaggerated their trunk forward flexion motion with increased propulsion speed, especially when accelerating from rest. The mean angle of trunk flexion was significantly larger during the ACC condition (16.0^{o}) than for the other constant speed conditions ($p<0.01$). Moreover, for all speed conditions, increased trunk flexion was observed at the early push phase, and reached a peak value during the follow-through stages. The largest trunk flexion angle (20.8^{o}) was found during the follow-through stage of recovery during the ACC condition. Trunk extension occurred at the hand return stage of the recovery phase to bring the trunk and upper limbs back for preparing the next stroke (Figure 6).

2.4.4. Main effect: speed

Trunk muscle intensity increased with increasing speed and start-up ($p<0.01$). Post-hoc tests showed that the intensity of the back muscle group (LT, IL and MU) during the ACC (17.9 % MVC) and FAST (15.4 % MVC) conditions was significantly higher as compared to pushing during the SLOW condition (9.1 % MVC, $p<0.01$) (Figure 7). A tendency of increasing muscle activity with a change in speed and acceleration from rest was found for the abdominal muscle group (RA, IO and EO) ($p=0.02$).

Figure 7 Main effect of speed for the abdominal and back muscle groups

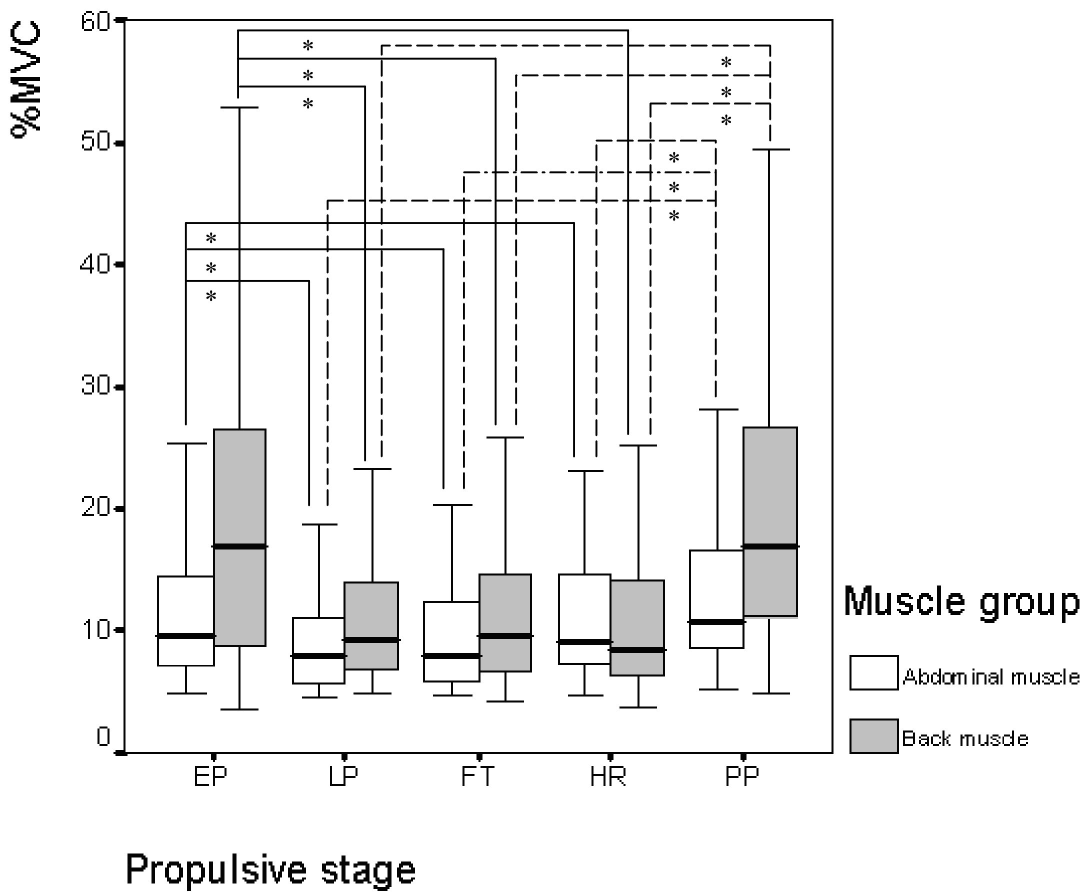


Solid lines represent the post-hoc comparisons between three different speed conditions
* denotes significant difference of muscle intensity between speed conditions

2.4.5. Main effect: propulsion stages.

A significant difference for the main effect of propulsion stages was also found (p<0.01). Post-hoc tests revealed that both abdominal (p<0.01) and back muscle groups (p<0.01) exhibited significantly higher activation at early push and pre-push stages (the beginning of push and late recovery) when compared to other three stages respectively (Figure 8). No significant interaction effect between speed and propulsion stages was found.

Figure 8 Main effect of propulsion stages for the abdominal and back muscle groups



Solid lines represent the post-hoc comparisons between **EP** and other three stages
Dashed lines represent the post-hoc comparisons between **PP** and other three stages
* denotes significant difference between propulsive stages
EP = early push; **LP** = late push; **FT** = follow-through; **HR** = hand return; **PP** = pre-push

2.5. DISCUSSION

2.5.1. Trunk muscle activation profile

The early and pre-push phases demanded more back and abdominal muscle recruitment compared to the other phases of propulsion. This cocontraction of both muscle groups likely provided the necessary trunk stability for generating propulsion forces. Moreover, this trunk muscle cocontraction may prevent the backwards trunk motion which has been observed among manual wheelchair users with limited trunk control due to SCI (Rice, Koontz, Boninger and Cooper 2004). Rice et al. observed the trunk motion of eighteen individuals with SCI ranging from T4 to L4 levels during wheelchair propulsion. They found that the trunk was moving backwards at the beginning of the push and concluded that reactive forces from the pushrim may cause backward motion of the trunk when trunk control is impaired. Furthermore, Koontz et al. investigated the influence of trunk movement patterns on mechanical effective forces (MEF) between wheelchair users with paraplegia and unimpaired participants (Koontz, Boninger, Rice, Yang and Cooper 2004). They reported that the wheelchair users with paraplegia not only exhibited greater backward trunk excursion during the push phase, they also propelled with less mechanical effective force than the unimpaired group. Backward trunk excursion increased at the faster speed and was accompanied by lower mechanical effective forces in the group with paraplegia.

The unimpaired subjects in the present study exhibited only trunk flexion, not extension, similar to the finding of unimpaired subjects by Koontz et al. The cocontraction of the highly active abdominal and back muscle groups may have provided adequate trunk stabilization to allow for initiating wheelchair propulsion without moving the trunk backward, thereby improving effective force application.

During the late push phase, back muscle activity declined with continuous activation of the abdominal muscle group. This may have allowed the trunk to flex forward while pushing the wheelchair. Trunk flexion increases the ability to transfer power to the pushrim and enhance the application of force on the pushrim to meet the physical demands of increased propulsion speed and acceleration (Sanderson and Sommer 1985). Trunk flexion also may improve the ability to reach the wheel for more effective propulsion since the body is moved forward and downward relative to the wheel. Another advantage to trunk flexion during propulsion is that the application of force is enhanced by gravity (Sanderson and Sommer 1985). Sanderson and Sommer have further hypothesized that any residual abdominal muscular strength could increase the amount of maximum trunk flexion. This assumption is verified by the present study. Kinematic data showed that the trunk started to flex forward during the late push phase, especially when pushing at a fast speed or during acceleration. At the same time, EMG data of the abdominal muscle group revealed increased activity, which may have allowed for continuous trunk flexion during the push phase.

Like the push phase, EMG data during the recovery phase showed cocontraction between back and abdominal muscles. Back muscle activity gradually increased during middle and late recovery phase (hand return/pre-push stage), particularly for the MU. Likewise, concentric contractions of the back muscles began when the trunk returned to an upright position in preparation for the next stroke. At the same time, abdominal muscles contracted eccentrically to slow down the backward motion of trunk and then contracted concentrically to flex the trunk at the pre-push stage in late recovery.

The intensity of abdominal and back muscle activity during wheelchair propulsion generally was low (averaged 10% to 16 % MVC, respectively) but with prolonged duration (as high as 100% of the entire PC). With increased propulsion speed, the average abdominal and back muscle intensity increased to 12% to 18 % MVC, respectively. Such above 10% MVC muscle intensity combined with a long duration of activity could lead to fatigue (Kahn, Favriou, Jouanin and Monod 1997), thereby limiting the individual's functional capacity to maintain a consistently high propulsion. Therefore, it is reasonable to expect that pushing a manual wheelchair at fast speed condition for long period of time is a very difficult task because it not only requires effective propulsion forces but also demands trunk and shoulder stabilization that could lead to muscle fatigue. Special attention should be given to other forms assistive technology to stabilize the trunk and reduce shoulder muscle effort.

2.5.2. Implication for persons with decreased trunk control

Previous studies have investigated a variety of devices, such as a chest belt or trunk orthosis, to stabilize the trunk among wheelchair users in an attempt to enhance their performance in activities of daily living (Curtis, Kindlin, Reich and White 1995; Allison and Singer 1997). In recent years, technology has become available to improve trunk stability and sitting posture by attaching a rigid backrest, modifying seat frame angles, and artificially stimulating paralyzed trunk muscles. Parent et al. (2000) reported that a rigid back support improved trunk stability and comfort for the user compared to a regular sling backrest. As a result, Parent et al. hypothesized that propulsion efficiency may be improved (Parent, Dansereau, Lacoste and Aissaoui 2000). Samuelsson et al. (2004) investigated the effect of two reclined positions of the seat frame with consistent back angles on wheelchair propulsion. They found that the change of inclination of the seat frame (12^0) significantly broadened the stroke angle and reduced push frequency during wheelchair propulsion on a treadmill (Samuelsson, Tropp, Nylander and Gerdle 2004). Maurer and Sprigle (2004) studied the effect of seat inclination on seat pressure among wheelchair users. They indicated that the effect of "squeezing" a manual wheelchair frame by seat inclination could provide better stability to people with impaired or absent trunk control while not significantly increasing seat interface pressure (Maurer and Sprigle 2004). With adequate stabilization of the trunk, manual wheelchair users may be able to improve propulsion force application. Therefore, customizing the wheelchair by either using a rigid backrest or reclining seat frame angles may be a compensatory strategy for people with poor trunk control.

Functional electrical stimulation (FES) is another possible method for providing trunk stability through the electrical activation of the otherwise paralyzed trunk musculature. Clinical applications of FES after spinal cord injury include standing, walking and hand grip through the electrical stimulation of the user's paralyzed muscles (Jaeger, Yarkony and Smith 1989; Yarkony, Jaeger, Roth, Kralj and Quintern 1990; Triolo, Bevelheimer, Eisenhower and Wormser 1995; Davis, Triolo, Uhlir, Bieri, Rohde, Lissy and Kukke 2001). The present study indicates that low back muscles could play an important functional role to stabilizing the trunk and preventing collapse while leaning further forward into the push. It is possible that electrical stimulation of the low back muscles may allow manual wheelchair users to adopt a trunk flexion propulsion style to enhance effective force application and compensate for fatigue. Furthermore, continuous electrical stimulation of both abdominal and back muscle groups may provide adequate trunk stabilization during propulsion. It might help people with impaired or absent trunk control to push wheelchairs at a fast speed or accelerate more easily than without stimulation. Further research is needed to investigate the potential benefit of FES on wheelchair propulsion in participants with SCI.

2.5.3. Limitations

There are a number of limitations that require consideration. First, the test wheelchair was not adjusted to the individual's anthropometry. The axle position of rear wheel was fixed which could result in varied seating positions among subjects. Studies have shown that seat position can affect propulsion biomechanics. (van der Woude, Veeger, Rozendal and Sargeant 1989; Masse, Lamontagne and O'Riain 1992; Hughes, Weimar, Sheth and Brubaker 1992; Boninger, Baldwin, Cooper, Koontz and Chan 2000). However, changing the seat position does not appear to affect trunk motion (van der Woude et al. 1989; Masse et al. 1992; Hughes et al. 1992; Boninger et al. 2000), which was the primary interest in the present study. Second, we used a low sling backrest on the test wheelchair (backrest height 20 cm) and instructed unimpaired participants not to lean on the backrest to determine the maximal activity demanded by the trunk musculature. Use of the backrest could have resulted in different trunk muscle activation profiles. Third, the trunk muscle activation profile was based on unimpaired participants. Individuals with SCI may compensate for decreased trunk stability by leaning on the backrest and/or recruiting intact muscles under volitional control, such as latissimus dorsi and trapezius, to stabilize their trunk and shoulder during wheelchair propulsion. A future study is needed to examine the recruitment patterns of residual muscles in persons with SCI.

2.6. CONCLUSION

The present study provided an understanding of the functional role of trunk musculature during wheelchair propulsion based on unimpaired subjects. The results showed a muscle activation profile of the trunk musculature at three different propulsion speed conditions. Both back muscle (LT, MU, and IL) and abdominal muscle (RA, IO, EO) groups illustrated the highest intensity during the pre-push and early push stages of the PC. Moreover, these two muscle groups cocontracted to provide sufficient trunk stability for the propulsion tasks. Customizing a personal wheelchair with a rigid backrest, modifying seat frame angles or artificially stimulating paralyzed trunk muscles may be options employed to increase trunk stability. As a result, propulsion performance may be improved. Further investigations on wheelchair propulsion performance among individuals with SCI by providing trunk stability through a variety of techniques are warranted.

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3. BIOMECHANICAL ANALYSIS OF FUNCTIONAL ELECTRICAL STIMULATION ON TRUNK MUSCULATURE DURING WHEELCHAIR PROPULSION: A PILOT STUDY

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ACKNOWLEDGEMENTS

This research was supported by the U.S. Department of Veterans Affairs, Rehabilitation Research and Development (B3043-C), and American Society of Biomechanics Graduate Student Grant-In-Aid (2004).

3.1. ABSTRACT:

Objective: To examine how surface electrical stimulation of trunk musculature influences the kinematic, kinetic and metabolic characteristics of wheelchair propulsion.

Methods: Twelve participants with spinal cord injury (SCI) were asked to propel their own wheelchairs on a dynamometer at a target speed of 1.35 m/sec for three five-minute trials. During a propulsion trial, one of three stimulation levels (HIGH, LOW and OFF) was randomly applied to the participant's abdominal and back muscle groups with a surface functional electrical stimulation (FES) device. Propulsion kinetics, kinematic and metabolic variables were obtained from SMART$^{\text{Wheels TM}}$, a 3-D OPTOTRAK motion analysis system and a SensorMedics 2900 Metabolic Cart respectively. Kinetic, kinematic and metabolic variables were recorded during three time intervals (30 seconds each) within a five-minute trial to examine the effects of stimulation. The differences amongst the variables due to stimulation level over three time intervals were examined by using a two-way repeated measures ANOVA.

Results: Participants with HIGH stimulation produced higher propulsion power output ($p=0.01$) and increased their gross mechanical efficiency (GME) ($p=0.06$) during wheelchair propulsion without a significant increase of energy expenditure compared to LOW stimulation and OFF for the three time intervals consistently over time. No statistical differences in other propulsion kinetic variables and upper limb and trunk motion between stimulation levels were found.

Conclusions: FES on the trunk musculature has potential advantages in helping manual wheelchair users with SCI improve propulsion efficiency.

KEYWORDS: Metabolic measurement, trunk stability, seat backrest, kinetics, kinematics.

3.2. INTRODUCTION

Functional electrical stimulation (FES) is a technique which artificially generates neural activity in order to overcome lost functions of paralyzed, incontinent or sensory impaired muscles. FES has been used extensively in individuals with spinal cord injury, primarily to induce ambulation, standing, hand grip and cycling exercise (Jaeger, Yarkony and Smith 1989; Yarkony, Jaeger, Roth, Kralj and Quintern 1990; Triolo, Bevelheimer, Eisenhower and Wormser 1995; Wieler, Stein, Ladouceur, Whittaker, Smith, Naaman, Barbeau, Bugaresti and Aimone 1999; Kukke and Triolo 2004). Studies with FES-assisted walking have demonstrated improvements in functional mobility and walking speed (Shimada, Sato, Abe, Kagaya, Ebata, Oba and Sato 1996; Kobetic, Triolo and Marsolais 1997; Klose, Jacobs, Broton, Guest, Needham-Shropshire, Lebwohl, Nash and Green 1997; Moynahan, Mullin, Cohn, Burns, Halden, Triolo and Betz 1996). Klose KJ et al. (1997) showed that at the end of 11 weeks of training, 16 users of Parastep, which is a commercially available FES assisted ambulation device, could ambulate independently with an average speed of 0.08 m/sec (Klose et al. 1997). Brissot et al. (2000) reported that thirteen Parastep users could achieve independent ambulation with a mean walking speed of 0.15 m/sec (Brissot, Gallien, Le Bot, Beaubras, Laisne, Beillot and Dassonville 2000). Similar findings were also reported by Jacobs et al. with a mean walking speed of 0.22 m/sec (Jacobs and Mahoney 2002).

However, FES-assisted walking speed is far from a comfortable walking speed of 1.5 m/sec (Bohannon 1997) or freely chosen wheelchair propulsion speed. Newman et al. indicated that the wheelchair propulsion ranged from 1.6 m/sec in low paraplegics to 0.9 m/sec in C-6 tetraplegics (Newsam, Mulroy, Gronley, Bontrager and Perry 1996). Mukenrjee et al. also pointed out the freely chosen speed for wheelchair propulsion was 0.9 m/sec (Mukherjee and

Samanta 2001). The low FES-assisted walking speed does not sufficiently allow for ambulation in activities of daily life. Thus, most users of FES-assisted walking systems use them for short distance walking, but still rely on a wheelchair as a primary means of mobility in daily activities (Kobetic, Triolo, Uhlir, Bieri, Wibowo, Polando, Marsolais, Davis and Ferguson 1999; Moynahan et al. 1996).

Because of its modest performance associated with high metabolic cost and cardiovascular strain, relatively high energy uptake has been perceived as another limitation of a FES-assisted walking system for mobility in daily life. Kobetic (1999) reported that an increased demand of energy consumption appeared during FES-assisted walking (Kobetic et al. 1999). Based on metabolic measurements from 11 Parastep users, Jacobs et al. (1997) showed that use of a FES-assisted walking device even at a self-selected comfortable pace could cause inappropriately high exercise intensity (Jacobs, Nash, Klose, Guest, Needham-Shropshire and Green 1997). The increased energy expenditure required with FES-assisted walking is very close to a voluntary arm cranking exercise (Jacobs and Mahoney 2002; Brissot et al. 2000). The high energy cost of FES-assisted walking ultimately reduces the extent to which individuals use it in the community and at home.

Manual wheelchair propulsion requires large static work from proximal shoulder muscle synergy and cocontraction to stabilize and adjust the shoulder girdle complex with respect to the trunk, for gripping and applying forces to the hand rim (van der Helm and Veeger 1996; van der Woude, Dallmeijer, Janssen and Veeger 2001; van der Woude, Veeger, Dallmeijer, Janssen and Rozendaal 2001; Vanlandewijck, Theisen and Daly 2001). It has been suggested that lack of trunk stability, leading to a less erect posture and poor support of the shoulder girdle complex, may limit production of maximal upper limb strength to push a wheelchair (Powers, Newsam, Gronley, Fontaine and Perry 1994). Furthermore, rapid movement of the upper limb during wheelchair propulsion produces a complex interplay of dynamic reactive forces acting on the shoulder and trunk. During propulsion, the dynamic reactive forces exerted on the trunk from the upper limbs, can cause the trunk to move backwards during the push phase of propulsion, referred to as paradoxical trunk movement (Rice, Koontz, Boninger and Cooper 2004). The occurrence of paradoxical trunk movement during propulsion is an important phenomena because it has been shown to occur at the beginning of the push phase and reduce the mechanical effectiveness of propulsion forces in SCI subjects in comparison to unimpaired subjects (Koontz, Boninger, Rice, Yang and Cooper 2004).

The functional role of the trunk during wheelchair propulsion was recently investigated by Yang et al. (Yang, Koontz, Triolo, Mercer and Boninger 2005a). They observed trunk muscle EMG activity among 14 unimpaired subjects during various propulsion speeds. They found that both back and abdominal muscles were most active in the pre-push and early push phase. The activity of these muscles increased just prior to hand contact and continued through initial contact. The combined activity of the trunk muscles is likely a preparatory trunk response to counteract dynamic reactive forces during propulsion (Aruin and Latash 1995). However, most manual wheelchair users (MWUs) who lose voluntary control of trunk musculature (e.g., individuals with high paraplegia due to SCI) are not able to recruit trunk stabilizing muscles. Consequently, the dynamic reactive forces exerted on the trunk result in inefficient paradoxical trunk movement during propulsion.

Using FES to augment trunk stability may improve the wheelchair mobility of persons with SCI. A preliminary study involving three persons with SCI who had an implanted FES-assisted walking system showed that continuous stimulation of the lumbar erector spinae through their FES systems during wheelchair propulsion resulted in higher resultant propulsion forces and torques accompanied by a trunk flexion posture (Triolo, Yang, Koontz, Nogan and Boninger 2005). Upon activation of the back muscles, implanted FES users could lean their trunk further into the push and increase propulsion forces.

The goal of this study was to examine whether a surface FES system, applied to the abdominal and back muscles of MWUs could improve propulsion technique and efficiency without a significant increase of energy expenditure. We hypothesized that using stimulation on trunk musculature would 1) produce a significant increase in propulsion force, torque, mechanical effective force, and power production, 2) allow for greater trunk flexion, longer push angles, and greater ranges of motion at the wrist, elbow and shoulder, and 3) generate no significant increase of energy expenditure during a five-minute propulsion trial compared to propulsion without stimulation.

3.3. METHODS

3.3.1. Subjects:

Twelve manual wheelchair users (10 male and 2 female, see Table 1 for subject demographics) provided informed consent in accordance with the procedures approved by the Institutional Review Board of Veterans Affair Medical Center prior to participation in the study.

Inclusion criteria were: 1) complete or incomplete spinal cord injury between C6 and T12; 2) use a manual wheelchair as a primary mode of mobility, and be 3) between the ages of 18 and 65 years. Exclusion criteria were: 1) previous history of upper extremity pain, 2) presence of a heart or lung condition that is worsened by pushing a wheelchair, and 3) pregnancy. Volitional trunk control was assessed by having the participants lean their trunk forward, backward and laterally unsupported and noting any loss of balance. Trunk control was noted as either present or absent (Table 1). Two subjects (S4, and S5) reported surgical fusions of the thoracic spine without implanted rods, and two subjects (S6, and S12) had cervical spine fusions with implanted rods. The rest of the participants did not have any fusion of the thoracic and lumbar spine or implanted rods to stabilize their spine.

Table 1 Subject characteristics

Subject	Gender	Handedness	Level of lesion	Age	Years post injury	Trunk control
S1	M	Right	T7 (ASIA-B)	48	23	absent
S2	M	Right	C7 (ASIA-A)	29	11	absent
S3	M	Right	T4 (ASIA-A)	43	24	absent
S4	F	Right	T10 (ASIA-A)	37	19	absent
S5	M	Right	T10 (ASIA-B)	45	23	absent
S6	M	Right	C6 incomplete (ASIA-C)	31	4	present
S7	M	Right	T9 incomplete (ASIA-B)	28	9	present
S8	M	Right	T6 (ASIA-A)	53	29	absent
S9	M	Right	T8 (ASIA-A)	56	11	absent
S10	F	Right	T2 (ASIA-A)	47	30	absent
S11	M	Right	T5 (ASIA-A)	43	9	absent
S12	M	Right	C6 (ASIA-A)	39	18	absent
Mean				41.6	17.5	
SD				9.05	8.5	

3.3.2. Surface FES device

In order to stimulate the abdominal and back muscle bilaterally at the same time, two commercially available double channel stimulators (EMS-5000 Electronic Muscle Stimulator, OrthoBionics Inc., Dallas, TX) were linked together using a custom circuit. Four pairs of self-adhering surface electrodes (Superior Silver Electrodes, size 2” × 2”, Uni-Patch, Wabasha, MN) were placed in the following positions: two pairs (1 right, 1 left) over the rectus abdominal muscles, two pairs (1 right, 1 left) over the multifidus muscles (Figure 9). The parameters of stimulation were set for asymmetrical biphasic waves of 30 Hz frequency, 300 µs pulse width, and up to the maximal amplitude of 80 mA depending on the participant’s tolerance levels. The stimulation activation ratio was set to 30 seconds of continuous burst stimulation with one-second stimulation off during the entire five minute propulsion trial.

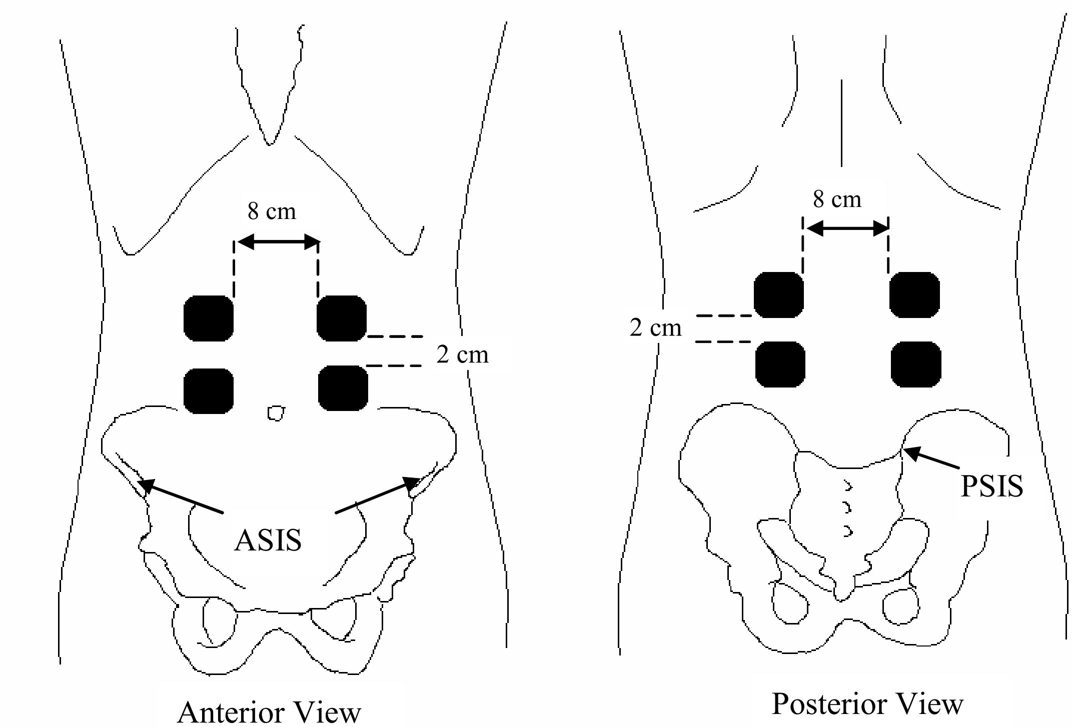


Figure 9 Surface stimulation electrode placements for abdominal and back muscles

Participants were given a supervised functional test prior to the experimental trials to determine the threshold response to surface electrical stimulation and maximal tolerable level of stimulation. The threshold response was defined as the minimal stimulation amplitude causing muscle contraction which was detectable by manual muscle palpation. The maximal tolerable level of stimulation was defined as the maximal tolerable amplitude reported by the participant or the maximal stimulation amplitude provided by the stimulator, which was 80 mA. After determination of the threshold and maximal tolerable levels, 50% and 25% of the difference between these two levels was used as HIGH and LOW stimulation intensity, respectively.

3.3.3. Kinetic/kinematic measurement system

Propulsion kinetics were obtained using a SMART$^{\text{Wheels TM}}$ (Three Rivers Holdings, Inc., Mesa, AZ) on both sides of the participant's own wheelchair to measure three dimensional forces and moments on the pushrim in a global reference system. Propulsion kinetic data were collected with a sample frequency of 240 Hz and filtered with an 8th order Butterworth low-pass filter, zero lag and 30 Hz cut-off frequency (Cooper, Robertson, VanSickle, Boninger and Shimada 1997). The propulsion cycle was comprised of push and recovery phases. The start and end of the push/recovery phase was determined by visual inspection of the presence/absence of propulsion forces and torques detected by the SMART$^{\text{Wheels TM}}$.

An OPTOTRAK 3020 three-dimensional motion analysis system (Northern Digital Inc., Ontario, Canada) and infrared-emitting diode (IRED) markers placed bilaterally on the subject's upper body (acromion process, lateral epicondyle, olecranon, radial styloid, ulnar styloid, and the head of the third metacarpal), and greater trochanter were used to record flexion/extension motion of the trunk, shoulder, and elbow (Figure 10). The motion system was synchronized with the SMART$^{Wheels\ TM}$ and the kinematic data were recorded at a sampling frequency of 60 Hz.

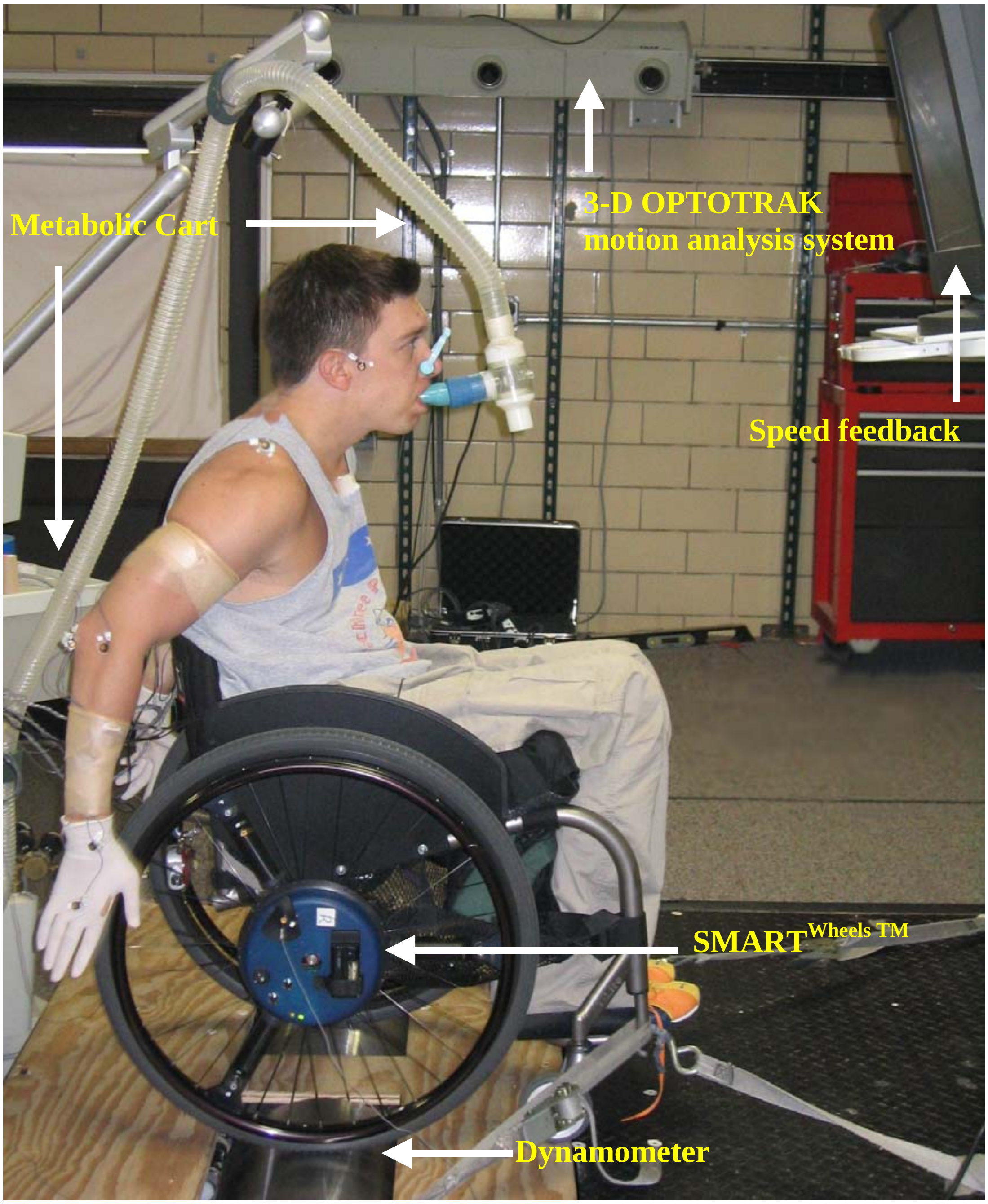


Figure 10 Experimental setup.

SMART$^{Wheels\ TM}$ and IRED markers used for kinetic and kinematic data acquisition. The mouthpiece with an open flow pneumotach was used for metabolic data collection.

3.3.4. Metabolic measurement

Metabolic data collection was performed by means of open-circuit spirometry using the SensorMedics 2900 Metabolic Measurement System (SensorMedics, Yorba Linda, CA). The analyzers were calibrated using known gases mixtures immediately prior to testing each participant. The participant was fitted with a nose clip and a standard mouthpiece with an open flow pneumotach attached to the SensorMedics 2900 system by means of tubing which did not inhibit the propulsion efforts.

3.3.5. Experimental Protocol

Participants performed three five-minute repeated wheelchair propulsion trials on an dynamometer system with independent rollers that simulated propulsion on a smooth level surface. Their personal wheelchairs were secured to the dynamometer system with a four-point tie down system. Participants were asked to push the wheelchair at target speed of 1.34 m/s and maintain this speed for five minutes. Propulsion speed was displayed on a 0.43 meter computer screen placed in front of them to provide visual speed feedback. For each propulsion trial, participants either propelled without stimulation (OFF) or with stimulation (HIGH, LOW) on both their abdominal and back muscles to examine the effect of FES during propulsion. The stimulation level order was randomly assigned. Before proceeding to the test trial, participants were asked to propel their wheelchairs on the dynamometers for several minutes with HIGH, LOW stimulation and stimulation OFF to familiarize with the experimental setup and the stimulation levels. At least five minutes of rest preceded each trial to avoid muscle fatigue.

During each five-minute propulsion trial, kinetic and kinematic data were collected for three time blocks of 30 seconds without the participant knowing when data was being collected. The first time block (T1) was the first 30 seconds of the propulsion trial, the second time block (T2) started at the middle of propulsion trial (2.00 min), and the last time block (T3) was initiated during the last minute (4.00 min) of the propulsion trial. Oxygen uptake ($\dot{V}O_2$, ml/min), and carbon dioxide output **(** $\dot{V}CO_2$, ml/min) were synchronized with kinetic/kinematic measurement and continuously recorded through the entire propulsion trial in 20-second intervals. Afterward, Metabolic data were averaged over two 20-second intervals which included same 30 seconds of propulsion data in each time block.

3.3.6. Data analysis

The mean value of $\dot{V}O_2$ and $\dot{V}CO_2$ during each time block (T1, T2 and T3) was analyzed by one-way repeated measures ANOVA with a Bonferroni post-hoc test (α =0.05) to determine steady-state metabolic response. The results showed that the mean value of $\dot{V}O_2$ and $\dot{V}CO_2$ during T1 were a significantly smaller than during T2 and T3 ($p<0.01$) (Table 2). There were no significant differences of the mean metabolic valuables between T2 and T3 ($p>0.1$). Therefore, metabolic variables and energy expenditure were only derived and analyzed for the T2 and T3 time intervals since subjects had not reached a steady-state condition during T1.

Table 2 Mean value of oxygen uptake and carbon dioxide output during propulsion trials

		T1 (0-0.30 min)		T2 (2-2.30 min)		T3 (4-4.30 min)	
		Mean	SD	Mean	SD	Mean	SD
$\dot{V}O_2$ (ml/min)	HIGH	494.29	128.36	830.97	293.99	855.24	290.79
	LOW	463.98	106.97	835.57	312.18	862.24	273.71
	OFF	458.13	85.86	822.67	265.85	843.55	296.91
Group mean		**472.13***	106.15	829.74	282.10	853.68	278.29
$\dot{V}CO_2$ (ml/min)	HIGH	419.06	117.80	699.58	289.48	753.32	316.81
	LOW	370.81	96.97	722.02	324.89	758.82	291.97
	OFF	381.48	77.99	724.79	266.82	751.52	307.45
Group mean		**390.45***	98.07	715.46	285.57	754.55	295.89

* denoted a significant difference between time intervals ($p<0.05$).

Kinetic data were linearly interpolated for synchronization with the kinematic data with collection rate of 60 Hz. Afterward, the peak kinetic and kinematic variables were determined from ten consecutive strokes during the middle of the each time block (T1, T2 and T3) and then averaged. Data from the left and right wheels were averaged since the kinetic and kinematic variables were highly correlated between sides ($r > 0.700$, $p<0.01$). Kinetic data from the SMART$^{\text{Wheels TM}}$ were further transformed to a force radial to the pushrim (F_r) and a force tangential to the pushrim (F_t) (Boninger, Cooper, Robertson and Rudy 1997a; Cooper et al. 1997). Mechanical effective force (MEF), the proportion of force at the pushrim that contributes to forward motion, was calculated as:

$$MEF = \frac{1}{n}\sum_{i=1}^{n}\frac{F_t^{\,2}(i)}{F^{\,2}(i)}$$

where $F = \sqrt{F_t^{\,2} + F_r^{\,2} + F_z^{\,2}} = \sqrt{F_x^{\,2} + F_y^{\,2} + F_z^{\,2}}$

The shoulder, elbow and wrist joint angles during propulsion were determined using a local coordinate system approach (Boninger, Cooper, Robertson and Shimada 1997b; Boninger, Cooper, Shimada and Rudy 1998; Cooper, Boninger, Shimada and Lawrence 1999). Trunk motion was measured using a reference line drawn between the acromion process and the greater trochanter in the sagittal plane (Vanlandewijck et al. 2001). Trunk flexion/extension was calculated by computing the angle between the reference line in the resting position while sitting in the test wheelchair and the same reference line during the propulsion trials. The paradoxical trunk movement was estimated as the distance of the trunk backward movement (mm) when the arm moved forward during the push phase. We concentrated on analyzing trunk excursion in the sagittal plane as the trunk has been shown to exhibit little motion in the coronal or transverse plane during wheelchair propulsion (Boninger et al. 1998).

The estimation of physiological energy expenditure (EE) during propulsion was calculated based on followed equation (Bursztein 1989; Arva, Fitzgerald, Cooper and Boninger 2001):

$$EE(Watts) = \frac{1440}{0.239*24*3600} \times \left\{ (3.941 \times \dot{V}O_2[\mathrm{ml/min}]) + (1.1 \times \dot{V}CO_2[\mathrm{ml/min}]) \right\}$$

The gross mechanical efficiency (GME) during wheelchair propulsion is defined as the ratio between externally produced energy (propulsion power output) and physiological energy expenditure:

$$GME(\%) = \frac{P_o}{EE} \times 100\%$$

where the propulsion power output was calculated from the measured propulsion torque applied on the pushrim (Mz), velocity (V_{rim}) and pushrim radius (R_{rim}) according to:

$$P_o(Watts) = \frac{Mz * V_{rim}}{R_{rim}}$$

3.3.7. General propulsion characteristics

Propulsion velocity was calculated based on SMART$^{Wheels\ TM}$ encoder reading that measured angular displacement during propulsion. Stroke frequency (*f*) was defined as the number of strokes that occurred per second. Start angle (SA) and end angle (EA) were defined as the angle between the line from the hand marker (the head of the third metacarpal) through the wheel axle, relative to the horizontal, at the start and the end of the push phase, respectively. Contact angle (CA) was defined as the angle between SA and EA.

3.3.8. Statistical analysis

To test the assumption of normality for analysis of variance (ANOVA), each dependent variable was screened for normality of distribution with the Wilk-Shapiro W statistic. There was no evidence that the normality was violated ($p>0.05$). Afterward, in order to examine the differences in surface FES stimulation levels on biomechanical and metabolic variables across the three time intervals, a two-way (stimulation levels × time intervals) repeated-measures analysis of variance (ANOVA) using the mixed models procedure with a Bonferroni post-hoc test based on a least-squares means (LSM) analysis was used. Mixed modeling (PROC MIXED) was used because the same subjects propelled their wheelchair with all stimulation levels. Mixed modeling allows for testing both random and fixed effects (Littell, Milliken, Stroup and Wolfinger 1996). Subjects were entered into the mixed model as the random factor and the fixed factors were stimulation levels (OFF, LOW, and HIGH) and time periods (T1, T2, and T3). Due to technical difficulties, one subject's metabolic data could not be processed. Therefore, metabolic data from 11 subjects were entered into the statistical model. All statistical analyses were performed using the SAS System for Windows 9.0 software package and SPSS 11.0. The level of statistical significance was set at $\alpha=0.05$.

3.4. RESULTS

3.4.1. Force application

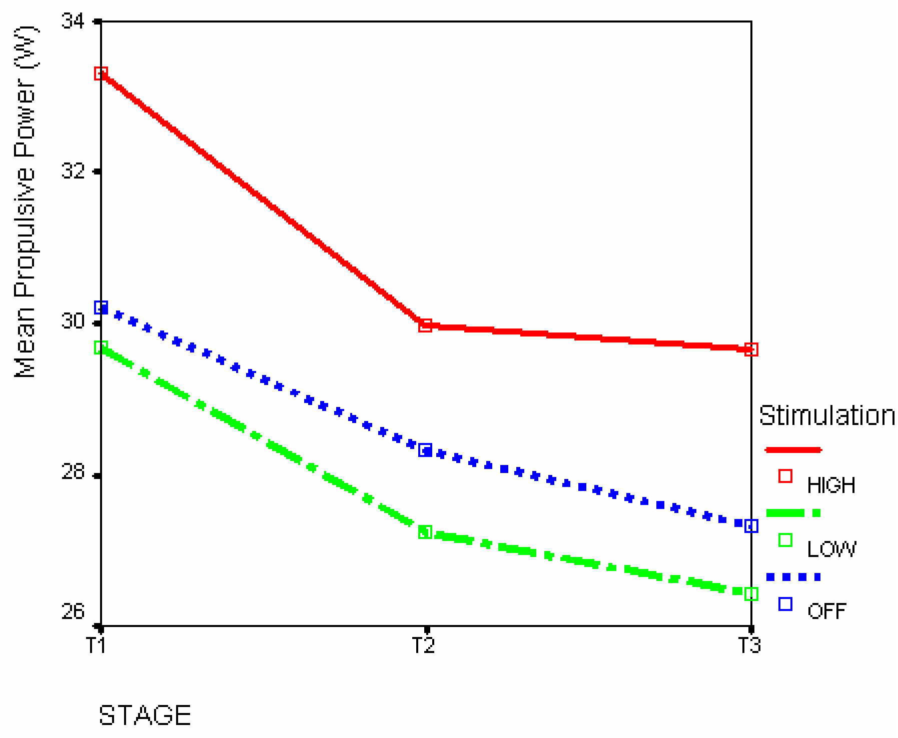
There were no differences in peak forces, peak torque, and mean MEF (Table 3). However, a main effect of stimulation levels on mean propulsion power output was found ($p=0.01$). Propulsion with HIGH stimulation resulted in significantly higher propulsion power than propulsion with LOW stimulation ($p=0.01$) and OFF ($p=0.08$). Power output decreased across all stimulation levels over time ($p < 0.01$) (Figure 11). When using HIGH stimulation, participants maintained a higher power output throughout the trial in comparison to LOW and OFF stimulation.

Table 3 Kinetic variables over time and main effect of stimulation levels

		T1 (0-0.30 min)		T2 (2-2.30 min)		T3 (4-4.30 min)		Stim. Main Effect (Pvalue)
		Mean	SD	Mean	SD	Mean	SD	
Peak resultant fore (N)	HIGH	74.74	17.26	77.07	19.52	74.50	19.48	0.17
	LOW	70.62	15.23	73.18	21.19	71.68	21.14	
	OFF	68.52	14.35	74.49	19.53	71.23	22.38	
Peak tangential force (N)	HIGH	58.61	16.38	57.82	18.48	55.94	18.21	0.23
	LOW	55.19	14.40	53.94	17.46	52.86	17.49	
	OFF	54.75	14.36	55.33	16.95	53.77	18.49	
Peak radial force (N)	HIGH	53.00	12.37	52.41	10.84	51.79	10.74	0.51
	LOW	49.23	9.42	52.30	14.60	51.06	14.25	
	OFF	47.29	9.17	53.26	14.05	50.28	14.99	
Peak torque (Nm)	HIGH	15.36	4.24	15.42	4.93	14.92	4.86	0.23
	LOW	14.83	3.86	14.39	4.66	14.10	4.67	
	OFF	15.10	4.34	14.76	4.52	14.34	4.93	
Mean power output (W)	HIGH	**33.32**	12.66	**29.98**	11.63	**29.66**	12.05	**0.01***
	LOW	**29.67**	11.48	**27.24**	11.68	**26.43**	12.02	
	OFF	**30.21**	11.00	**28.32**	11.23	**27.32**	11.89	
Mean MEF (%)	HIGH	0.55	0.21	0.48	0.11	0.51	0.15	0.52
	LOW	0.56	0.22	0.49	0.15	0.48	0.13	
	OFF	0.58	0.18	0.48	0.15	0.53	0.16	

Highlighted numbers indicate significant difference due to main effect of time interval ($p<0.05$)

Figure 11 The mean propulsion power output over time



3.4.2. General propulsion variables

The general propulsion variables are listed in Table 4. The participants propelled with similar average propulsion velocity, stroke frequency, start angle, end angle, and contact angle between stimulation levels. However, a significant decrease in stroke frequency ($p<0.01$) with longer contact angle ($p=0.06$) across all stimulation levels over time was found. This change may be considered as fatigue-related change.

Table 4 General propulsion characteristics over time and main effect of stimulation levels

		T1 (0-0.30 min)		T2 (2-2.30 min)		T3 (4-4.30 min)		Stim. Main Effect (Pvalue)
		Mean	SD	Mean	SD	Mean	SD	
Mean velocity (m./sec)	HIGH	1.20	0.34	1.12	0.33	1.16	0.35	0.31
	LOW	1.12	0.33	1.08	0.37	1.09	0.36	
	OFF	1.13	0.30	1.12	0.34	1.09	0.34	
Frequency (stroke/sec)	HIGH	**1.05**	0.20	**0.97**	0.19	**0.96**	0.18	0.25
	LOW	**1.00**	0.20	**0.99**	0.14	**0.95**	0.19	
	OFF	**1.01**	0.17	**0.94**	0.14	**0.92**	0.19	
Start angle (^{O})	HIGH	120.15	19.67	120.17	17.35	122.37	16.46	0.22
	LOW	119.11	17.43	117.53	21.19	119.05	17.96	
	OFF	121.31	18.68	122.92	20.09	121.58	17.22	
End angle (^{O})	HIGH	32.16	7.33	31.45	8.89	30.60	8.61	0.71
	LOW	31.89	9.57	32.06	11.76	30.28	7.96	
	OFF	32.90	7.67	29.35	7.63	29.60	7.19	
Contact angle (^{O})	HIGH	87.99	22.69	88.72	22.21	91.77	19.95	0.18
	LOW	87.22	22.45	85.47	26.74	88.78	20.12	
	OFF	88.42	22.46	93.56	21.79	91.98	16.85	

Highlighted numbers indicate significant difference due to main effect of time interval ($p<0.05$)

3.4.3. Kinematic parameters

Trunk motion in the sagittal plane during propulsion showed no differences between stimulation levels. Participants propelled their wheelchairs with an average peak trunk flexion angle of 5.6 – 8.5 degrees. Shoulder, elbow and wrist motion did not vary with stimulation levels (Table 5). Small paradoxical trunk movement was observed (distance range from 20.9 – 24.6 mm); however, no significant stimulation-related change in the paradoxical trunk movement was found.

Table 5 Kinematic variables over time and main effect of stimulation levels

		T1 (0-0.30 min)		T2 (2-2.30 min)		T3 (4-4.30 min)		Stim. Main Effect (Pvalue)
		Mean	SD	Mean	SD	Mean	SD	
Mean peak trunk flexion angle (^{O})	HIGH	6.74	7.43	7.36	6.16	8.19	7.22	0.93
	LOW	7.21	7.36	8.49	7.61	7.77	7.10	
	OFF	5.64	5.86	8.27	7.10	8.35	7.64	
Mean trunk paradoxical movement (mm)	HIGH	20.87	16.85	24.63	18.00	21.54	19.01	0.38
	LOW	24.36	22.83	23.54	17.45	23.02	16.75	
	OFF	23.69	21.67	23.22	20.36	24.49	22.69	
Peak shoulder flexion (^{O})	HIGH	16.51	8.34	17.00	9.39	19.08	8.36	0.29
	LOW	16.31	8.33	15.60	8.27	17.35	8.21	
	OFF	15.71	8.10	19.14	7.56	19.11	7.64	
Peak shoulder extension (^{O})	HIGH	37.42	6.81	38.17	7.65	39.01	7.38	0.86
	LOW	37.92	6.82	36.39	8.41	37.91	8.01	
	OFF	38.69	5.86	37.34	7.55	38.02	7.52	
ROM of shoulder (^{O})	HIGH	53.93	9.01	55.18	11.59	58.09	8.93	0.29
	LOW	54.23	9.32	51.99	12.77	55.26	9.82	
	OFF	54.40	9.35	56.48	9.13	57.12	9.04	
Peak elbow flexion (^{O})	HIGH	105.23	5.71	104.86	7.64	104.36	6.66	0.92
	LOW	104.83	6.48	105.60	8.26	105.06	7.37	
	OFF	104.48	6.00	104.84	8.17	104.98	6.85	
Peak elbow extension (^{O})	HIGH	146.10	6.58	146.17	6.88	148.42	6.31	0.14
	LOW	146.07	6.10	145.22	7.48	147.62	6.67	
	OFF	145.50	5.20	149.46	5.56	149.43	5.41	
ROM of elbow (^{O})	HIGH	40.87	9.90	41.31	13.31	44.06	12.05	0.21
	LOW	41.24	10.00	39.62	14.22	42.56	11.32	
	OFF	41.03	9.47	44.61	11.58	44.44	10.49	
Peak wrist flexion (^{O})	HIGH	10.79	9.52	16.01	22.68	9.74	8.57	0.19
	LOW	10.75	10.33	11.14	8.61	11.54	11.04	
	OFF	12.42	10.26	18.26	24.54	13.63	13.33	
Peak wrist extension (^{O})	HIGH	29.17	11.71	36.64	21.30	31.41	10.30	0.65
	LOW	29.87	13.58	30.97	12.71	31.91	11.91	
	OFF	28.97	11.41	35.82	23.09	29.94	10.84	
ROM of wrist (^{O})	HIGH	39.24	7.76	52.50	38.78	41.15	6.40	0.37
	LOW	40.14	7.41	41.43	9.62	42.82	7.43	
	OFF	40.76	7.26	53.81	42.21	42.93	7.85	

3.4.4. Physiological energy expenditure

Physiological responses to surface FES during propulsion are displayed in Table 6. Mean $\dot{V}O_2$ and $\dot{V}CO_2$ did not differ significantly between stimulation levels across the time intervals. The mean value of oxygen uptake ($\dot{V}O_2$, ml · kg^{-1} · min^{-1}), which was normalized by subject weight, also showed no significant increase between stimulation levels.

3.4.5. Gross mechanical efficiency (GME)

A trend of difference in GME between the stimulation levels over time was found ($p=0.06$) (Table 6). Propulsion with HIGH stimulation resulted in a marginally significant increase in GME compared to LOW stimulation ($p=0.08$) and OFF ($p=0.07$). A decrease in mechanical efficiency regardless of stimulation levels was observed between T2 and T3 ($p=0.05$) (Figure 12).

Table 6 Metabolic variables over time and main effect of stimulation levels

		T2 (2-2.30 min)		T3 (4-4.30 min)		Stim. Main Effect (Pvalue)
		Mean	SD	Mean	SD	
$\dot{V}O_2$ (ml/min)	HIGH	830.97	293.99	855.24	290.79	0.57
	LOW	835.57	312.18	862.24	273.71	
	OFF	822.67	265.85	843.55	296.91	
Normalized $\dot{V}O_2$ (ml/kg-min)	HIGH	11.23	2.90	11.56	2.69	0.71
	LOW	11.26	3.05	11.71	2.50	
	OFF	11.14	2.32	11.40	2.86	
$\dot{V}CO_2$ (ml/min)	HIGH	699.58	289.48	753.32	316.81	0.81
	LOW	722.02	324.89	758.82	291.97	
	OFF	724.79	266.82	751.52	307.45	
Energy expenditure (W)	HIGH	281.89	102.42	292.68	103.46	0.78
	LOW	284.88	109.85	295.03	97.02	
	OFF	281.55	92.19	289.33	104.20	
GME (%)	HIGH	**10.81**	6.22	**9.86**	4.45	**0.06***
	LOW	**9.85**	6.14	**8.69**	3.71	
	OFF	**9.86**	4.78	**9.13**	4.23	

Highlighted numbers indicate significant difference due to main effect of time interval ($p<0.05$)

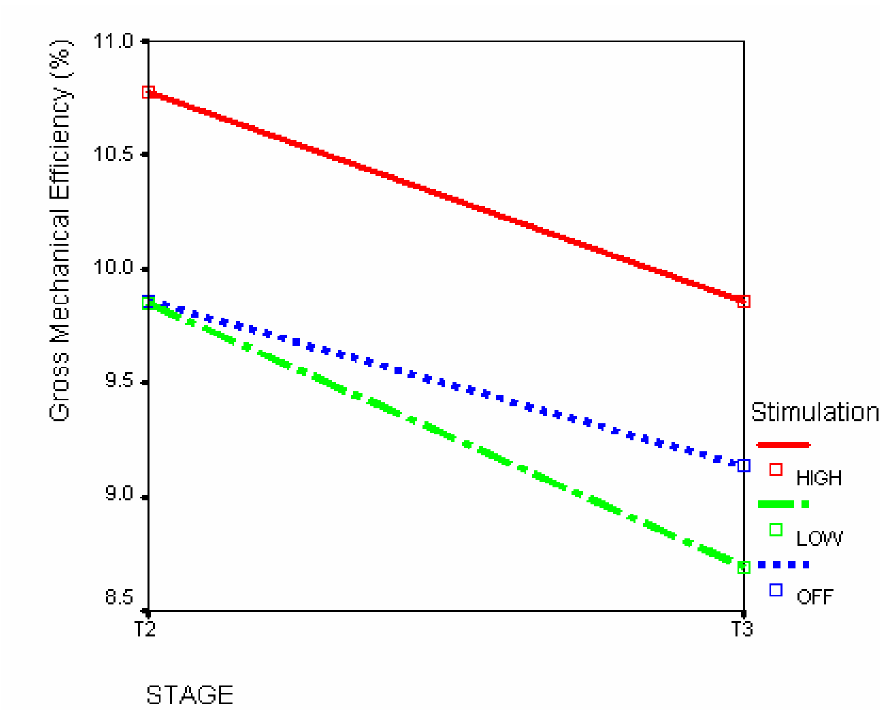


Figure 12 Gross mechanical efficiency over T2 and T3 intervals.

3.5. DISCUSSION

3.5.1. Effect of stimulation

This study is the first experimental study to investigate if surface electrical stimulation of the trunk musculature affects wheelchair propulsion biomechanics and physiological responses. The results showed that participants with HIGH stimulation had consistently higher power output and GME across the three time intervals. During wheelchair propulsion, in order to produce power on the pushrim, shoulder stabilization is needed to control arm movements and transfer power from the limbs to the pushrim (van der Helm and Veeger 1996). When more shoulder muscle effort is needed for stabilization, more energy is consumed without additional contribution to external propulsion power. Through using electrical stimulation on trunk musculature to provide a better base of support of the shoulder girdle complex, MWUs may be able to use shoulder muscles as primary movers rather than stabilizers to increase power production. The effects of trunk electrical stimulation on shoulder EMG activity during wheelchair propulsion is currently under investigation (Yang, Koontz, Triolo, Rice and Boninger 2005b).

3.5.2. Propulsion power output

Participants with HIGH stimulation showed higher propulsion power than propulsion with LOW and OFF. Based on the definition of power, this increase may have resulted from larger torque in combination with a faster speed as noted in Table 3, and Table 4. Although we attempted to control speed, subjects had a tendency to push closer to the target with HIGH stimulation. The propulsion torque and speed under HIGH condition was larger and faster than other two conditions (LOW and OFF), but did not reach statistical differences ($p=0.23$, and $p=0.31$ respectively). As a result of the combination of propulsion torque and speed, power production was increased under HIGH condition. Also, it is worth noting that propulsion speed for all stimulation conditions was lower than the target speed of 1.34 m/s. This speed may have been too difficult or uncomfortable for subjects to attain and maintain the velocity throughout the entire five minutes trial.

Furthermore, the increase of propulsion power could be associated with the Hawthorne effect. Although the stimulation level order was randomly assigned without the participant knowing which one they would receive, participants reported that they felt a vibration or pulse feeling around the stimulation area while receiving HIGH stimulation and had less or no feeling with LOW condition. This occurred even if their sensory function was not intact at that particular body region.

3.5.3. Upper body motion

The peak angles of the wrist, elbow and shoulder in the current study were similar to those presented in previous studies (Bednarczyk and Sanderson 1994; Rao, Bontrager, Gronley, Newsam and Perry 1996; Boninger et al. 1997a; Boninger et al. 1998; Veeger, Meershoek, van der Woude and Langenhoff 1998; Finley, Rodgers, Rasch, McQuade and Keyser 2002). We did not find significant stimulation-related changes in either upper limb or trunk motion during propulsion in the present study. We expected that stimulation would allow participants to adopt a more trunk flexed position thereby increasing range of shoulder, elbow and wrist motion during propulsion compared to stimulation OFF. However, this hypothesis was not supported by the data. One explanation of this unexpected finding might be that the level of surface electrical stimulation in present study was not strong enough to cause measurable changes in these variables during propulsion. In the Triolo et al. (2005) study (Appendix A), recipients of an implanted neuroprosthesis with continuous stimulation of their lumbar erector spinae while propelling the wheelchair at 0.9 m/sec showed larger trunk flexion angles ($16.2 \pm 9.9\,^{o}$) compared to the participants with HIGH stimulation in the present study ($7.4 \pm 6.8\,^{o}$). The differences in trunk angles between these two studies might be attributed to different electrical stimulation techniques used to augment trunk stability during propulsion. As opposed to the implanted electrodes which target the subcutaneous tissue structures directly, surface electrodes require stronger currents in order to penetrate the body's tissues to contract the muscles. Although 50% of maximal tolerated level was used as a HIGH stimulation during trials, such stimulation intensity may not be sufficient to stabilize the trunk and cause significant changes. Perhaps for this reason, no difference in propulsion kinetics was found between the LOW stimulation and OFF conditions with surface stimulation in this study.

Another reason could be that subjects had a short time to acclimate to FES. The implanted subjects in the Triolo study had been using FES on a daily basis and were able to tune the system for the optimal amount of trunk stiffness needed for a given activity. On the other hand, participants in the current study were long-term wheelchair users (on average 17 years) with no prior experience using FES on their trunk musculature and were likely resistant to modifying their propulsion technique in the short-term.

3.5.4. Physiologic energy expenditure

Participants in the present study did not show a significant increase in energy expenditure when receiving stimulation on their trunk musculature during wheelchair propulsion. The propulsion power was achieved by upper body musculature, primarily the shoulder. Therefore, the majority of oxygen cost and energy expenditure resulted from the voluntary upper limb movements rather than FES-induced trunk muscle contraction. As a result, participants did not increase their energy expenditure when using FES.

3.5.5. Gross mechanical efficiency

Participants using HIGH stimulation produced significantly more power output without an increase in energy expenditure compared to LOW stimulation and OFF. As a result, GME was higher for HIGH stimulation and remained higher than the other two conditions throughout the five-minute trial. The GME found in the present study (range from 8.7 to 10.8 %) was slightly higher than results in earlier literature (2-10 %) (van der Woude, Veeger, Rozendal, van Ingen Schenau, Rooth and van Nierop 1988; van der Woude, Hendrich, Veeger, van Ingen Schenau, Rozendal, de Groot and Hollander 1988; Veeger, van der Woude and Rozendal 1989; Veeger, van der Woude and Rozendal 1992). Perhaps this is because experienced wheelchair subjects were enrolled in the present study compared to the unimpaired subjects in earlier literature (van der Woude et al. 1989; Veeger et al. 1989; Veeger et al. 1992). Wheelchair type and setup could also have contributed to the higher level of efficiency. The testing wheelchair used in the present study was the subject's own wheelchair and eleven out of twelve subjects used ultralight wheelchairs. Subjects' wheelchairs may have been appropriately adjusted to optimize fit and propulsion technique. Studies showed that propulsion efficiency was highly influenced by wheelchair configuration, such as seat height and axle position, in addition to individual propelling technique (Masse, Lamontagne and O'Riain 1992; Hughes, Weimar, Sheth and Brubaker 1992; Boninger, Souza, Cooper, Fitzgerald, Koontz and Fay 2002; Boninger, Baldwin, Cooper, Koontz and Chan 2000; van der Woude, Veeger, Rozendal and Sargeant 1989).

The present study provides early evidence that trunk FES can increase propulsion efficiency during submaximal propulsion. FES trunk stimulation may have a greater impact on other more demanding propulsion tasks such as propelling up a ramp or curb ascents and traversing outdoor terrain. Individuals who currently have an FES-assisted walking system rely on a wheelchair as their primary mode of locomotion. Modifications of their existing FES system to include programming for trunk stability may help these users improve their propulsion efficiency.

3.5.6. Alternatives to FES

FES is one way to increase propulsion efficiency and power output. Customizing the wheelchair by either using a rigid backrest or reclining seat frame angles may result in similar outcomes. Parent et al. (2000) reported that a rigid back support improved trunk stability and comfort for the user compared to a regular sling backrest (Parent, Dansereau, Lacoste and Aissaoui 2000). As a result, Parent et al. hypothesized that propulsion efficiency may be improved. Samuelsson et al. (2004) investigated the effect of two reclined positions of the seat frame with consistent back angles on wheelchair propulsion (Samuelsson, Tropp, Nylander and Gerdle 2004). They found that the change of inclination of the seat frame (12^0) significantly broadened the stroke angle and reduced push frequency during wheelchair propulsion on a treadmill. Using a combination of strategies (e.g. FES, rigid back rest, ultralight wheelchairs etc) is a comprehensive approach to maximizing propulsion efficiency and minimizing joint stress and fatigue.

3.5.7. Limitations

Muscle fatigue is a known side effect of FES. In reality, wheelchair users often propel for several minutes or longer (Hoover, Cooper, Ding, Koontz, Cooper, Fitzgerald and Boninger 2004). Participants using stimulation to stabilize their trunk could gain some benefits at the beginning, but the efficiency might decrease over time as shown in the present results (Figure 4). With advanced sensing and programming, a FES device could possibly be synchronized with the propulsion cycle to avoid continuous stimulation on trunk musculature and only provide stimulation during pre-push and early push phase of the propulsion cycle when trunk stability is most important (Yang et al. 2005a). Alternatively, FES could be used for challenging propulsion tasks of short duration, such as pushing up a ramp or across thick carpet.

Another potential limitation is the manufacturing limits of the surface FES device in this study. The surface FES device used in this study is a FDA approved commercial device, in which the maximal stimulation intensity was limited due to safety considerations. Four of twelve participants were able to comfortably tolerate the maximum stimulation intensity from the FES device. It is likely that these individuals could tolerate more stimulation intensity than the device could deliver. In order to remain consistent with experimental protocol, we still used a 25%, and 50% differential between the threshold and maximal stimulation intensity which the device could provide as LOW and HIGH stimulation levels. The setting of LOW stimulation intensity on these four subjects may not have been sufficient to cause an effective trunk muscle contraction due to a ceiling effect of the device. A secondary nonparametric analysis using Friedman repeated measures ANOVA was done to examine the influence of this ceiling effect. The result indicated that these four participants with LOW stimulation showed no statistical difference on propulsion power ($p>0.1$) compared to the OFF condition. The rest of eight participants who were not affected by the ceiling effect showed a statistically larger propulsion power production ($p=0.04$) for LOW stimulation compared to OFF condition. The LOW stimulation intensity for the four participants was not high enough to cause differences, thereby reducing the group mean of propulsion power for the LOW condition.

The lower power output found for LOW condition compared to without stimulation may indicate that the insufficient low stimulation intensity on trunk musculature hindered rather than helped participants. Low intensity stimulation could have resulted in only minimal muscle contractile activity that interfered with the subject's propulsion performance. A FES device with a higher stimulation output (e.g. a FES with implantable electrodes or a custom FES system that can be made to deliver higher voltages for this particular application) might be needed to lead to significant effects on biomechanical variables during propulsion.

The effects of FES might be likely influenced by injury levels. Individuals with high paraplegia might gain more benefits of FES on trunk musculature than those with low paraplegia. In a secondary analysis, subjects were groups as two groups: high paraplegia (injury levels between C6 and T4, n=5) and low paraplegia (injury levels between T5 and T10m, n=7) to compare the effects of FES on each group using Friedman repeated measures ANOVA respectively. The results showed that propulsion power and other kinetic variables both increased under HIGH stimulation compared to LOW stimulation and stimulation OFF condition can be observed on both group, but did not reach statistical differences. Individuals with high paraplegia most likely produced larger propulsion power for HIGH condition than other two stimulation levels ($p=0.07$). On the other hand, individuals with low paraplegia showed less increased tendency of propulsion power production for HIGH condition ($p=0.36$).

This is a heterogeneous subject group whose level of injuries ranged between C6 and T10 level. One subject whose injury level is C6 complete is an outlier from the rest of the subjects (Figure13). However, kinetic data from this subject did not significantly affect on the normality of data distribution ($p>0.12$). Therefore, this subject was still included into the statistical analysis model. Besides, results of the main effect of stimulation levels and time intervals were based on within-subject comparisons. Therefore, much of the variability due to subject's characteristics was control for within-subject comparisons.

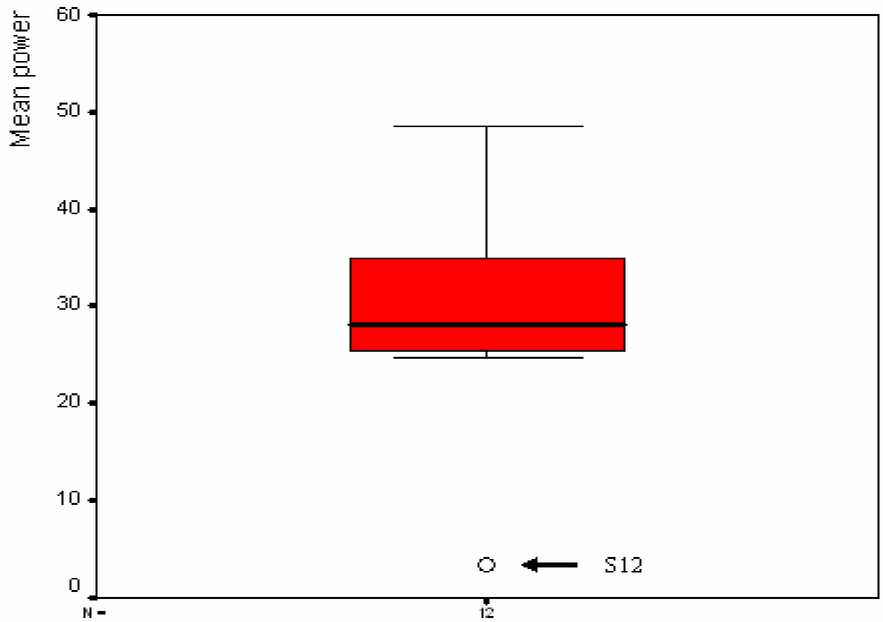


Figure 13 Boxplot of mean power output among 12 subjects

In the present study, we did not quantify trunk stability was gained with FES device. Instead, we did manual palpation to verify abdominal/back muscle contraction. Future studies should evaluate the extent of trunk control due to electrical stimulation. Furthermore, most subjects in present study had no previous experience using surface FES on their trunk. Subjects were given several minutes to familiarize themselves with the different stimulation levels during propulsion before the experimental trials. However, this short period of practice time with stimulation may not be long enough for subjects to get used to propelling their wheelchair with FES. A future study should investigate prolonged use of surface FES, which may help participants without trunk control realize and gain the benefits of FES on trunk stability.

After spinal cord injuries, muscles normally supplied by intact motoneurons from spinal cord segments at or below the injury site are paralyzed and undergo atrophy. Since the participants were long-term wheelchair users (on average 17 years of post injuries), their trunk musculature below the injury site due to disuse atrophy could have less response of muscle contraction induced by FES. In order to avoid the effect of disuse atrophy, a training program with low-frequency electrical stimulation via implanted or skin surface electrode of the trunk muscles can be used for preparing these muscles prior to the application of FES during wheelchair propulsion. An electrical stimulation training program has been widely used to train muscles and counteract disuse atrophy (Kralj and Bajd 1989; Rodgers, Glaser, Figoni, Hooker, Ezenwa, Collins, Mathews, Suryaprasad and Gupta 1991; Gordon and Mao 1994). This training not only conditioned the muscles but allowed individuals to become more experienced and comfortable with FES while performing t functional movements.

3.6. CONCLUSION

The present study demonstrates that MWUs who use trunk FES can generate more propulsion power and increase GME during wheelchair propulsion. Because of its biomechanical advantage and higher GME performance, trunk FES has the potential advantage of helping individuals with SCI negotiate demanding propulsion tasks such as ramp and curb ascents, and traversing outdoor terrain. Further research is needed for an advanced FES stimulation device with synchronization with the propulsion cycle, thereby improving propulsion power and efficiency while minimizing muscle fatigue.

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4. ELECTROMYOGRAPHIC ANALYSIS OF SHOULDER MUSCLE DURING WHEELCHAIR PROPULSION WITH FUNCTIONAL ELECTRICAL STIMULATION ON TRUNK MUSCULATURE: A PILOT STUDY

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ACKNOWLEDGEMENTS

This research was supported by the U.S. Department of Veterans Affairs, Rehabilitation Research and Development (B3043-C), and American Society of Biomechanics Graduate Student Grant-In-Aid (2004).

4.1. ABSTRACT

Objective: To investigate the influence of surface electrical stimulation of trunk musculature on shoulder muscle recruitment patterns during wheelchair propulsion.

Methods: Eleven wheelchair users with spinal cord injury (SCI) were asked to push their own wheelchairs on a dynamometer at a target speed of 1.35 m/sec for three five-minute trials. During a propulsion trial, one of three stimulation levels (HIGH, LOW and OFF) was randomly applied to the participant's abdominal and back muscle groups with a surface functional electrical stimulation (FES) device. The surface electromyographic (sEMG) activity of six shoulder muscles and corresponding propulsion kinetics were recorded during three time intervals (30 seconds each) within a five-minute trial. The differences amongst the sEMG and kinetic variables due to stimulation level were examined using a two-way repeated measures ANOVA.

Results: Pectoralis major, anterior deltoid, biceps brachii and triceps brachii served as prime movers during push phase. Middle and posterior deltoid acted as shoulder stabilizers. No differences were found in shoulder EMG activation patterns between stimulation levels; however, pectoralis major and biceps showed fatigue-related increases in muscle activity and duration ($p<0.05$). Participants with HIGH stimulation generated higher propulsion power outputs ($p = 0.017$) than the other two stimulation conditions regardless of no significant changes in shoulder muscle activation.

Conclusions: Trunk FES may help individuals to generate propulsion power without placing additional demands on shoulder musculature. With trunk FES, the functional role of the shoulder may shift from stabilizers to a prime movers contributing more directly to propulsion.

Keywords: muscle recruitment; trunk stability; muscle mover

4.2. INTRODUCTION

Manual wheelchair propulsion requires large static work from proximal shoulder muscle synergy and cocontraction to stabilize and adjust the shoulder girdle complex with respect to the trunk, for gripping and applying force to the hand rim during the push phase of propulsion (van der Helm and Veeger 1996; van der Woude, Dallmeijer, Janssen and Veeger 2001a; van der Woude, Veeger, Dallmeijer, Janssen and Rozendaal 2001b; Vanlandewijck, Theisen and Daly 2001). It has been suggested that lack of trunk stability, leading to a less erect posture and poor support of the shoulder girdle complex, may limit production of maximal upper limb strength to push a wheelchair (Powers, Newsam, Gronley, Fontaine and Perry 1994). Furthermore, rapid movement of the upper limb during wheelchair propulsion produces a complex interplay of dynamic reactive forces acting on the shoulder and trunk. During propulsion, the dynamic reactive forces exerted on the trunk from the upper limbs can cause the trunk to move backwards during the push phase of propulsion, a phenomenon referred to as paradoxical trunk movement (Rice, Koontz, Boninger and Cooper 2004). The occurrence of paradoxical trunk movement during propulsion is an important phenomena because it has been shown to reduce mechanical effective propulsion forces (Koontz, Boninger, Rice, Yang and Cooper 2004).

Trunk stability to counteract the effect of the dynamic reactive forces during propulsion is initiated prior to hand contact with the pushrim. This trunk stability response can be referred to as an anticipatory postural response (Aruin and Latash 1995). Studies evaluating the anticipatory response of the trunk muscles associated with movement of the upper limb indicated contraction of either the erector spinae (ES) prior to upper limb flexion (Zattara and Bouisset 1988; Friedli, Hallett and Simon 1984; Aruin and Latash 1995) or contraction of the rectus abdominis (RA) preceding upper limb extension (Friedli et al. 1984; Aruin and Latash 1995). Yang et al.

investigated back and abdominal muscle activation patterns among 14 unimpaired individuals during wheelchair propulsion under different speed conditions (Yang, Koontz, Triolo, Mercer and Boninger 2005b). They found that both back and abdominal muscles were most active in the pre-push and early push phase. The fact that the activity of these muscles increased just prior to hand contact and continued through initial contact provides insight into preparatory trunk response during propulsion. However, most manual wheelchair users who lose voluntary control of trunk musculature (e.g., individuals with high paraplegia due to SCI) are not able to recruit trunk stabilizing muscles. Consequently, the dynamic reactive forces exerted on the trunk result in inefficient paradoxical trunk movement during propulsion.

Individuals with paralysis of the lower extremities due to SCI rely on their upper limbs to push their wheelchairs for mobility. The power during wheelchair propulsion originates from the musculature of the upper limb and shoulder. Previous studies have investigated the intensity and duration of shoulder muscle electromyographic activity during wheelchair propulsion. (Harburn and Spaulding 1986; Schantz, Bjorkman, Sandberg and Andersson 1999; Mulroy, Farrokhi, Newsam and Perry 2004). Harburn and Spaulding reported higher intensity shoulder muscle activity in subjects with tetraplegia compared to subjects with paraplegia and an unimpaired subject group. One possible explanation could be that individuals with higher level SCI must use the shoulders more to compensate for poor trunk stability during propulsion (Harburn and Spaulding 1986). Schantz et al. found significant differences in the pattern of muscle activation between individuals with paraplegia and tetraplegia (Schantz et al. 1999). Greater volitional control of the trunk and arm muscles allowed individuals with paraplegia to have a longer push phase and muscle activation duration during propulsion. However, differences in the intensity of shoulder muscle activity were not reported in this study. More recently, Mulroy et al. investigated the muscle activation pattern between four different SCI level groups (Mulroy et al. 2004). They reported that subjects with paraplegia with and without trunk control showed similar patterns of shoulder muscle activation in response to the demands of wheelchair propulsion. They suggested that the wheelchair backrest used adequately stabilized the trunk in the absence of trunk control, thereby resulting in a similar shoulder muscle activation pattern.

Previous studies have investigated a variety of devices for improving trunk stability during propulsion and sitting posture, such as a rigid backrest, inclination of seat frame angles, or artificially stimulating paralyzed trunk muscles (Parent, Dansereau, Lacoste and Aissaoui 2000; Samuelsson, Tropp, Nylander and Gerdle 2004; Triolo, Yang, Koontz, Nogan and Boninger 2005). However, few studies have examined the effects of these devices on shoulder muscle intensity and duration during propulsion. Masse et al. found that a lower wheelchair seat position resulted in less shoulder muscle activity along with lower stroke frequency during wheelchair propulsion (Masse, Lamontagne and O'Riain 1992). Trunk movement which may reflect stability of trunk did not show a clear change between different seat positions.

The purpose of this study was to investigate shoulder muscle activation and duration in response to varying intensity levels of trunk stimulation during wheelchair propulsion. It was hypothesized that surface electrical stimulation applied to abdominal and back muscles in individuals with SCI would provide a better base of support for the shoulder girdle complex thereby reducing the intensity of shoulder muscle activity and duration to achieve the same propulsion demand. The results of this study might provide insight into the potential application of surface electrical stimulation of trunk muscles as a preventative mechanism for minimizing the risk of shoulder pain and injury in long-term wheelchair users.

4.3. METHODS

4.3.1. Subjects:

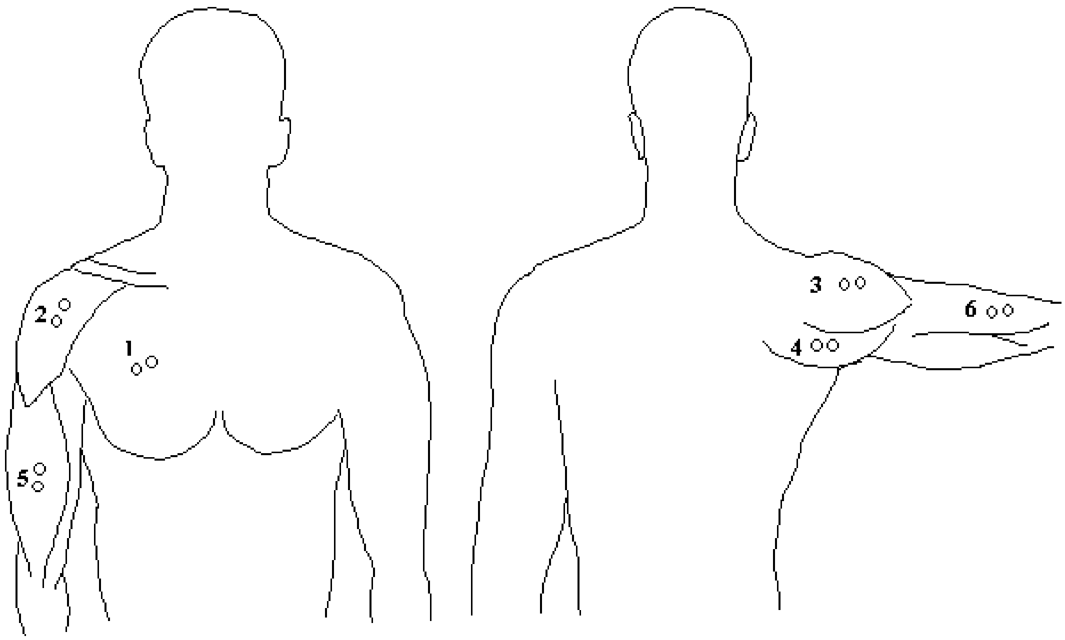
Eleven manual wheelchair users (Table 7) provided informed consent in accordance with the procedures approved by the Institutional Review Board of Veterans Affair Medical Center prior to participation in the study. Inclusion criteria were: 1) complete or incomplete SCI between C6 and T12; 2) use a manual wheelchair as a primary mode of mobility, and be 3) between the ages of 18 and 65 years. Exclusion criteria were: 1) previous history of upper extremity pain, 2) presence of a heart or lung condition that is worsened by pushing a wheelchair, and 3) pregnancy. Volitional trunk control was assessed by having the participants lean their trunk forward, backward and laterally unsupported and noting any loss of balance. Trunk control was noted as either present or absent (Table 7). Two subjects (S4, and S5) reported surgical fusions of the thoracic spine without implanted rods, and other subject (S6) had cervical spine fusions with implanted rods. The rest of the participants did not have any fusion of the thoracic and lumbar spine or implanted rods to stabilize their spine.

Table 7 Subject characteristics

Subject	Gender	Handedness	Level of lesion	Age	Years post injury	Trunk control
S1	M	Right	T7 (ASIA-B)	48	5	absent
S2	M	Right	C7 (ASIA-A)	29	5	absent
S3	M	Right	T4 (ASIA-A)	43	20	absent
S4	F	Right	T10 (ASIA-A)	37	23	absent
S5	M	Right	T10 (ASIA-B)	45	11	absent
S6	M	Right	C6 incomplete (ASIA-C)	31	24	present
S7	M	Right	T9 incomplete (ASIA-B)	28	19	present
S8	M	Right	T6 (ASIA-A)	53	23	absent
S9	M	Right	T8 (ASIA-A)	56	4	absent
S10	F	Right	T2 (ASIA-A)	47	9	absent
S11	M	Right	T5 (ASIA-A)	43	29	absent
Mean				41.9	17.4	
SD				9.6	8.9	

4.3.2. Electromyography

The surface electromyographic (sEMG) activity of six shoulder muscles were measured using a TELEMYO 2400T (Noraxon U.S.A. Inc., Scottsdale, AZ) with a bandwidth of 150 to 500 Hz. The data were then sampled a rate of 1500 Hz and digitized using MyoResearch XP Master software (Noraxon U.S.A. Inc., Scottsdale, AZ). Six pairs of Ag-AgCl surface electrodes (Blue Sensor M-00-S, MedicoTest Inc., Denmark) were used in a bipolar configuration with a 2-cm interelectrode distance on the participant’s dominant shoulder. Electrode location was placed on anterior, middle, and posterior portions of the deltoid, the sternal portion of pectoralis major, biceps brachii, and triceps brachii based on the guidelines published by Basmajian and Blumenstein (Basmajian and Blumenstein 1980) and verified with isolated manual muscle tests (Kendall, McCreary and Provance 1993) (Figure 14).



(1) Pectoralis Major (PM), (2) Anterior Deltoid (AD), (3) Middle Deltoid (MD),
(4) Posterior Deltoid (PD), (5) Biceps Brachii (BB), and (6) Triceps Brachii (TB)

Figure 14 The electrode placement

The electrodes for the sternal portion of pectoralis major (PM) were placed two fingers' breaths above the nipple line, and the lateral electrode slightly lower than the medial one. For the anterior deltoid (AD), the electrodes were placed vertically within an elongated oval deltoid below the lateral end of the clavicle. The middle deltoid (MD) electrodes were placed along the midline of the lateral surface of the arm, and located below the lateral margin of the acromion approximately a quarter of the distance from the acromion to the elbow. For the posterior deltoid (PD), the electrodes were placed in the area about two fingerbreadths behind the angle of the acromion. The electrodes of the biceps brachii (BB) were placed over the belly of the greatest bulge of the muscle. The triceps brachii (TB) electrodes were placed within a small oval area located at a finger's breadth lateral to the midline and 50% of the distance between the acromion process and the olecranon process. The ground electrode was attached to the sternal notch. Prior to electrode attachment, the skin surface was debrided and cleaned with alcohol to enhance the EMG signal.

For each muscle, maximal voluntary contraction (MVC) EMG signals were recorded and used for normalizing raw EMG signal during the experimental trials. When subjects were lying on the mat table, the following standard manual muscle testing positions were performed to assess the maximum effort of each muscle:

- PM: a combination of shoulder flexion and adduction against resistance at 90^{o} shoulder flexion and 90^{o} abduction.
- AD: shoulder forward flexion against resistance at 45^{o} shoulder flexion.
- MD: shoulder abduction against resistance at 45^{o} shoulder abduction.
- PD: shoulder extension against resistance at 45^{o} shoulder flexion.
- BB: elbow flexion against resistance at 135^{o} elbow flexion.
- TB: elbow extension against resistance at 135^{o} elbow flexion.

4.3.3. Surface FES device

In order to stimulate the abdominal and back muscle bilaterally at the same time, two commercially available double channel stimulators (EMS-5000 Electronic Muscle Stimulator, OrthoBionics Inc., Dallas, TX) were linked together using a custom circuit. Four pairs of self-adhering surface electrodes (Superior Silver Electrodes, size 2" × 2", Uni-Patch, Wabasha, MN) were placed in the following positions: two pairs (1 right, 1 left) over the rectus abdominal muscles, two pairs (1 right, 1 left) over the multifidus muscles (Figure 15). The parameters of stimulation were set for asymmetrical biphasic waves of 30 Hz frequency, 300 μs pulse width, and up to the maximal amplitude of 80 mA depending on the participant's tolerance levels. The stimulation activation ratio was set to 30 seconds of continuous burst stimulation with one-second stimulation off during the entire five minute propulsion trial.

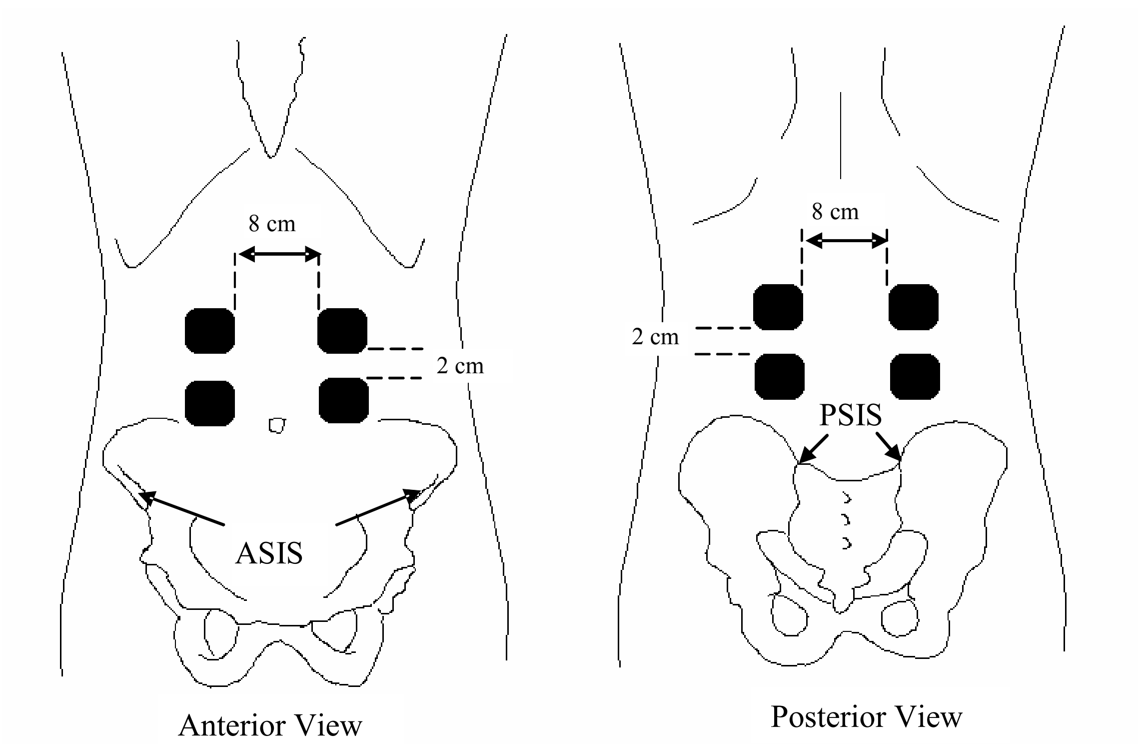


Figure 15 Surface stimulation electrode placements for abdominal and back muscles

Participants were given a supervised functional test prior to the experimental trials to determine the threshold response to surface electrical stimulation and maximal tolerable level of stimulation. The threshold response was defined as the minimal stimulation amplitude causing muscle contraction which was detectable by manual muscle palpation. The maximal tolerable level of stimulation was defined as the maximal tolerable amplitude reported by the participant or the maximal stimulation amplitude provided by the stimulator, which was 80 mA. After determination of the threshold and maximal tolerable levels, 50% and 25% of the difference between these two levels was used as HIGH and LOW stimulation intensity, respectively.

4.3.4. Kinetic measurement system

A SMART$^{Wheels\ TM}$ (Three Rivers Holdings, Inc., Mesa, AZ), force and torque sensing wheel, was fitted to the participant's own wheelchair on the dominant side to measure three dimensional forces and moments on the pushrim in a global reference system. Propulsion kinetic data during the experimental trials were collected with a sample frequency of 240 Hz and filtered with an 8th order Butterworth low-pass filter, zero lag and 30 Hz cut-off frequency (Cooper, Robertson, VanSickle, Boninger and Shimada 1997). The propulsion cycle was comprised of push and recovery phases. The start and end of the push phase was determined by visual inspection of the presence/absence of propulsion forces and torques detected by SMART$^{Wheels\ TM}$.

4.3.5. Experimental Protocol

Participants performed three 5-minute repeated wheelchair propulsion trials on an independent dynamometer system simulated to that of propulsion on a smooth level surface. Their personal wheelchairs were secured to the dynamometer system with a four-point tie down system. Participants were asked to push the wheelchair at target speed of 1.34 m/s and maintain this speed for five minutes. Propulsion speed was displayed on a 0.43-meter computer screen placed in front of them to provide visual speed feedback. For each propulsion trial, participants either propelled without stimulation (OFF) or with stimulation (HIGH, LOW) on both their abdominal and back muscles to examine the effect of FES during propulsion. The stimulation level order was randomly assigned. Before proceeding to the test trial, participants were asked to propel their wheelchairs on the dynamometers for several minutes with HIGH, LOW stimulation and stimulation OFF to familiarize with the experimental setup and the stimulation levels. At least five minutes of rest preceded each trial to avoid muscle fatigue.

During each 5-minute propulsion trial, shoulder muscle EMG activity, and propulsion kinetic data were collected synchronously for three time blocks of 30 seconds without the participant knowing when data were being collected. The first time block (T1) was the first 30 seconds of the propulsion trial, the second time block (T2) started at the middle of propulsion trial (2.00 min), and the last time block (T3) was initiated during the last minute (4.00 min) of the propulsion trial.

4.3.6. Data analysis

EMG data sampled at 1500 Hz were full wave rectified and smoothed with a 4^{th} order Butterworth low-pass filter (6 Hz cut-off) to obtain a linear envelope (Winter 1990). Afterward, EMG data were linearly interpolated for synchronization with the kinetic data with collection rate of 240 Hz. In order to compare muscle activity across subjects during propulsion trials, EMG signals during propulsion were normalized as %MVC for each muscle. The PC time was normalized to 100% for each subject. Significant EMG activity was defined as activity with an intensity of at least 5% MVC and for longer than 5% of the entire propulsion cycle (PC) (Mulroy, Gronley, Newsam and Perry 1996; Mulroy et al. 2004). The duration of EMG activity spent in the push or recovery phase was expressed as a percentage of the entire PC. Integrated

EMG (iEMG), which represents the area under the EMG waveform was determined for both push and recovery phases.

Kinetic data from the SMART^{WheelsTM} on the subject's dominant side were further transformed to a force radial to the pushrim (Fr) and a force tangential to the pushrim (Ft) (Boninger, Cooper, Robertson and Shimada 1997; Cooper et al. 1997). Furthermore, the propulsion power output was calculated from the measured propulsion torque applied on the pushrim (Mz), velocity (V_{rim}) and pushrim radius (R_{rim}) according to:

$$P_o(Watts) = \frac{Mz * V_{rim}}{R_{rim}}$$

For each time block (T1, T2 and T3), the peak kinetic and EMG variables were determined for ten consecutive strokes and then averaged.

4.3.7. General propulsion characteristics

Propulsion velocity was calculated based on the SMART$^{Wheels\ TM}$ encoder that measured angular displacement during propulsion. Stroke frequency (*f*) was defined as the number of strokes that occurred per second. Push time (PT) was defined as the time spent on the pushrim during the push phase. Recovery time (RT) was defined as the time spent during the recovery phase, and stroke time (ST) was defined as the time spent completing an entire propulsion cycle.

4.3.8. Statistical Analysis

Some shoulder muscle EMG data during the push and recovery phases showed a non-normal distribution and periods of inactivity which was consistent with prior studies by Mulroy et al. (Mulroy et al. 1996; Mulroy et al. 2004). Therefore, the median intensity for each muscle during the push and recovery phase was determined for each subject. Afterward, each EMG variable was screened for normality of distribution amongst subject group with the Wilk-Shapiro W statistic to validate the assumption of normality (α=0.05) before it was added to the mixed model. There was no evidence that the normality was violated ($p>0.05$). In order to examine the differences in surface FES stimulation levels on each shoulder muscle and propulsion kinetic data across the three time intervals, a two-way (stimulation levels × time intervals) repeated-measures analysis of variance (ANOVA) using the mixed models procedure with a Bonferroni post-hoc test based on a least-squares means (LSM) analysis was used. The level of statistical significance was set at $p < 0.05$. Mixed modeling (PROC MIXED) was used because the same subjects propelled their wheelchair with all stimulation levels. Mixed modeling allows for testing both random and fixed effects (Littell, Milliken, Stroup and Wolfinger 1996). In the mixed model, subjects were entered as the random factor and the fixed factors were stimulation levels (OFF, HIGH, and LOW) and time periods (T1, T2, and T3).

Another advantage of using mixed modeling (PROC MIXED) is to retain all subject data for cases where missing data for a trial are present. This is different than a traditional repeated measures ANOVA test (PROC GLM), which omits all of the subject's data if he/she does not have complete data. Due to technical difficulties, one subject's shoulder EMG data during the T3 interval with stimulation OFF could not be processed, and another subject's PM EMG data for all three conditions was lost due to the failure of a surface electrode during experimental trials. Therefore, the total number of trials analyzed was n=27 (3 time intervals × 9 subjects) for PM and n=30 (3 time intervals × 10 subjects) for the other muscles for stimulation OFF condition. With the exception of PM, stimulation HIGH and LOW resulted in n=33 (3 time intervals × 11 subjects) trials of shoulder EMG data. The mixed-model test is valid only if the data is missing as a result of random occurrence. Since there was no systematic reason for missing data for the three conditions in this study, this assumption was met. All statistical analyses were performed using the SAS System for Windows 9.0 software package. The level of statistical significance was set at $\alpha=0.05$.

4.4. RESULTS:

4.4.1. General propulsion characteristics

No significant stimulation-related changes in propulsion speed, frequency, PT, RT, and ST were found in present results (Table 8). Regardless of the stimulation levels, there was a trend for increasing PT and ST, and reduced stroke frequency over time ($p<0.05$). These change may be fatigue-related.

Table 8 Propulsion variables over time and main effect of stimulation levels

		T1 (0-0.30 min)		T2 (2-2.30 min)		T3 (4-4.30 min)		Stim. Main Effect (Pvalue)
		Mean	SD	Mean	SD	Mean	SD	
Mean velocity (m./sec)	HIGH	1.28	0.25	1.21	0.23	1.25	0.24	0.09
	LOW	1.16	0.24	1.13	0.26	1.16	0.29	
	OFF	1.19	0.17	1.19	0.23	1.17	0.25	
Frequency (stroke/sec)	HIGH	**1.09**	0.17	**1.01**	0.15	**1.00**	0.13	0.10
	LOW	**1.04**	0.17	**1.00**	0.14	**0.99**	0.16	
	OFF	**1.04**	0.13	**0.97**	0.10	**0.96**	0.14	
Push time (sec)	HIGH	**0.43**	0.09	**0.49**	0.08	**0.48**	0.07	0.35
	LOW	**0.47**	0.11	**0.49**	0.10	**0.50**	0.12	
	OFF	**0.46**	0.10	**0.48**	0.07	**0.49**	0.09	
Recovery time (sec)	HIGH	0.51	0.10	0.52	0.15	0.54	0.13	0.08
	LOW	0.52	0.08	0.53	0.09	0.52	0.10	
	OFF	0.52	0.08	0.55	0.09	0.57	0.12	
Stroke time (sec)	HIGH	**0.94**	0.15	**1.01**	0.17	**1.02**	0.15	0.21
	LOW	**0.98**	0.15	**1.02**	0.15	**1.03**	0.16	
	OFF	**0.98**	0.13	**1.03**	0.10	**1.06**	0.15	
Peak resultant fore (N)	HIGH	78.68	15.18	76.19	20.72	79.14	15.90	0.26
	LOW	71.74	12.23	77.00	18.86	75.16	18.45	
	OFF	72.46	12.24	78.65	15.16	73.53	19.50	
Peak tangential force (N)	HIGH	64.05	14.96	59.68	20.89	62.20	18.80	0.29
	LOW	60.14	12.24	60.20	16.33	58.73	19.06	
	OFF	59.35	13.54	61.17	15.02	59.06	18.39	
Peak radial force (N)	HIGH	54.40	13.09	50.00	14.58	51.57	10.30	0.43
	LOW	47.44	11.69	51.72	14.09	49.64	13.60	
	OFF	49.10	11.33	54.39	11.46	48.88	13.21	
Peak torque (Nm)	HIGH	17.08	3.99	15.92	5.57	16.59	5.01	0.29
	LOW	16.04	3.26	16.05	4.36	15.66	5.08	
	OFF	15.83	3.61	16.31	4.01	15.75	4.91	
Mean power output (W)	HIGH	36.93	7.74	32.71	8.17	33.52	9.44	**0.02***
	LOW	32.19	7.59	30.56	9.43	29.85	11.10	
	OFF	33.04	6.92	31.57	7.95	30.89	10.28	

Highlighted numbers indicate significant difference due to main effect of time interval ($p<0.05$)

4.4.2. Force application.

There were no differences in peak propulsion forces and torque production across the three time intervals (Table 8). However, a main effect of stimulation level on mean propulsion power output was found ($p = 0.02$). HIGH stimulation resulted in higher propulsion power than LOW stimulation ($p<0.01$) or OFF ($p= 0.03$). There was also a trend ($p = 0.06$) for decreasing averaged power output across all stimulation levels over time.

4.4.3. Shoulder muscle activity

Participants showed a similar pattern of shoulder muscle activity between stimulation levels during the propulsion cycle (Figure 16, 17 and 18). During propulsion, the PM initiated its activity in late recovery phase and ceased activity in the middle of the push phase. The AD showed continuous activity throughout the entire propulsion cycle. The MD and PD were also active throughout the propulsion cycle. These two muscles gradually increased their intensity in the early phase of push, reached their peak value in early recovery phase, and then showed decreased intensity in late recovery. The BB showed two peaks of activity during the propulsion cycle that started in late recovery phase and remained active until the early push. The TB showed an increased activity in the middle of the push phase and then decreased in intensity during the recovery phase.

Figure 16 Group average shoulder muscle activation patterns with HIGH stimulation

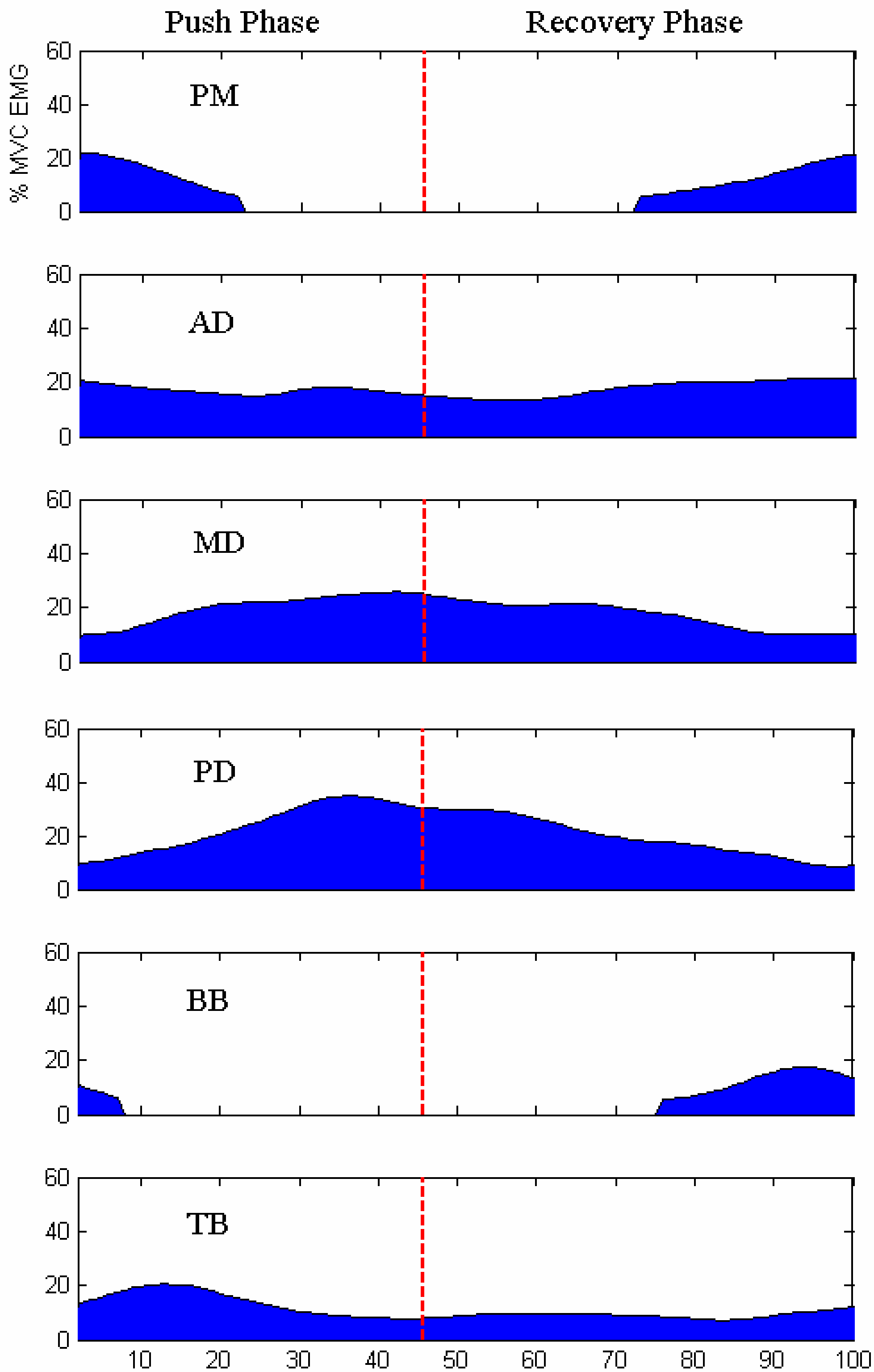


Figure 17 Group average shoulder muscle activation patterns with LOW stimulation

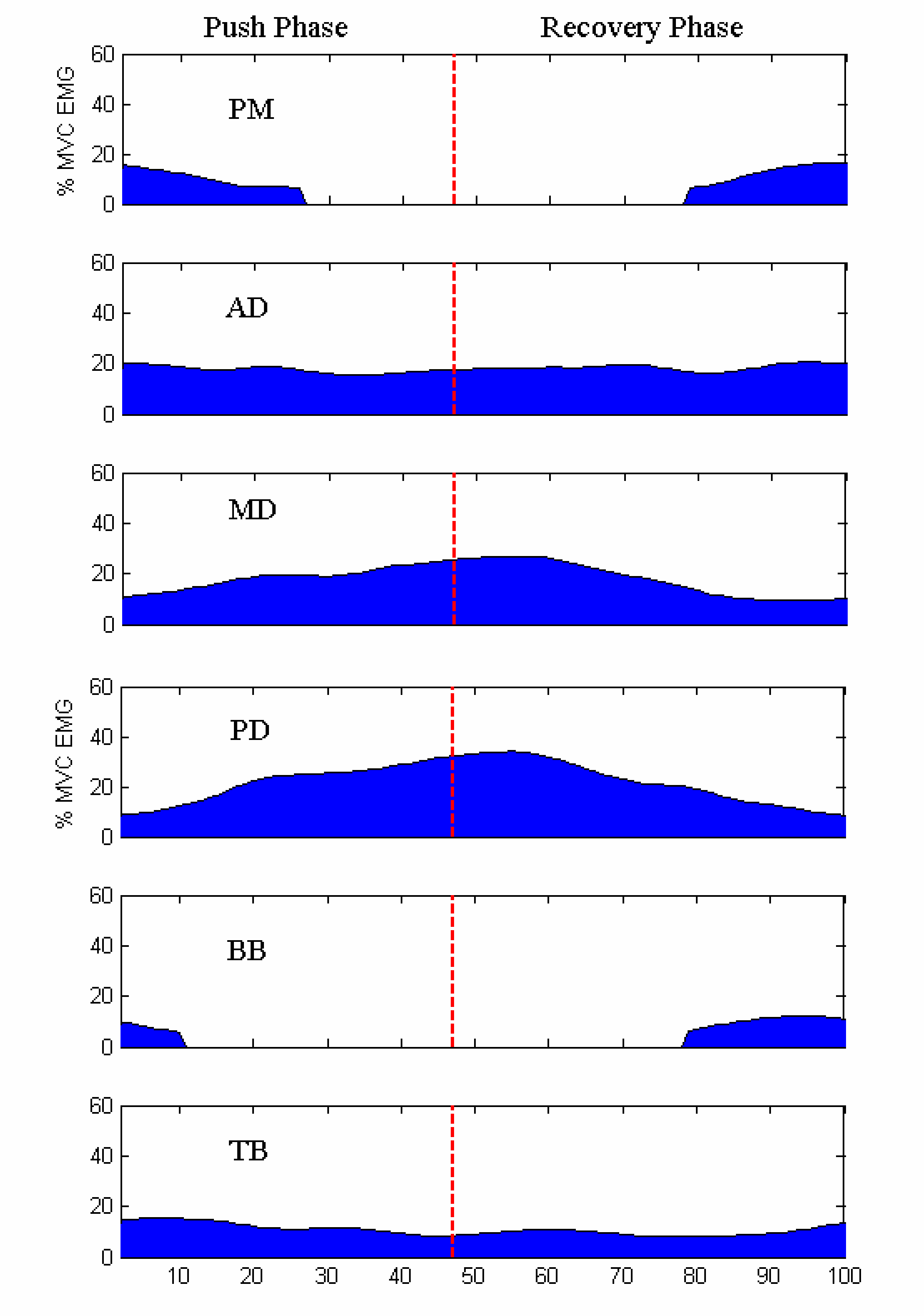
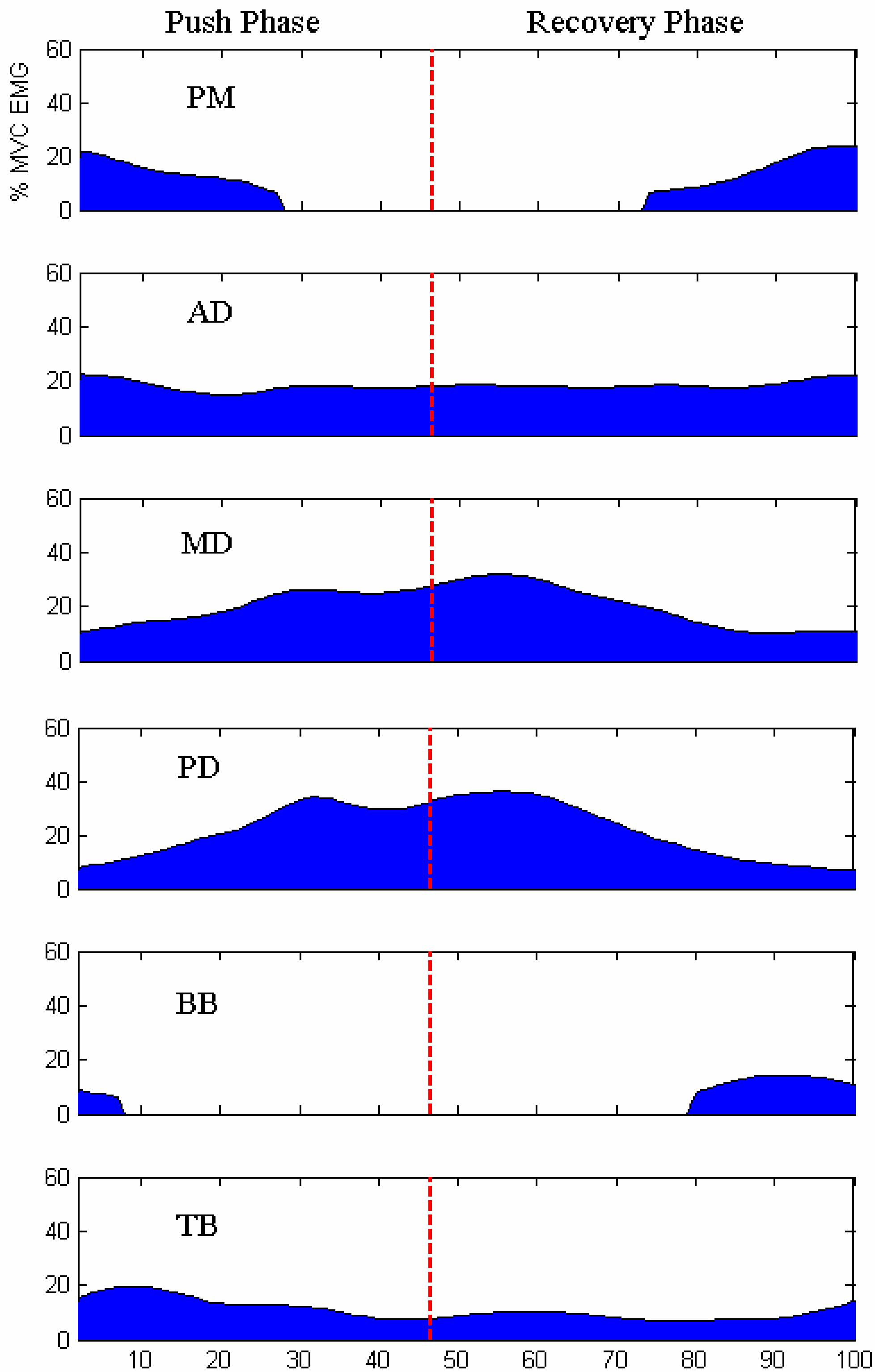


Figure 18 Group average shoulder muscle activation patterns with stimulation OFF



4.4.4. Main effect of FES

Shoulder muscle activity (Table 9 & 10) during propulsion did not vary with stimulation levels. With or without stimulation, participants propelled their wheelchairs with a similar shoulder muscle activity pattern over time. There was a significant difference found in average median intensity and iEMG of the PM between the three time intervals ($p<0.05$). Participants at T3 interval displayed larger median intensity ($p=0.02$) and iEMG ($p=0.03$) of PM during the push phase compared to the T1 interval (Table 10). A similar finding was also observed for the BB. The iEMG ($p = 0.05$) and duration ($p = 0.05$) during the push phase was significantly larger and longer at T3 than T1 (Table 10). These findings could be indicative of muscle fatigue. No interaction effect between the stimulation levels or time intervals was found for any muscle.

Table 9 Shoulder EMG activity during push phase and main effect of stimulation levels

		T1 (0-0.30 min)		T2 (2-2.30 min)		T3 (4-4.30 min)		Stim. Main Effect (Pvalue)
		Mean	SD	Mean	SD	Mean	SD	
Anterior deltoid								
Median intensity (%MVC)	HIGH	16.13	9.0	19.15	13.35	17.58	11.46	0.48
	LOW	17.44	11.4	18.92	14.33	20.15	18.32	
	OFF	17.41	12.4	16.00	10.33	18.19	16.02	
Duration (% cycle)	HIGH	43.55	3.5	46.64	8.27	42.36	7.00	0.57
	LOW	44.18	4.5	43.91	9.36	46.60	6.60	
	OFF	41.82	4.6	41.64	9.01	45.64	6.47	
iEMG (%MVC-%cycle)	HIGH	7.24	5.1	9.04	5.08	8.25	4.99	0.83
	LOW	7.78	7.7	7.91	5.44	9.80	9.24	
	OFF	7.46	12.0	7.42	4.57	9.06	6.12	
Middle deltoid								
Median intensity (%MVC)	HIGH	21.98	12.30	22.90	16.49	20.43	12.75	0.63
	LOW	20.27	13.60	20.51	13.20	21.84	17.26	
	OFF	21.98	12.30	21.97	11.72	21.56	14.93	
Duration (% cycle)	HIGH	36.73	9.34	40.18	11.78	41.00	8.81	0.74
	LOW	36.45	11.39	37.82	9.45	41.90	6.61	
	OFF	36.73	9.34	41.09	7.40	40.55	8.00	
iEMG (%MVC-%cycle)	HIGH	8.85	5.49	9.17	6.12	8.69	4.42	0.77
	LOW	8.06	5.83	8.04	4.85	9.72	6.39	
	OFF	8.85	5.49	8.77	3.96	8.59	4.11	
Posterior deltoid								
Median intensity (%MVC)	HIGH	26.03	17.51	21.25	12.54	24.77	12.64	0.41
	LOW	24.13	18.53	25.13	13.63	25.90	15.17	
	OFF	25.73	19.78	28.09	15.61	27.00	12.54	
Duration (% cycle)	HIGH	42.27	6.26	43.45	8.27	41.27	6.59	0.46
	LOW	40.64	8.25	40.27	6.53	42.10	8.70	
	OFF	40.73	9.02	43.09	7.71	38.73	8.76	
iEMG (%MVC-%cycle)	HIGH	10.59	6.10	11.28	5.62	10.94	4.51	0.74
	LOW	9.73	7.18	9.76	3.77	11.67	6.01	
	OFF	10.34	6.84	11.36	2.97	10.57	3.46	

Table 10 Shoulder EMG activity during push phase and main effect of stimulation levels

		T1 (0-0.30 min)		T2 (2-2.30 min)		T3 (4-4.30 min)		Stim. Main Effect (Pvalue)
		Mean	SD	Mean	SD	Mean	SD	
Pectoralis major								
Median intensity (%MVC)	HIGH	**13.35**	7.99	**15.29**	6.60	**16.71**	9.06	0.25
	LOW	**12.97**	7.31	**18.79**	10.17	**19.11**	12.36	
	OFF	**14.06**	5.95	**19.35**	11.24	**22.66**	12.23	
Duration (% cycle)	HIGH	**22.00**	14.04	**28.30**	12.49	**31.70**	11.15	0.67
	LOW	**22.70**	14.18	**28.80**	10.59	**29.78**	11.56	
	OFF	**29.90**	12.69	**28.20**	9.00	**29.10**	8.16	
iEMG (%MVC-%cycle)	HIGH	**3.49**	2.96	**4.95**	2.92	**6.94**	3.45	0.96
	LOW	**3.18**	2.62	**5.57**	4.47	**5.89**	4.57	
	OFF	**4.10**	2.13	**4.94**	1.88	**6.23**	2.74	
Biceps brachii								
Median intensity (%MVC)	HIGH	6.44	4.92	9.59	8.43	8.69	8.20	0.13
	LOW	7.13	7.63	7.74	6.56	6.20	8.49	
	OFF	5.39	5.10	5.30	2.85	6.67	4.08	
Duration (% cycle)	HIGH	**8.91**	8.62	**13.00**	11.60	**13.09**	11.85	0.84
	LOW	**9.09**	11.43	**10.09**	10.60	**12.20**	18.29	
	OFF	**7.73**	8.61	**11.36**	13.02	**14.27**	12.54	
iEMG (%MVC-%cycle)	HIGH	**0.97**	1.12	**1.92**	2.25	**2.05**	2.77	0.12
	LOW	**1.27**	2.22	**1.36**	2.08	**1.62**	2.64	
	OFF	**0.71**	0.85	**1.10**	1.69	**1.34**	1.29	
Triceps brachii								
Median intensity (%MVC)	HIGH	15.31	10.26	15.75	9.65	18.65	12.91	0.65
	LOW	14.97	8.46	15.42	7.58	18.51	9.69	
	OFF	16.32	10.48	17.85	9.61	17.74	8.73	
Duration (% cycle)	HIGH	33.27	14.31	34.64	15.09	37.36	12.78	0.42
	LOW	32.27	15.73	37.27	13.77	35.80	14.64	
	OFF	35.82	9.90	38.73	12.79	37.00	10.78	
iEMG (%MVC-%cycle)	HIGH	5.94	4.27	6.04	3.96	6.65	3.15	0.82
	LOW	5.61	3.49	6.14	3.07	6.64	4.55	
	OFF	5.76	2.91	6.97	3.85	6.56	3.26	

Highlighted numbers indicate significant differences due to main effect of time interval ($p<0.05$)

4.5. DISCUSSION

4.5.1. Propulsion kinetics

In this study, surface stimulation was applied to paralyzed back and abdominal musculature to augment trunk stability during propulsion. The results showed that HIGH stimulation resulted in higher power output and velocity compared to the other two stimulation levels consistently across the three time intervals. This finding was consistent with our previous findings (Yang, Koontz, Triolo, Mercer and Boninger 2005a). By inducing trunk muscle cocontraction through electrical stimulation, propulsion torque from the upper limbs may transfer to the pushrim more effectively, but also propulsion speed increased. As a result, the power output may have resulted from larger torque in combination with a faster speed.

4.5.2. Shoulder muscle activity

The patterns of shoulder muscle activity in the present study were similar to those in previous studies. The PM and AD appeared to function as prime movers during the push phase (Masse et al. 1992; Mulroy et al. 1996; Schantz et al. 1999). AD also showed continuous activation during the recovery phase. The AD muscle likely served as a shoulder stabilizer during the recovery phase (Masse et al. 1992; Schantz et al. 1999).

During the push phase, the activity of the MD and PD muscles act to stabilize the shoulders, and extend the arm back to prepare for the next stroke during the recovery phase (Harburn and Spaulding 1986; Masse et al. 1992; Schantz et al. 1999). Of all of the six shoulder muscles monitored, the MD and PD were the most consistently active with moderate intensity during the entire propulsion cycle. These two muscles appeared to have an important role during wheelchair propulsion. Endurance training incorporating these two muscle groups may be advantageous in the design of future intervention programs.

The BB is another prime mover to bring the arm upward and forward in the early push phase (pull motion) and was used to flex the elbow during late recovery and in preparation for the next stroke (Harburn and Spaulding 1986; Masse et al. 1992; Mulroy et al. 1996; Schantz et al. 1999). However, BB was also inactive in some subjects during the push phase which was consistent with a prior report by Schantz et al. (Harburn and Spaulding 1986; Masse et al. 1992; Mulroy et al. 1996; Schantz et al. 1999). For these subjects no pull motion occurred during the push phase.

The TB was active during the push phase of propulsion and served as another prime mover to push forward and downward on the rim. In seven subjects, TB muscle activity started in the early push phase and ceased at the end of the recovery phase (Figure 19) consistent with findings by Mulroy et al. and Rodgers et al. (Mulroy et al. 1996; Rodgers, Gayle, Figoni, Kobayashi, Lieh and Glaser 1994). For the three subjects not following this trend, their activity pattern was similar to that of the BB which showed two peaks of activity during an entire propulsion cycle, one at the beginning and one at the end of the recovery (Figure 20). Their activation time was followed by the BB. These different activation patterns could be explained by individual variation in subjects' propulsion patterns which have been reported by Boninger et al. (Boninger, Souza, Cooper, Fitzgerald, Koontz and Fay 2002). Further investigation may be warranted to investigate the effect of propulsion pattern on the TB's muscle activation patterns.

Figure 19 Triceps brachii activation patterns among 8 subjects

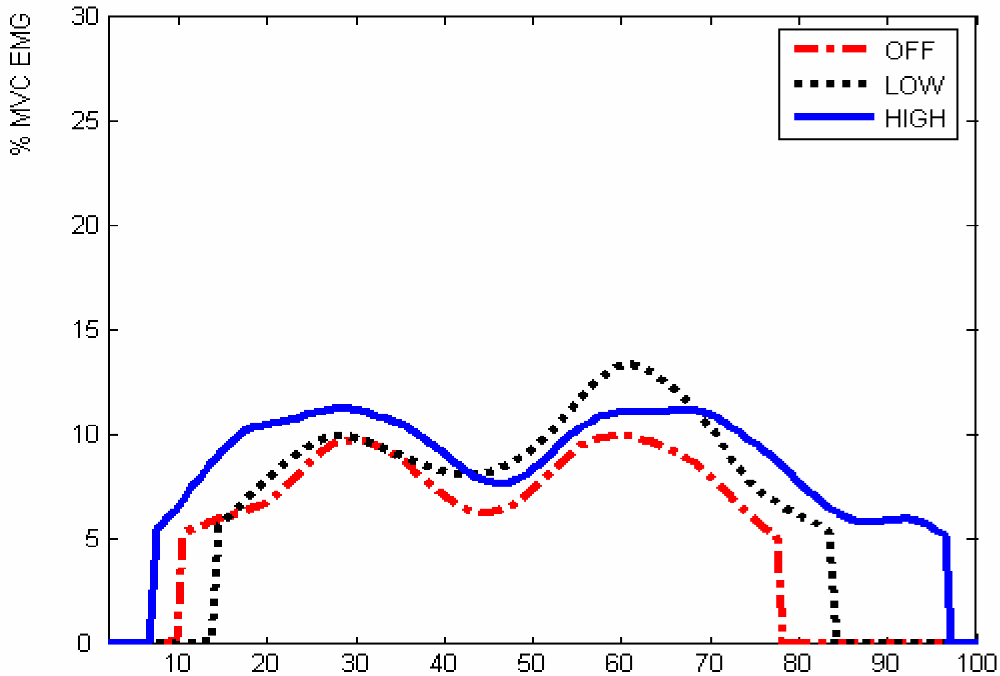
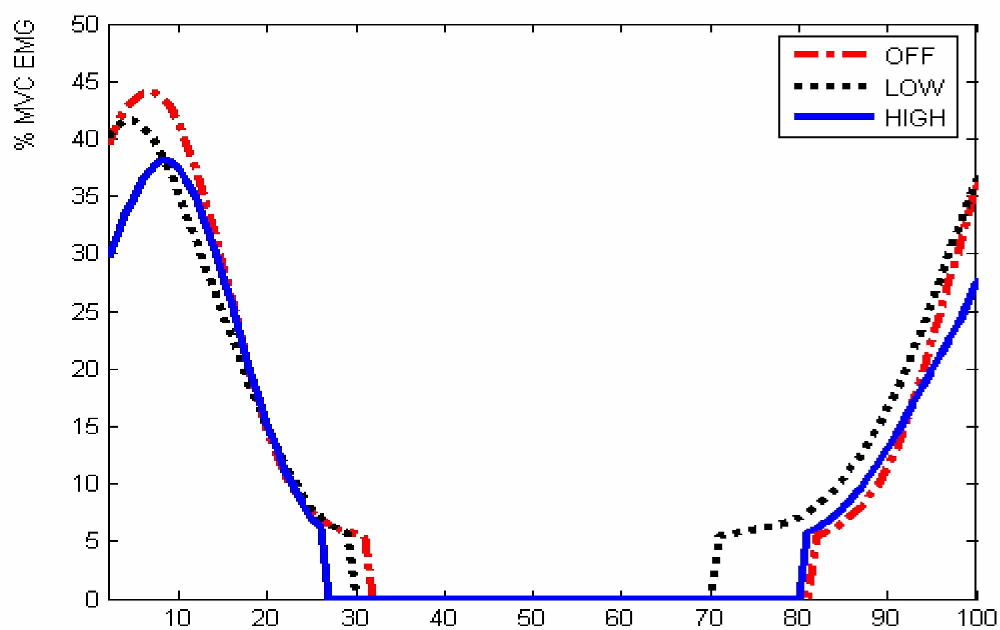


Figure 20 Triceps brachii activation patterns among 3 subjects



4.5.3. Effect of stimulation

Shoulder EMG activity patterns were similar amongst the three FES stimulation levels. With or without stimulation to stabilize their trunk, participants propelled their wheelchairs with similar shoulder muscle activation patterns. However, participants using HIGH stimulation produced greater power output and faster velocities with similar shoulder muscle intensity and duration. It may be likely that muscles which were acting to stabilize the joint to compensate for weak trunk musculature are being used more to move the arm. As a result, propulsion power may increase with no change seen in shoulder muscle intensity and duration.

In order to produce power on the pushrim during wheelchair propulsion, shoulder stabilization is needed to control arm movements and transfer power from the limbs to the pushrim (van der Helm and Veeger 1996; van der Woude et al. 2001a). The shoulder plays an integral role in facilitating power production to the push rim. In order to deliver power to the pushrim, proximal shoulder stability is vital. However, because of the functional anatomy of the shoulder, it is by nature a highly flexible and potentially unstable joint. If power is to be transferred optimally from the shoulder, down through the arm to the push rim, the shoulder itself must be stable and able to allocate more of its activity and expenditure towards delivery of propulsion forces and less towards active stabilization. As the shoulder is perhaps the foundation or origin of the power produced during a propulsion stroke, any instability can cause undesired or unintended movements resulting in an inefficient transfer of power from the shoulder down through the arm and wrist.

Muscle can function as a segment mover and also a joint stabilizer, providing force and power for the movement and balance of the musculoskeletal structures (An 2002). Guo et al. used a power flow model to illustrate the characteristics of mechanical energy and power flow of upper limb during wheelchair propulsion (Guo, Su, Wu and An 2003). They indicated that the proximal parts of the upper limb and trunk act as propulsion movers and stabilizers to push the wheelchair. The proximal shoulder muscles must work to stabilize the joint and produce propulsion movement at the same time. Therefore, the estimated power supplied from the proximal part of upper limb is often greater than the actual mechanical requirement. With appropriate stability of the proximal part of shoulder and trunk (e.g., through stimulation, appropriate postural support or wheelchair modification), less muscle force will be needed to stabilize the shoulder and more force can be used to move the joint, thereby increasing the propulsion power on the pushrim.

To further investigate this theory, changes in EMG activity could be documented from other muscles which act as primary stabilizers (e.g. rotator cuffs, or intact abdominal/back muscles) while providing external stability of the trunk through a FES device during propulsion. Another alternative may be to replicate the present study with unimpaired subjects. While controlling the propulsion power output, it can be expected that shoulder EMG activities from unimpaired subjects who had sufficient trunk stability might show less muscle activity intensity compared to the same shoulder muscles from impaired subjects who lack trunk stability.

4.5.4. Effect of time intervals

Participants showed a trend for decreasing averaged power output ($p=0.06$), and an increased PT ($p=0.013$), ST ($p=0.003$) and higher stroke frequency ($p=0.002$) over the 5-minute propulsion trial. These changes may be a result of muscle fatigue. At the same time, the PM and BB, which are two of prime movers during push phase at T3 showed higher intensity, larger iEMG and longer duration compared to T1 (fresh stage) ($p<0.05$). The increase in muscle active intensity and duration could be a compensatory strategy when an individual becomes fatigued. This finding is in agreement with results by Rogers et al. (Rodgers et al. 1994). They found that fatigue resulted in prolonged EMG activity during propulsion, but they did not indicate whether the intensity of muscle activity increased or not.

A majority of FES users use FES for exercise, standing and short distance walking (Moynahan, Mullin, Cohn, Burns, Halden, Triolo and Betz 1996; Kobetic, Triolo, Uhlir, Bieri, Wibowo, Polando, Marsolais, Davis and Ferguson 1999). This study indicates another potential use and benefit of FES. Propelling a wheelchair for longer than 10 to 20 minutes placed significant demands on shoulder musculature (Mulroy et al. 1996). Manual wheelchair users often propel for several minutes or more at a time (Hoover, Cooper, Ding, Koontz, Cooper, Fitzgerald and Boninger 2004). Prolonged manual wheelchair use can lead to pain and repetitive strain injury (RSI) in the upper extremities (Subbarao, Klopfstein and Turpin 1995; Nichols, Norman and Ennis 1979; Pentland and Twomey 1991; Dalyan, Cardenas and Gerard 1999; Boninger, Towers, Cooper, Dicianno and Munin 2001). The shoulder is the most commonly reported site of musculoskeletal injury in MWUs. Using electrical stimulation to help stabilize the trunk during propulsion may enable individuals to produce greater propulsion power without placing increased demands on shoulder musculature, thereby reducing potential shoulder injury due to long term wheelchair use. However, continuous electrical stimulation on the trunk musculature for several minutes would cause muscle fatigue. Therefore, with advanced sensing and programming, a FES device could possibly be synchronized with the propulsion cycle to avoid continuous stimulation on trunk musculature and only provide stimulation during pre-push and early push phase of the propulsion cycle when trunk stability is most important (Yang et al. 2005b). Alternatively, FES could be used for challenging propulsion tasks of short duration, such as pushing up a ramp or across thick carpet.

4.5.5. Limitations:

With surface electrodes only superficial muscle activity can be measured. Other muscles, such as teres minor, supraspinatus and the subscapularis, also play an important role during wheelchair propulsion (Mulroy et al. 1996). These muscles were not recorded in present study because they are too deep to measure accurately with surface EMG. Use of fine-wire EMG would be necessary for studying the activity of these muscles during propulsion.

Another potential limitation is the manufacturing limits of the surface FES device in this study. The surface FES device used in this study is a FDA approved commercial device, in which the maximal stimulation intensity was limited due to safety considerations. Four of twelve participants were able to comfortably tolerate the maximum stimulation intensity from the FES device. It is likely that these individuals could tolerate more stimulation intensity than the device could deliver. In order to remain consistent with experimental protocol, we still used a 25%, and 50% differential between the threshold and maximal stimulation intensity which the device could provide as LOW and HIGH stimulation levels. The setting of LOW stimulation intensity on these four subjects may not have been sufficient to cause an effective trunk muscle contraction due to a ceiling effect of the device. A secondary nonparametric analysis using Friedman repeated measures ANOVA was done to examine the influence of this ceiling effect. The result indicated that these four participants with LOW stimulation showed no statistical difference on propulsion power ($p>0.1$) compared to the OFF condition. The rest of seven participants who were not affected by the ceiling effect showed a statistically larger propulsion power production ($p=0.04$) for LOW stimulation compared to OFF condition. The LOW stimulation intensity for the four participants was not high enough to cause differences, thereby reducing the group mean of propulsion power for the LOW condition.

In the present study, we did not quantify trunk stability was gained with FES device. Instead, we did manual palpation to verify abdominal/back muscle contraction. Future studies should evaluate the extent of trunk control due to electrical stimulation. Furthermore, most subjects in present study had no previous experience using surface FES on their trunk. Subjects were given several minutes to familiarize themselves with the different stimulation levels during propulsion before the experimental trials. However, this short period of practice time with stimulation may not be long enough for subjects to get used to propelling their wheelchair with FES. A future study should investigate prolonged use of surface FES, which may help participants without trunk control realize and gain the benefits of FES on trunk stability.

4.6. CONCLUSION

The patterns of shoulder muscle activation in the present study were consistent with other studies on wheelchair propulsion. No significant stimulation-related changes on shoulder EMG activation patterns were found. However, trunk FES has advantage of helping individuals with SCI generate propulsion power. With trunk stimulation, less muscle effort may be necessary to stabilize the joint and more effort can be devoted more directly to propulsion. Therefore, trunk FES may help individuals to generate propulsion power without placing additional demands on shoulder musculature.

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5. CONCLUSIONS

An underlying goal for this dissertation was to investigate if augmenting trunk stability through surface electrical stimulation would improve wheelchair propulsion efficiency, in terms of propulsion kinetics, upper limb range of motion, and gross mechanical efficiency, while reducing the demand on the shoulder musculature. This first study provided an understanding of the functional role of trunk musculature during wheelchair propulsion based on unimpaired subjects. The results described a muscle activation profile of the trunk musculature at three different propulsion speed conditions. Both back and abdominal muscle groups illustrated the highest intensity during the pre-push and early push stages of the propulsion cycle. Moreover, these two muscle groups cocontracted to provide sufficient trunk stability for the propulsion tasks. The results suggest that it may be worthwhile to augment trunk stability through electrical stimulation on the trunk musculature of wheelchair users, through surface stimulation or modification of existing implanted FES systems, in order to improve wheelchair propulsion performance.

The second study indicated that MWUs with a trunk FES device could generate more propulsion power and increase gross mechanical efficiency during wheelchair propulsion. Because of its biomechanical advantage and higher gross mechanical efficiency performance, a user with a trunk FES device may be able to easily negotiate more demanding propulsion tasks such as ramps and curb ascents, and traversing outdoor terrain. It is worth noting that no statistical differences in upper limb motion or other kinetic variables between stimulation levels were found. It is possible that the stimulation intensity of the current surface FES-induced trunk stability device was not strong enough to stabilize the trunk and cause changes among these variables. Surface FES devices with higher stimulation intensity could be used in future studies to further examine the effect on wheelchair propulsion kinetics and kinematics.

The shoulder muscle activation pattern in the third study was consistent with other studies reporting EMG activity during wheelchair propulsion. No significant stimulation-related changes on shoulder EMG activation patterns were found. However, trunk FES has the advantage of helping individuals with SCI generate propulsion power without altering the demand of shoulder musculature. Some shoulder muscles may decrease the demand on activity to act as stabilizers and divert this muscular activity to propulsion power. Therefore, MWUs with augmented trunk stability through electrical stimulation may generate increased propulsion power without placing additional demands on shoulder musculature.

Most users of FES systems use them for exercise, standing and short distance walking. Based on the findings on this dissertation, augmenting trunk stability through electrical stimulation on trunk musculature helps individuals with SCI to generate increased propulsion power and efficiency without placing additional demands on shoulder musculature thereby reducing potential shoulder injury due to long-term wheelchair use. Therefore, it could be possible to integrate a user’s existing FES system for daily wheelchair use in addition to the specialized uses described above. For MWUs who do not use an FES system, customizing the wheelchair by either using a rigid backrest or reclining seat frame may be another compensatory strategy for dealing with the loss of trunk stability. With adequate stabilization of the trunk, MWUs may improve their propulsion performance.

5.1. LIMITATIONS

There are a number of limitations in this dissertation that require consideration. During the first study, the test wheelchair was not adjusted to the individual's anthropometry and a low sling backrest (height 20 cm) was used on the test wheelchair. Although unimpaired participants were instructed not to lean on the backrest during recording the activity of trunk musculature, use of the backrest could have resulted in different trunk muscle activation profiles.

Muscle fatigue is a known side effect of FES. This limitation affects the second study where a constant stimulation was applied to the subjected trunk musculature. Participants using stimulation to stabilize their trunk could gain some benefits at the beginning, but the efficiency might decrease over time. With advanced sensing and programming, a FES device could possibly be synchronized with the propulsion cycle to avoid continuous stimulation on trunk musculature and only provide stimulation during pre-push and early push phase of the propulsion cycle when trunk stability is most important as documented in the first study. Alternatively, FES could be used for challenging propulsion tasks of short duration, such as pushing up a ramp or across thick carpet.

Another potential limitation is the manufacturing limits of the surface FES device in this study. The surface FES device used in this study is a FDA approved commercial device, in which the maximal stimulation intensity was limited due to safety considerations. Four of twelve participants were able to comfortably tolerate the maximum stimulation intensity from the FES device. It is likely that these individuals could tolerate more stimulation intensity than the device could deliver. The setting of LOW stimulation intensity on these four subjects may not have been sufficient to cause differences, thereby reducing the group mean of propulsion power for the LOW condition.

The lower power output found for LOW condition compared to without stimulation may indicate that the insufficient low stimulation intensity on trunk musculature hindered rather than helped participants. Low intensity stimulation could have resulted in only minimal muscle contractile activity that interfered with the subject's propulsion performance. A FES device with a higher stimulation output (e.g. a FES with implantable electrodes or a custom FES system that can be made to deliver higher voltages for this particular application) might be needed to lead to significant effects on biomechanical variables during propulsion.

In the present study, we did not quantify trunk stability was gained with FES device. Instead, we did manual palpation to verify abdominal/back muscle contraction. Future studies should evaluate the extent of trunk control due to electrical stimulation. Furthermore, most subjects in present study had no previous experience using surface FES on their trunk. Subjects were given several minutes to familiarize themselves with the different stimulation levels during propulsion before the experimental trials. However, this short period of practice time with stimulation may not be long enough for subjects to get used to propelling their wheelchair with FES. A future study should investigate prolonged use of surface FES, which may help participants without trunk control realize and gain the benefits of FES on trunk stability.

During the third study, only six superficial shoulder muscles were measured and as such the results were limited. For other deep muscles (e.g. rotator cuff muscles), use of fine-wire

EMG would be needed. Secondly, the participant had a short time to acclimate to the FES system, which may have impacted their propulsion biomechanics. Prolonged use of a FES-induced trunk stability device on a daily basis in the community may help manual wheelchair users without trunk control ability to realize the full benefits, thereby modifying propulsion pattern and increasing propulsion efficiency.

5.2. FUTURE WORK

This dissertation work is the first pilot study to investigate if augmenting trunk stability through surface electrical stimulation affects wheelchair propulsion biomechanics, shoulder EMG and physiological responses. The results can help researchers to evaluate the feasibility of FES on trunk musculature during wheelchair propulsion. However, a size was recruited in this dissertation work, thereby having limitations in the generalization capability of the current results. Based on a technical report by D'Amico et al, we conducted a statistical power analysis to detect the statistical power of the current dissertation work by analyzing peak propulsion forces, which is one of key variables but did not show significant differences between stimulation levels. The preliminary result showed that there is a low statistical power (0.14). Thus, the chance to detect a significant change in peak propulsion force between stimulation levels based on the current study sample size (n=12) was small. Nonsignificant outcomes of current study may simply mean that the available evidence is not strong enough to reject the null hypothesis. The effect of FES on wheelchair biomechanics may potentially exist. We further estimated an adequate sample size based on the current research design. The results showed that at least 93 subjects should be recruited to allow for a statistical power = 0.8 with a significance level (α) = 0.05 based on the current research design. However, it is very difficult to conduct a study with this kind large sample size at single laboratory setting within a certain time frame. A large multicenter study with less complicated study design might be proposed in a future research project to validate the generalization of effect of FES on trunk musculature during wheelchair propulsion.

Future studies could potentially build on this dissertation research in other ways as well. Customizing the wheelchair by either using a rigid backrest or reclining seat frame angles may result in similar outcomes of trunk stability compared to electrical stimulation on trunk musculature. Prospective studies could examine propulsion efficiency ,upper extremity joint loading and muscle fatigue with other strategies (e.g. rigid back rest, ultralight wheelchairs with reclined seat frame).

The change in the functional role of shoulder muscle hypothesized from the results of the third study remains unclear and needs to be further verified. With appropriate stability of the proximal part of shoulder and trunk (e.g., through stimulation, appropriate postural support or wheelchair modification), shoulder muscle force could be used more effectively to push the wheelchair rather than to stabilize the shoulder, thereby minimizing the risk of shoulder pain and injury due to overuse. It may be interesting to replicate the present study with unimpaired subjects. While controlling the propulsion power output, it can be expected that shoulder EMG activities from unimpaired subjects who had sufficient trunk stability might show less muscle activity intensity compared to the same shoulder muscles from impaired subjects who lack trunk stability.

After spinal cord injuries, muscles normally supplied by intact motoneurons from spinal cord segments at or below the injury site are paralyzed and undergo atrophy. Since the participants were long-term wheelchair users (on average 17 years of post injuries), their trunk musculature below the injury site due to disuse atrophy could have less response of muscle contraction induced by FES. In order to avoid the effect of disuse atrophy, a training program with low-frequency electrical stimulation via implanted or skin surface electrode of the trunk muscles can be used for preparing these muscles prior to the application of FES during wheelchair propulsion. An electrical stimulation training program has been widely used to train muscles and counteract disuse atrophy (Kralj and Bajd 1989; Rodgers, Glaser, Figoni, Hooker, Ezenwa, Collins, Mathews, Suryaprasad and Gupta 1991; Gordon and Mao 1994). This training not only conditioned the muscles but allowed individuals to become more experienced and comfortable with FES while performing t functional movements.

Muscle fatigue could be a potential problem resulting from continuous stimulation to trunk musculature. With advanced sensing and programming, a FES device could possibly be synchronized with the propulsion cycle to avoid muscle fatigue and only provide stimulation during pre-push and early push phase of the propulsion cycle when trunk stability is most important. Based on the EMG profile from the third study, the biceps EMG activity was shown to behave similarly to the trunk muscle activation patterns: two peaks of activity during an entire propulsion cycle, one at the beginning and one at the end of the recovery. It may be possible to use the EMG signal from the biceps brachii to trigger a FES device, stimulating the trunk muscles to provide trunk stability during pre-push and early push phase of the propulsion cycle.

As a final recommendation for future work, effects of trunk FES could be investigated during challenging propulsion tasks of short duration, such as pushing up a ramp or across thick carpet. Future studies could explore the potential advantage of FES system during wheelchair propulsion in activities for daily living. This is currently under investigation by authors.

5.3. BIBLIOGRAPHY

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APPENDICES

APPENDIX A: THE EFFECT OF FUNCTIONAL ELECTRICAL STIMULATION DURING WHEELCHAIR PROPULSION

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INTRODUCTION

Functional electrical stimulation (FES) can provide people with SCI with activities of daily living [1]. Bilateral activation of the paraspinal muscles in implanted FES users can improve their seated posture [2], thereby may improve their propulsion efficiency. The goal of this preliminary investigation is to quantify the effect of stimulating the lumbar trunk on propulsion biomechanics by using implanted FES system. The results can provide the insight of the benefit of FES during wheelchair propulsion.

METHODS

Subjects: Three subjects (two male and one female) who had received implanted standing CWRU/VA neuroprosthesis at least one year without any medical complications provided consents to participate in this study. Their age, weight and years of wheelchair use were 40.5 $\pm$ 9.3 years old, 1.72 $\pm$ 0.05 meter, and 5.8 $\pm$ 0.7 years respectively.

Experimental protocol: Subjects' own wheelchairs were fitted bilaterally with SMART$^{\text{WheelsTM}}$ (Three Rivers Holdings, ILL., Mesa, AZ), and secured to a dynamometer with a four-point tie down system. An OPTOTRACK motion analysis system (Northern Digital Inc., Ontario, Canada) was synchronized with the kinetic system to record subject's kinematic data during trials. Subjects were asked to propel their wheelchairs at a steady-state speed of 0.9, and

1.8 m/s for one minute while 20 seconds of data were collected. Real-time propulsion speed was displayed on a computer screen in front of the subjects. All propulsion trials were repeated three times: two with electrical stimulation ON (at 50% and 25% maximal recruitment), and one with electrical stimulation OFF. The order of stimulation was randomly assigned. To minimize fatigue, at least one-minute of rest was provided between trials.

Data analysis: For each stroke, the start and end of the push phase was determined by the presence/absence of forces detected by SMART$^{\text{WheelsTM}}$. The kinetic data were collected at 240 Hz and linearly interpolated for synchronization with the kinematic data with collection rate of 60 Hz. Since data from both sides were highly correlated ($r^2 = 0.89$; $p < 0.01$), average values of both sides were obtained on all biomechanical variables over ten continuous strokes. Descriptive analyses were reported for each speed condition separately. Due to one subject cannot reach the target speed at 1.8 m/s during study, only data from two subjects were reported under this speed condition.

RESULTS AND DISCUSSION

Table 1 summarized the biomechanical variables while propelling with and without FES. The results showed that continuous activation of the paraspinal muscles have better propulsion performance than without. Propulsion efficiency, the percentage of the resultant force leading to effective forward propulsion, is generally higher while stimulation was given over all speed conditions. Subjects with stimulation ON generally produced higher propulsive force with longer stroke cadence. Their trunks were most likely able to lean forward during pushing wheelchair. This ability of trunk leaning may help the subjects to transfer of power from the upper extremities to the pushrim, thereby increase propulsion efficiency [3]. Although low level (25%)

stimulation activation showed less advantages than high level (50%) based on present results, but it could cause less muscle fatigue with the use over long period of time. Low activation level also could allow some freedoms to oscillate the trunk in comparison with high level. These cons of low activation may need to further investigated. Due to limited people who had received implanted FES system so far, this present study is limited on small sample size. A future study with using surface FES system, a noninvasive system, may be needed to allow increased sampling pool and elicit the benefit of FES on wheelchair propulsion.

CONCLUSIONS

Stabilizing the trunk by continuous stimulation of the lumbar erector spinae appears to improve manual wheelchair propulsion in present study. With activation of back muscle, implanted FES users were able to lean forward thereby increase propulsion efficiency. A future study with large sample size is need to verify the benefit of FES on wheelchair propulsion.

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ACKNOWLEDGEMENTS

VA Rehab R&D (B3043-C) & ASB'2004 Graduate Student-In-Aid.

Table 1: Summary of the effects of trunk stimulation on manual wheelchair propulsion**.**

Speed	0.9m/s (n=3)					1.8m/s (n=2)				
Stimulation level	Cadence (stroke/sec)	Max force (N)	Moments (N-M)	Efficiency (%)	Trunk angle (^{O})	Cadence (stroke/sec)	Max force (N)	Moments (N-M)	Efficiency (%)	Trunk angle (^{O})
OFF	1.20± 0.2	68.0± 3.3	6.85± 0.6	0.59±0.04	1.9±1.2	1.32± 0.01	90.4± 3.7	8.24± 2.4	0.51± 0.06	4.03±10.5
25 %	1.26±0.2	68.4± 3.2	6.33± 0.5	0.55±0.02	19.9±14	1.40± 0.07	89.5± 6.1	8.22± 2.5	0.55± 0.04	18.4± 9.7
50 %	1.20± 0.2	70.1± 3.2	6.98± 0.9	0.62±0.05	16.2±9.9	1.39± 0.13	97.8± 4.7	8.47± 2.9	0.55± 0.07	18.0±15.7

APPENDIX B: MATLAB PROGRAMS FOR STUDY #1

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
function FESCONTROL6
num_sub=input('Input number of subjects to run:','s');
num_sub=str2num(num_sub);
ID_matrix=[];

for i=1:num_sub
  ID = input('enter patient 4 digit ID [ex: p3b3]: ', 's');
  ID_matrix=[ID_matrix;ID];
end;

[number,c]=size(ID_matrix);
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%Loading individual data %%%%%%%%%%%%%%%%%%%%

for gm=1:number
   cd('S:\Protocols\Trunk Stimulation\DATA\Subject_data')
   rawID=ID_matrix(gm,:);
   subj_name=rawID(1:4);
   cd(subj_name)
%%%%%%%%%%%%%%%%Loading SW data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

   cd('Clean_FM');
   if gm==1
      disp('Please select a FORCE DATA file.');
      dataname_force=uigetfile('*.*','Please select a FORCE DATA file.');
      sw=load(dataname_force);
      side=dataname_force(7);
      condition=dataname_force(6);
      speed=dataname_force(8);
         if length(sw)<4800
            sw(length(sw):4800,:)=0;
            swdata=sw;
         else
            swdata=sw(1:4800,:);
         end;
   else
      sw=load([subj_name ,'w',condition,side,speed]);
         if length(sw)<4800
            sw(length(sw):4800,:)=0;
            swdata=sw;
         else
            swdata=sw(1:4800,:);
         end;
   end;


%%%%%%%%%%%%%%%%Loading MO data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

   cd ..
   cd('Clean_MO\SW');
   modata=load([subj_name ,'m',condition,'b',speed]);

%%%%%%%%%%%%%%%%Loading sepo motion data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
   sepo_name=[subj_name,'msp1'];
   subj_sepo=load(sepo_name);

   cd ..
```

```
    cd ..

    cd('Clean_EMG')
    fname=[subj_name, 'e',condition,'b',speed];
    emgdata=load([fname]);
    emgdata=emgdata(1:20000,:);
    fname=[subj_name, 'rest'];
    restemg=load([fname]);
    fname=[subj_name, 'abmvc'];
    abmvc=load([fname]);
    fname=[subj_name, 'bkmvc'];
    bkmvc=load([fname]);

    cd('S:\Students\Yusheng\Trunk_Stim_Project\Data_analysis\control');


%%%%%%%%%%%%%%%%%%%%%%%change the sampling rate %%%%%%%%%%%%%%%%%%%%%%
    [raw,col]=size(emgdata);
     F=[]; F1=[]; F2=[]; F3=[]; F4=[];
    for n=1:col
       F=emgdata(:,n);
       Fnew(n,:)= spline(1:20000,F,1:(20000/4800):20000); %%20000 is EMG sample# 4800 is SW sample #
    end;
    newemg=Fnew';
    newsw=swdata;

%%%%%%%%%%%%%%%determine onpercentage based on 14 subjects%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%% the answer was given based ontimep2 m.file %%%%%%%%%%%%%%%%
    if condition=='1' & speed=='2';
       onper=50;
       offper=50;
       onper_1=5;
       offper_1=57;
       offper_2=98;
       offper_3=100;
    elseif condition=='1' & speed=='4';
       onper=45;
       offper=55;
       onper_1=2;
       offper_1=53;
       offper_2=97;
       offper_3=100;
    end;

%%%%%%%%%%%%%%%%%%trunk angle during pushing %%%%%%%%%%%%%%%%%%%%%%%%
    [trkang] = trunkang(side, modata, subj_sepo);
    F1=trkang;
    F1new= spline(1:1200,F1,1:(1200/4800):1200);
    newtrkang=F1new';
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%% spline the emg data based on the push and recovery phase%%%%%%%%%%%%%%%%%%%%
%%% Then calcuate the EMG %MVC for each phase based on group mean %%%%%%%%%%%%%%%%%%%%
       mxnum=max(swdata(:,7));
       if mxnum<10
          mxstroke=mxnum-1;
       else
          mxstroke=10;
       end;

    for j=1:mxstroke;          %%pick up for 10 strokes%%
          on=min(find(newsw(:,7)==j));
          off=max(find(newsw(:,7)==j));
```

```
  strokend=max(find(newsw(:,7)==-j));
  F3=newtrkang(on:strokend);
  F4=[spline(1:length(F3),F3,1:(length(F3)/100):length(F3))];
%%%%%%% trunk angle during each phase %%%%%%%%%
  trkon_1(j) =median(F4(1:onper_1));
  trkon_2(j) =median(F4(onper_1+1:onper));
  trkoff_1(j)=median(F4(onper+1:offper_1));
  trkoff_2(j)=median(F4(offper_1+1:offper_2));
  trkoff_3(j)=median(F4(offper_2+1:offper_3));
%%%%%%% determine the significant EMG activity >5% propulsion cycle %%%%%%%%%%%%%%%%%%%%%%%%
  for n=2:8
     F=newemg(on:strokend,n);
     rawF2=[spline(1:length(F),F,1:(length(F)/100):length(F))];
     for g=3:length(rawF2)-2;
        if rawF2(g-2:g+2)<5
           F2(g)=0;
        else
           F2(g)=rawF2(g);
        end;
     end;
     for g=1:2;
        if rawF2(g:g+1)<5
           F2(g)=0;
        else
           F2(g)=rawF2(g);
        end;
     end;
     for g=98:100;
        if rawF2(g-1:100)<5
           F2(g)=0;
        else
           F2(g)=rawF2(g);
        end;
     end;
     %%%% Median intensity of EMG activation%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
     on_1 =median(nonzeros(F2(1:onper_1)));
     on_2 =median(nonzeros(F2(onper_1+1:onper)));
     off_1=median(nonzeros(F2(onper+1:offper_1)));
     off_2=median(nonzeros(F2(offper_1+1:offper_2)));
     off_3=median(nonzeros(F2(offper_2+1:offper_3)));

     xon_1 =max(F2(1:onper_1));
     xon_2 =max(F2(onper_1+1:onper));
     xoff_1=max(F2(onper+1:offper_1));
     xoff_2=max(F2(offper_1+1:offper_2));
     xoff_3=max(F2(offper_2+1:offper_3));
     %%%% Duration of EMG activation%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
     don_1 =sum(find(F2(1:onper_1))>0);
     don_2 =sum(find(F2(onper_1+1:onper))>0);
     doff_1=sum(find(F2(onper+1:offper_1))>0);
     doff_2=sum(find(F2(offper_1+1:offper_2))>0);
     doff_3=sum(find(F2(offper_2+1:offper_3))>0);

     if n==2
        pRAon_1(j,:) =on_1;
        pRAon_2(j,:) =on_2;
        pRAoff_1(j,:)=off_1;
        pRAoff_2(j,:)=off_2;
        pRAoff_3(j,:)=off_3;
        %%% Max MVC %%%
        xRAon_1(j,:) =xon_1;
        xRAon_2(j,:) =xon_2;
```

```
        xRAoff_1(j,:)=xoff_1;
        xRAoff_2(j,:)=xoff_2;
        xRAoff_3(j,:)=xoff_3;
        %%%EMG durnation %%%
        RAdon_1(j,:) =don_1;
        RAdon_2(j,:) =don_2;
        RAdoff_1(j,:)=doff_1;
        RAdoff_2(j,:)=doff_2;
        RAdoff_3(j,:)=doff_3;
    elseif n==3
        pEOon_1(j,:) =on_1;
        pEOon_2(j,:) =on_2;
        pEOoff_1(j,:)=off_1;
        pEOoff_2(j,:)=off_2;
        pEOoff_3(j,:)=off_3;
        %%% Max MVC %%%
        xEOon_1(j,:) =xon_1;
        xEOon_2(j,:) =xon_2;
        xEOoff_1(j,:)=xoff_1;
        xEOoff_2(j,:)=xoff_2;
        xEOoff_3(j,:)=xoff_3;
        %%%EMG durnation %%%
        EOdon_1(j,:) =don_1;
        EOdon_2(j,:) =don_2;
        EOdoff_1(j,:)=doff_1;
        EOdoff_2(j,:)=doff_2;
        EOdoff_3(j,:)=doff_3;

    elseif n==4
        pIOon_1(j,:) =on_1;
        pIOon_2(j,:) =on_2;
        pIOoff_1(j,:)=off_1;
        pIOoff_2(j,:)=off_2;
        pIOoff_3(j,:)=off_3;
        %%% Max MVC %%%
        xIOon_1(j,:) =xon_1;
        xIOon_2(j,:) =xon_2;
        xIOoff_1(j,:)=xoff_1;
        xIOoff_2(j,:)=xoff_2;
        xIOoff_3(j,:)=xoff_3;
        %%%EMG durnation %%%
        IOdon_1 (j,:)=don_1;
        IOdon_2 (j,:)=don_2;
        IOdoff_1(j,:)=doff_1;
        IOdoff_2(j,:)=doff_2;
        IOdoff_3(j,:)=doff_3;
    elseif n==6
        pLTon_1(j,:) =on_1;
        pLTon_2(j,:) =on_2;
        pLToff_1(j,:)=off_1;
        pLToff_2(j,:)=off_2;
        pLToff_3(j,:)=off_3;
        %%% Max MVC %%%
        xLTon_1(j,:) =xon_1;
        xLTon_2(j,:) =xon_2;
        xLToff_1(j,:)=xoff_1;
        xLToff_2(j,:)=xoff_2;
        xLToff_3(j,:)=xoff_3;
        %%%EMG durnation %%%
        LTdon_1 (j,:)=don_1;
        LTdon_2 (j,:)=don_2;
        LTdoff_1(j,:)=doff_1;
```

```
            LTdoff_2(j,:)=doff_2;
            LTdoff_3(j,:)=doff_3;
        elseif n==7
            pILon_1(j,:) =on_1;
            pILon_2(j,:) =on_2;
            pILoff_1(j,:)=off_1;
            pILoff_2(j,:)=off_2;
            pILoff_3(j,:)=off_3;
            %%% Max MVC %%%
            xILon_1(j,:) =xon_1;
            xILon_2(j,:) =xon_2;
            xILoff_1(j,:)=xoff_1;
            xILoff_2(j,:)=xoff_2;
            xILoff_3(j,:)=xoff_3;
            %%%EMG durnation %%%
            ILdon_1(j,:) =don_1;
            ILdon_2(j,:) =don_2;
            ILdoff_1(j,:)=doff_1;
            ILdoff_2(j,:)=doff_2;
            ILdoff_3(j,:)=doff_3;
        elseif n==8
            pMUon_1(j,:) =on_1;
            pMUon_2(j,:) =on_2;
            pMUoff_1(j,:)=off_1;
            pMUoff_2(j,:)=off_2;
            pMUoff_3(j,:)=off_3;
            %%% Max MVC %%%
            xMUon_1(j,:) =xon_1;
            xMUon_2(j,:) =xon_2;
            xMUoff_1(j,:)=xoff_1;
            xMUoff_2(j,:)=xoff_2;
            xMUoff_3(j,:)=xoff_3;
            %%%EMG durnation %%%
            MUdon_1(j,:) =don_1;
            MUdon_2(j,:) =don_2;
            MUdoff_1(j,:)=doff_1;
            MUdoff_2(j,:)=doff_2;
            MUdoff_3(j,:)=doff_3;
        end;
    end;
end;

%%%%%%%%%%averaged median EMG for each muscle during push phase%%%%%%%%%%%%
%%%%%%%%%%% push phase was divided by early push and late push stages %%%%%%%%%
avRAon_1=mean(pRAon_1(find(isnan(pRAon_1)==0)));
avRAon_2=mean(pRAon_2(find(isnan(pRAon_2)==0)));
avEOon_1=mean(pEOon_1(find(isnan(pEOon_1)==0)));
avEOon_2=mean(pEOon_2(find(isnan(pEOon_2)==0)));
avIOon_1=mean(pIOon_1(find(isnan(pIOon_1)==0)));
avIOon_2=mean(pIOon_2(find(isnan(pIOon_2)==0)));
avLTon_1=mean(pLTon_1(find(isnan(pLTon_1)==0)));
avLTon_2=mean(pLTon_2(find(isnan(pLTon_2)==0)));
avILon_1=mean(pILon_1(find(isnan(pILon_1)==0)));
avILon_2=mean(pILon_2(find(isnan(pILon_2)==0)));
avMUon_1=mean(pMUon_1(find(isnan(pMUon_1)==0)));
avMUon_2=mean(pMUon_2(find(isnan(pMUon_2)==0)));
%%%%%%%%%%averaged max EMG for each muscle during push phase%%%%%%%%%%%%
mxRAon_1=mean(nonzeros(xRAon_1)); mxRAon_2=mean(nonzeros(xRAon_2));
mxEOon_1=mean(nonzeros(xEOon_1)); mxEOon_2=mean(nonzeros(xEOon_2));
mxIOon_1=mean(nonzeros(xIOon_1)); mxIOon_2=mean(nonzeros(xIOon_2));
mxLTon_1=mean(nonzeros(xLTon_1)); mxLTon_2=mean(nonzeros(xLTon_2));
mxILon_1=mean(nonzeros(xILon_1)); mxILon_2=mean(nonzeros(xILon_2));
```

```
mxMUon_1=mean(nonzeros(xMUon_1)); mxMUon_2=mean(nonzeros(xMUon_2));
%%%%%%%%%%averaged median EMG for each muscle during recovery phase%%%%%%%%%%%%
%%%%%%%%%%recovery phase was divided by three stages %%%%%%%%%%%%%%%%%%%
avRAoff_1=mean(pRAoff_1(find(isnan(pRAoff_1)==0)));
avRAoff_2=mean(pRAoff_2(find(isnan(pRAoff_2)==0)));
avRAoff_3=mean(pRAoff_3(find(isnan(pRAoff_3)==0)));
avEOoff_1=mean(pEOoff_1(find(isnan(pEOoff_1)==0)));
avEOoff_2=mean(pEOoff_2(find(isnan(pEOoff_2)==0)));
avEOoff_3=mean(pEOoff_3(find(isnan(pEOoff_3)==0)));
avIOoff_1=mean(pIOoff_1(find(isnan(pIOoff_1)==0)));
avIOoff_2=mean(pIOoff_2(find(isnan(pIOoff_2)==0)));
avIOoff_3=mean(pIOoff_3(find(isnan(pIOoff_3)==0)));
avLToff_1=mean(pLToff_1(find(isnan(pLToff_1)==0)));
avLToff_2=mean(pLToff_2(find(isnan(pLToff_2)==0)));
avLToff_3=mean(pLToff_3(find(isnan(pLToff_3)==0)));
avILoff_1=mean(pILoff_1(find(isnan(pILoff_1)==0)));
avILoff_2=mean(pILoff_2(find(isnan(pILoff_2)==0)));
avILoff_3=mean(pILoff_3(find(isnan(pILoff_3)==0)));
avMUoff_1=mean(pMUoff_1(find(isnan(pMUoff_1)==0)));
avMUoff_2=mean(pMUoff_2(find(isnan(pMUoff_2)==0)));
avMUoff_3=mean(pMUoff_3(find(isnan(pMUoff_3)==0)));

mxRAoff_1=mean(nonzeros(xRAoff_1));mxRAoff_2=mean(nonzeros(xRAoff_2)); mxRAoff_3=mean(nonzeros(xRAoff_3));
mxEOoff_1=mean(nonzeros(xEOoff_1)); mxEOoff_2=mean(nonzeros(xEOoff_2)); mxEOoff_3=mean(nonzeros(xEOoff_3));
mxIOoff_1=mean(nonzeros(xIOoff_1)); mxIOoff_2=mean(nonzeros(xIOoff_2)); mxIOoff_3=mean(nonzeros(xIOoff_3));
mxLToff_1=mean(nonzeros(xLToff_1)); mxLToff_2=mean(nonzeros(xLToff_2)); mxLToff_3=mean(nonzeros(xLToff_3));
mxILoff_1=mean(nonzeros(xILoff_1)); mxILoff_2=mean(nonzeros(xILoff_2)); mxILoff_3=mean(nonzeros(xILoff_3));
mxMUoff_1=mean(nonzeros(xMUoff_1)); mxMUoff_2=mean(nonzeros(xMUoff_2));
mxMUoff_3=mean(nonzeros(xMUoff_3));

%%%Duration for each phase %%%%%%%%%%%%%%%%%%%%%%%%%%

avRAdon_1=mean(RAdon_1);avRAdon_2=mean(RAdon_2);avRAdoff_1=mean(RAdoff_1);avRAdoff_2=mean(RAdoff_2);avRAdoff_3=mean(RAdoff_3);

avEOdon_1=mean(EOdon_1);avEOdon_2=mean(EOdon_2);avEOdoff_1=mean(EOdoff_1);avEOdoff_2=mean(EOdoff_2);avEOdoff_3=mean(EOdoff_3);

avIOdon_1=mean(IOdon_1);avIOdon_2=mean(IOdon_2);avIOdoff_1=mean(IOdoff_1);avIOdoff_2=mean(IOdoff_2);avIOdoff_3=mean(IOdoff_3);

avLTdon_1=mean(LTdon_1);avLTdon_2=mean(LTdon_2);avLTdoff_1=mean(LTdoff_1);avLTdoff_2=mean(LTdoff_2);avLTdoff_3=mean(LTdoff_3);

avILdon_1=mean(ILdon_1);avILdon_2=mean(ILdon_2);avILdoff_1=mean(ILdoff_1);avILdoff_2=mean(ILdoff_2);avILdoff_3=mean(ILdoff_3);

avMUdon_1=mean(MUdon_1);avMUdon_2=mean(MUdon_2);avMUdoff_1=mean(MUdoff_1);avMUdoff_2=mean(MUdoff_2);avMUdoff_3=mean(MUdoff_3);

%%%%%%%%%%Trunk angle %%%%%%%%%%%%%%%%%%%%%%%%
avtrkon_1=mean(trkon_1); avtrkon_2=mean(trkon_2);
avtrkoff_1=mean(trkoff_1); avtrkoff_2=mean(trkoff_2);  avtrkoff_3=mean(trkoff_3);
%%%%%%%%%% Final output%%%%%%%%%%%%%%%%%%%%%%%%
emgphase=[ avtrkon_1 avtrkon_2 avtrkoff_1 avtrkoff_2 avtrkoff_3...
        avRAon_1 avRAon_2 avRAoff_1 avRAoff_2 avRAoff_3 ...
        avEOon_1 avEOon_2 avEOoff_1 avEOoff_2 avEOoff_3 ...
        avIOon_1 avIOon_2 avIOoff_1 avIOoff_2 avIOoff_3 ...
        avLTon_1 avLTon_2 avLToff_1 avLToff_2 avLToff_3 ...
        avILon_1 avILon_2 avILoff_1 avILoff_2 avILoff_3 ...
        avMUon_1 avMUon_2 avMUoff_1 avMUoff_2 avMUoff_3 ...
        avRAdon_1 avRAdon_2 avRAdoff_1 avRAdoff_2 avRAdoff_3...
```

```
            avEOdon_1 avEOdon_2 avEOdoff_1 avEOdoff_2 avEOdoff_3...
            avIOdon_1 avIOdon_2 avIOdoff_1 avIOdoff_2 avIOdoff_3...
            avLTdon_1 avLTdon_2 avLTdoff_1 avLTdoff_2 avLTdoff_3...
            avILdon_1 avILdon_2 avILdoff_1 avILdoff_2 avILdoff_3...
            avMUdon_1 avMUdon_2 avMUdoff_1 avMUdoff_2 avMUdoff_3...
            mxRAon_1 mxRAon_2 mxRAoff_1 mxRAoff_2 mxRAoff_3 ...
            mxEOon_1 mxEOon_2 mxEOoff_1 mxEOoff_2 mxEOoff_3 ...
            mxIOon_1 mxIOon_2 mxIOoff_1 mxIOoff_2 mxIOoff_3 ...
            mxLTon_1 mxLTon_2 mxLToff_1 mxLToff_2 mxLToff_3 ...
            mxILon_1 mxILon_2 mxILoff_1 mxILoff_2 mxILoff_3 ...
            mxMUon_1 mxMUon_2 mxMUoff_1 mxMUoff_2 mxMUoff_3 ];
  outcome(gm,:)=emgphase;
  emgphase=[];
end;
%%%%%%%%%%%%%% save the file as  *.txt %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
[FILENAME, PATHNAME] = uiputfile('*.txt', 'Save As');
   cd \
   eval(['cd ' PATHNAME]);
   fid1=fopen(FILENAME,'w');

   for i=1:gm
     fprintf(fid1,'%c%c%c%c \t',ID_matrix(i,:));

     for j_1=1:length(outcome)
       fprintf(fid1, '%f \t', outcome(i,j_1));
     end;

     fprintf(fid1,'\n');
  end;
  fclose(fid1);

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%% Trunk angle calculation %%%%%%%%%%%%%%%%%%%%%%%%%%

function [trkang] = trunkang(side, modata, subj_sepo)
%%%%%%%%%%%%%%%% load marker position%%%%%%%%%%%%%%%%%%%%%%%%%%
if side=='l'
asisx=modata(:,65);asisy=modata(:,66);asisz=modata(:,67);
psisx=modata(:,68);psisy=modata(:,69);psisz=modata(:,70);
acrox=modata(:,14);acroy=modata(:,15);acroz=modata(:,16);
elseif side=='r'
asisx=modata(:,71); asisy=modata(:,72); asisz=modata(:,73);
psisx=modata(:,74);psisy=modata(:,75);psisz=modata(:,76);
acrox=modata(:,41);acroy=modata(:,42);acroz=modata(:,43);
end;
acro=[acrox,acroy,acroz];
hip=[(asisx+psisx)/2, (asisy+psisy)/2, (asisz+psisz)/2];
%%%%%%%%%%%%%%%% trunk motion vector %%%%%%%%%%%%%%%%%%%%%%%%%%
trunk=acro(:,1:2)-hip(:,1:2);

modata=[];
modata=subj_sepo;

if side=='l'
asisx=modata(:,65);asisy=modata(:,66);asisz=modata(:,67);
psisx=modata(:,68);psisy=modata(:,69);psisz=modata(:,70);
acrox=modata(:,14);acroy=modata(:,15);acroz=modata(:,16);
elseif side=='r'
asisx=modata(:,71); asisy=modata(:,72); asisz=modata(:,73);
psisx=modata(:,74);psisy=modata(:,75);psisz=modata(:,76);
acrox=modata(:,41);acroy=modata(:,42);acroz=modata(:,43);
end;
```

```
acro=[acrox,acroy,acroz];
hip=[(asisx+psisx)/2, (asisy+psisy)/2, (asisz+psisz)/2];
sepo=acro(:,1:2)-hip(:,1:2);
%%%%%%%%%%%%%%%% trunk reference vector %%%%%%%%%%%%%%%%%%%%

for i=1:length(trunk)
   ang=acos(dot(sepo(i,:),trunk(i,:))/(norm(sepo(i,:))*norm(trunk(i,:))));
   if trunk(i,1)>sepo(i,1);
      theta(i)=ang;
   elseif trunk(i,1)<sepo(i,1);
       theta(i)=-ang;
   end;
end;
%%%%%%%%%%%%%%%% trunk angle %%%%%%%%%%%%%%%%%%%%%%%%%%%
trkang=theta*180/pi;


%%%%%%%%%%%% Determine five stages during a propulsion cycle based on Newsam study%%%%%%
%%%%%%%%%%%% Hand 3thMP was used to determine each stage %%%%%%%%%%%% %%%%%%

num_sub=input('Input number of subjects to run:','s');
num_sub=str2num(num_sub);
ID_matrix=[];

for i=1:num_sub
  ID = input('enter patient 4 digit ID [ex: p3b3]: ', 's');
  ID_matrix=[ID_matrix;ID];
end;

%%%%%%%%%%%%%%%Loading FM data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
[number,c]=size(ID_matrix);
for gm=1:number
   cd('S:\Protocols\Trunk Stimulation\DATA\Subject_data')
   rawID=ID_matrix(gm,:);
   subj_name=rawID(1:4);
   cd(rawID)
   cd('Clean_FM');
   if gm==1
      disp('Please select a FORCE DATA file.');
      dataname_force=uigetfile('*.*','Please select a FORCE DATA file.');
      sw=load(dataname_force);
      side=dataname_force(7);
      condition=dataname_force(6);
      speed=dataname_force(8);
         if length(sw)<4800
            sw(length(sw):4800,:)=0;
            swdata=sw;
         else
            swdata=sw(1:4800,:);
         end;
   else
      sw=load([subj_name ,'w',condition,side,speed]);
         if length(sw)<4800
            sw(length(sw):4800,:)=0;
            swdata=sw;
         else
            swdata=sw(1:4800,:);
         end;
   end;

%%%%%%%%%%%%%%Loading MO data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
```

```
    cd ..
    cd('Clean_MO\SW');
    modata=load([subj_name ,'m',condition,'b',speed]);
    cd('S:\Students\Yusheng\Trunk_Stim_Project\Data_analysis\control');
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%change the sampling rate %%%%%%%%%%%%
[raw,col]=size(swdata);
    for n=1:col
        F1=swdata(:,n);
        F1new(n,:)= spline(1:4800,F1,1:(4800/1200):4800);
    end;
    newsw=F1new';
    mz=newsw(:,6);
    fy=newsw(:,2);

    [raw,col]=size(modata);
    for n=1:col
        F2=modata(:,n);
        F2new(n,:)= spline(1:1200,F2,1:(1200/1200):1200);
    end;
    newmodata=F2new';

    if side=='l'
        latx=newmodata(:,17);laty=newmodata(:,18);latz=newmodata(:,19);
        lat=[latx,laty,latz];
        thirdmpx=newmodata(:,29);thirdmpy=newmodata(:,30);thirdmpz=newmodata(:,31);
        thirdmp=[thirdmpx,thirdmpy,thirdmpz];
    elseif side=='r'
        latx=newmodata(:,44);laty=newmodata(:,45);latz=newmodata(:,46);
        lat=[latx,laty,latz];
        thirdmpx=newmodata(:,56);thirdmpy=newmodata(:,57);thirdmpz=newmodata(:,58);
        thirdmp=[thirdmpx,thirdmpy,thirdmpz];
    end;


    mxnum=max(swdata(:,7));
    if mxnum<10
        mxstroke=mxnum-1;
    else
        mxstroke=10;
    end;
    for j=1:mxstroke;          %%pick up for 10 strokes%%
        on=min(find(newsw(:,7)==j));
        off=max(find(newsw(:,7)==j));
        strokend=max(find(newsw(:,7)==-j));
        timediff(j)=strokend-on;
        ontime(j)=((off-on)/timediff(j)); %percentage of ontime
        qoff=off-round(length(on:off)/4);
        pkpton=max(find(diff(thirdmpy(on:qoff))>0))+on; %%%pkpton == peak point during push phase%%
        pkptoff=find(thirdmpx==max(thirdmpx(off:strokend))); %% pkptoff == peak point of MP3X during recovery phase%%
        lwptoff=find(thirdmpx==min(thirdmpx(off:strokend))); %% lwptoff == lowest point of MP3X during recovery phase%%
        if pkpton-on>0;
            onph_1(j)=(pkpton-on)/(strokend-on);
            onph_2(j)=(off-pkpton)/(strokend-on);
        else
            onph_1(j)=0;
            onph_2(j)=(off-on)/(strokend-on);
        end;

        if pkptoff-off>0;
            offph_1(j)= (pkptoff-off)/(strokend-on);
        else
            offph_1(j)=0;
```

```
        end;

        if strokend-lwptoff>0;
            offph_3(j)= (strokend-lwptoff)/(strokend-on);
        else
            offph_3(j)=0;
        end;

            offph_2(j)=(lwptoff-pkptoff)/(strokend-on);
    end;
    phase=[onph_1' onph_2' offph_1' offph_2' offph_3']  %%%propulsion V phase %%%

    ontimep(gm,:)=mean(phase);
end;
```

APPENDIX C: SAS PROGRAMS FOR STUDY #1

```
proc print data=MVC;
run;
PROC RANK data=MVC;
   BY speed;
   VAR avMVC;
   RANKS ravMVC;
   RUN;
 PROC PRINT;
   TITLE2 'ORIGINAL AND RANKED VALUES OF YIELD';
   RUN;
 PROC ANOVA; CLASSES phase speed;
   MODEL ravMVC = phase speed;
   TITLE2 'FRIEDMAN''S TWO-WAY NON-PARAMETRIC ANOVA';
   RUN;
proc print data=MVC;
run;
PROC RANK data=MVC;
   BY phase;
   VAR avMVC;
   RANKS ravMVC;
   RUN;
 PROC PRINT;
   TITLE2 'ORIGINAL AND RANKED VALUES OF YIELD';
   RUN;
 PROC ANOVA; CLASSES phase speed;
   MODEL ravMVC = phase speed;
   TITLE2 'FRIEDMAN''S TWO-WAY NON-PARAMETRIC ANOVA';
   RUN;
```

APPENDIX D: MATLAB PROGRAMS FOR STUDY #2

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
function fescase
   ID = input('enter patient 4 digit ID [ex: p3b3]: ', 's');
for nnn=1:2
   %load data
   cd('S:\Protocols\Trunk Stimulation\DATA\Subject_data')
   cd(ID);

   cd('Clean_FM');

   if nnn==1
      condition = input('enter stim condition [ex: 1; 3; 5]: ', 's');
      speed = '3';
      subj_name=ID;
      side='r'
      sw=load([subj_name ,'w',condition,'r3', speed]);
   elseif nnn==2
      sw=load([subj_name ,'w',condition,'l3' speed]);
      side='l'
   end;
   cd ..
   cd('Clean_MO');
   modata=load([subj_name ,'m',condition,'b3', speed]);

%%%%%%%%%%%%%%%Loading sepo motion data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
   sepo_name=[subj_name,'msp3'];
   subj_sepo=load(sepo_name);
%%%%%%%%%%%%%%%Loading SW data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
   cd('S:\Students\Yusheng\Trunk_Stim_Project\Data_analysis\case');

%%%%%%%%%%%%%%%check sholder markers%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

plot(modata(:,41),'r');
title('r--right side marker')
hold;
plot(modata(:,14),'b');
plot(modata(:,8),'k');
side2 = input('enter which side markers is sin curve [ex: r/l]: ', 's');
if side2=='l';
   side=='l';
else
   side=='r';
end;
close all;
%%%%%%%%%%%%%%%%%%%%%%%%%%%change SW sampling 240Hz to 60Hz%%%%%%%%%%%%%%%%%%

   if length(sw)<7200
       sw(length(sw):7200,:)=0;
       whl=sw;
   else
       whl=sw(1:7200,:);
   end;

   swout=[];
   F2=whl;
   for m=1:8
      F=spline(1:7200,F2(:,m),1:(7200/1800):7200);
```

```
        swout=[swout; F];
    end;
    swout=swout';
%%%%%%%%%%%%%%%%%%%%Convert to Linear Velocity %%%%%%%%%%%%%%%%%%%%%%%%%%%%%

[velout] = velrim(side, whl); %% velout=[Fx Fy Fz Mx My Mz onoff vel power]%%

%%%%%%%%%%%%%%%%% Cadence/Velocity %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%cadence--stroke/s, vel--m/s
data_force=whl;
rawvel=velout(:,8);
power=-velout(:,9);
[rows cols]=size(data_force);

j=1;
k=1;
for i=1:rows
    if i-1==0
        %do nothing
    elseif data_force(i,7)>data_force(i-1,7) %step on
        step(j,1)=i;
        j=j+1;
    elseif data_force(i,7)<data_force(i-1,7) %step off
        step(k,2)=i-1;
        k=k+1;
    end %end if data_force
end %end for i
%%start push --step: column#1  end of push---step: column#2
%building matrix of hand coordinates at on/off time locations
[orows ocols]=size(step);

for i=1:orows-1 %for each stroke...
    stroke_interval=(step(i+1,1)-step(i,1));
    push_interval=(step(i,2)-step(i,1));
    pushtime(i)=push_interval*(1/240);
    totaltime(i)=stroke_interval*(1/240);
    pushpercent(i)=push_interval/stroke_interval; %percentage of push phase%%
    cadence(i)=1/(stroke_interval*(1/240));  %%sample rate 240 Hz, stroke/persecond%%
    avrimvel(i)=mean(rawvel(step(i,1):step(i+1,1))); %%start push to the next push%%
    maxrimvel(i)=max(rawvel(step(i,1):step(i+1,1))); %%start push to the next push%%
    minrimvel(i)=min(rawvel(step(i,1):step(i+1,1))); %%start push to the next push%%
    maxp(i)=max(power(step(i,1):step(i,2)));
    meanp(i)=mean(power(step(i,1):step(i,2)));
    work(i)=abs(sum(power(step(i,1):step(i,2))))*1/240;
end

%%Note!! Following kinetic variables was based on 60Hz sample rate%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%% Start/Stop/Contact angle %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

[startang,stopang,contang]=pushang(modata, swout, side);

%%%%%%%%%%%%%%%%%%%%%% kinetic variable %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
      Fx=swout(:,1);
      Fy=swout(:,2);
      Fz=swout(:,3);
      Mx=swout(:,4);
      My=swout(:,5);
      Mz=swout(:,6);
      onoff=swout(:,7);

      FR=sqrt(Fx.^2 + Fy.^2 + Fz.^2);
      r=0.2667; %%radian of pushrim
```

```
    Ft=Mz./r;
    Fr=sqrt(abs(FR.^2 - Ft.^2 - Fz.^2));
    fef=Ft.^2./FR.^2;

    hubfm=[FR Ft Fr Fz Mx My Mz onoff fef];

%%%%%%%%%%%%%%%%%%force calculation%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
      [output]=kinoutput(hubfm);
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
          meanfhub=output(1:4,:); %%mean of FR, ft, fr, fz%%
          maxfhub=output(5:8,:); %%max of FR, ft, fr, fz %%
          mzhub=output(9:10,:);%%mean Mz and Max Mz %%
          rorhub=output(11,:);
          avfefhub=output(12,:);
          mphub=output(13,:);


%%%%%%%%%%%%%%%%%%%%% kinematic variable (wrist-elbow flex/extension) %%%%%%%%%%%%%%%%%%
[wristROM, elROM] = WR_ROM(side, modata, subj_sepo);
%%%%%%%%%%%%%%%%%%%%% kinematic variable (shoulder flex/extension) %%%%%%%%%%%%%%%%%%%%
[shoulderout]= shoulderang(side, modata, subj_sepo);
%%%%%%%%%%%%%%%%%%%%% kinematic variable (trunk flex/extension) %%%%%%%%%%%%%%%%%%%%%%
[trkang, deacrox] = trunkang(side, modata, subj_sepo);

[b,a]=butter(2,5/30);
[raws,columns]=size(trkang);
filttrkang=filtfilt(b,a,trkang);
filtdeacrox=filtfilt(b,a,deacrox);
trkang=filttrkang;
deacrox=filtdeacrox;

onoff=swout(:,7);
  for j=1:max(onoff)-1;
    start=min(find(onoff==j));
    endstroke=max(find(onoff==j));
    endcycle=min(find(onoff==j+1));
    midpt=round(start+(endstroke-start)/2);
    wr_ext(j)=min(wristROM(start:endstroke,1)); %%extenison(-)
    wr_flx(j)=max(wristROM(start:endstroke,1));%%flexion(+)

    wr_uln(j)=min(wristROM(start:endstroke,2)); %%uln(-)
    wr_rad(j)=max(wristROM(start:endstroke,2));%%rad(+)

    el_min(j)=min(elROM(start:endstroke));  %%180 degrees means elbow natural position%%
    el_max(j)=max(elROM(start:endstroke));

    sh_flx(j)=max(shoulderout(start:endstroke,1));
    sh_ext(j)=min(shoulderout(start:endstroke,1));

    sh_abd(j)=max(shoulderout(start:endstroke,2));
    sh_add(j)=min(shoulderout(start:endstroke,2));

    tk_flx(j)=max(trkang(start:endcycle));
    tk_ext(j)=min(trkang(start:endcycle));
    tk_on(j)=mean(trkang(start:endstroke));
    trkdiff=diff(trkang(start:midpt));
    trkdiff2=diff(trkang(midpt+1:endstroke));
    tk_paradx(j)=sum(trkdiff(find(trkdiff<0)));     %%%trunk paradx moment at the beginning of stroke%%%
    tk_paradx2(j)=sum(trkdiff2(find(trkdiff2<0)));  %%%trunk paradx moment at the late phase of stroke%%%

    acrodiff=diff(deacrox(start:midpt));
    acrodiff2=diff(deacrox(midpt+1:endstroke));
```

```
      acro_paradx(j)=sum(acrodiff(find(acrodiff<0))); %%% beginning of stroke%%%
      acro_paradx2(j)=sum(acrodiff2(find(acrodiff2<0)));
  end;
  wr_flxROM=wr_flx-wr_ext;
  wr_radROM=wr_rad-wr_uln;
  el_flxROM=el_max-el_min;
  sh_flxROM=sh_flx-sh_ext;
  sh_abdROM=sh_abd-sh_add;
  tk_flxROM=tk_flx-tk_ext;
  tk_paradx3=tk_paradx+tk_paradx2;          %%%trunk paradx moment during the entired stroke%%%
  acro_paradx3=acro_paradx+acro_paradx2;

%%%%%%%%%%%%%%%%%%%% Save data for 10 strokes %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

[rows cols]=size(avrimvel);
if cols>20;
   col=11:20;
else
   col=4:13;
end;

      finalout=[startang(:,col); stopang(:,col); contang(:,col); pushtime(:,col);...
             totaltime(:,col); pushpercent(:,col); cadence(:,col); avrimvel(:,col); ...
             maxrimvel(:,col); minrimvel(:,col); ...
             avfefhub(:,col); meanfhub(2:4,col); maxfhub(2:4,col);...
             maxp(:,col); meanp(:,col); work(:,col);...
             abs(wr_ext(col)); wr_flx(col); wr_flxROM(col); abs(wr_uln(col)); wr_rad(col); wr_radROM(col); ...
             el_min(col); el_max(col); el_flxROM(col); sh_flx(col); abs(sh_ext(col)); sh_flxROM(col);...
             sh_abd(col); sh_add(col); sh_abdROM(col); tk_flx(col); tk_ext(col); tk_flxROM(col); tk_on(col); tk_paradx(col);
tk_paradx2(col); tk_paradx3(col); acro_paradx(col); acro_paradx2(col); acro_paradx3(col); col];  %63 variables%%

%%%%%%%%%%%%%% plot(PFA forces & Mz(SW)%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

     figure(4)
     plot(Mz)
     plottitle=['Mz(SW) for : ',subj_name];
     TITLE(plottitle);

     figure(5)
     plot(trkang,'k')
     hold;
     plot(swout(:,7),'b');
     TITLE('TRUNK ANGLE');
     pause;

     close all;
end;

%%%%%%%%%%%%%% Velocity calculation & Propulsion power %%%%%%%%%%%%%%%%%%%%%%%%%%

function [velout] = velrim(side, whl)
r=0.3048; %%tire radius%%
   if side=='r'
      rwhl=whl;
      opti=rwhl(:,8);
      onoff=rwhl(:,7);
   else
      lwhl=whl;
      opti=lwhl(:,8);
      onoff=lwhl(:,7);
   end;
```

```
%ploto = opti;
dop=[];
dop=diff(opti);
count=1;
badpt=[];
badpt=find(abs(dop)>5);

x=0;

while count > 0
   for i=1:length(badpt)
      if badpt(i)==1;
         dop(badpt(i))=dop(badpt(i)+count);
      elseif badpt(i)==2;
         dop(badpt(i))=dop(badpt(i)-1);
      elseif badpt(i)==length(dop);
         dop(badpt(i))=dop(badpt(i)-count);
      elseif badpt(i)==length(dop)-1;
         dop(badpt(i))=dop(badpt(i)-count);
      else
         dop(badpt(i))=(dop(badpt(i)-count)+dop(badpt(i)+count))/2;
      end;
   end;

   badpt=find(abs(dop)>5);
   if isempty(badpt)==1;
      count=0;
   else
      count=count+1;
   end;
end;

lastpt=dop(length(dop));
dop=[dop;lastpt];

onrim=find(onoff~=0);
offrim=find(onoff==0);
off1 = length(find(offrim < length(onoff)/2));
off2 = length(find(offrim > length(onoff)/2));
off1 = zeros(1,off1);
off2 = zeros(1,off2);

meanv=[];
for i=1:length(onrim)

   meanv(i)=abs(mean(dop((onrim(i)-30):(onrim(i)+30))));
end;
meanv=meanv*(pi/180);
time=1/240;
angvel=meanv/time;
vel=(meanv/time)*r;
vel=[off1 vel off2];
angvel=[off1 angvel off2];

if side=='r'
   Fx=rwhl(:,1);
   Fy=rwhl(:,2);
   Fz=rwhl(:,3);
   Mx=rwhl(:,4);
   My=rwhl(:,5);
   Mz=rwhl(:,6);
   powerout=Mz'.*angvel;
```

```
    vel=vel';
    velout=[Fx Fy Fz Mx My Mz onoff vel powerout'];
  else
    Fx=lwhl(:,1);
    Fy=lwhl(:,2);
    Fz=lwhl(:,3);
    Mx=lwhl(:,4);
    My=lwhl(:,5);
    Mz=lwhl(:,6);
    powerout=Mz'.*angvel;
    vel=vel';
    velout=[Fx Fy Fz Mx My Mz onoff vel powerout'];
 end;

%%Push angle, Contact angle, and End angle Calculation %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%This m-file is only based on Kinematic data to calculate these three angle.  %%%%%%%%%%%%%%%%%%%%%%
%%The second method is to use encoder angle to do these.  please refer to Biocal m-file%%
%%Modified by Yusheng 08/29 2004%%
function [startang,stopang,contang]=pushang(modata, swout, side);

 if side=='r'
    thirdmpx=modata(:,56);thirdmpy=modata(:,57);thirdmpz=modata(:,58);
    thirdmp=[thirdmpx,thirdmpy,thirdmpz];
    imatrix=ones(length(modata),1);
    hubx=imatrix*mean(modata(:,59));
    huby=imatrix*mean(modata(:,60));
    hubz=imatrix*mean(modata(:,61));
    hub=[hubx,huby,hubz];
 elseif side=='l'
    thirdmpx=modata(:,29);thirdmpy=modata(:,30);thirdmpz=modata(:,31);
    thirdmp=[thirdmpx,thirdmpy,thirdmpz];
    imatrix=ones(length(modata),1);
    hubx=imatrix*mean(modata(:,32));
    huby=imatrix*mean(modata(:,33));
    hubz=imatrix*mean(modata(:,34));
    hub=[hubx,huby,hubz];
 end;

onoff=swout(:,7);
rstart=[];
rstop=[];
lstart=[];
lstop=[];
  for j=1:max(onoff);

    start=min(find(onoff==j));
    endstroke=max(find(onoff==j));

    thirdmpx_start=thirdmpx(start);
    thirdmpy_start=thirdmpy(start);
    thirdmpx_stop=thirdmpx(endstroke);
    thirdmpy_stop=thirdmpy(endstroke);
    xhub=mean(hubx); yhub=mean(huby);

    start_theta=atan(abs(thirdmpx_start-xhub)/abs(thirdmpy_start-yhub));
    stop_theta=atan(abs(thirdmpx_stop-xhub)/abs(thirdmpy_stop-yhub));

    %% The 0 degree reference point is 3 oclock%%
   if  thirdmpx_start-hubx<0;
      start_theta=90+start_theta*(180/pi);
   else
      start_theta=90-start_theta*(180/pi);
```

```
    end

    if  thirdmpx_stop-hubx<0;
        stop_theta=90+stop_theta*(180/pi);
    else
        stop_theta=90-stop_theta*(180/pi);
    end;

    contang(j)=start_theta-stop_theta;
    startang(j)=start_theta;
    stopang(j)=stop_theta;
 end;

strt_stp_ang=[startang,stopang,contang];

%%%%%%%%%%%%%%%%%%%%% Kinetic variables caculation %%%%%%%%%%%%%%%%%%%%%%%%%%%
function [output]=kinoutput(final)

     fm=final;
     onoff=fm(:,8);
     fR=fm(:,1);
     ft=-(fm(:,2));
     fr=fm(:,3);
     fz=abs(fm(:,4));
     mx=fm(:,5);
     my=fm(:,6);
     mz=-fm(:,7);
     RORF=diff(fR)/(1/60);
     RORFt=diff(ft)/(1/60);
     RORFr=diff(fr)/(1/60);
     fef=fm(:,9);

   for j=1:max(onoff); %pick up all strokes %%
         on=min(find(onoff==j));
         off=max(find(onoff==j));
         point=round((off-on)*5/100);
         midpt=round(median(on:off));
         %% pick up the 90% of phuse phase for fef%%
         newon=on+point;
         newoff=off-point;
         allavfef(j)=mean(fef(newon:newoff));
         raw_mp=sqrt(mx(newon:newoff).^2+my(newon:newoff).^2);
         allmaxmp(j)=max(raw_mp);

         %% pick up the 50% of phuse phase for ror%%
         allmaxROR(j)=max(RORF(on:midpt));
         %maxRORFt=max(RORFt(on:midpt));
         %maxRORFr=max(RORFr(on:midpt));

         %% Total push forces %%
         allfR(j)=mean(fR(on:off));
         allft(j)=mean(ft(on:off));
         allfr(j)=mean(fr(on:off));
         allfz(j)=mean(fz(on:off));
         allmz(j)=mean(mz(on:off));

         allmaxfR(j)=max(fR(on:off));
         allmaxft(j)=max(ft(on:off));
         allmaxfr(j)=max(fr(on:off));
         allmaxfz(j)=max(fz(on:off));
         allmaxmz(j)=max(mz(on:off));
     end;
```

```matlab
    output=[allfR; allft; allfr; allfz; allmaxfR; allmaxft; allmaxfr; allmaxfz; allmz; allmaxmz; allmaxROR; allavfef; allmaxmp];

%%%%%%%%%%%%%%%%%%%%%% Trunk angles caculation %%%%%%%%%%%%%%%%%%%%%%%%%

function [trkang, deacrox] = trunkang(side, modata, subj_sepo)

if side=='l'
acrox=modata(:,14);acroy=modata(:,15);acroz=modata(:,16);
imatrix=ones(length(modata),1);
sepohipx=imatrix*mean(subj_sepo(:,35));
sepohipy=imatrix*mean(subj_sepo(:,36));
sepohipz=imatrix*mean(subj_sepo(:,37));

hubx=imatrix*mean(modata(:,32));
huby=imatrix*mean(modata(:,33));
hubz=imatrix*mean(modata(:,34));
elseif side=='r'
acrox=modata(:,41);acroy=modata(:,42);acroz=modata(:,43);
imatrix=ones(length(modata),1);
sepohipx=imatrix*mean(subj_sepo(:,62));
sepohipy=imatrix*mean(subj_sepo(:,63));
sepohipz=imatrix*mean(subj_sepo(:,64));

hubx=imatrix*mean(modata(:,59));
huby=imatrix*mean(modata(:,60));
hubz=imatrix*mean(modata(:,61));
hub=[hubx,huby,hubz];%%centerhub%%
end;

referenced_hipx = sepohipx;
referenced_hipy = sepohipy;
referenced_hip=[referenced_hipx, referenced_hipy];

acro=[acrox,acroy,acroz];
moacrox=acrox;
trunk=acro(:,1:2)-referenced_hip(:,1:2);

modata=[];
modata=subj_sepo;

if side=='l'
acrox=imatrix*mean(modata(:,14));
acroy=imatrix*mean(modata(:,15));
acroz=imatrix*mean(modata(:,16));
elseif side=='r'
acrox=imatrix*mean(modata(:,41));
acroy=imatrix*mean(modata(:,42));
acroz=imatrix*mean(modata(:,43));

end;
acro=[acrox,acroy,acroz];
sepoacrox=acrox;
sepo=acro(:,1:2)-referenced_hip(:,1:2);


for i=1:length(trunk)
  ang=acos(dot(sepo(i,:),trunk(i,:))/(norm(sepo(i,:))*norm(trunk(i,:))));
  if trunk(i,1)>sepo(i,1);
    theta(i)=ang;
  elseif trunk(i,1)<sepo(i,1);
     theta(i)=-ang;
```

```
    end;
  end;

deacrox=moacrox-sepoacrox;
trkang=theta*180/pi;

%%%%%%%%%%%%%%%%%%%%%% Upper limb angles caculation %%%%%%%%%%%%%%%%%%%%%%%%%%

function [wristROM, elROM] = WR_ROM(side, modata, subj_sepo);

 kinem_spline=modata;
[r,c]=size(kinem_spline);
[b,a]=butter(2,7/30);  % 4th order butterworth with  7 Hz frequency cutoff
FilterATD=[];

    for i=1:c;
      AtdArray=filtfilt(b,a,kinem_spline(:,i));
      FilterATD=[FilterATD,AtdArray];
    end;

kin=(FilterATD);

[flexangle_mean_r,flexangle_mean_l,radangle_mean_r,radangle_mean_l,proangle_mean_r,proangle_mean_l]=wristangle_setp1(
side,subj_sepo);

   if side=='r'
    thirdmpx=kin(:,56);thirdmpy=kin(:,57);thirdmpz=kin(:,58);
    thirdmp=[thirdmpx,thirdmpy,thirdmpz];
     radx=kin(:,53);rady=kin(:,54);radz=kin(:,55);
    rad=[radx,rady,radz];
    ulnx=kin(:,50);ulny=kin(:,51);ulnz=kin(:,52);
    uln=[ulnx,ulny,ulnz];
    olecx=kin(:,47);olecy=kin(:,48);olecz=kin(:,49);
    olec=[olecx,olecy,olecz];
    latx=kin(:,44);laty=kin(:,45);latz=kin(:,46);
    lat=[latx,laty,latz];
    acrox=kin(:,41);acroy=kin(:,42);acroz=kin(:,43);
    acro=[acrox,acroy,acroz];

    else
    thirdmpx=kin(:,29);thirdmpy=kin(:,30);thirdmpz=kin(:,31);
    thirdmp=[thirdmpx,thirdmpy,thirdmpz];
    radx=kin(:,26);rady=kin(:,27);radz=kin(:,28);
    rad=[radx,rady,radz];
    ulnx=kin(:,23);ulny=kin(:,24);ulnz=kin(:,25);
    uln=[ulnx,ulny,ulnz];
    olecx=kin(:,20);olecy=kin(:,21);olecz=kin(:,22);
    olec=[olecx,olecy,olecz];
    latx=kin(:,17);laty=kin(:,18);latz=kin(:,19);
    lat=[latx,laty,latz];
    acrox=kin(:,14);acroy=kin(:,15);acroz=kin(:,16);
    acro=[acrox,acroy,acroz];

  end;

  flexangle=[]; n=[];endstroke=[]; abc=[]; deviation=[]; radangle=[]; proangle=[];k=[];time=[];
  v_acro_lat_raw=[]; v_acro_lat_convert=[]; EL_flexangle=[];
  dt=1/60; %sampling rate---Kis 60 Hz
for n=1:1:length(kin)
    vector12=(rad([n],:)-uln([n],:));   %%direction ulnar--->rad%%
    Unit_vector12=norm(vector12,2);
    Vhand_i=vector12/Unit_vector12;
```

```
midpwc=(rad([n],:)+uln([n],:))./2;
vector3midpwc=(thirdmp([n],:)-midpwc);   %%direction wrist-midpoint--->thirdmp%
Unit_vector3midpwc=norm(vector3midpwc,2);
Vhand_j=vector3midpwc/Unit_vector3midpwc;  %%the first raw vector in j direction%

vectork=cross(Vhand_i,Vhand_j);
Unit_vectork=norm(vectork,2);
Vhand_k=vectork/Unit_vectork;
%%Vhand_j=cross(Vhand_k,Vhand_i); %% =the new vector in j direction
Vhand_i=cross(Vhand_j,Vhand_k);


vector52=(uln([n],:)-lat([n],:));
R=[Vhand_i; Vhand_j; Vhand_k];
HV52=R*vector52';
Unite_HV52=norm(HV52);
abc=HV52/Unite_HV52;
beta=atan2(-abc(3),abc(2));
flexangle=[flexangle,beta];
%%positive indicate the flexion, negative indicate the extension %%

thetar=atan2(abc(1),abc(2)); %% b==y(face side), a==x(close side)%%
radangle=[radangle,thetar];

vector62=(acro([n],:)-lat([n],:));
sup_vector=cross(vector52,vector62);
Unite_vector62=norm(sup_vector);
Unite=sup_vector/Unite_vector62;

abc_sup=R*Unite';
delta=atan2(abc_sup(1),-abc_sup(3));
proangle=[proangle,delta];

%%%%%% Elbow flex/extenion angle%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
 v_lat_acro=(acro([n],:)-lat([n],:));
 Unit_v_lat_acro=norm(v_lat_acro);

 v_lat_uln=(uln([n],:)-lat([n],:));
 Unit_v_lat_uln=norm(v_lat_uln);

 EL_thetar_raw=dot(v_lat_acro,v_lat_uln)/(Unit_v_lat_acro*Unit_v_lat_uln);
 EL_thetar=acos(EL_thetar_raw);
 EL_flexangle=[EL_flexangle, EL_thetar];

end;

 if side=='r'
     flexangle_r=flexangle-flexangle_mean_r; %substract setpo angles
     radangle_r=radangle-radangle_mean_r;
     proangle_r=proangle-proangle_mean_r;
     flexangle_l=[];
     radangle_l=[];
     proangle_l=[];
     wristROM=[flexangle_r' radangle_r' proangle_r']*180/pi;
     elROM=[EL_flexangle]*180/pi;

 else
     flexangle_l=(-flexangle)-flexangle_mean_l;
     radangle_l=radangle-radangle_mean_l;
     proangle_l=(-proangle)-proangle_mean_l;
     flexangle_r=[];
```

```
        radangle_r=[];
        proangle_r=[];
        wristROM=[flexangle_l' radangle_l' proangle_l']*180/pi;
        elROM=[EL_flexangle]*180/pi;
    end;
%% positive indicate the flexion, negative indicate the extension %%
%% positive indicate the radial devication
%% positive indicate the pronation
```

APPENDIX F: SAS PROGRAMS FOR STUDY #2

```
proc print data=fes;
run;
proc mixed data=fes;
    class ID stage condition;
    model maxFtotalpfa = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model maxftothub = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model maxfrhub = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model maxfzhub = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model mzhub = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model maxmzhub = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model avfefhub = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID condition stage ;
    model meanpower = condition|stage;
    repeated condition / subject=ID type=cs;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model work = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model el_ROM = stage|condition;
```

```
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model el_min = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model el_max = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model sh_flxROM = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model sh_ext = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model shabdROM = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model maxwrflx = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model maxwrext = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model wrflxROM = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model tk_rom = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model tk_avon = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
```

```
run;
proc mixed data=fes;
     class ID stage condition;
     model acroparadox1 = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model sumacroparadox = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model startang = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model stopang = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model contang = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model pusttime = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model cadence = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model rimvel = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model mathpwer = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model GME = stage|condition;
     repeated stage condition / subject=ID type=un@un;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
```

```
    class ID stage condition;
    model VO2 = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model VCO2 = stage|condition;
    repeated stage condition / subject=ID type=un@un;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
```

APPENDIX G: MATLAB PROGRAMS FOR STUDY #3

```
%%%%function FESEMG3 %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
num_sub=input('enter number of subject: ', 's');
num_sub=str2num(num_sub);
ID_matrix=[];

for i=1:num_sub
  ID = input('enter patient 4 digit ID [ex: p3b3]: ', 's');
  ID_matrix=[ID_matrix;ID];
end;

[number,c]=size(ID_matrix);
%%%%%%%%%%%%%%%%%%%%%%load individual data %%%%%%%%%%%%%%%%%%%%%%%%%%%%%

for gm=1:number
   cd('S:\Protocols\Trunk Stimulation\DATA\Subject_data')
   rawID=ID_matrix(gm,:);
   subj_name=rawID(1:4);
   cd(subj_name)
   %%%%%%%%%%%%%%%%%Loading SW data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

   cd('Clean_FM');
   if gm==1
      disp('Please select a FORCE DATA file.');
      dataname_force=uigetfile('*.*','Please select a FORCE DATA file.');
      sw=load(dataname_force);
      side=dataname_force(7);
      condition=dataname_force(6);
      if length(dataname_force)>8;
         speed=dataname_force(8:10);
      else
         speed=dataname_force(8);
      end;

         if length(sw)<7200
            sw(length(sw):7200,:)=0;
            swdata=sw;
         else
            swdata=sw(1:7200,:);
         end;
   else
      sw=load([subj_name ,'w',condition,side,speed]);
         if length(sw)<7200
            sw(length(sw):7200,:)=0;
            swdata=sw;
         else
            swdata=sw(1:7200,:);
         end;
   end;


%%%%%%%%%%%%%%%%%Loading MO data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%

   cd ..
   cd('Clean_MO');
   modata=load([subj_name ,'m',condition,'b',speed]);

%%%%%%%%%%%%%%%%%Loading sepo motion data%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
   sepo_name=[subj_name,'msp1'];
   subj_sepo=load(sepo_name);
```

```
cd ..

cd('Clean_EMG')
fname=[subj_name, 'e',condition,'b',speed];
emgdata=load([fname]);
emgdata=emgdata(1:45000,:);

referemg=load([subj_name, 'e',condition,'b3']);

cd('S:\Students\Yusheng\Trunk_Stim_Project\Data_analysis\case\emgcase');

if condition==3;
   emgdata=emgdata;
else
   plot(referemg(1:4000,2));
   [x,y]=ginput(2);
   emgdata(:,2)=emgdata(:,2)-mean(referemg(x(1):x(2),2));
   emgdata(:,8)=emgdata(:,8)-mean(referemg(x(1):x(2),8));
   emgdata(:,9)=emgdata(:,9)-mean(referemg(x(1):x(2),9));
end;

%%%%%%%%%%%%%%%%%%%%Convert to Linear Velocity %%%%%%%%%%%%%%%%%%%%%%%%

[velout] = velrim(side, swdata); %% velout=[Fx Fy Fz Mx My Mz onoff vel power]%%

%%%%%%%%%%%%%%%%%%%%% Cadence/Velocity %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%cadence--stroke/s, vel--m/s
data_force=swdata;
rawvel=velout(:,8);
power=-velout(:,9);
[rows cols]=size(data_force);

j=1;
k=1;
for i=1:rows
   if i-1==0
      %do nothing
   elseif data_force(i,7)>data_force(i-1,7) %step on
      step(j,1)=i;
      j=j+1;
   elseif data_force(i,7)<data_force(i-1,7) %step off
      step(k,2)=i-1;
      k=k+1;
   end %end if data_force
end %end for i
%%start push --step: column#1  end of push---step: column#2
%building matrix of hand coordinates at on/off time locations
[orows ocols]=size(step);

for i=1:orows-1 %for each stroke...
   stroke_interval=(step(i+1,1)-step(i,1));
   push_interval=(step(i,2)-step(i,1));
   pushtime(i)=push_interval*(1/240);
   totaltime(i)=stroke_interval*(1/240);
   pushpercent(i)=push_interval/stroke_interval; %percentage of push phase%%
   cadence(i)=1/(stroke_interval*(1/240));  %%sample rate 240 Hz, stroke/persecond%%
   avrimvel(i)=mean(rawvel(step(i,1):step(i+1,1))); %%start push to the next push%%
   maxrimvel(i)=max(rawvel(step(i,1):step(i+1,1))); %%start push to the next push%%
   minrimvel(i)=min(rawvel(step(i,1):step(i+1,1))); %%start push to the next push%%
   power2=power(step(i,1):step(i+1,1))*cadence(i);
   maxp2(i)=max(power2);  %max. watts per second%%
```

```
        meanp2(i)=mean(power2); %%av. Watts per second%%
    end


 %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%Change the EMG sampling rate %%%%%%%%%%%%%%%%
    [raw,col]=size(emgdata);
     F=[]; F1=[]; F2=[]; F3=[]; F4=[];
    for n=1:col
        F=emgdata(:,n);
        Fnew(n,:)= spline(1:45000,F,1:(45000/7200):45000); %%45000 is EMG sample#  7200is SW sample #
    end;
    newemg=Fnew';

    newsw=swdata;

      Fx=swdata(:,1);
      Fy=swdata(:,2);
      Fz=swdata(:,3);
      Mx=swdata(:,4);
      My=swdata(:,5);
      Mz=swdata(:,6);
      onoff=swdata(:,7);

      FR=sqrt(Fx.^2 + Fy.^2 + Fz.^2);
      r=0.2667; %%radian of pushrim
      Ft=Mz./r;
      Fr=sqrt(abs(FR.^2 - Ft.^2 - Fz.^2));
      fef=Ft.^2./FR.^2;
      final=[FR Ft Fr Fz Mx My Mz onoff fef];

%%%%%%%%%kinetic data calculation %%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
      [output]=kinoutput(final);
%%%%%%%%%% spline the emg data based on the push and recovery phase%%%%%%%%%
      [cols]=max(swdata(:,7));
    if cols>20;
        col=11;
    else
        col=4;
    end;
    pm=[]; da=[]; dm=[]; dp=[]; bi=[]; tr=[]; ld=[]; mt=[];

      for j=col:col+9;          %%pick up for strokes%%
          on=min(find(newsw(:,7)==j));
          off=max(find(newsw(:,7)==j));
          strokend=min(find(newsw(:,7)==j+1));
          timediff(j)=strokend-on;
          ontime(j)=(((off-on)+1)/timediff(j))*100;
          F=newemg(on:strokend,:);

              for n=2:9
                  F2=[spline(1:length(F(:,n)),F(:,n),1:(length(F(:,n))/100):length(F(:,n)))];
                  if n==2;
                      pm=[pm;F2];
                  elseif n==3
                      da=[da;F2];
                  elseif n==4
                      dm=[dm;F2];
                  elseif n==5
                      dp=[dp;F2];
                  elseif n==6
                      bi=[bi;F2];
                  elseif n==7
```

```
            tr=[tr;F2];
          elseif n==8
            ld=[ld;F2];
          elseif n==9
            mt=[mt;F2];
          end;
        end;
  end;

avontime=round(mean(nonzeros(ontime)));
avpm=mean(pm); avda=mean(da); avdm=mean(dm); avdp=mean(dp);
avbi=mean(bi); avtr=mean(tr); avld=mean(ld); avmt=mean(mt);
avemg=[avpm; avda; avdm; avdp; avbi; avtr; avld; avmt];
for n=1:8;
    EMG=avemg(n,:);
        for g=3:length(EMG)-2;
          if EMG(g-2:g+2)<5
            F2(g)=0;
          else
            F2(g)=EMG(g);
          end;
        end;

        for g=1:2;
          if EMG(g:g+1)<5
            F2(g)=0;
          else
             F2(g)=EMG(g);
          end;
        end;
        for g=98:100;
          if EMG(g-1:100)<5
            F2(g)=0;
          else
            F2(g)=EMG(g);
          end;
        end;
      if n==1;
        avfpm=F2;
      elseif n==2
        avfda=F2;
      elseif n==3
        avfdm=F2;
      elseif n==4;
        avfdp=F2;
      elseif n==5;
        avfbi=F2;
      elseif n==6;
        avftr=F2;
      elseif n==7;
        avfld=F2;
      elseif n==8;
        avfmt=F2;
      end
end;
avfemg=[avfpm; avfda; avfdm; avfdp; avfbi; avftr; avfld; avfmt];
plot(avfemg');

for n=1:8
   emgmedon(n)=median(nonzeros(avfemg(n,1:avontime)));
   emgmedoff(n)=median(nonzeros(avfemg(n,avontime+1:100)));
   emgmeanon(n)=mean(nonzeros(avfemg(n,1:avontime)));
```

```
        emgmeanoff(n)=mean(nonzeros(avfemg(n,avontime+1:100)));
        emgmaxon(n)=max(nonzeros(avfemg(n,1:avontime)));
        emgmaxoff(n)=max(nonzeros(avfemg(n,avontime+1:100)));
        pkpush=max(avfemg(n,1:avontime));
        pkrecov=max(avfemg(n,avontime+1:100));

        %%%%%%%%%%%below variables is to get the duration, onset/cessation time
        %%%%%%%%%%%based on percentage of propulsion cycle
        %%%%%%%%%%%report each muscle individually %%%%%%%%%%%%%%%%%%%%%%%%%
        onpk(n)=find(avfemg(n,:)==pkpush);  %% find the peak MVC during push%%
        offpk(n)=find(avfemg(n,:)==pkrecov);
        onper(n)=length(nonzeros(avfemg(n,1:avontime)));    %%% find duration of push during a propulsive cycle%%
        offper(n)=length(nonzeros(avfemg(n,avontime+1:100)));
        if length(nonzeros(avfemg(n,1:avontime)))==avontime
           emgoff(n)=avontime;
        else
           emgoff(n)=min(find(avfemg(n,1:avontime)==0));
        end;
        if length(nonzeros(avfemg(n,avontime+1:100)))==100-avontime;
           emgon(n)=avontime;
        else
           emgon(n)=max(find(avfemg(n,:)==0));
        end;
    end;
    duration=onper+offper;
    rawmatrix=[emgmedon; emgmeanon; emgmaxon; onper; onpk; duration; emgmedoff; emgmeanoff; emgmaxoff; offper; offpk;
duration];
    rimmatrix=[mean(avrimvel(col:col+9)) mean(cadence(col:col+9)) mean(meanp2(col:col+9)) mean(maxp2(col:col+9))];
    kinmatrix=[mean(output(1,col:col+9)) mean(output(2,col:col+9)) mean(output(3,col:col+9)) mean(output(5,col:col+9)) ...
             mean(output(6,col:col+9)) mean(output(7,col:col+9)) mean(output(9,col:col+9)) mean(output(10,col:col+9))];


     emgphase=[ rawmatrix; rimmatrix kinmatrix ];

end;

[FILENAME, PATHNAME] = uiputfile('*.txt', 'Save As');
     cd \
     eval(['cd ' PATHNAME]);
     fid1=fopen(FILENAME,'w');

     for i=1:gm
         fprintf(fid1,'%c%c%c%c \t',ID_matrix(i,:));

         for j_1=1:length(outcome)
             fprintf(fid1, '%f \t', outcome(i,j_1));
         end;

         fprintf(fid1,'\n');
     end;
    fclose(fid1);
```

APPENDIX H: SAS PROGRAMS FOR STUDY #3

```
proc print data=fes;
run;
proc mixed data=fes;
     class ID stage condition;
     model  medemg= stage|condition;
     repeated condition / subject=ID type=cs;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  medemg= stage|condition;
     repeated stage / subject=ID type=cs;
            lsmeans stage / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  avemg= stage|condition;
     repeated condition / subject=ID type=cs;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  iEMGon = stage|condition;
     repeated condition / subject=ID type=cs;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  iEMGon = stage|condition;
     repeated stage / subject=ID type=cs;
            lsmeans stage / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  onper = stage|condition;
     repeated condition / subject=ID type=cs;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  duration = stage|condition;
     repeated condition / subject=ID type=cs;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  onper = stage|condition;
     repeated stage / subject=ID type=cs;
            lsmeans stage / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  medemgoff= stage|condition;
     repeated condition / subject=ID type=cs;
            lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
     class ID stage condition;
     model  medemgoff= stage|condition;
     repeated stage / subject=ID type=cs;
```

```
        lsmeans stage / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model  avemgoff= stage|condition;
    repeated condition / subject=ID type=cs;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model  iEMGoff = stage|condition;
    repeated condition / subject=ID type=cs;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model  iEMGoff = stage|condition;
    repeated stage / subject=ID type=cs;
        lsmeans stage / pdiff adjust=BON alpha=.05;
run;
proc mixed data=fes;
    class ID stage condition;
    model  offper = stage|condition;
    repeated condition / subject=ID type=cs;
        lsmeans condition / pdiff adjust=BON alpha=.05;
run;
```