MUSCLE PERFORMANCE USING MC-SENSOR DURING FES-EVOKED CONTRACTIONS IN INDIVIDUALS WITH SPINAL CORD INJURY

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MUSCLE PERFORMANCE USING MC-SENSOR DURING FES-EVOKED CONTRACTIONS IN INDIVIDUALS WITH SPINAL CORD INJURY

Muscle contractions induced by functional electrical stimulation (FES) can contribute in rapid muscle fatigue. Muscle fatigue is one of the main limiting factors in FES. Spinal Cord Injury (SCI) patients experience fatigue in muscle early in prolonged FES-evoked activities than normal person. Due to the absence or compromised proprioceptive response that characterizes paralyzed muscle activities, it is crucial to monitor muscular forces generated during FES-evoked activity. An MMG-based muscle contraction (MC) sensor is presented in this research in which it was attached to the skin surface, used to monitor paralyzed muscle tension throughout a fatiguing functional electrical stimulation evoked contraction. As the MC sensor tip exerts compression pressure, which causes a dent on the skin and the intermediate layer above the muscle and also the muscle itself, the force measured by the sensor tip is proportional to muscle tension. Nine individuals with complete spinal lesion volunteered to participate in two separate trials, one to investigate MC-torque correlation while the other to investigate fatigue. The quadriceps muscle contraction under isometric conditions on Biodex Dynamometer was induced by electrical current in randomized order using Rehastim stimulator for the SCI participants, while measurements with MC sensor was recorded. Among SCI, MC-sensor-predicted measures of dynamometer torques, including the signal peak (SP) and signal average (SA), were highly associated with isometric knee extension peak torque (SP: $r = 0.91$, $p< 0.0001$), and average torque (SA: $r = 0.89$, $p< 0.0001$), respectively. Bland-Altman (BA) analyses with Lin's concordance (ρC) revealed good association between MC-sensor-predicted peak muscle torques (SP; $\rho C = 0.91$) and average muscle torques (SA; $\rho C = 0.89$) with the equivalent dynamometer measures, over a range of FES current amplitudes. The relationship of dynamometer torques and predicted MC torques during repetitive FES-evoked muscle contraction to fatigue were moderately associated (SP: $r = 0.80$, $p < 0.0001$; SA: $r = 0.77$; $p < 0.0001$), with BA associations between the two devices fair-moderate (SP; $\rho C = 0.70$: SA; $\rho C = 0.30$). In comparison, the MC sensor responses in able-bodied voluntary contractions had lower correlations with torque ($R^2 = 26\%$) over a range of isokinetic speeds, while the relationship between signal average of dyna-torque and MC-voltage were highly correlated ($R^2 = 85\%$). These findings demonstrated that a skin-surface muscle mechanomyography sensor was an accurate proxy for electrically-evoked muscle contraction torque when directly measured during isometric dynamometry in individuals with SCI. This study demonstrates the novel application of MC sensor in FES-evoked contraction application with individuals with SCI, which may be the highly sought-after solution for contraction feedback directly at the muscle level.

ABSTRAK

Kontraksi otot yang disebabkan oleh simulasi elektrik berfungsi (FES) cenderung menyebabkan keletihan otot dengan lebih cepat. Keletihan otot adalah salah satu faktor batasan utama dalam FES. Pesakit yang cedera saraf tulang belakang (SCI) mengalami keletihan otot lebih awal daripada FES berpanjangan daripada orang biasa. Oleh kerana kekurangan maklum balas proprioceptif yang memperlihatkan aktiviti otot,maka amat penting untuk memantau daya otot yang dijana semasa aktiviti FES. Sensor kontraksi otot (MC) yang berasaskan MMG yang digunakan dalam kajian ini diletakkan pada permukaan kulit, dan ia digunakan untuk memantau ketegangan otot lumpuh sepanjang simulasi elektrik berfungsi yang akan merangsang kontraksi. Oleh kerana hujung sensor MC menimbulkan tekanan mampat yang menyebabkan tekanan pada kulit dan lapisan di atas otot dan juga otot itu sendiri, maka daya yang diukur oleh hujung sensor adalah berkadar dengan ketegangan otot. Sembilan individu dengancedera saraf tulang belakang lengkap bersetuju untuk menyertai. Kontraksi otot quadriceps dalam keadaan isometrik menggunakan Biodex Dynamometer yang disimulasi melalui arus elektrik (dalam susunan rawak) menggunakan stimulator Rehastim sementara pengukuran dengan sensor MC direkodkan. Langkah-langkah yang diramalkan tor-dynamometer, termasuk puncak signal (SP) dan puratasignal (SA), menunjukkan persamaan apabila dikaitkan dengan tork puncak isometrik (SP: $r = 0.91$, $p <0.0001$) dan purata tork rata : $r = 0.89$, $p <0.0001$). Analisis Bland-Altman (BA) dengan concordance Lin (ρC) menunjukkan hubungan yang baik antara tork otot puncak yang MC-diramalkan (SP; $\rho C = 0.91$) dan purata tork (SA; $\rho C = 0.89$) semasa pelbagai amplitud FES. Hubungan tork dynamometer dan torsi MC yang diramalkan semasa pengulangan FES-menyebabkan pengecutan otot kepada

keletihan adalah berkaitan sederhana (SP: $r = 0.80$, $p < 0.0001$; SA: $r = 0.77$; $p < 0.0001$), dengan persatuan BA antara kedua-dua peranti adil-sederhana (SP; $\rho C = 0.70$: SA; $\rho C = 0.30$). Sebagai perbandingan, tindak balas sensor MC dalam kontraksi tanpa simulasi FES yang kepada peserta yang bukan paraplegic yang lebih rendah dengan tork ($R^2 = 26\%$) semasa kelajuan isokinetic yang pelbagai, manakala hubungan antara purata isyarat tork dyna dan tegangan MC sangat berkorelasi tinggi ($R^2 = 85\%$).Penemuan ini menunjukkan bahawa sensor MCadalah pilihan yang tepat untuk tork kontraksi otot yang diransangkan secara elektrik apabila diukur terus semasa dynamometri isometrik dalam individu dengan SCI. Kajian ini menunjukkan penerapan baru sensor MC dalam aplikasi penguncupan FES-menimbulkan dengan individu dengan SCI, yang mungkin menjadi penyelesaian yang sangat dicari untuk maklum balas penguncupan secara langsung pada tahap otot.

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LIST OF ABBREVIATIONS

AP	Action Potential
ASIA	American Spinal Injury Association
BA	Bland-Altman
dyna-torque	Dynanometer Torque
EMG	electromyogram
F	Force
FES	Functional Electrical Stimulation
FF	Fatigable Fibers
FR	Fatigue-Resistance
Fs	Vector sum of force
MAPE	Mean absolute percentage errors
MC	Muscle contraction sensor
MMG	Mechanomyography
MU	Motor Unit
NIRS	Near-infrared spectroscopy
NSCISC	National Spinal Cord Injury Statistical Center
r	Correlation coefficient
r^2	Coefficient of Determination
SA	Signal Area
SCI	Spinal Cord Injury
SD	Standard Deviation
SEE	Standard Error Estimate
sEMG	surface electromyogram
SP	Signal peak
VMG	Vibromyograpy
ρ_C	Lin's concordance correlation coefficient

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CHAPTER 1: INTRODUCTION

1.1 Overview

Functional electrical stimulation (FES) training are very popular among spinal cord injury (SCI) patients especially for rehabilitation purposes. Through doing this FES training, they will improve their muscle function and lead to a better health in the future (Ryan et al., 2013). During FES-evoked muscle contraction, muscles fatigue occurs rapidly especially in SCI individuals (Ibitoye et al., 2014) due to their synchronous pattern of motor unit recruitment and the preferential stimulation of large diameter neurons innervating fast-fatiguing muscle fibers (Beck et al., 2004). Mechanomyography (MMG) records mechanical activity of muscles using different types of sensors in order to monitor muscle vibrations as the muscle fibres move (Islam et al., 2013; Orizio et al., 2003). The amplitude of MMG portrays the strength of the muscle where it will continue to decline throughout fatiguing contraction (Sarillee et al., 2014). Current fatigue monitor, Surface Electromyography (sEMG), it hampered by stimulation artifact during FES (Chesler & Durfee, 1997) and it is not easy to analyze (Raez, Hussain, & Mohd-Yasin, 2006). A relatively novel muscle-contraction sensor (MC) was first introduced in 2011 (Dordević, Stančin, Meglič, Milutinović, & Tomažič, 2011) from the family of MMG-based sensors, and MC-sensor is based on the principle of measuring skin-surface tension when a muscle is recruited. MC data is relatively easy to analyse because its measurement is based on muscle tension changes, the signal pattern correlates directly with the instantaneous torque profile compared to other MMG signals (Dordević et al., 2011; M. O. Ibitoye, Hamzaid, Hasnan, et al., 2016).

1.2 Motivation

Given the fact of population size in the U.S is currently 314 million people and 32 million in Malaysia. Based on National Spinal Cord Injury Statistical Center (NSCISC), Spinal Cord Injury's annual incidence is reported to be approximately 54 cases per million populations in the U.S. or approximately 17,000 new SCI cases each year (Jain et al., 2015; National Spinal Cord Injury Statistical Center, 2019). Based on Ibrahim et al. (2013) the majority of people with SCI based on a single center study on epidemiology of SCI in Kuala Lumpur Hospital are young men (under 40 years of age) of low socio-economic status. SCI individuals will necessitate a prolonged period of Rehabilitation through physiotherapy and occupational therapy (Ibrahim et al., 2013). Help to deal with the many psychosocial issues like to prevent pressure (bed) sores; catheterization of the bladder which is tubes in the bladder to drain the urine, and education for the use of the wheelchair; is also other forms of rehabilitation and these patients will need to get help if appropriate.

The SCI patients also require neurological rehabilitation (Kitago & Krakauer, 2013). The patient will relearn the skills, which disoriented when they damage part of their brain. For example, movement skills that loss must be relearn through neurological rehabilitation. "Brain plasticity" is the known as the ability of brain to re-learn the lost functions due to injury (L. Cai, Chan, Yan, & Peng, 2014). Repetition-based task is the way of brain re-learning process. For the patient to do the skill voluntarily, the brain will need the motor function to be repeatedly to regain the lost skills (Dietz & Fouad, 2014). The physiotherapist must move the SCI individual's paralyzed hand or leg in a repeated way during the neurological rehabilitation which the activities may takes weeks or months. Nowadays, to assist physiotherapists in performing the rehabilitation repetition-based tasks, Functional Electrical Stimulation has been used (Rushton, 1997).

FES is one of the clinical neurostimulation applications used to restore the lost or damaged function in SCI rehabilitation instead from occupational therapy and conventional physiotherapy (Rushton, 1997). In restoring or assisting function in individuals with chronic spinal cord injury studies using FES have shown promising results. By using a computer system and electrodes, FES device to generate muscle contraction; uses to deliver small bursts of a low-level electrical current to paralyzed muscles. In FES previous studies, many researchers believe that as the stimulation proceeds, the muscle dynamics remain the same. However, the dynamics change during muscle contraction in practice.

Muscle contractions induced by functional electrical stimulation tend to result in rapid muscle fatigue. "Muscle fatigue is a complicated phenomenon, and it is often a combination of excitation, contractile and ischemic fatigue" based on Erfanian, Chizeck, & Hashemi (1996). SCI patients cannot identify muscle fatigue as electrical stimulated continued as most of them has lost their sensory and motor functions. Thus, the electrical stimulation patterns using FES will manually have adjusted by the patients or the therapist in order to monitor fatigue from muscle performances. Other than that, the reflexes of SCI patient will affect the stimulate torque level. Thus, accurate muscle force results are crucial for better FES activities. Considering muscle fatigue as an important factor, a fatigue-monitoring device must be included as one crucial component in any FES activities.

Possible muscle fatigue sensors are a surface EMG (Chesler & Durfee, 1997; Mizrahi, Levy, Ring, Isakov, & Liberson, 1994), mechanomyography (M. T. Tarata, 2003), and near-infrared spectroscopy (NIRS) (Yoshitake, Ue, Miyazaki, & Moritani, 2001).

Measuring of fatigue by EMG, MMG, and NIRS, which by the electrical, mechanical, and metabolic activities in the contracting muscle respectively.

1.3 Problem Statement

Muscle fatigue monitoring mostly done for promising health benefits of FES-evoked exercise. Fatigue indicator may benefit the Physiotherapy and SCI patient while doing the FES for rehabilitation so that the risks of injury because of fatigue can be avoidable in future and the training time can be adjustable based on muscle performance analysis of the SCI patient.

EMG can be contaminated by electrical noise (Chesler & Durfee, 1997) and studies was carried out on extensive comparison between EMG and MMG in voluntary isometric contraction. Tarata stated that, the mechanical components of the muscle (such as torque and contraction speed) are more closely related to the muscle function than to its electrical characteristics (M. Tarata, Spaepen, & Puers, 2001).

The analysis of the MMG tendency curve displayed in the results of the study by Tarata on 2003, indicated a downward trend of the mean normalized values in the development of muscle fatigue (M. T. Tarata, 2003). By applying the protocol developed for the present study, it was possible to conclude that MMG could be applied concomitantly to FES because it does not suffer electrical interference and can be used during functional movements and muscle contraction generated by FES.

A different type of MMG which is MC-sensor is expected to further demonstrate MMG as fatigue indicator during FES stimulation. MC-sensor chosen over other muscle sensors; is that MC data is relatively easy to analyse because its measurement is based on muscle tension changes, the signal pattern correlates directly with the instantaneous torque profile compared to other MMG signals (Dordević et al., 2011; M. O. Ibitoye, Hamzaid, Hasnan, et al., 2016).

However, the ability and sensitivity of the MC sensor in detecting electrically-induced muscle fatigue (i.e. a reduced force generating a capacity of muscle at an unchanged stimulation current) in persons with SCI remains an important and unanswered. If the MC sensor can predict muscle torque accurately, it might be deployed in mission-critical daily functional tasks for neurological populations.

1.4 Objectives

The purpose of this research study is to investigate the sensitivity of MC-Sensor to monitor muscle performance, and muscle fatigue during FES-evoked contraction on a larger sample of size inclusive in people with Spinal Cord Injury. If the processed data analyzed of MC-sensor voltage dc is proportional to torque output during the experiment on SCI patient, the MC-sensor is sensitive. Thus, it can be validated as a suitable sensor to collect muscle performances, and muscle fatigue during FES.

In particular, the objectives of this research are:

1) To investigate MC sensor responses versus muscle torque measurements during prolonged, electrically evoked, fatiguing muscle contractions in individuals with motor-complete SCI.
2) To determine degree of agreement between body-worn MC-sensor with directly measured isometric knee extension torques derived from a commercial dynamometer in order to act as proxy to each other.

1.5 Significant of the study

Since there was no reported studies on the ability and sensitivity of the MC sensor in detecting muscle fatigue during electrically-induced contraction in persons with SCI, thus, findings of this study would demonstrate the first application of MC sensor in neurological research. If the MC sensor can predict muscle torque accurately, it can be used as a tool for monitoring the muscle state in critical functional tasks among neurological populations.

1.6 Dissertation Organization

This dissertation consists of six chapters, which are Introduction, Literature Review, Methodology, Results, Discussion, and Conclusion.

Chapter 1 is the Introduction. It explains the general idea of the study in brief. This chapter also contains the problem statement, research objectives, significance of the study and dissertation organization.

Chapter 2 is the Literature Review. It mainly addresses the critical analysis of the previous relevant studies in relation to the present study.

Chapter 3 is the Methodology. This chapter describes the participants, protocols, and materials used in the study. Participants recruited for this study are SCI individuals with specific inclusion and exclusion criteria. Two experimental protocols conducted in this research are explained.

Chapter 4 is the Results. It contains all the findings of the current study. This chapter describes the results obtained from method of validation and method of agreement.

Chapter 5 is the Discussion. This chapter discusses the findings of the current study and clarifies the findings of the current research with previous studies.

Chapter 6 is the Conclusion. This chapter summarizes the findings of the current study and how it can be applied especially in the rehabilitation field.

CHAPTER 2: LITERATURE REVIEW

This chapter contains a critical review of currently available literature related to the current study. This chapter are divided into seven sections. The first section explains spinal cord injury in detail along with its classification. The second section describes functional electrical stimulation in the rehabilitation field. The third section explains how muscle fatigue happens during FES exercise. The fourth and fifth section on the other hand describes surface electromyogram, and MMG and their role in measuring muscle fatigue respectively and their limitations. The sixth section explained the novelty of MC-sensor and portrays previous studies that use MC sensor.

2.1 Spinal Cord Injury

"The spinal cord is a major bundle of neural cells (neurons and glia) and nerve pathways (axons) that extend from the base of the brain to the lower back" based on Caroline, Elling, & Smith, (2014). "Spinal cord is primary information highway that receives sensory information from the skin, joints, internal organs, and muscles of the trunk, arms, and legs. Then the information is relay upward to the brain and downward from the brain to other body systems" (NINDS, 2015). Brain communication with the rest of the body below the level of lesion, which interrupts the signal communication, by impairment of the spinal cord is called spinal cord injury (SCI). It is caused by spinal cord damage from trauma such as motor vehicle accident, sports or fall, that stretches, compresses or severe this tissue or by diseases may damage the neurological tissue of the spinal cord (Lynch & Popovic, 2008). "Injuries can happen at any level of the spinal cord. The severity of the damage to the nervous tissue and the injured segment of the cord would determine which body functions are compromised or lost. Physiological consequences to parts of the body controlled by nerves at and below the level of the injury” as explained through article *Spinal Cord Injury: Hope Through Research* (NINDS, 2015) may also cause by injury to a part of the spinal cord. Injuries in the thoracic region usually affect the chest and the legs and result in paraplegia. “The vertebrae in the lower back between the thoracic vertebrae, where the ribs attach, and the pelvis (hip bone), are the Lumbar vertebrae. The sacral vertebrae run from the pelvis to the end of the spinal column. Injuries to the five Lumbar vertebra (L-1 thru L-5) and similarly to the five Sacral Vertebrae (S-1 thru S-5) generally result in some loss of functioning in the hips and legs” based on Sarkodie-Gyan (2006).

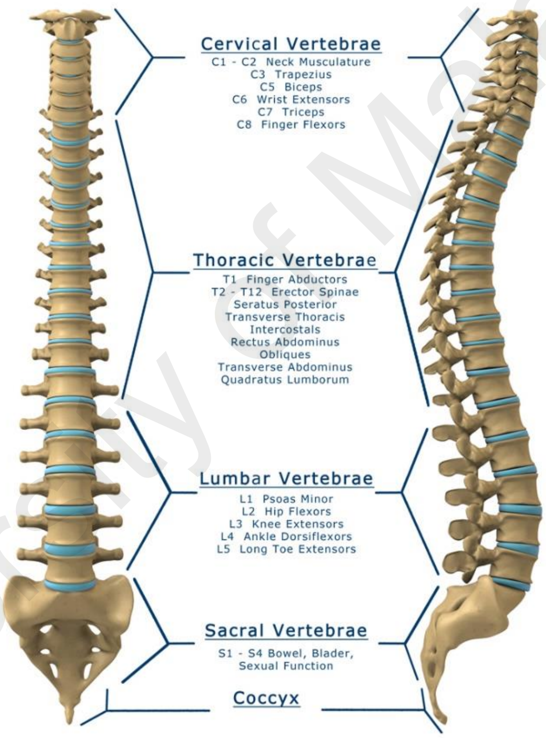


Figure 2.1 Function of the spinal cord for body part transaction cord level. Retrieved from (NINDS, 2015)

As describe from Pneumaticos, Triantafyllopoulos, & Lasanianos (2015), “SCIs are also described as complete and incomplete, and an incomplete injury is further classified into four subsections”. SCIs classification by the American Spinal Injury Association (ASIA) shown in Table 2.1.

Table 2.1 Classification of Spinal Cord Injury. Retrieved from Sisto, Druin, & Sliwinski (2009)

Classification	Description
A	Complete: No feeling or movement of the areas of your body that are controlled by your lowest sacral nerves. This means you do not have to feel around the anus or control of the muscle that closes the anus. People with complete SCI do not have control of bowel and bladder function.
B	Incomplete: Feeling but no movement below the level of injury, including sacral segments that control bowel and bladder function.
C	Incomplete: Feeling and movement below the level of injury. More than half of key muscles can move, but not against gravity. Moving against gravity means moving up, for example, raising your hand to your mouth when you are sitting up.
D	Incomplete: Feeling and movement below the level of injury. More than half of key muscles can move against gravity.
E	Feeling and movement are normal.

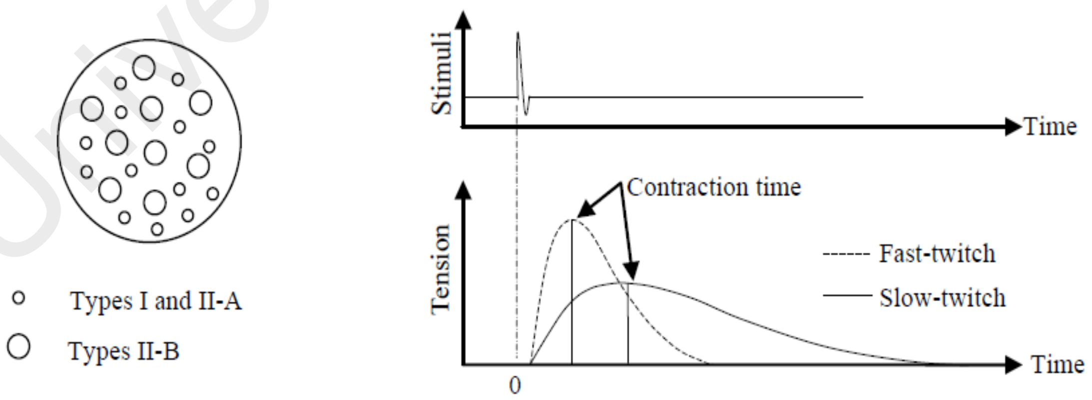
It is important for SCI patients who would benefit most from any future therapeutic strategies to maintain their body in the best condition to improve their health and lessen the risk of SCI related diseases. Based on scientists’ opinion, advances in research will someday repair the spinal cord injury.

2.1.1 Skeletal Muscle

Skeletal muscles are named based on their connection to the skeletal bones. Skeletal muscles are specifically composed of three fiber types. “Slow twitch fiber that is found in high proportions in the postural muscles of the back, where slow sustained contractions are necessary is a Type I. Fast twitch fiber is a Type II and may be subdivided into two subtypes: Fatigue-Resistance (FR) fibers (type II-A), and Fatigable Fibers (FF) (type II-B)” based on Horch & Dhillon (2004). Main properties of the muscle fibers differences are listed Table 2.2 and Figure 2.2 is showing the relative size and the contractile properties of different muscle fiber types.

Table 2.2 Important properties of different muscle fibers.

Fiber type/specifications	Type I	Type II-A	Type II-B
Firing stimuli threshold level	High	High	Low
Associated nerve axon size	Small	Small	Large
Twitch speed	Slow	Fast	Fast
Fatigability	FR	FR	FF



(a) (b)

Figure 2.2 (a) Relative size muscle fibers. (b) Contractile characteristics of fast-twitch and slow-twitch muscle fibers as shown. The contraction time is the time from onset of the stimulus to the instant of peak force. (Shalaby, 2011)

2.12 Motor Unit (MU)

The smallest unit for force generation is called Motor Unit (MU). When it is recruited by the brain, force generation contributed by each fiber of a muscle. One motoneuron can branch into tens, hundreds, or even a thousand branches. From each branch terminating at a different muscle fiber of the same type. A motor unit consists of an alpha-motoneuron cell body within the spinal cord, the motor axon, and all branches plus all muscle fibers it innervates. A simplified schematic diagram showing a motor unit is given in Figure 2.3.

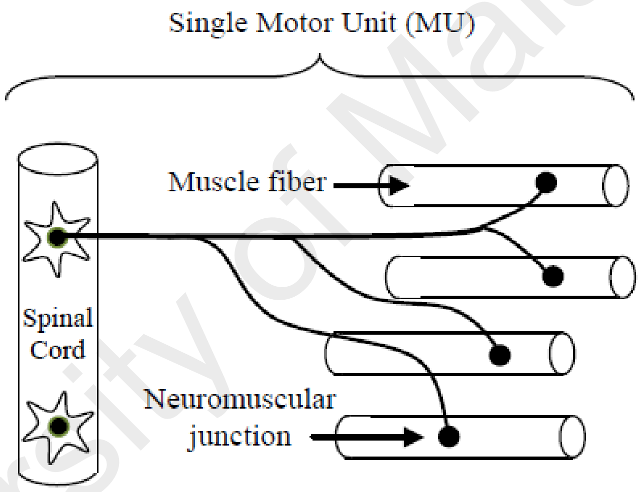


Figure 2.3 The motor unit consists of a single alpha-motoneuron and all the muscle fibers it innervates. Adapted from Latash (2008)

An electrical signal from the central nervous system must first reach the alpha-motoneuron, which is responsible for initiating the muscle contraction for MU fibers to be contracted. A series of electrophysiological and electrochemical processes take place as the contraction signal spreads from the alpha-motoneuron across the muscle fiber. This produces an electrically measurable de-polarization and re-polarization event known as the Action Potential (AP). The AP is important as it carry information within the body by means of diffusion. The intensity of muscle contraction is controlled by how often the nerve impulses arrive and innervate muscle fibers, i.e. as the AP arrives more often the muscle contracts harder (Latash, 2008).

2.1.3 Muscle Contraction

A skeletal muscle contract as long as neural signals activated it. “Neural signals transform from the muscles into mechanical outputs, for instance muscle force or length. The muscle contractile mechanism is complex and is summarized by the "sliding filament" theory” based on Huxley (1957). Through the "crossbridge" cycling in a sarcomere between actin and myosin at the molecular level, where contraction muscle is induced. Brief explanation of neuromuscular system based on J. Forehand (2009),“where the brain sends a neural signal, in the form of a neural action potential, which then propagates along the motor neuron axons. When the neural action potential reaches the neuromuscular junction, the depolarization produced in the nerve endplate and then in the postsynaptic receptors on the muscle gives rise to a muscle action potential”.

Other than that, J. Forehand (2009) also explained that “When the muscle action potential propagates along the membrane and along the T-tubules, calcium is released across the entire sarcomere. Calcium release makes the active site of the thin actin bind with the myosin head (or cross-bridge), and then pulls the actin sliding along the myosin toward the center of the sarcomere. At the same time, the cross-bridge rotates, allowing for the attachment to a new active site at a shorter length. When a new neural action potential arrives this process is repeated, resulting in progressive myofibril contraction. The entire muscle fiber and then the entire muscle consequently contract. If no action potential arrives the cross-bridge cycle is terminated due to the decrease in calcium release. The muscle relaxes and returns to the rest state as a result”.

2.1.4 Impairment of Motor Function

SCI, cerebral injury, multiple sclerosis or other disorders in neuro resulting the disability of motor function.

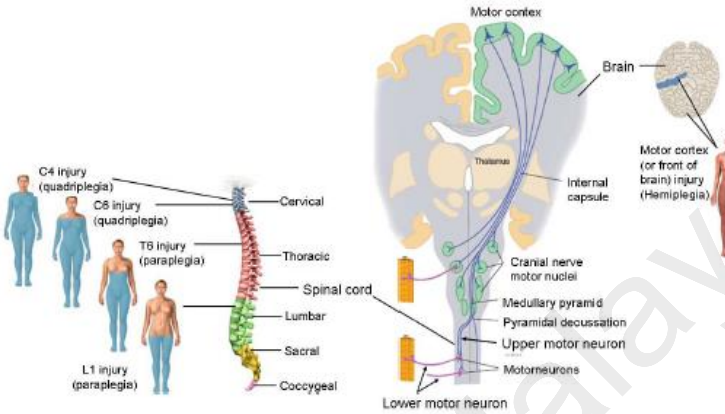


Figure 2.4 Lower motor neurons (purple) directly serve for the effectors. Left: SCI refers to any injury within the spinal cord, leading to quadri/paraplegia at different levels. Right: Hemiplegia caused by various injuries. Retrieved from Brodal (2004).

As shown in Figure 2.4, "Lower motor neurons are the neurons that directly serve for the effectors (such as skeletal muscles). They are either from the cranial nerve in the brain stem or from motor nerves in the spinal cord. On the other hand, the upper motor neurons originating from the motor cortex or the brain stem are not directly responsible for stimulating effectors but connect the brain to the appropriate level in the spinal cord. From this spinal segment, the nerve signals propagate to the effectors by means of lower motor neurons" which explained by Singh. Therefore, "motor neuron lesion is usually divided into upper motor neuron lesion and lower motor neuron lesion with respect to different injury sites and resultant symptoms. An upper motor neuron lesion is usually accompanied by innervated and paralyzed muscle, while a lower motor neuron lesion results in denervated and paralyzed muscle. SCI refers to any damage to the spinal cord from trauma, loss of normal blood supply, compression from tumor or inflammation" based on Zhang, Hayashibe, Fraisse, & Guiraud (2011). SCI severity is roughly qualified as "complete" or "incomplete" and classified according to the American Spinal Injury Association (ASIA) scale. Higher level injuries imply greater functional deficits and impairment of personal mobility (Triolo et al., 1996).

"Five classifications means there are no motor or sensory function (ASIA A), only sensory function remaining (ASIA B), some sensory and motor preservation (ASIA C), useful motor function (ASIA D) and normal function (ASIA E) (NINDS, 2015) as in Table 2.1. Based on Figure 2.4, complete paralysis in all upper and lower limbs (quadriplegia) or in lower limbs only (paraplegia) resulted from injuries at different sites in cerebral column. Brodal et. al also explained that "Hemiplegia can be resulted by various injuries, for instance, traumatic injury to the brain which caused to opposite side of body being affected".

2.2 Functional electrical stimulation

Functional electrical stimulation is one of the clinical neurostimulation applications used to restore the lost or damaged function in SCI rehabilitation instead from occupational therapy and conventional physiotherapy (Rushton, 1997). In restoring or assisting function in individuals with chronic spinal cord injury, it has been shown in studies using functional electric stimulation that there were promising results. By using a computer system and electrodes, FES device to generate muscle contraction; the system delivers small bursts of a low-level electrical current to the paralyzed muscles. Researchers are working to improve the electrode and computer interfaces, so they can produce more natural yet complex movements. FES is being used to restore breathing without a ventilator, cough unassisted, enhance bladder and bowel control, increase hand movement and grasping, and improve blood flow to the skin. Several studies continue to investigate the use of FES in restoring or improving function following a spinal cord injury. Researchers are using FES to help individuals with SCI move their paralyzed muscles to restore voluntary grasping function (Ethier, Oby, Bauman, & Miller, 2012).

Others are attempting to stimulate recovery with the goal of being able to grasp objects without stimulation once the treatment program is completed. This approach was shown to be useful for people who have had a stroke (A. Thrasher, Zivanovic, McIlroy, & Popovic, 2008). Based on Lynch & Popovic (2008), by varying the tension produced in the flexor and extensor muscles that actuate the joint, the angle of a joint, or the torque about that joint, can be regulated. The flexor muscles are the hamstrings group, while the extensor muscles are the quadriceps group for the knee joint. The hamstrings flex the knee to a bent position while the quadriceps extend the knee and straighten the leg. The intact neurological system produces tetanic contractions, which are characterized by sustained, constant tension, by stimulating each motor unit at a frequency of 6–8 Hz. Adjacent motor units are stimulated sequentially so that the overall muscle produces a tetanic contraction.

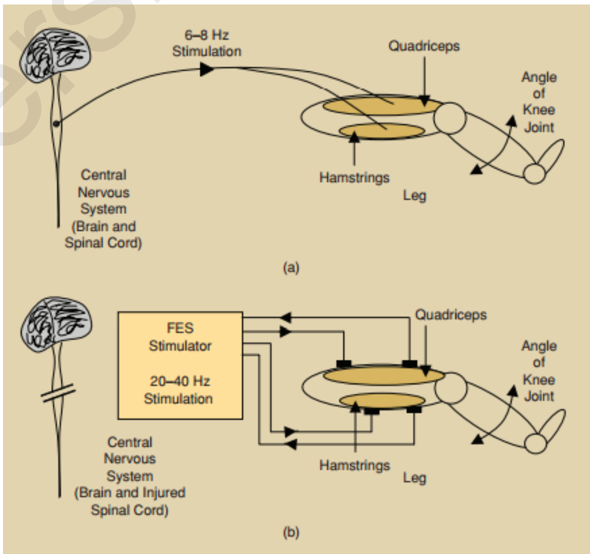


Figure 2.5 Contraction production of (a) intact neurological and (b) injured spinal cord individuals (Lynch & Popovic, 2008)

If the muscle tension produced by the tetanic contraction is sufficiently high, the knee angle changes, as shown in Figure 2.5, a functional electrical stimulation (FES) system can produce tetanic contractions in a spinal cord injured subject.

2.2.1 Electrode tissue interface

Based on Baker (2000), an electrical stimulation pulse applied by an electric field or current between the stimulation electrodes lead a current of ions to flow and diffuse through the skin and flow through the tissues underneath the skin that include muscles (motor points) and nerves at the electrode-tissue interface in transcutaneous electrical stimulation. Type of stimulation signal pulse polarity will affect the polarity of the electrode. When using Monophasic stimulation signal one of the electrodes will be negative in polarity (active electrode) and the other will be positive in polarity (indifferent electrode). When using Alternating or Bi-phasic signals it is not possible to decide about the polarity of the electrode, as polarity at each electrode alternate from positive to negative or vise-versa.

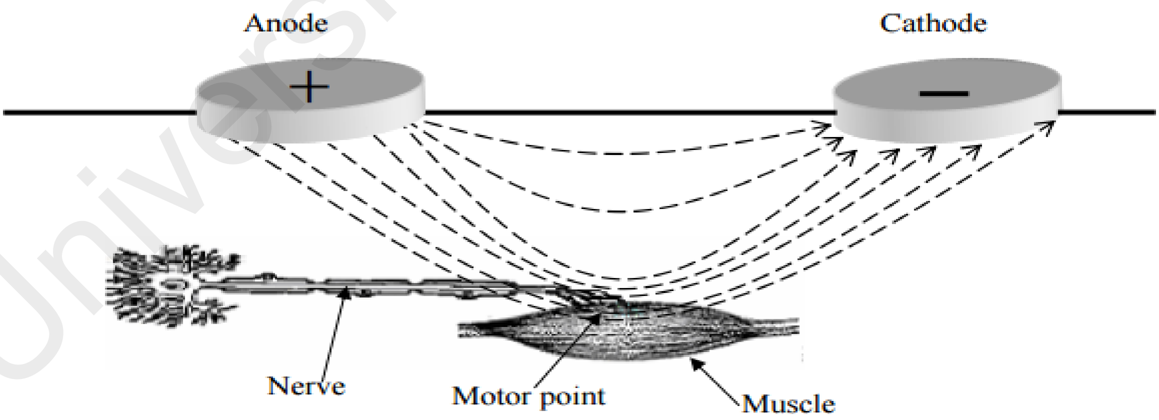


Figure 2.6 Muscle stimulation by applying electric field between two electrodes, stimulation applied near to the motor point of the nerve. Adapted from Baker (2000)

Figure 2.6 which adapted from Baker (2000) show transcutaneous stimulation mechanism simplified diagram. “At the positive electrodes the positive ions at the electrolyte interface are attracting the negative ions. At the negative electrode, the negative ions attract the migrating positive ions. This causes electrical charged particles to flow. It is the movement of potassium and sodium ions across the axon membrane that causes the action potential and as a result, a nerve impulse is produced. “This nerve impulse causes the muscle that this nerve supplies to contract (Baker, 2000). The depolarisation or the excitation of the nerve axon occurs principally at the cathode, hyperpolarisation occurring at the anode, near the cathode the potential is lowered. If a nerve axon is in the neighbourhood, the potential difference across the membrane might be lowered sufficiently to cause an action potential. Critical to whether a given stimulus causes an action potential or not is the impedance between the electrodes and the skin, the orientation and size of the electrodes and the stimulation signal type”. To stimulate muscle, the placement of the electrodes is very important. For instance, the placement of electrodes must be over the selected muscle motor point, which is responsible for specific movement.

2.2.2 Limitation with FES to neurological rehabilitation

FES can evoke effective muscle contractions; however, serious restriction exists. With FES, the action potentials of different motor units are triggered simultaneously. The recruitment of MUs by the CNS is different which activates motor units asynchronously. “According to the size principle, the physiological recruitment of motor units within a muscle proceeds from slow-twitch motor units which are FR, to the fast-twitch ones FF. Thus, the FR motoneurons are mostly responsible to produce low muscle contraction” based on Shalaby (2011). In FES recruitment, low muscle contraction is achievable by using low FES intensity. Referring to Table 2.2, low FES levels activate the lower firing stimuli threshold MUs, which are close to the stimulation electrodes. Unfortunately, those are mainly the fatigable MUs (type II-B). As the intensity of FES increases the fatigue resistance MUs type I and type II-A (with higher firing thresholds) start to fire, as well as the deep type II-B motor units. In addition to this, a constant FES for the same electrode position activates the same motor units all the time. This results in an early fatigue of the electrically stimulated muscles. Thus, the recruitment with FES starting with fatigable MUs is thought to be inverted.

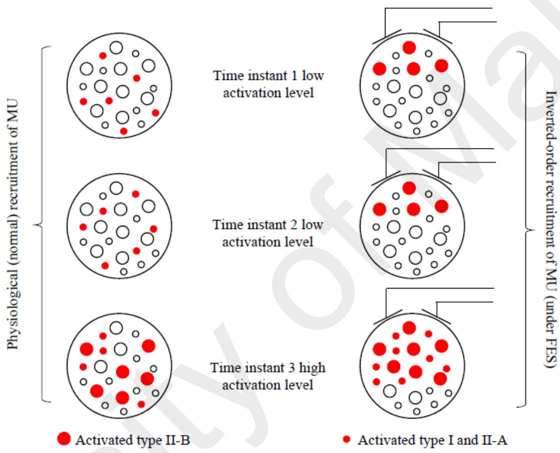


Figure 2.7 Natural muscle activation versus FES artificial muscle activation. Adapted from Schauer (2017)

2.3 Muscle Fatigue with FES

Untrained innervated (paralyzed limbs of patients with spinal cord) muscles can respond vigorously to electrical stimulation, however quickly fatigue; they may take to recover fully for hours from such fatiguing stimulation. As we know, muscle fatigue and weaken with use, however, electrical stimulation muscle fatigue seems to be intense.

"There are two groups of reasons: firstly, there are various ways in which paralyzed, innervated muscles are different from normal; and secondly, there are various ways in which FES is wasteful of the resources of the muscle” based on Rushton (1997). These differences are summarized in Table 2.3.

Table 2.3 Causes and prevention fatigue in stimulated muscle (Rushton, 1997).

Extra fatigue in FES of paralysed muscle	**Reasons for excess fatigue**	**Treatment of excess fatigue**
Changes in muscle because of paralysis and disuse	loss of fatigue-resistant fibre types	training programme
	reduction in capillary density	training programme
Inefficiencies inherent in FES achieved by stimulation of motor nerves or root	reverse recruitment	partial anodal block
	synchronous twitches	multichannel stimulation
	no duty cycling	multichannel stimulation
	excess force production	force feedback
		partial anodal block
		multichannel stimulation

Muscles show increasing signs of fatigue as they progressively expend their available energy resources. If muscles are artificially stimulated to the point of fatigue, their response changes nonlinearly as their energy stores are depleted. Eventually, the muscle will no longer be able to produce tension (T. A. Thrasher, Graham, & Popovic, 2004). Figure 2.8 shows four FES-evoked fatigue patterns that were seen in a single SCI subject at different days. Similar fatigue pattern also shown in study by Dzulkifli, Hamzaid, Davis, & Hasnan (2018). There is a rise in the initial frequency when the muscle is fatigued (Figure 2.9).

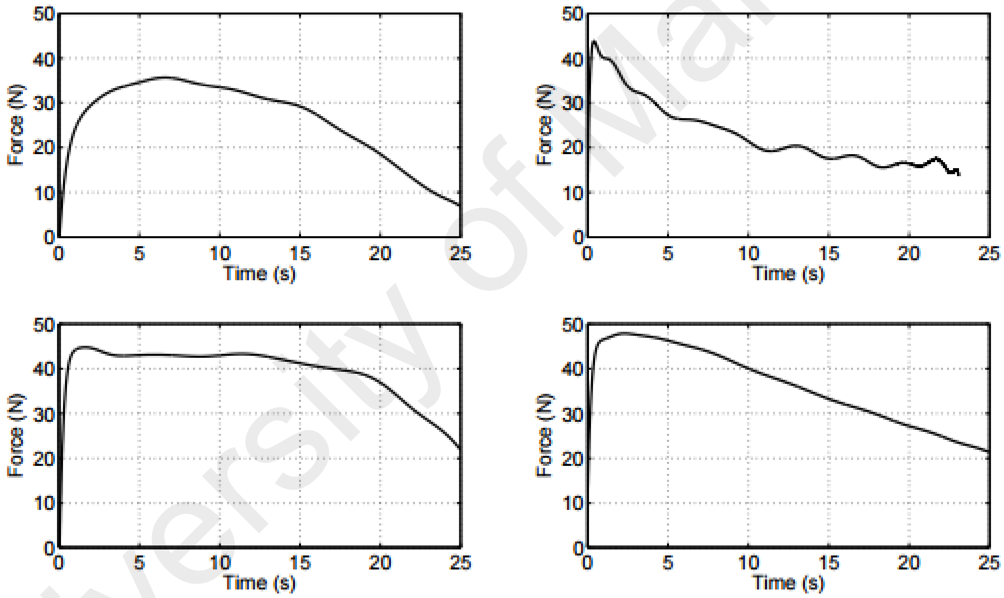


Figure 2.8 Force exerted by quadriceps versus time. These plots show the force exerted by the quadriceps muscle group when subjected to electrical stimulation. The data were collected on different days from an individual who had sustained a SCI resulting in complete paralysis of the lower extremities. Four different force decay profiles are shown, representing the wide range of force profiles that be the muscle fatigues in response to stimulation. Adapted from (T. A. Thrasher et al., 2004)**. The order of days in the different graphs were unspecified, portraying random force profile in different days.**

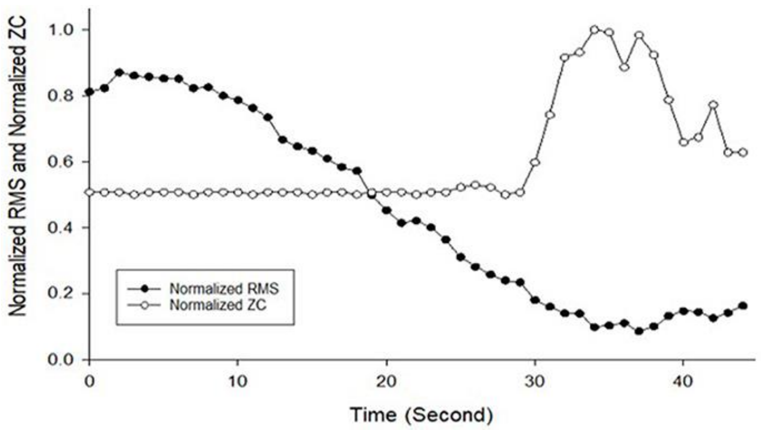


Figure 2.9 Normalized Fatigue MMG RMS and Normalized MMG ZC against time (Dzulkifli et al., 2018)

2.4 Muscle fatigue & Muscle contraction monitoring

In FES previous studies, many researchers believe that as the stimulation proceeds, the muscle dynamics remain the same. However, the dynamics change during muscle contraction in practice. Muscle contractions induced by functional electrical stimulation (FES) tend to result in rapid muscle fatigue. “Muscle fatigue is a complicated phenomenon, and it is often a combination of excitation, contractile and ischemic fatigue” based on Erfanian et al. (1996). The behavior of muscle tends to be distinguished based in response to various factors in FES. For instance, “with the same stimulation level, different stimulation patterns (e.g., high frequency or low frequency stimulations), different muscle fiber types (e.g., fast or slow muscles), and different muscle contraction conditions (e.g., dynamic or static contractions),” explained by Zhang et al. (2011) which it isresulted dissimilar muscle fatigue behaviors. Muscle fatigue is hence considered as a protocol-specific, muscle-specific, task-specific, and even subject-specific phenomenon.

Other than that, SCI patients cannot identify muscle fatigue as electrical stimulated continued as most of them has lost their sensory and motor functions. Thus, the electrical stimulation patterns using FES will manually adjusted by the patients or the therapist in order to monitor fatigue from muscle performances. Other than that, the reflexes of SCI patient will affect the stimulate torque level. Thus, accurate muscle force results are crucial for better FES activities. Muscle fatigability should be considered as additional important criteria as the limiting factor in FES activities.

2.4.1 Predicting Muscle Fatigue with FES

Even though there is attempting to predicting muscle fatigue with FES, however, muscle fatigue is in fact not easy to reduce effectively. Previous research is commonly about how to detect fatigue on when electrical stimulation caused the muscle to fatigue. Fatigue monitoring is important especially in SCI patients, which their paralyzed muscles cause them to suffer lack of sensory feedback. Based on previous studies from the experimental measurement, physiological and interpretation of mathematical, a lot of fatigue models has developed. For instance, a “biomechanical model was developed to predict the shank motion induced by FES” by Riener, Quintern, & Schmidt (1996). “Individual model parameters were identified by specific experiments. A four-parameter nonlinear equation was proposed to extract fatigue indices for predicting torque decrease in paraplegic subjects during an isometric sustained contraction” studied by Rabischong & Chavet (1997). Whilst “A five-element musculotendon model was developed to predict the force generation capacity of the activated muscle, and a fatigue recovery function, based on the metabolic profile, was introduced into this model” by Mizrahi, Seelenfreund, Isakov, & Susak (1997). In Z. Cai, Bai, & Shields (2010), a “Wiener-Hammerstein model was proposed to predict FES-induced muscle force in unfatigued and fatigued muscle, and this work was verified by stimulating the Soleus muscle in SCI patients”.

2.4.2 Limitation with current fatigue monitor- sEMG

Muscle fatigue monitoring mostly was done to improve health benefits of FES-evoked exercise. Fatigue indicator may benefit the physiotherapy as well as the SCI patients while doing the FES for rehabilitation so that the risks of injury due to fatigue can be avoided in the future and the training time can be adjusted based on muscle performance analysis of the SCI patient. Possible muscle fatigue sensors are surface electromyogram (EMG) (Chesler & Durfee, 1997; Mizrahi et al., 1994), vibromyograpy (VMG) (M. T. Tarata, 2003) and near-infrared spectroscopy (NIRS) (Yoshitake et al., 2001). Measuring of fatigue by EMG, VMG, and NIRS, which by the electrical, mechanical, and metabolic activities in the contracting muscle respectively. Previous studies have attempted to predict force/torque variations with fatigue based on eEMG. "An exponential function was proposed to predict force of the FES-activated quadriceps muscles from eEMG Peak-to-Peak (PTP)" (Mizrahi et al., 1994). As fatigue index, PTP was suggested for prolonged cycling speed (Chen & Nan-Ying Yu, 1997). Chen and Yu proposed "a hyperbolic modeling of PTP to dynamic cycling numbers under intermittent FES".

Other than that, under continuous stimulation in paraplegic subjects; a high correlation between EMG and knee torque reported by Erfanian et al. (1996). Erfanian et.al other works also proposed "a predictive model of muscle force production under an isometric percutaneous continuous FES system" (Erfanian, Chizeck, & Hashemi, 1998). Force prediction and EMG performance from the electrical stimulation suggested that using EMG signals to predict muscle torque. However, based on the research by Al-Mulla, Sepulveda, & Colley (2011), they explained that "EMG data is relatively harder to analyze because the measurement was based on the electrical thus, the data interfereswith the electrical properties of contact between skin and electrode". Other than that, based on a study by M. T. Tarata (2003), EMG can be contaminated by electrical noise and carried out an extensive comparison between EMG and mechanomyography (MMG) in voluntary isometric contraction. Researchers M. Tarata et al. (2001) stated that, the mechanical components of the muscle (such as torque and contraction speed) are more closely related to the muscle function than to its electrical characteristics. The analysis of the MMG tendency curve displayed in the results indicated a downward trend of the mean normalized values in the development of muscle fatigue (M. T. Tarata, 2003). However, because the coefficient of determination of the adjustment was very low (R^2=0.1949), this parameter could not be considered an indication of muscle fatigue during the application of the FES under the same conditions of this study. By applying the protocol developed for the present study, it was possible to conclude that MMG could be applied concomitantly with FES because it does not suffer electrical interference and can be used during functional movements and muscle contraction generated by FES. Hopefully, a MC-sensor which is one of MMG-based sensor can demonstrate MMG as suitable fatigue monitor during FES stimulation.

2.5 Basic principles of MC-sensor

The measurement methods of skeletal muscle tension can be described through selective manner and completely non-invasive sensors. The muscle contraction (MC) methods propose the measurements the force on subject's skin above the desired skeletal muscle. Tension of muscle changed as skeletal muscle activities these varying levels of contractile force are able to produce different levels of tension in skeletal muscle.

The basic structure of the MC sensor is illustrated in Figure 2.10. "The MC sensor consists of a sensor tip (1), force meter (2), and supporting part (3). The sensor is attached to the subject's skin surface (4)above the intermediate layer (5) and the skeletal muscle being measured (6)" retrieved from Dordević et al. (2011)

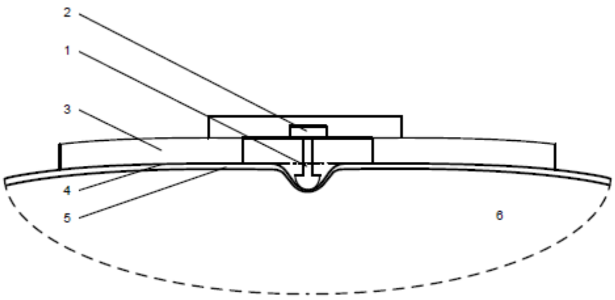


Figure 2.10 “MC sensor for determining the mechanical and physiological properties of skeletal muscles (1): sensor tip; (2): force meter; (3): supporting part; (4): skin surface;” retrieved from (Dordević et al., 2011)

“The sensor is designed with sensor tip above so that the pressure on subject’s skin surface and the intermediate layer” based on (Dordević et al., 2011). The sensor designed that way, so the sensor tip compresses the skin surface and applying pressure on the desired measure skeletal muscle. Whilst this placement of sensor is non-invasive, however, the sensor tip must be suitable shaped, so it can push down onto subject’s skin. A simplified diagram in Figure 2.11 explained of the principle of the MC measuring method.

The measured muscle tension produces the force F in the direction along the muscle surface. This force causes the intermediate layer and skin to press on the sensor tip. Based on (Dordević et al., 2011), “The vector sum of all forces produces the force Fs in the direction along the sensor tip. This force was then measured with the force meter. From the illustration on Figure 2.11, this force would be equal to:

$$Fs = 2F \cos \alpha$$

Where α is the angle between the directions of F and Fs. The measured force is still proportional to the force produced by the muscle tension, If the pressure produced by the sensor tip remains constant (i.e., the angle α between the forces does not change). Through the sensor supporting part, the sensor was attached to the skin of the measured subject skeletal muscle. However, the thickness and the elasticity of the skin, and the tissue under the skin differs on every subject.

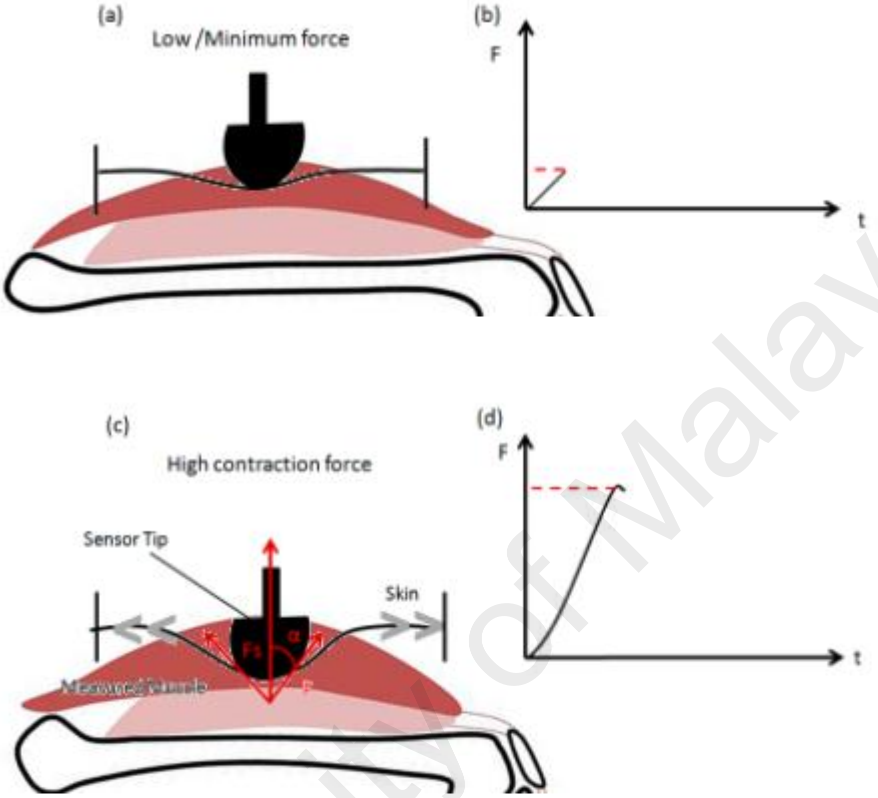


Figure 2.11 A simplified representation of the MC measuring principle for the determination of the mechanical and physiological properties of skeletal muscles (a): sensor tip;(b) tension-signal (c): measured muscle; (d): tension-signal.

Thus, calibration of the device for each subject separately must be done to measure the absolute value of muscle force. However, Dordević et al. (2011) explained that "on the dynamical changes of the muscle force can be observed without the need for individual calibration. When the force on the sensor tip is large, it causes deformation (stretching) of the intermediate layer and consequently diminishes the deepening of the indentation produced by the sensor tip”. This reduction increases the angle α between the muscle and sensor forces and reduces the sensitivity of the sensor, causing a certain amount of nonlinearity. Finally, if the force is great enough, the deepening would disappear altogether, and the sensor would come to saturation.

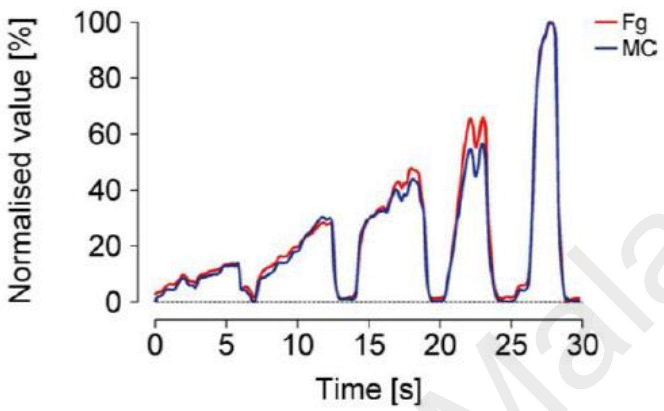


Figure 2.12 The Fg and MC variables are normalized to the maximal value (Dordević et al., 2011).

2.5.1 MC vs Force relationship

Based on previous study by Dordević et al. (2011), it shows that the strong relationship between Force (Fg) and MC and the strength of the linear relationship for all Fg and MC data (21 subjects able bodied) is measured.

The coefficient of determination (R^2) calculation was 0.85. Other than that, another study by Đorđević, Tomažič, Narici, Pišot, & Meglič (2014) tested the mechanical properties of a “Muscle Contraction” (MC) sensor during isometric muscle contraction in different length/tension conditions. The MC sensor was attached so that it indents the skin overlying a muscle group and detects varying degrees of tension during muscular contraction. They have compared MC sensor readings over the biceps brachii (BB) muscle to dynamometric measurements of force of elbow flexion together with recordings of surface EMG signal of BB during isometric contractions at 15° and 90° of elbow flexion. Statistical correlation between MC signal and force was very high at 15° (r = 0.976) and 90° (r = 0.966) across the complete time domain. In this work, we want to propose the use of Muscle Contraction (MC) sensor as the new and reliable fatigue indicator during FES stimulation.

A few related studies that used MC-sensor to measure muscle activity during their own respective experiment protocol tabulated in Table 2.4. It shows that the MC-sensor can effectively use in muscle measurement.

Table 2.4 Summary of MC-sensor study and respective finding

Author(s) (year)	Study	Emphasized finding(s)
(Meglič et al., 2019)	The Piezo-resistive MC Sensor is a Fast and Accurate Sensor for the Measurement of Mechanical Muscle Activity	The MC sensor readings were compared to tensiomyographic (TMG) measurements. The signals obtained by MC only partially matched the TMG measurements, largely due to the faster response time of the MC sensor.
(Krašna, Đorđević, Hribernik, & Trajkovski, 2017)	A Novel Approach to Measuring Muscle Mechanics in Vehicle Collision Conditions.	The observed differences in EMG and MC sensor collected signals indicate that the MC sensor offers an additional insight into the analysis of the neck muscle load and activity in impact conditions.
(Tanaka, Imawaka, Srdjan, & Tsunoda, 2016)	Evaluation of mechanical properties of knee extensor and patellar tendon during squatting by muscle contraction sensor method	Muscle activities of the knee extension during squatting showed a significant correlation to the MC signal. From these

		results, it is considered, that the method using an MC sensor is effective for evaluating the changes in dynamic tension in muscle-tendon complex during squatting
(Đorđević et al., 2014)	In-vivo measurement of muscle tension: dynamic properties of the MC sensor during isometric muscle contraction.	Statistical correlation between MC signal and force was very high indicates high linearity and good dynamic properties of MC sensor signal when compared to elbow flexion force
(Ng, Pourmajidian, & Hamzaid, 2014)	Evaluation of isokinetic muscle performance using a novel mechanomyogram sensor	A stronger association was observed between MC-sensor and Biodex, compared to MC-EMG and Biodex-EMG

2.6 Summary

Even there is a lot of research using new MMG based sensor; MC sensor in detecting muscle performance in different experiment protocol and environment, however, there is no previous research that studied the ability and sensitivity of the MC sensor in detecting electrically induced muscle fatigue. Muscle fatigue (i.e. a reduced force generating a capacity of muscle at an unchanged stimulation current) in persons with SCI still unresolved. Thus, this study aims for; if the MC sensor can predict muscle torque accurately, it might be located in daily functional tasks for neurological populations.

CHAPTER 3: METHODOLOGY

3.1 Introduction

This chapter describes the protocols and materials used in the study. It explains the participants' information, experimental setup, data collection, processing and analysis for SCI participants for two experiment protocols. Figure 3.1 shows the flow chart of the research methodology.

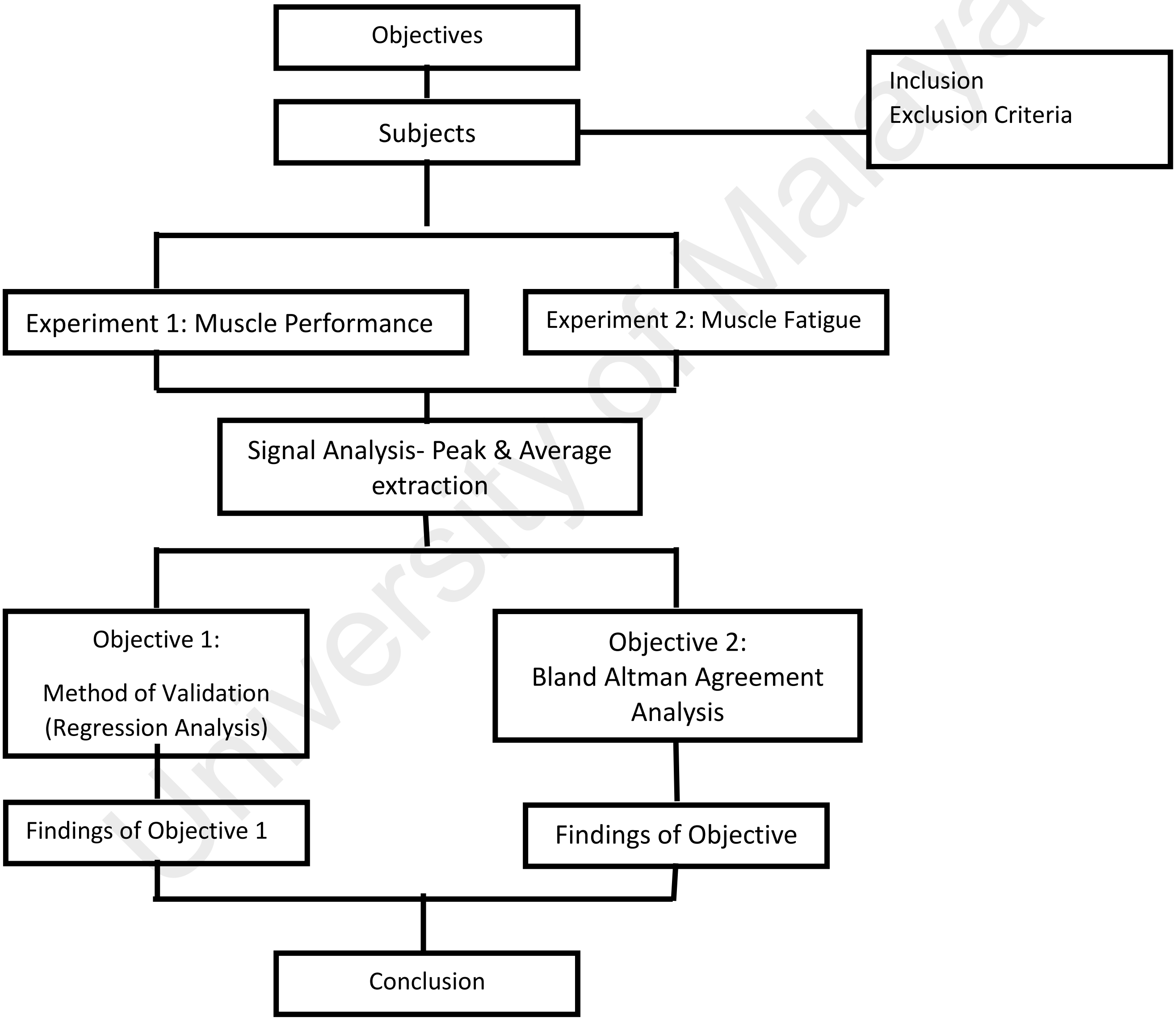


Figure 3.1 Research methodology workflow

3.2 Instrument

3.2.1 Stimulator and electrodes

Rehastim Stimulator (Reha-Stim Medtec GmbH &Co. Berlin, Germany) was used to induce electrical current to the quadriceps muscle belly of the subject's dominant leg. Some stimulation parameter of FES is set to constant throughout the study to maintain a consistent and clear response by eliminating its effect on the data collected as the parameter variability is not the focus of this study. Table 3.1 is the stimulation current variability and pulse width used in this research.

Figure 3.2 Rehastim- Stimulator used

Table 3.1 Stimulation Parameter Constant

Parameter	Value/Type
Current	Variable = 70mA, 80mA, 90mA 100mA,110mA
Pulse Width	200 µs

In this research, contraction in muscles is generated using short electrical pulses train shown in Figure 3.3.

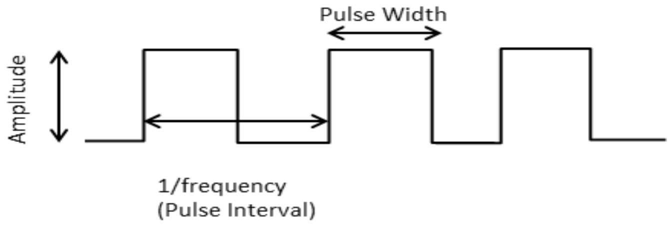


Figure 3.3 Stimulation Pulse Train

3.2.2 Electrodes

Based on Rushton (1997), "Electrodes are necessary to transduce electron current in leads into ionic current in tissues. They must do this without generating excessive amounts of toxic ionic species at the electrode–tissue interface, or excessive changes of local tissue pH. Electrodes are used to aim the stimulating current at the desired tissue, avoiding other excitable tissues. For example, only the longitudinal component of axonal current will stimulate nerve fibres, so the direction of current in the volume conductor is a crucial variable. Obviously, two electrodes are required to complete the current circuit, but when one of them is placed close to the excitable tissue and the other is remote the stimulation arrangement is referred to as monopolar. When both are close, the arrangement is called bipolar. Bipolar stimulation results in better current direction. However, multichannel stimulation requires more electrodes to be placed, with more electrode leads; monopolar channels can share the same return electrode and lead. For FES, particularly when using surface electrodes, the fewer leads the better. There are many types of electrodes that commonly used in FES fields. FES applied via transcutaneous electrodes is a common rehabilitation technique for assisting grasp in patients with central nervous system lesions.

Thus, in this study, two transcutaneous gel-backed electrodes (5cm x 9cm) were used as neuromuscular stimulating electrodes over the subjects' quadriceps muscles.

(a) (b)

Figure 3.4 Electrode used for quadriceps muscle (a) in sealed bag

The stimulating electrodes were connected to the FES system. Electrode (Figure 3.4) used and sizes were fixed for all subjects. Measuring electrode used in this research is the Muscle Contractions (MC) sensor electrode. The MC sensor used to measure the output of the subject muscle activities.

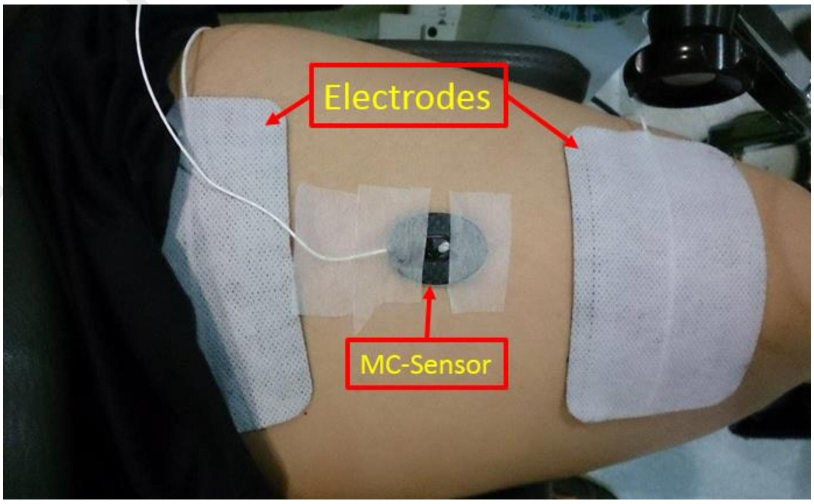


Figure 3.5 Placement of sensor and electrodes

3.2.3 Biodex Dynamometers Machine

Figure 3.6 Biodex Dynamometers system. Retrieved from

Biodex Medical Systems to capture the torque produced during FES contraction. This device is equipped with very friendly user software and makes this plug and play device easy to handle. The output data was captured using 1kHz sampling rate but the data generated on the log file is at 100Hz. By setting the testing mode in isometric, the leg will be fixed in the angle position where any torque produced during FES contraction will exert force on the machine and the value was captured. The system is provided with a computer (IBM) compatible device that collects displays, stores the data and control the movement of the dynamometer.

3.3 Research Design

The steps of the test were explained for each subject to allow the subject to be oriented and familiar with the testing protocol. Calibration of the unit was performed prior to use according to the manufacturer guidelines.

3.3.1 Subject

Nine healthy spinal-cord injured (SCI) individuals (44.6±13.4 years of age) with complete motor and sensory lesion, American Spinal Injury Association Impairment Scale (ASIA) A or B below T7 and T11 neurological segment volunteered to participate in this study. Exclusion criteria are subjects with uncontrolled hypertension, unstable angina pectoris or other medical conditions that would interfere with testing (Table 3.2). Subjects should be able to follow simple commands, their intelligence within the level that enables the subjects to follow instructions of the researchers.

Experiment protocol was approved by University of Malaya Ethics Committee (Ref No.:1003.14(1)) and all participants were provided with written informed consent. The participants recruitment was done by physiotherapist and the participant also have been participated in various studies that investigated muscle fatigue effect in FES-evoked knee extension and standing among people with SCI (Dzulkifli, Hamzaid, Davis, & Hasnan, 2018; Ibitoye, Hamzaid, Hasnan, et al., 2016). The subject number selected in this study was based on other studies with similar objectives, who recruited between 4 to 19 subjects (**8±3.9**), which agrees with our population sample size (Table 3.3) (Deley, Denuziller, Babault, & Taylor, 2015; Karu, Durfee, & Barzilai, 1995; Olusola Ibitoye et al., 2016).

The estimation of sample size also been determined by using G-Power Analysis Software (Faul, Erdfelder, Lang, & Buchner, 2007). In order to achieve high coefficient of determination of at least 0.85 (*MC-sensor's previous study results*), thus, to produce reliable results, effect size of 0.92 was needed to determine the sample size required. 0.80 used as power of correlation, so that if the experiment is repeat with different subjects, there will be 80% probability the results produced be similar. From the following parameters, GPower Analysis determined minimal required value for sample size is 3 subjects.

Table 3.2 Exclusion and inclusion criteria for SCI participants

Inclusion Criteria	Exclusion Criteria
Outpatient, 1 year and above	Skin allergies to electric stimulation or electrodes
Age 20-60 years old.	uncontrolled hypertension
SCI C5 and above, ASIA A & B	unstable angina pectoris
Able to understand simple instructions	Morbid obesity (BMI > 40)
	Pre-existing bone conditions (e.g. fractures, osteoporosis)

Table 3.3 SCI Study Group Demography

Subject	Age (year)	Gender	Height (m)	Body Weight (kg)	Spinal Lesion Level	Injury Classification (AIS *)	Years Since Injury (year)
1	28	M	1.71	62.4	C7	B	14
2	35	M	1.61	81.0	T10–T11	A	18
3	59	M	1.79	71.6	C6–C7	B	2.5
4	47	F	1.62	82.0	T4	B	24
5	58	M	1.73	80.0	C5–C7	B	4
6	33	M	1.71	44.0	C5–C6	A	13
7	59	M	1.72	60.5	T10–T11	A	13
8	34	M	1.70	75.9	C6	A	17
9	21	F	1.60	48.0	T10	B	1

3.3.2 Experimental Protocol

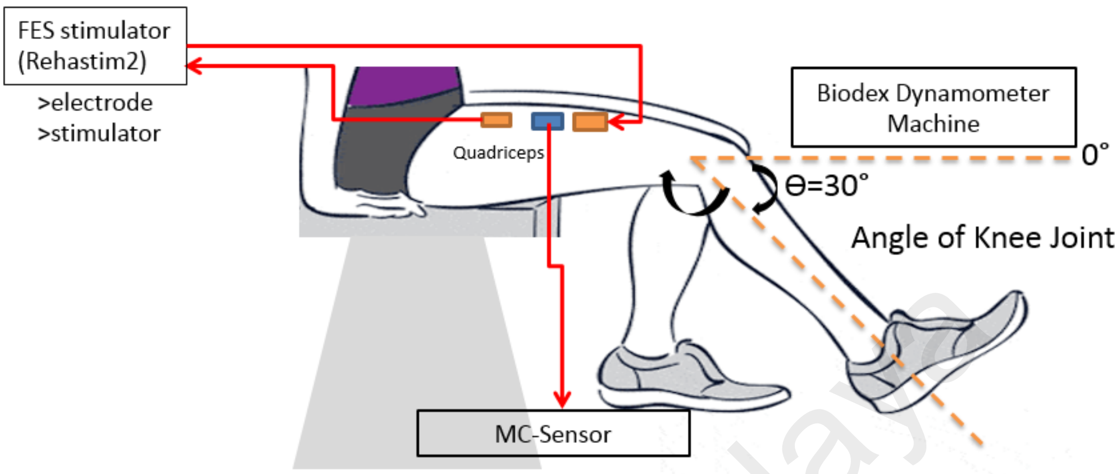


Figure 3.7 Instrumentation for data collection of voltage from MC sensor during FES contraction and torque

The measurement was performed on the quadriceps muscle for: muscle performance, and muscle fatigue monitoring using MC-sensor during FES activity. Stimulation parameters were set constant (200 µs pulse width, 35 Hz frequency) and the induced current would vary based on the experiment protocol. Two transcutaneous gel-backed electrodes (5cm x 9cm) were used as neuromuscular stimulating electrodes over the subjects' quadriceps muscles and were connected to the FES output stage. The MC-sensor was placed on the skin surface above the targeted skeletal muscle using double sided adhesive patches which enabled fixation of sensors to the skin of the measured subject. The sensor tip is the important part of the sensor as once it compressed on the skin surface, the skin above the muscle will be a dent, and the muscle force, in turn,exerts back to the sensor tip. Thus, the force measured by the sensor tip is proportional to muscle tension.

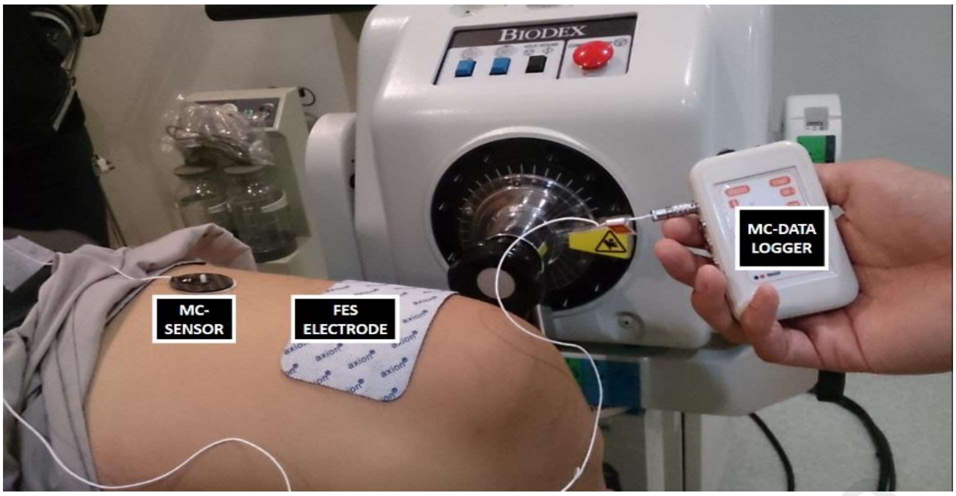


Figure 3.8 SCI individual seated at Biodex dynamometer to measure isometric knee extension torque as muscle performed during FES-evoked contraction and the placement of neuromuscular stimulating electrodes and MC-sensor on subject's quadriceps muscle.

The SCI subject was seated on Biodex Dynamometer System 4 Pro (Biodex Medical Systems, USA) with the back supported, and knee joint angle at 30° (Figure 3.7). The measurement was performed on the quadriceps muscle (Figure 3.8**)** for muscle performance and muscle fatigue monitoring using MC-sensor during FES activity. Each subject participated in two experiments. The first one aims to investigate the correlation of muscle evoked torque and MC-sensor measurements. This is important to show whether the MC is responsive to different levels of muscle performance in SCI individuals. The objective of the second experiment was to investigate the sensitivity of MC as muscle fatigue sensor. In this experiment, the reducing muscle torque throughout a session of FES-evoked isometric contraction was correlated with the MC sensor measurements. The instruction for the participants as in below.

3.3.2.1 Experiment 1: Muscle performance

In this case, the FES stimulation current level was set to be variable. The testing protocol is illustrated in Figure 3.9. There were 5 contractions during one session with the 30s to

60s duration separating the consecutive bursts. The participants was to perform 5 sessions in which different values of current level (in mA) were delivered in each contraction for 2 seconds with 30 seconds rest in random orders in each session. Randomization of stimulation current order was done to prevent the selection bias and ensured against the accidental bias. Thus, when fatigue occurs during Experiment 1 the fatigue would most unlikely be due to the different FES current levels, but rather due to the prolonged stimulation.

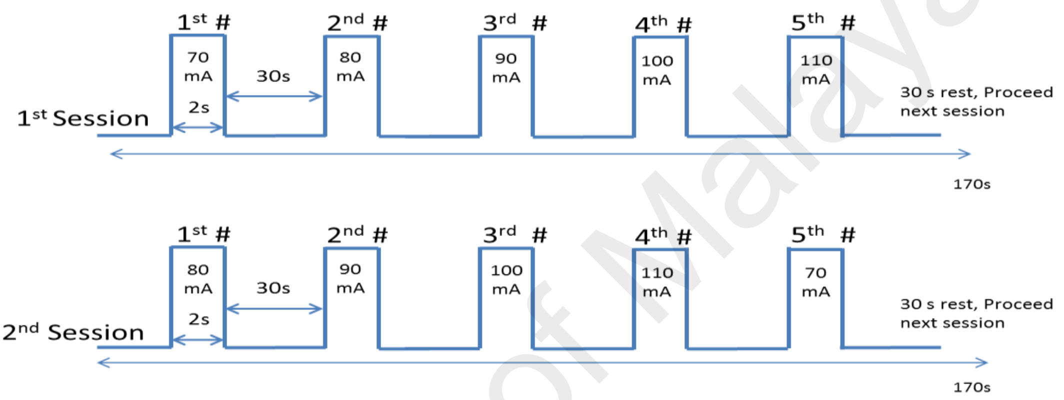


Figure 3.9 Diagram illustrating the muscle performance protocol. # indicates 'contraction'

3.3.2.2 Experiment 2: Muscle Fatigue

After 15 minutes rest from experiment 1, the subjects performed up to 30 consecutive 4 seconds isometric contractions at 30° knee joint angle on the Biodex 3 dynamometer with 10 seconds rest and 100mA constant value of current stimulation throughout.

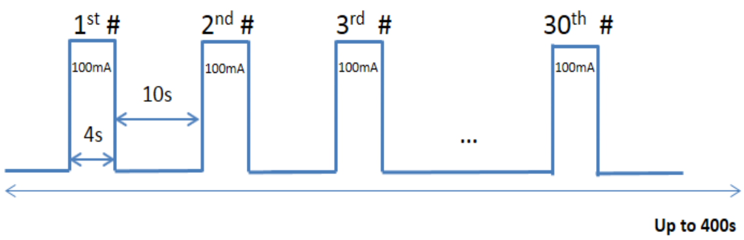


Figure 3.10 Muscle fatigue measurement. # *indicates 'contraction'*

The testing protocol is illustrated in Table 3.4.

Table 3.4 Schema illustrating the muscle stimulation current protocol (mA)

Session / Set	1	2	3	4	5
1	70	80	90	100	110
2	80	90	100	110	70
3	90	100	110	70	80
4	100	110	70	80	90
5	110	70	80	90	100

3.4 Data Analysis

The measured muscle tension produces the force F in the direction along the muscle surface. This force causes the intermediate layer and skin to press on the sensor tip. The vector sum of all forces produces the force Fs in the direction along the sensor tip and α is the angle between the directions if F and Fs. Collected measurements data from the MC date logger were analyzed and exported using Sensmotion software (TMG-BMC Ltd., Ljubljana, Slovenia). Changes of muscle tension during FES-evoked contractions were measured, and the difference in muscle contraction intensities resulted in different amounts of muscle tension. The MC signal was sampled at 5 kHz and Gaussian-filtered with cutoff frequency 10kHz (Đorđević, Stančin, Meglič, Milutinović, & Tomažič, 2011).

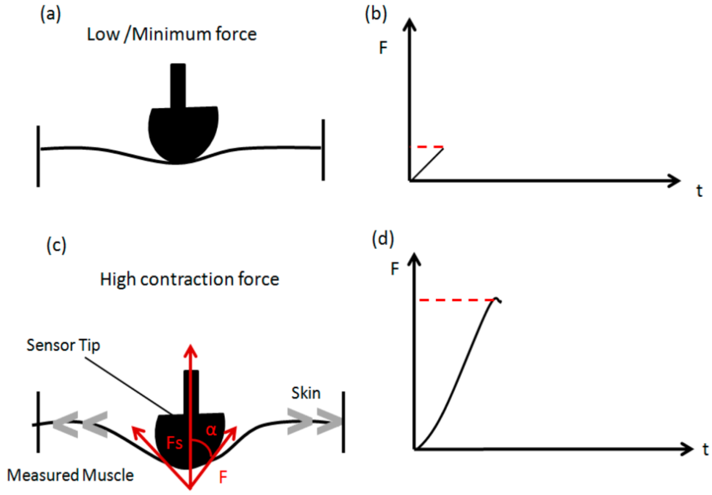


Figure 3.11 Working principle of the MC sensor in detecting variation in muscle force during contraction. (a,b) Measured force signal during low contraction (c,d) Measured force signal during high contraction.

3.4.1 Statistical Analysis

From the Biodex dynamometer, two variables were extracted to characterize muscle performance. Dynamometer peak torque was the highest knee extension torque observed in each muscle contraction (Figure 3.12). Dynamometer average torque was expressed as the 'area under the torque-time curve' for each muscle contraction (Figure 3.12). From the MC-Voltage for each FES-induced muscle contraction, two variables were derived—Signal Peak (SP) and Signal Average (SA).

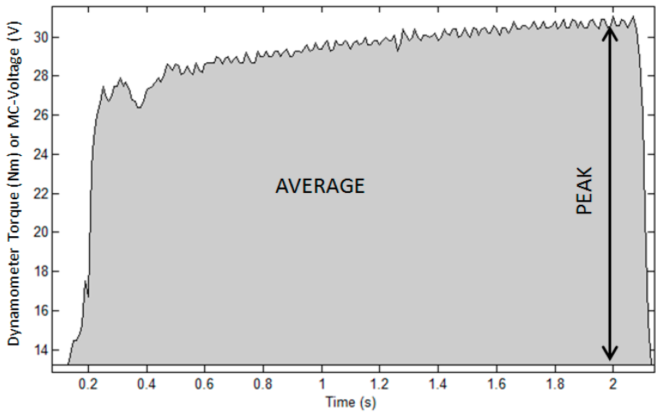


Figure 3.12 Schematic representation of a single FES-muscle contraction for derivation of dynamometer peak torque and average torque.

SP was the highest amplitude of the MC signal (V), and SA was the area under the voltage-time curve for each muscle contraction. These were investigated as possible proxies for dynamometer peak torque and average torque, respectively. Then, to derive a prediction equation between SP and SA from MC-voltage, and the respective dynamometer peak and average torques (N·m), "bootstrapping" statistical analysis was performed. Initially, 40% of the dataset, stratified by current amplitude (on Table 3.3), was randomly selected to derive an initial relationship for each individual fitting the linear regression:

$$\mathbf{T}_{(SP/SA)} = \beta_0 + \beta_1 \cdot MC_{(SP/SA)}$$

where:

- T = dynamometer torque (N·m)
- β_0 = y-intercept
- β_1 = slope of the line
- MC = sensor output (V)
- SP = Signal Peak
- SA = Signal Average

Hence, for everyone's regression equation, muscle torques (predicted MC torque) was estimated from MC-sensor voltages from the whole data set, for each subject.Before descriptive statistical analyses and regression modeling were performed, the dyna-torque and MC voltage data were normalized to each individual's maximum muscle stimulation contraction for the simultaneous analysis of dyna-torque and MC voltage signals.

Then, the mean absolute percentage errors (MAPE) between dynamometer measured torque and MC-predicted torque were calculated:

MAPE (%) = mean difference dyna-torque and predicted MC torque × 100/mean dyna-torque.

3.4.2 Method of validation

To determine the relationship between the torque produced by biodex and the measured MC-signal during both experiments, the signal peak (SP) and Signal Area (SA) was compared with each contraction. The strength of the linear correlation between isometric knee extension torque and MC was evaluated with a Correlation coefficient (r). High values ($0.97 \leq r \leq 1$) Correlation coefficient (r) indicates strong relationship. Consequently, the degree of association between isometric knee extension torque and predicted MC torque SP/SA was evaluated with the Coefficient of Determination (R^2).

3.4.3 Method of agreement

To investigate the level of agreement between the SP/SA dyna-torque and SP/SA-predicted MC torque, Bland-Altman plots were constructed with 95% confidence limits of agreement and both bias (mean difference) and Lin's concordance correlation coefficient (ρ_C) were calculated. All data were analyzed and plotted using Microsoft Excel (Microsoft Corporation, Redmond, WA, USA) and SigmaPlot 13.0 (SYSTAT Software Inc., San Jose, CA, USA).

CHAPTER 4: RESULTS

This chapter presents all the findings of the study. The results are presented in graphs and tables. This chapter are divided into two main sections associated with the correlations between

4.1 Pre-processing MC data

The elimination of noise signal of MC is using low pass Gaussian-filter function in Matlab R2010 with cutoff frequency 10kHz. From the results, the filtered data is cleaner and easier to analyze.

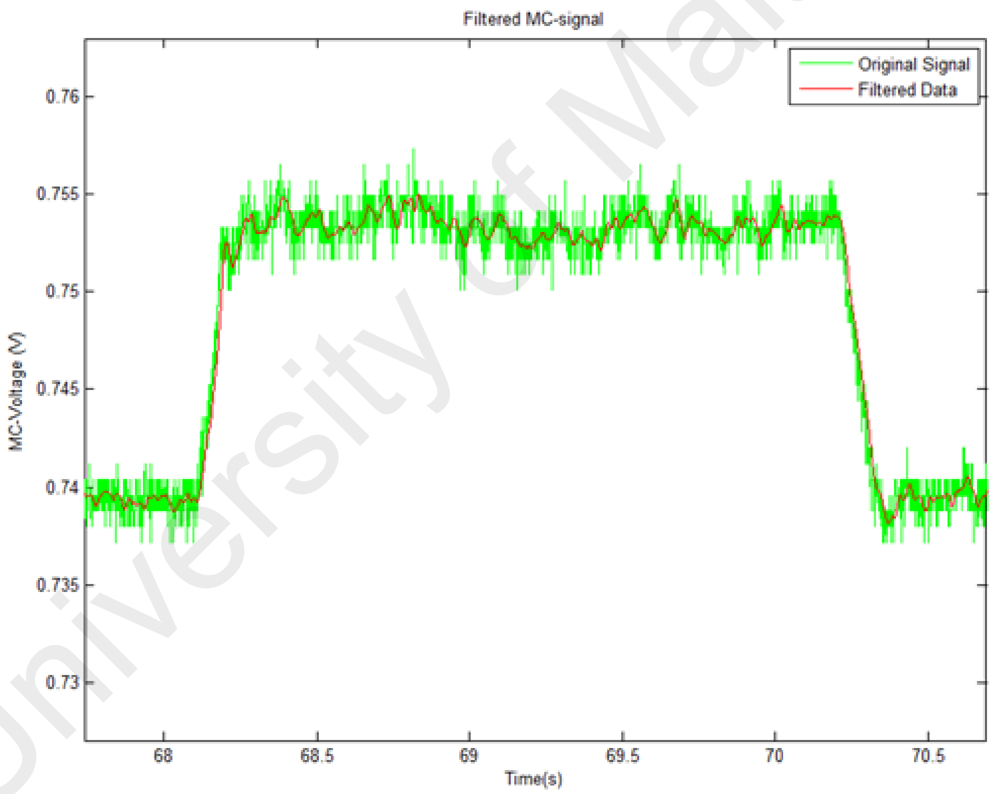


Figure 4.1 Elimination of noise signal of MC using filters

4.1.2 Signal Recording of MC-sensor and Dynanometer

The recording of MC-signal and isometric knee torque was recorded simultaneously. The simultaneous recording of normalized MC and Torque (Figure 4.2) indicates that the amplitude of MC rises with the rise of the evoked muscle torque. From the simultaneous recording, it shows that when the muscle contract as induced by FES, the MC can record any change in tension in the targeted muscle as the knee extended and produced torque.

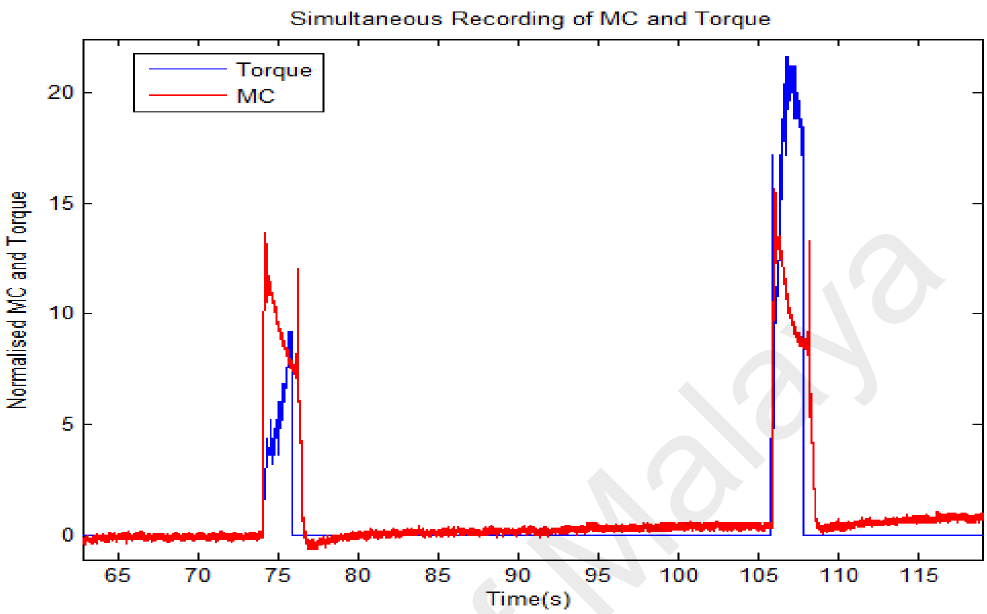


Figure 4.2 Simultaneous recording of MC-signal and isometric knee torque during FES-evoked contraction

The signal outcome of Experiment 2, which investigated muscle fatigue, are as described next. Figure 4.3 shows the mean values of normalized MC and normalized torque in 9 SCI participants and their 30 repetitive contractions. The results showed the amplitude value of MC and torque decreases throughout the repetitive FES-evoked muscle contraction. The correlation values of MC-signal and knee extension torque are presented in Table 4.1.

Simultaneous measurement of exemplar normalized dyna-torque and normalized MC-voltage signals revealed that the amplitude of MC rose concurrently with the increase of FES-evoked muscle torque (Figure 4.3). When the muscle contraction was elicited by the neuromuscular stimulator, the MC recorded a change in tension rectus femoris muscle around the knee joint. One participant's data was not included in the statistical analyses, because of ‘outlier’ findings, however this data have been deliberated upon in the Discussion section.

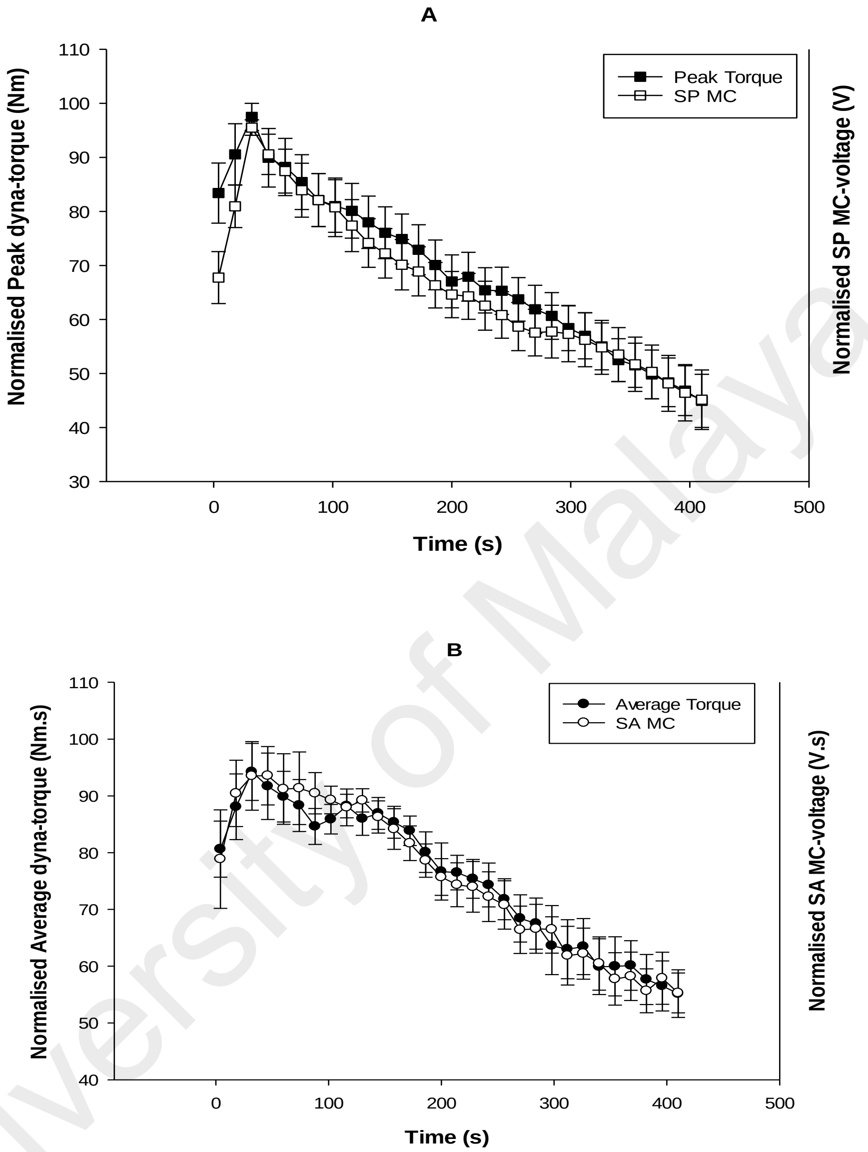


Figure 4.3 Rectus femoris fatigues in SCI participants represented by their mean normalized dyna-torque and mean normalized MC voltages during FES-evoked contractions (A) Peak dyna-torque and SP MC voltage throughout 30 contractions over 400 s; (B) Average dyna-torque and the SA MC-voltage throughout 30 contractions over 400 s. Data are mean ± standard error.

4.2 Method of Validation

Figure 4.4 and Figure 4.5 shows the linear regression of peak dyna-torque and the MC-torque SP with 95% confidence bands; and linear regression of average dyna-torque and the MC-torque SA with 95% confidence bands; respectively during muscle performance (Figure 4.4) and muscle fatigue (Figure 4.4). The derivation of these variables was described in Figure 3.12 in the methodology section.

Table 4.1 Muscle performance and muscle fatigue analysis for signal peak and signal average relationships of dyna-torque and predicted MC torque.

		N	***p***	***R^2***	***SEE***
Performance					
	SP	225	<0.0001	83%	4.53
	SA	225	<0.0001	80%	17.17
Fatigue					
	SP	304	<0.0001	64%	6.35
	SA	304	<0.0001	64%	30.15

N—Total; *p*—*p* value; R^2—Correlation of determination; SEE—Standard Error Estimate.

The relationship of MC-signal and Torque are presented in Table 4.1. In Figure 4.4 and Table 4.1, a comparison between the measured peak/average dyna-torques and the predicted SP/SA MC-torques were shown to have a strong linear relationship with a high degree of association (peak dyna-torque with SP MC-torque r=0.91, p<0.0001; average dyna-torque with SA MC-torque r=0.89, p<0.0001). Figure 4.4 also revealed that the MC sensor could measure muscle contraction values at low stimulation currents even when no dyna-torque could be recorded.

4.2.1 Muscle Performance

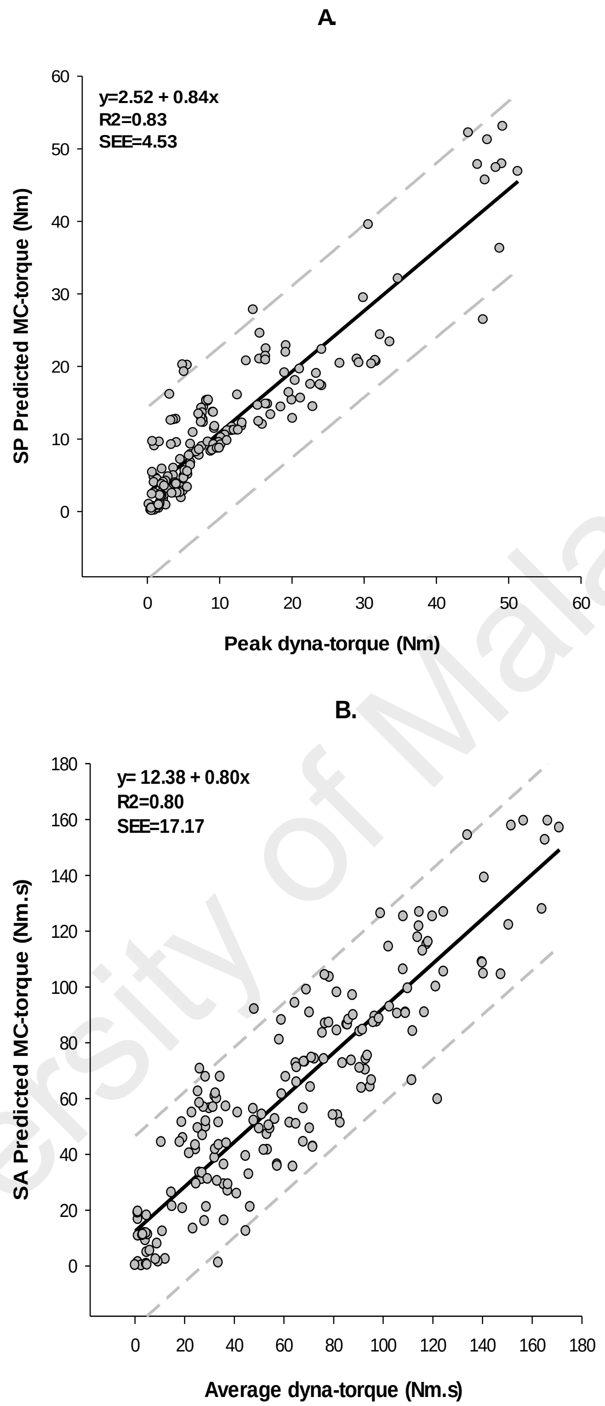


Figure 4.4 (A) Linear regression scatter plots of peak dyna-torque and the MC-torque SP with 95% confidence bands; B) Linear regression scatter plots of average dyna-torque and the MC-torque SA with 95% confidence bands.

4.2.2 Muscle Fatigue

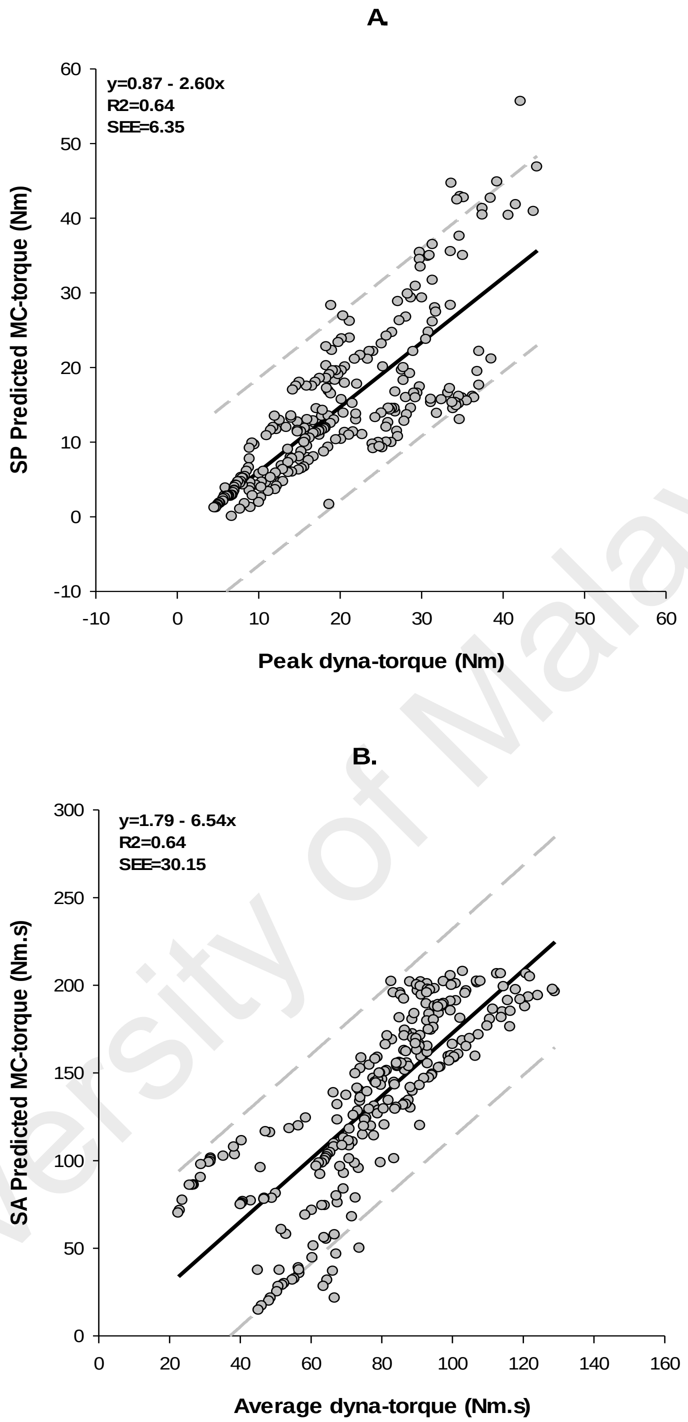


Figure 4.5 (A) Linear regression scatter plots of Peak dyna-torque and the SP predicted MC-torque with 95% confidence band. (B) Linear regression scatter plots of Average dyna-torque and the SA predicted MC-torque with 95% confidence band.

The high correlation between dynamometer peak torques during fatiguing muscle contractions and SP MC predicted torques (Figure 4.5A; $r = 0.80$; $p<0.0001$), and between dynamometer average torques over time and SA MC torques (Figure 4.5B; $r=0.77$; $p<0.0001$), suggested the MC sensor well-represented FES-evoked muscle fatigue, acting as a close proxy for dynamometer measurements.

4.3 Bland Altman as Agreement Analysis

Level of agreement between the SP/SA dyna-torque and SP/SA-predicted MC torque, Bland-Altman plots were constructed with 95% confidence limits of agreement and both bias (mean difference) and Lin's concordance correlation coefficient (ρ_C) were also plotted in Figure 4.6 and Figure 4.7 during muscle performance and muscle fatigue experiment, respectively. The agreement of MC and Dynanometer technique are presented in Table 4.2.

Table 4.2 Muscle performance and muscle fatigue analysis for signal peak and signal average relationships of dyna-torque and predicted MC torque.

		N	***Bias***	ρ_C
Performance				
	SP	225	0.69	0.91
	SA	225	-0.03	0.89
Fatigue				
	SP	304	-5.27	0.70
	SA	304	55.62	0.30

N—Total; *Bias— mean difference*; ρ_C —Lin's Concordance correlation coefficient

4.3.1 Muscle Performance

As Bland-Altman plots portrayed in Figure 4.6 (A-B) and Table 4.2 the measured peak and average dyna-torques and predicted SP/SA MC-torques had small bias and standard deviation (SD), which were SP: bias =0.69; SD=4.92, SA: bias =-0.03; SD=19.14, and moderate-high concordance (ρC). MAPE was in the moderate range.

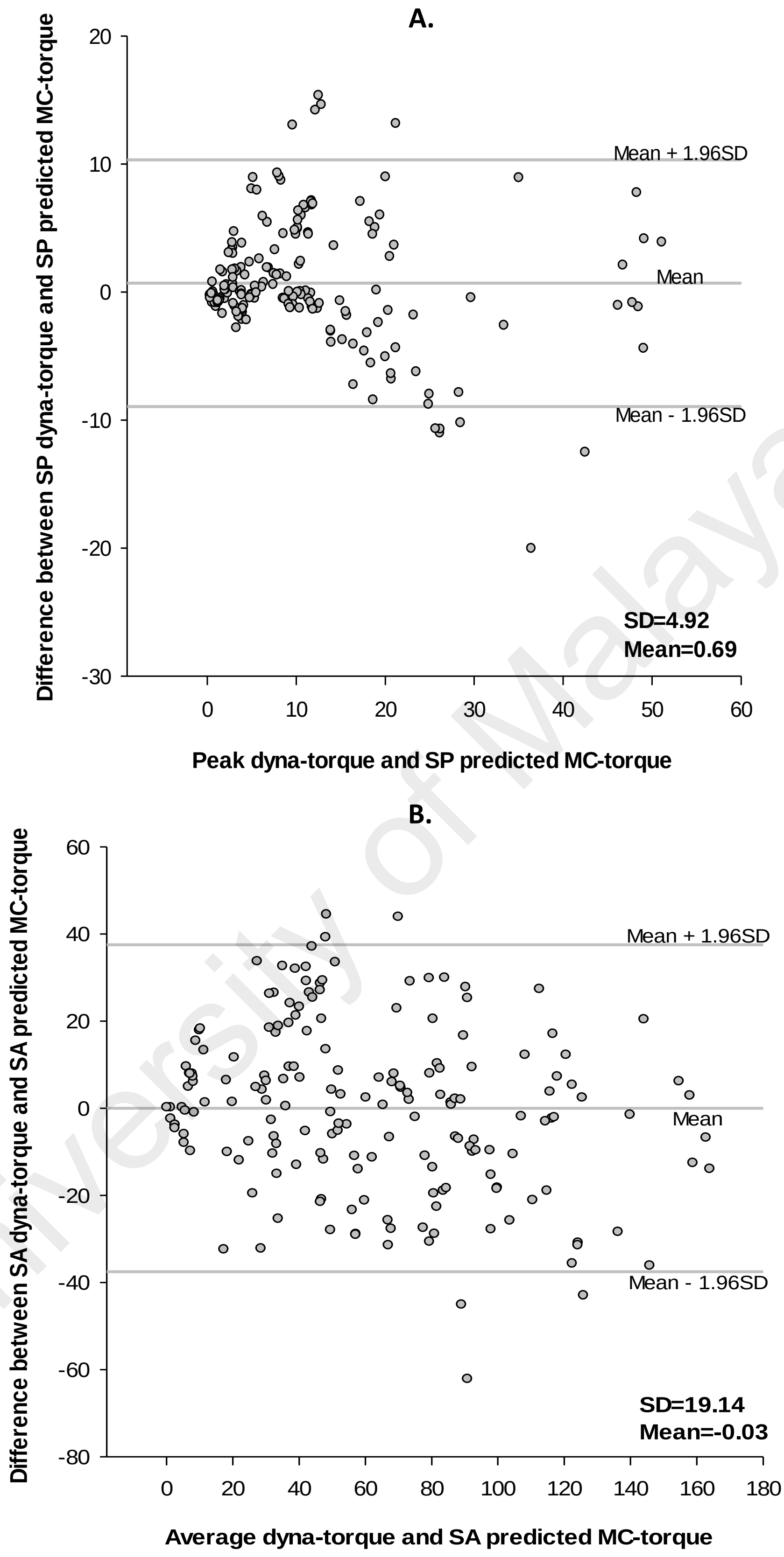
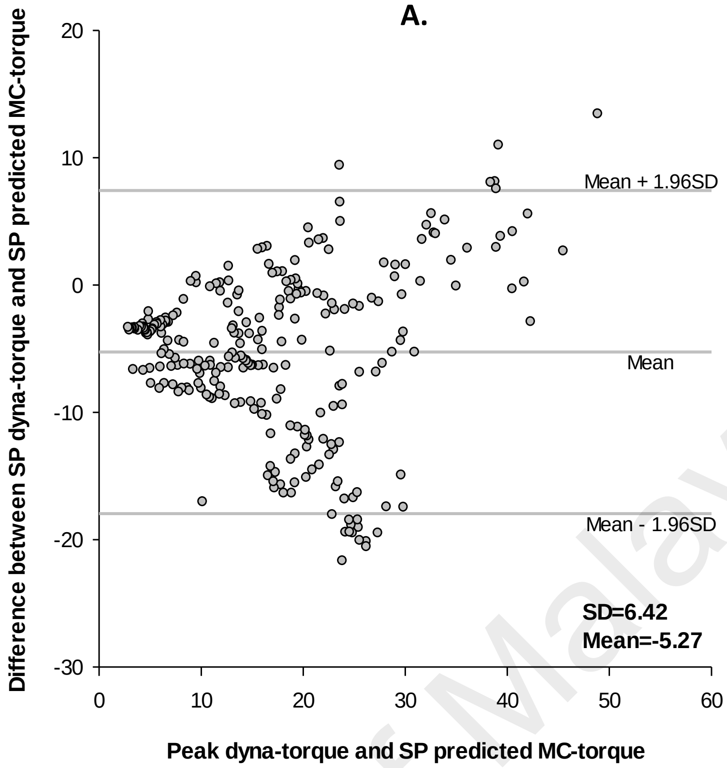


Figure 4.6 (A) Bland-Altman plots of peak dyna-torque and the Predicted MC-torque SP; (B) Bland-Altman plots of average dyna-torque and the Predicted MC-torque SA.

4.3.2 Muscle Fatigue



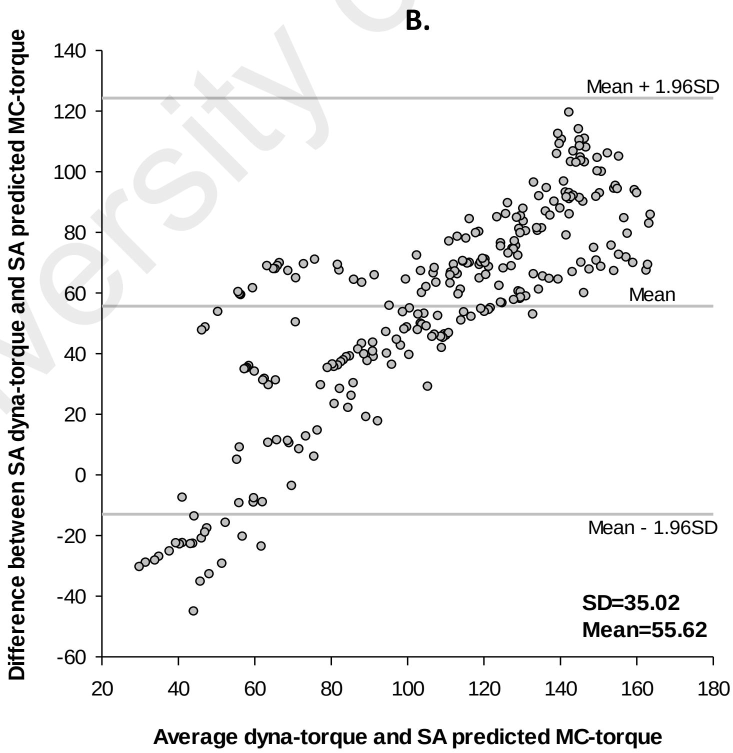


Figure 4.7 (A) Bland–Altman plots of Peak dyna-torque and the SP predicted MC-torque. (B) Bland–Altman plots of Average dyna-torque and the SA predicted MC-torque

From the Bland-Altman plots (Figure 4.7(A-B)) and Table 4.2, the measured peak/average dyna-torques and predicted SP/SA MC-torques with bias and standard deviation, were SP: 5.27 (6.47); SA: 55.62 (35.02), and fair-moderate concordance (ρ_C). MAPE was in the fair-moderate range. Some 'trend-biases' were visually noted in BA analysis of peak/average dyna-torques and the respective SP/SA MC estimated torques during the muscle fatigue trial.

CHAPTER 5: DISCUSSION

This chapter discusses all the findings in the study. It is divided into two sections; Method of Validation and Method of Agreement (Bland Altman). Based on the objective of the study, the present study sought to:

1) To investigate MC sensor responses versus muscle torque measurements during prolonged, electrically evoked, fatiguing muscle contractions in individuals with motor-complete SCI.

2) To determine degree of agreement between body-worn MC-sensor with directly measured isometric knee extension torques derived from a commercial dynamometer in order to act as proxy to each other.

Our study essentially found that Spinal Cord Injured individual's muscles can respond vigorously to electrical stimulation, however, quickly leads to fatigue (Figure 4.3). This agrees with other studies in the literature, which also investigated prolonged FES-evoked contractions in individuals with SCI. During their post-SCI rehabilitation FES exercises may lead to rapid fatigue, whereby their muscles are not able to maintain a constant work production due to decrease in muscle force (Enoka & Duchateau, 2008; T. A. Thrasher et al., 2004). The condition of rapid onset muscle fatigue may even lead to muscle damage (Dugan & Frontera, 2000). Even though there are other muscle fatigue monitoring techniques including dynamometer, EMG, NIRS and other force sensors currently available, the MC sensor may be a practical choice for clinicians due its reliability and ease of use compared to other techniques (M. O. Ibitoye, Hamzaid, Abdul Wahab, et al., 2016; M. Tarata et al., 2001). Such use of MC-sensor has been previously proposed by Dordevic and colleagues (Dordević et al., 2011) who demonstrated a strong correlation between MC voltage and force in 21 able-bodied subjects (R^2=85%).

However, the MC sensor as fatigue proxy, especially during FES-evoked contractions in SCI patients, has not been previously investigated. Thus, the findings of this current study are the first application of MC-sensor for estimating muscle peak forces and muscle fatigue rates in a neurological population. Figure 4.3 shows the incerement in the dynamometer torque and MC-voltage in first 3 initial contraction (meaning an increase in strength), and the rapid decrease of torque values observed in the results (fatigue). This is because it takes a short period of time for muscles of paraplegic to reach their peak force generating capacity under FES.This kind of results also shown in previous studies (Adam, Geoffrey, & Milos, 2005; Faller, Nogueira Neto, Button, & Nohama, 2009; Phinyomark, Angkoon; Thongpanja, Sirinee; Hu, Huosheng; Phukpattaranont, Pornchai; Limsaku, 2012).

5.1 Method of Validation

This study sought to determine the sensitivity and accuracy of MC as muscle contraction sensor and determine whether it might be a good fatigue proxy for FES-evoked muscle contractions. It was observed that even though the start and end time as well as duration of MC and torque signal were the same, the peak values of MC and torque were not simultaneous. The findings reported a strong dyna-torque vs predicted MC-torque relationship, with r = 0.91 (R^2=83%) in muscle performance over a range of FES current amplitudes and r = 0.80 (R^2=64%) under conditions of progressive muscle fatigue.

This linear relationship suggested that the MC sensor could be used as a suitable proxy for muscle tension of individuals with SCI during FES rehabilitation training and demonstrated a satisfactory representation of muscle performance during progressive fatigue. While the signal was highly correlated with the measured dyna-torque, there was some variability that might have been due to the skin and underlying adipose tissue of different individuals. Additionally, in the fatigue experiment, some subjects' dynamometer measurements could not be detected, yet the MC sensor still recorded skin-muscle deformation of muscle contraction. A possible cause may have been the thickness of skin tissue or that some individuals with SCI might have fatigued earlier than others, thus no contraction torque could be detected by the Biodex dynamometer.

5.2 Method of Agreement: Bland-Altman

Traditional regression analyses revealed a strong linear correlation between dyna-torque and the predicted MC-torque. However, this does not necessarily translate into a strong case for agreement between the two techniques. Thus, Bland and Altman (J.M.Bland & D.G.Altman, 1986) analyses were performed to quantify the degree of agreement between the two methods of measuring muscle contraction (i.e. using dynamometer and MC-sensor).

There was a strong agreement between of dyna-torque and predicted MC-torque during muscle performance over a range of FES current amplitudes, with ρ_C= 0.91 for SP and ρ_C= 0.89 for SA, respectively. Figure 4.6 (A-B) illustrated the difference in muscle contraction torque measured by a dynamometer versus predicted by the MC-sensor. Across the range of muscle forces measured in these subjects, the mean difference (SP/SA dyna-torque – SP/SA predicted MC-torque) were 0.69 for SP and -0.03 for SA. Most of the measured muscle contractions were clustered close to the mean difference line, and lay within the 95% limits of agreement, suggesting a strong trend for agreement between dyna-torque and predicted MC torque (Bland & Altman, 2003).

Indeed, the limits of agreement for both SP and SA were narrow enough for one to be confident that the small body-worn MC-sensor could be used in the place of a large laboratory-based dynamometer for measuring muscle contractions within a clinical environment. Under conditions of progressive muscle fatigue, the agreement analysis between the peak/average dyna-torque and SP/SA predicted MC-torque revealed ρ_C = 0.70 for SP, but was quite low for SA at ρ_C= 0.30. Even though the measured contractions were clustered within ±1.96 SD of mean difference (Figure 4.7B), the 95% limits of agreement were much wider for SA (between -13.01 and 124.26 N.m) crossing cipher which suggested that that high intra-individual and between-subject variability rendered the MC-Sensor less acceptable to estimate peak and average torques during muscle fatigue, in our sample of SCI individuals.

In addition, visual inspection of the BA analyses for the fatigue trials (Figure 4.7A and 4.7B) revealed "fan-spread bias" and "linear-bias" in the data. This suggested that the difference between measured dyna-torque and predicted MC torque during the muscle fatigue trial (comprising thirty FES-induced consecutive 4s isometric contractions) varied with the magnitude of torque output.

5.3 Mean Absolute Percentage Error (MAPE)

The MAPE measured the prediction "accuracy" of signal peak (SP) and signal average (SA) from MC-torque. With MAPE ranging from 49%-77% during muscle performance and muscle fatigue experimental trials, there were moderate-good levels of prediction for dyna-torque in these individuals. These findings suggested that the small MC sensor might be more sensitive under low FES current amplitudes and low-torque, a conventional laboratory dynamometer in recording evoked muscle contractions in persons with SCI. The findings also revealed that the MC-sensor might also be used as a muscle fatigue monitor (but noting the 'bias' caveat previously stated), adding its utility to the other commonly used sensors of fatigue such as sEMG and vibromyography

(Chesler & Durfee, 1997; Mizrahi et al., 1994). Since MC sensor measurements are based on skin-muscle tension, its detection threshold was low enough to quantify ‘weak’ contractions (Figure 4.4), even under conditions of low torque muscle activity when the dynamometer could not.

In this current study, one of the 9 SCI participants had a history of neuromuscular ‘spasticity’. This individual demonstrated ‘different’ predicted muscle torque based on the MC sensor from the fatigue trial (Part 2) than from the performance trial (Part 1). This ‘outlier’ finding suggested an additional caveat, whereby the MC sensor may lead to inaccurate estimates of “true” muscle forces in individuals with particular muscle conditions such as spasticity, rigidity or contracture. Figure 5.1 illustrated where this particular subject’s predicted MC data during the fatigue trial portrayed a slope much steeper than the other subjects. The reason might have been muscle stiffness or torpor, which may have been due to lack of regular stretching and/or FES-training before the experiment was conducted. Yet, even this ‘unusual’ MC relationship managed to distinguish and estimate the difference in low torque production.

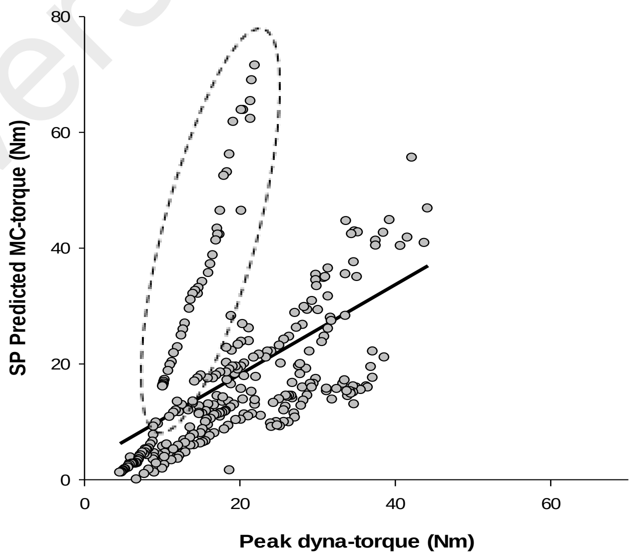


Figure 5.1 Regression analysis between Peak dyna-torque and SP predicted MC during the muscle fatigue experiment; includes 1 subject (data in dashed circle).

5.4 Advantages of MC Sensor

One advantage of the MC-sensor over other muscle sensors is that MC data is relatively easy to analyze because its measurement is based on muscle tension changes, the signal pattern correlates directly with the instantaneous torque profile (portrayed in Figure 4.2) compared to other MMG signals (M. O. Ibitoye, Hamzaid, Hasnan, et al., 2016).

Additionally, the MC signal is not hampered by any electrical properties in contact between the skin and the electrode (Al-Mulla et al., 2011; M. Ibitoye, Hamzaid, Zuniga, Hasnan, & Wahab, 2014). Tarata and co-workers reported that the mechanical components of the muscle (such as torque and contraction speed) are more closely related to the muscle function than to its electrical characteristics. Thus, the measurement of muscle activity does not require such a complex laboratory setup as the Biodex dynamometer, and the SCI users' muscle performance could be monitored over a wider range of daily rehabilitation and physical therapy tasks with the MC-sensor.

5.5 Effects of Force-Velocity Alteration on Measurements of Isokinetic Voluntary Muscle Performance

A study was conducted on the evaluation of Isokinetic Voluntary Muscle performance using the MC sensor where thirteen healthy male subjects performed a series of isokinetic knee extensions at nine different speeds. These velocities were selected to assess the sensor discrimination of muscle forces at "strength speeds" (0-120 °/s), endurance speeds" (180-300 °/s) and "power speeds" (400, 500 °/s). Surface EMG and Biodex Isokinetic dynamometer torque output were collected as a comparative reference to evaluate the reliability of MC-sensor.

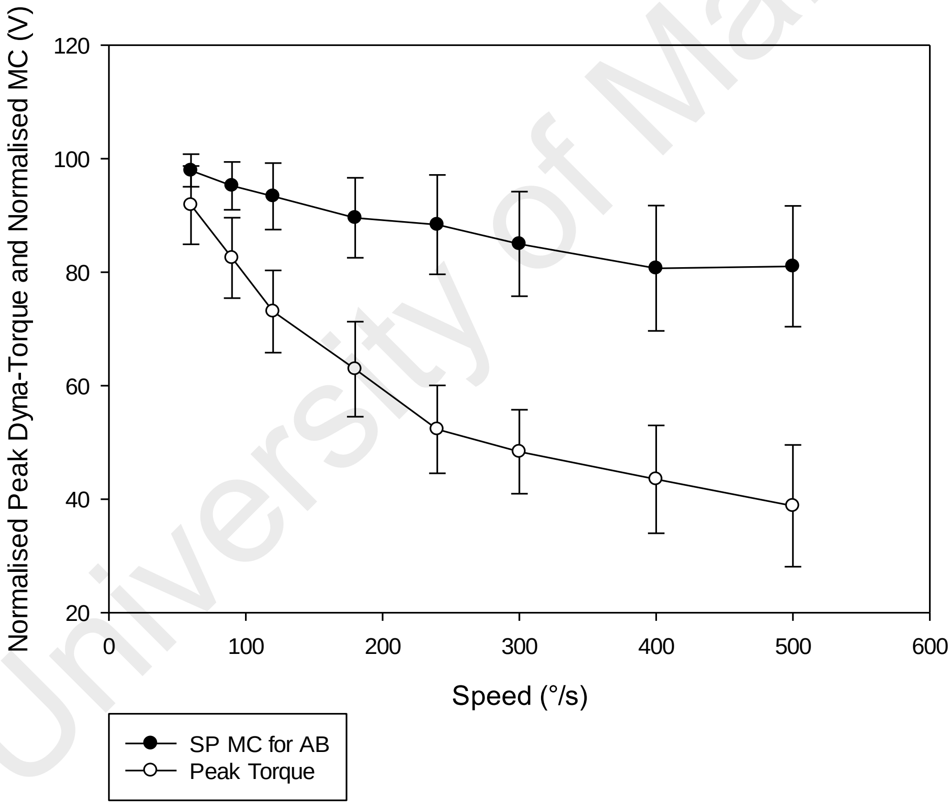


Figure 5.2 Peak torque of AB participants represented by their mean normalized dyna-torque and mean normalized MC voltages during FES-evoked contractions with different isokinetic dynamometer speed. Data are mean ± standard error.

Based on results from this study, as shown on Figure 5.2 and 5.3, the MC sensor also can detect the different torque contraction on able bodied participant during series of isokinetic knee extensions at nine different speeds. The study also reported dyna-torque vs MC-voltage relationship, with R^2=26% in muscle performance over a range of isokinetic speeds and relationship of Signal Average of dyna-Torque vs MC-voltage with R^2=85%.

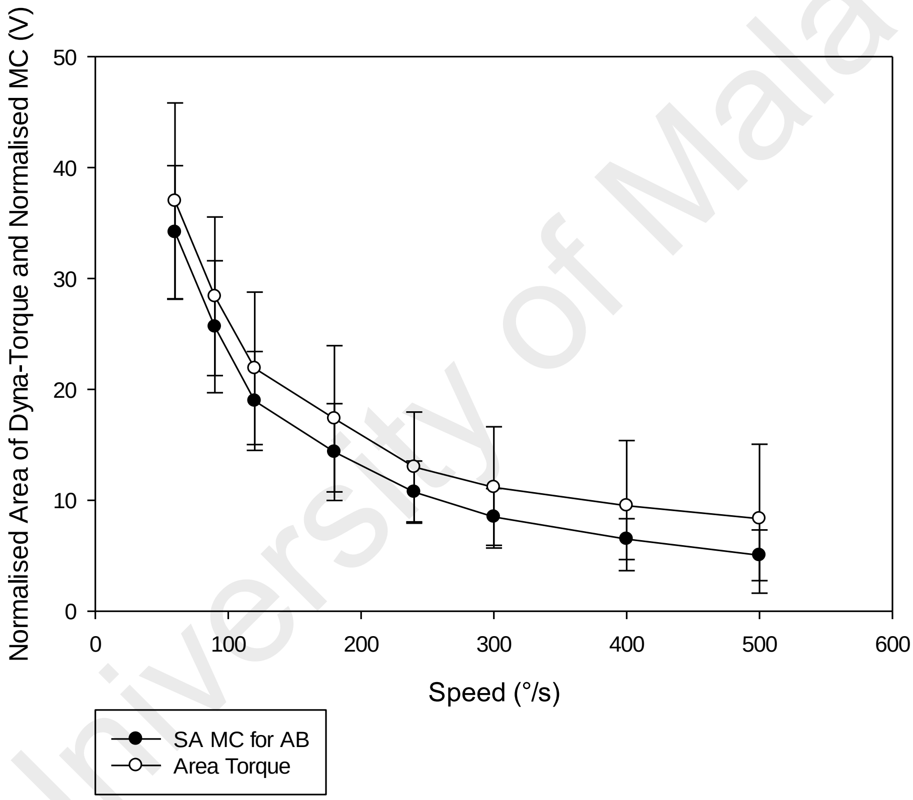


Figure 5.3 Average Signal of AB participants represented by their mean normalized dyna-torque and mean normalized MC voltages during FES-evoked contractions with different isokinetic dynamometer speed. Data are mean ± standard error.

It can also be concluded that the MC sensor is suitable to be use on able bodied person with FES stimulation. The isokinetic knee joint torque measurement system can be used to develop novel muscle activation strategies to optimise isolated muscle performance, which plays a key role in fundamental research on activation patterns and development of appropriate stimulation patterns. This will make human powered mobile cycling a more realistic recreational option and opens the potential to increase the beneficial effects on cardiopulmonary and musculoskeletal health. The of MC sensor as muscle performance sensor on SCI patient during FES cycling should be taken into consideration on next study.

CHAPTER 6: CONCLUSION

Individuals who live with SCI seek FES systems that provide standing balance, torso control, and walking, as well as systems to prevent pressure sores while sitting. Physicians and therapists who work with SCI individuals echo these requests. However, the rehabilitation by prolonged FES was limited by muscle fatigue. In order to consider muscle fatigue for muscle force/torque control, various fatigue models have been proposed as a supplement to the physiological muscle model.

In this study, novel sensor investigated the sensitivity of small body worn MC-sensor as to monitor muscle fatigue and muscle performance in SCI patients are justified so that its principle can be used later in the development of FES. The strong linear correlation between MC-sensor torque measurements and dynamometer isometric knee torque suggested that the MC-sensor was able to detect different muscle contraction levels and a fatiguing contraction in FES-evoked activity among individuals with SCI.

As from the discussion and results, the MC-sensor is easy to analyse the signals from muscle performance and fatigue monitoring activities. Other than that, the MC is also small and easy to carry, thus MC-sensor could be used in the place of a large laboratory-based dynamometer for measuring muscle contractions within a clinical environment.

Contrariwise, the MC is a relatively new sensor, and at the time of this current study, the MC-sensor's data could only be saved in a data logger and not be transferred wirelessly in real-time during the study. Future technical developments must achieve wireless, real-time portrayal of MC peak and average estimates of muscle performance, for clinical efficacy.

Different protocols to validate the MC-sensor as muscle performance and muscle fatigue monitor with a greater number of SCI participants and in different settings are also recommended for future investigation, including the effects of environmental factors such as temperature and humidity.

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