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Contribution to the evaluation of muscle fatigue model and recovery model

Deep Seth

► **To cite this version:**

Deep Seth. Contribution to the evaluation of muscle fatigue model and recovery model. Robotics [cs.RO]. Ecole Centrale de Nantes (ECN), 2016. English. NNT: . tel-01526369

HAL Id: tel-01526369

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Thèse de Doctorat

Deep SETH

*Mémoire présenté en vue de l'obtention du
grade de Docteur de l'Ecole Centrale de Nantes
sous le label de L'Université Nantes Angers Le Mans*

École doctorale : Sciences et technologies de l'information, et mathématiques

Discipline : Automatique, productique et robotique

Unité de recherche : *Institut de recherche en communications et cybernétique de Nantes*

Soutenue le 11 juillet 2016

Contribution to the evaluation of muscle fatigue model and recovery model

JURY

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Abstract

Automation has changed the working conditions in industries and manufacturing process, but still many industries needs manual operations and handling for various tasks. These manual operations lead to work-related Musculoskeletal Disorders (MSD) which is one of the major problems for industrial workers. Muscle fatigue is one of the reason leads to Musculoskeletal Disorder. The companies have to pay attention on this issue due to the new laws on penibility or repetitive tasks. In this thesis, we are focusing on the development of an adequate and realistic dynamic muscle fatigue and recovery model for dynamic work posture, its applications and validation through experiments. Firstly, we have introduced a new dynamic muscle fatigue model with a newly introduced co-contraction factor to predict the physical strength in dynamic conditions. A recovery model is used to predict the recovery after fatigue. Both the models are first validated theoretically and compared with previous models and then validated experimentally. In theoretical analysis models are compared using regression methods. Secondly, experiments were conducted on 10 subjects for push pull motion of the arm to study the muscle strength and rate of recovery. Electromyography (EMG) technique is used to analyze the muscle activity. At last, the experimental data are used to validate muscle fatigue and recovery model. It has been found that with the increase in muscle fatigue co-contraction index decreases and most of the subjects followed the exponential function predicted by fatigue model. With the use of co-contraction factor dynamic maximum endurance time (DMET) decreases by 25.9% as compared to R.Ma's Model. At last, the normalized values of fatigue rate and co-contraction factor are used to predict the DMET for various work loads. The fatigue and recovery model proposed in this thesis can be useful in analyzing the muscle fatigue and recovery parameters, to analyze human posture and in improving working ergonomics conditions.

Keywords :

Muscle fatigue, musculoskeletal disorder (MSD), maximum voluntary contraction (MVC), dynamic muscle fatigue model, strength recovery, recovery model, co-contraction, electromyography (EMG), dynamic maximum endurance time (DMET).

Résumé

L'automatisation a changé les conditions de travail dans les industries en changeant les processus de fabrication. Cependant, il existe encore beaucoup d'industries demandant des opérations manuelles ou de manutention pour diverses tâches. Ces opérations manuelles conduisent à l'apparition de troubles musculo-squelettiques (TMS) qui sont l'un des problèmes majeurs pour les travailleurs. La fatigue musculaire est l'une des causes des troubles musculo-squelettiques. Les entreprises doivent faire attention à ce problème en raison des nouvelles lois sur la pénibilité ou les tâches répétitives. Dans cette thèse, nous nous concentrons sur le développement d'un modèle dynamique de fatigue et de récupération musculaire pour des postures de travail dynamique, ses applications et sa validation par le biais d'expériences. Tout d'abord, nous avons introduit un nouveau modèle de fatigue musculaire dynamique incluant un facteur de co-contraction pour prédire la variation de la force physique dans des conditions dynamiques. Un modèle de récupération est utilisé pour prédire la récupération après la fatigue. Les deux modèles sont d'abord validés théoriquement et comparés avec les modèles précédents, puis validés expérimentalement. En théorie, les modèles d'analyse sont comparés en utilisant la méthode de régression. Deuxièmement, des expériences ont été menées sur 10 sujets pour le mouvement push-pull du bras pour étudier la force musculaire et le taux de récupération. Des signaux d'électromyographies (EMG) sont utilisés pour analyser l'activité musculaire et valider les modèles de fatigue musculaire et de récupération. Il a été constaté que lorsque la fatigue musculaire augmente, l'indice de co-contraction diminue. Aussi, la courbe de fatigue de la plupart des sujets a suivi la fonction exponentielle décroissante prédite par le modèle de fatigue. Avec l'utilisation du facteur de co-contraction, dynamique, le temps d'endurance maximale (DMET) diminue de 25,9 % par rapport au modèle de R. Ma. Enfin, les valeurs normalisées du taux de fatigue et le

facteur co-contraction sont utilisés pour prédire la DMET pour diverses charges de travail. Le modèle de fatigue et de récupération proposé dans cette thèse peut être utile dans l'analyse des paramètres de fatigue musculaire et de récupération, d'analyser la posture humaine et dans l'amélioration de l'ergonomie des postes de travail.

Mots-clés :

Fatigue musculaire, troubles musculo-squelettiques (TMS/MSD), contraction maximale spontanée, modèle de la fatigue musculaire dynamique, modèle de récupération, co-contraction, électromyographie (EMG), temps d'endurance maximale dynamique (DMET).

Acknowledgement

First of all, I am sincerely thankful to Mr. Michel Malabre for their welcome in l'Institut de Recherche en Communications et Cybernétique de Nantes, where I performed all my works. I also thank Mr. Philippe Wenger, head of Team Design Methods in Mechanics. All I would like to thank 'Mr. Damien Chablat', 'Mr. Fouad bennis' and 'Mme. Sophie Sakka' for their direction and supervision in my thesis. I want to express my gratitude for their job availability, their professional coaching, their scientific qualities input, and their kindness. I will never forget their welcome and their management during my work in France. I am thankful 'Marc Jubeau' and 'Antoine Nordez' from STAPS, University of Nantes, for their help in preparing and conducting the experiments. I am also thankful to the laboratory STAPS for providing the experiment equipments and facilities for my research.

I am also thankful to the reviewers for their valuable comments and suggestions to improve my thesis report.

This project is financially supported by Erasmus Mundus HERITAGE Project funded by the European Commission and Coordinated by Ecole Centrale de Nantes, France.

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Chapter 1

General Introduction

1.1 Motivation and Purpose

Manual handling is an important area of ergonomic intervention because of its association with a very large number of work related injuries and compensation claims. In the NOHSC code, Manual handling is defined as *any activity requiring the use of force to lift, lower, push, pull, carry or otherwise move, hold or restrain any animate or inanimate object*. Many jobs in the construction and manufacturing industries involve repetitive movements. Risks of manual handling include those associated with the working environment, the task, the load and the individual's capacity. The spine, discs, muscles and ligaments are particularly vulnerable to the work related injuries. Automation has increased its significance in many industries and manufacturing processes, but still manual work is required in many processes. In many studies it was found that industrial repetitive task lead to work related Musculoskeletal Disorders (MSD) problem. Fatigue caused by physical load is one of the main reasons responsible for MSD.

Computer aided tools are in huge demand with the increasing global manufacturing competition and changing customer demands. In the 80's, conventional tools were used to study the human ergonomics in different working condition. However, physical fatigue is not considered and modeled in commercial ergonomic tools. Physical fatigue models for static and dynamic conditions are proposed in some studies, but they are not so realistic because of their limitations. Co-contraction is the activation of the group of muscle to hold a position or load. Co-contraction is one of the factors which are not included in the previous studies. So our objective is to develop and experimentally validate a more realistic muscle fatigue model so that later it can be integrated into computer aided tools to prevent MSD.

1.2 Problem and Objective

In the study of muscle fatigue, muscle tiredness and muscle fatigue are two similar terms. It is a sense of weariness or exhaustion that you feel when using the muscle. The muscle isn't genuinely weaker, it can still do its job, but it takes you more effort to manage it. Sometimes the muscle starts off normally, but tires very quickly and takes longer to recover than normal. The recovery of muscle to its normal strength after fatigue depends on the fatiguability of the considered muscle. The evaluation of muscle fatigue and recovery is the main concern in the ergonomics.

Muscle fatigue models proposed by Freund [1], Liu [2], xia [3] and silva [4] did not explain the effect co-contraction in their study and also lack experimental validation for the same. L.Ma (2008) [5] and R.Ma (2012) [6] proposed a dynamic muscle fatigue model and validated it experimentally on Static and dynamic working posture respectively. But these models also lack in explaining the effect of muscle co-contraction. They have also introduced recovery model to study the effect of recovery time on the performance of human beings, which needs experimental validation for dynamic conditions. These models are defined for simple working posture for static and dynamic conditions and did not explain the validity of their models for complex motion. The more detailed description of these models with other models are in Chapter 2.

The objective of this research is to develop and validate a more realistic dynamic muscle fatigue model. We have an objective to include a co-contraction factor in the proposed model. Further, we aim to extend and develop the model for more complex motion and validate experimentally. The evaluation of fatigue parameters, recovery parameters, co-contraction between the muscle and muscle strength are the main problems that we want to address and solve by muscle fatigue model. In this research we are focusing on the muscles of the human arm for the analysis of muscle fatigue and recovery. The overall purpose of this project is to develop and validate a muscle fatigue model which can be useful in computer aided industrial tools to analyze different working posture with fatigue and recovery parameters.

1.3 Structure of the Thesis

Muscle fatigue and recovery models are the main focus area of this thesis. Firstly, Chapter 2 is the literature survey of the muscle fatigue models: the muscle fatigue phenomenon is described, literature survey on various static and dynamic models, the overall state of the art is exposed, insisting particularly on the problems of muscle fatigue and recovery. The solutions proposed in the literature are analyzed, leading to the fact that effect of co-contraction is not considered in the study and in some cases experimental validations are also required. Based on this observation, we propose a new dynamic muscle fatigue model with the introduction of a co-contraction factor in the model. This leads to the experimental validation of our muscle fatigue and recovery model.

In Chapter 3, the methodology for a new dynamic muscle fatigue model with various

hypothesis is defined. This chapter gives the detailed description of the muscle fatigue model with trajectory and velocity behavior for a simple dynamic working posture. The detailed description about the co-contraction factor is done in this chapter. The theoretical validation of the new dynamic muscle fatigue model and its comparison with the previous model is done to validate its application for different body joints. This model is applicable to simple dynamic and static postures.

Experimental validation of the muscle fatigue model is done with three different case studies in Chapter 4. The dynamic muscle fatigue and recovery model is validated through three experiments. The motion used in the experimental case study are Flexion and extension. The experimental protocol is defined for the case study. The method for data processing of raw electromyography data (for Bicep, Tricep, and Trapezius muscle) and mechanical data (torque, position, and velocity) is also explained in details.

The results obtained from the previous chapters are described, analyzed and discussed in Chapter 5. This concerns the evaluation of muscle activity, muscle fatigue and co-contraction, Comparison of a case studies, effect of recovery and validation of the superposition principle for complex motion is done to the expected behavior. At last, conclusions and perspectives for the overall research work are presented in Chapter 6. In the same chapter, we have presented the future perspectives or scopes of the dynamic muscle fatigue model. We also describe the possible application and proposed the development of the dynamic muscle fatigue model for complex motion in dynamic conditions.

1.4 Main Contribution of the Thesis

The main contribution of the thesis is in the field of industrial ergonomics and biomechanics. The thesis represents the evaluation and validation of new dynamic muscle fatigue model. It also includes the global analysis of muscle fatigue for the complex motion along a joint using the newly proposed superposition model. The newly introduced co-contraction factor in the dynamic muscle fatigue model makes it more realistic towards the study of muscle fatigue. The electromyography (EMG) technique is used to analyze the co-contraction between the muscles and corresponding fatigue. Most of the models are applicable to either static or dynamic work posture, but this model can be applicable on both type of work posture. This model can be used in computer aided ergonomic tools in industries for fatigue analysis at different working postures.

The thesis also contributes in analyzing the recovery model for a case study which helps us to determine the recovery rate and parameters according to ergonomic condition. The torque, position and velocity simulation help us to analyze the actual work requirements.

The proposed dynamic muscle fatigue and recovery model takes co-contraction into consideration which makes this model more realistic in comparison with other models. The co-contraction factor which is introduced in the model shows significant effect on the model by changing the fatigue rate in comparison to R.Ma's and L.Ma's models. The experimental validation of this model for different working postures and complex motions

makes it global muscle fatigue model which can be applicable to both static and dynamic work posture. A model for dynamic maximum endurance time (DMET) is also validated in this thesis. The results shows that DMET calculated through this model is much near to the experimental data unlike R.Ma (DMET) and L.Ma. (MET) model. This gives the safer value of endurance time in dynamic situations and helps in designing better ergonomic work environment and work postures.

Chapter 2

Muscle Fatigue and Literature Survey

2.1 Human Musculo-skeletal Movement

2.1.1 Musculo-skeletal System

The system of muscles is responsible for the movement of the human body. Each muscle is a discrete organ constructed of skeletal muscle tissue, blood vessels, tendons, and nerves. The heart, digestive organs, and blood vessels also have muscle tissue.

There are approximately 640 named muscles, attached to the bones of the skeletal system that make up roughly half of a persons body weight. Almost every muscle constitutes one part of a pair of identical bilateral muscles, found on both sides, resulting in approximately 320 pairs of muscles. The muscles of the human body can be divided into a number of groups which include muscles relating to the head and neck, muscles of the torso or trunk, muscles of the upper limbs, and muscles of the lower limbs. From standard anatomical position, the action refers to the activity of each muscle. In other positions, other actions may be performed. The pictorial representation of the important muscles of the whole body in anterior and posterior view respectively are shown in figures 2.1 and 2.2.

Most of the skeletal muscles are attached to two bones through tendons (Figure 2.8). Tendons are tough bands that firmly attach muscles to bones [7]. Tendons are very strong and are woven into the coverings of both muscles and bones.

The movement in muscle consists in shortening their length, pulling on tendons, and moving bones closer to each other. *Origin* is called as the place on the stationary bone that is connected via tendons to the muscle [8]. *Insertion* is the place on the moving bone that is connected to the muscle via tendons. The fleshy part i.e., the belly of the muscle in between the tendons does the actual contraction.

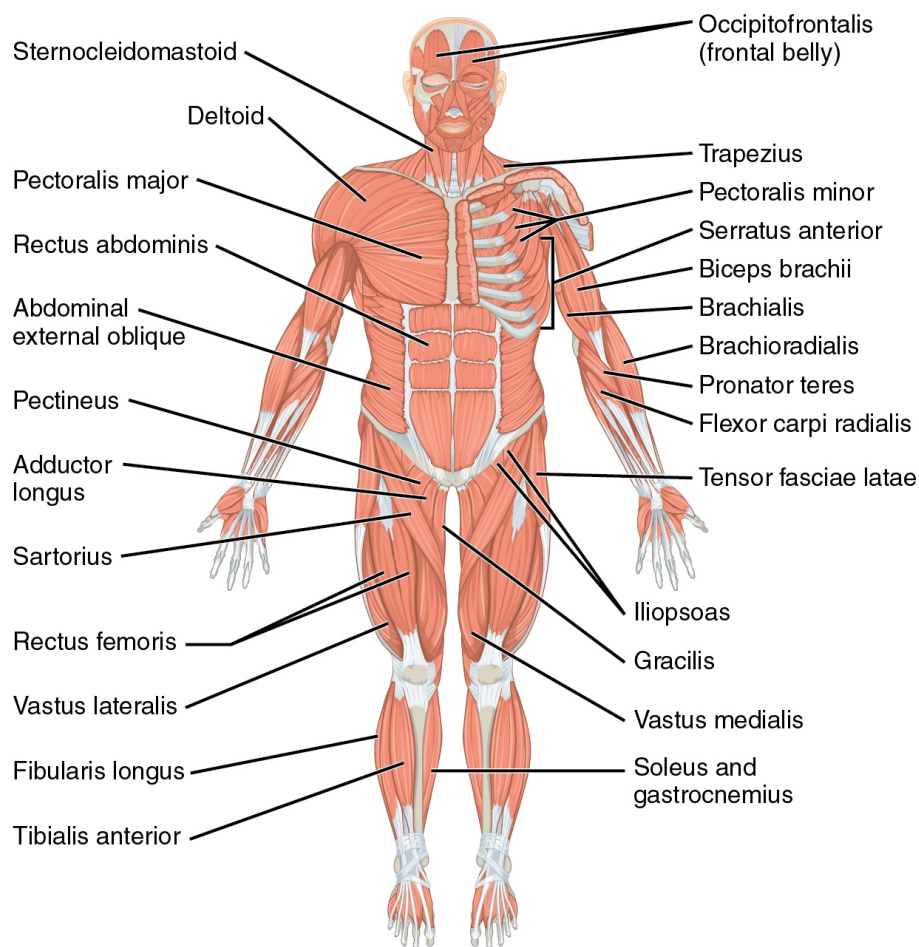


Figure 2.1: Major muscles of the body. Right side: Superficial; left side: deep (anterior view) [9]

To achieve movements in the body, skeletal muscles rarely work by themselves. Most often they work in groups to produce precise movements. The muscle that produces any particular movement of the body is known as an agonist or prime mover. The agonist pairs with an antagonist muscle to produce the reverse effect on the same bones [8]. For example, the biceps muscle flexes the arm at the elbow. As the antagonist for this motion, the triceps muscle extends the arm at the elbow. When the triceps is extending the arm, the biceps would be considered as the antagonist. Between the antagonist and agonist muscles activity there are always some transition activities between both muscles known as *Co-contraction*.

In this thesis most work is done for the arm movement. The muscles of the arm and hand are specifically designed to meet the body's diverse needs of strength, speed, and precision while completing many complex daily manipulation tasks [10].

For the flexion and extension of the forearm, at the elbow joint, the muscles of the upper arm are responsible [10]. Flexion of the forearm is achieved by a group of three muscles: the brachialis, biceps brachii, and brachioradialis (Figure 2.1). The flexor muscles are all located on the anterior side of the upper arm and extend from the humerus and scapula to the ulna and radius of the forearm (Figure 2.2). Also, the biceps brachii

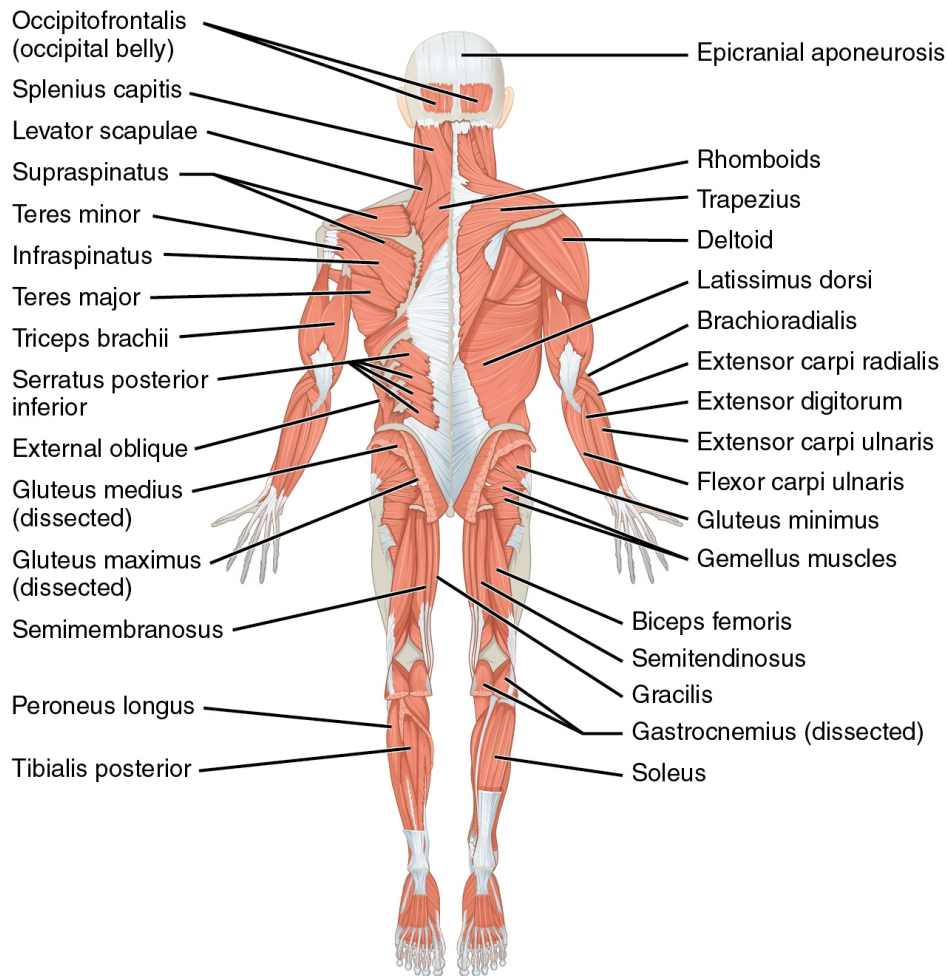


Figure 2.2: Major muscles of the body. Right side: Superficial; left side: deep (posterior view) [9]

operates as a supinator of the forearm by rotating the radius and moving the palm of the hand. The triceps brachii, acts as an extensor of the forearm at the elbow and the humerus at the shoulder.

Most of the muscles that move the wrist, hand, and fingers are located in the forearm. The muscles on the anterior side of the forearm, the flexor carpi radialis and flexor digitorum superficialis, form the flexor group which flexes the hand at the wrist.

The supination (anterior rotation) and pronation (posterior rotation) are the motion from the muscles of the forearm and hand. The biceps brachii of the upper arm and the supinator muscle of the forearm are responsible for supination. Likewise, the pronator teres of the forearm is responsible for pronation. The origin of both supinator and pronator teres muscles are on the humerus and ulna and insert on opposite sides of the radius to roll the wrist in reverse side.

2.1.2 Muscle Activity

All muscle contractions do not produce movement. For example isometric exercise or isometrics are a type of strength training in which the joint angle and muscle length do not change during contraction. During isometric contraction when we try to move body part without any movement, light contractions can cause tension in the muscle without exerting enough force. When people tense their bodies due to stress, they are performing an isometric contraction. To Hold an object still and maintain the same posture are also the result of isometric contractions. A contraction that does produce movement is an isotonic contraction. Isotonic contractions are required to develop muscle mass through weight lifting [7].

Movement of the body parts is the the main function of the muscular system. Muscles tissue in the body that has the ability to contract and move the other parts of the body. The other functions of the muscles is the maintenance of posture and body position [8]. Most of the time muscles contract to hold the body still or in a particular position instead of causing movement. The muscles that hold up the body throughout the day have greater endurance and can do this without becoming very tired.

Human body muscles work together with joints and bones to form a lever system where the muscle acts as a linear force. The bone acts as the lever. The joint acts as the fulcrum and the object can be considered as the load.

Muscles are made of Muscle fibers. Production of force and movement is realized by contraction of muscle fibers driven by nervous system commands as shown in figure 2.3. The main functional unit of muscles is motor units which contain motoneurons and muscle fibers. The number of motor units depends on the size and function of muscles, ranging from few for small muscles to thousands for larger.

To control or activate motor units brain sends command signal pulses to motor units. The command is in the form of electrical impulse as shown in figure 2.3 [2]. When the command unit gets impulse more than the threshold, then the motor units activate and control muscle fiber of this motor unit to contract synchronously [2].

If the stimulus is strong enough, then it triggers an action potential and required number of muscle fibers of the motor unit get activated. But if the stimulus is not strong enough, then no action potential will be triggered and so no motor in the fiber unit get excited. We can say that there is no state between active and not active motor units.

The number of motor units involved in contraction and the amount of stimulus from the nervous system can control the strength of a muscle contraction. A motor unit can be contracted by a single nerve impulse briefly before relaxing. This small contraction is known as a *twitch contraction*. If in short period of time the motor neuron provides several signals, the duration and strength of the muscle contraction increases and this phenomenon is known as *temporal summation*. If in rapid succession, the motor neuron provides many nerve impulses, the muscle may enter the state complete and lasting contraction. A muscle will remain in this condition until the nerve signal rate slows down or the muscle becomes too fatigued to maintain the state.

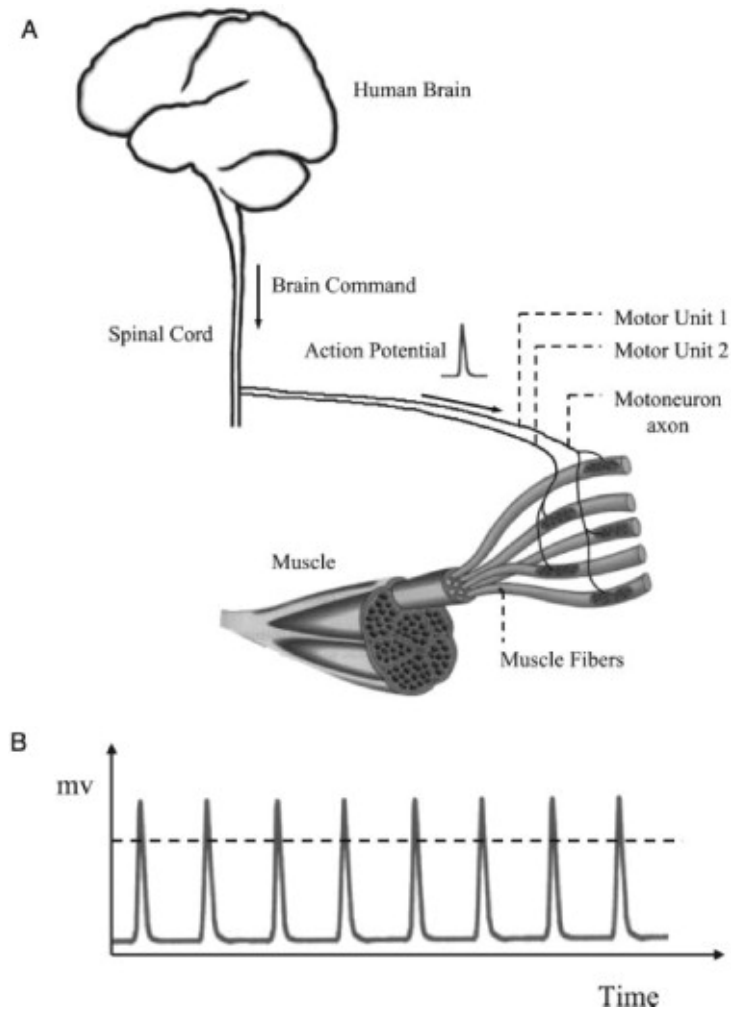


Figure 2.3: Muscle control Activity: (A) Schematic illustration of the human nervous system and muscle. The brain sends down a command (voluntary drive) through the spinal cord and peripheral nerves to muscle. Muscle is made of motor units. A motor unit contains a motoneuron and the muscle fibers it innervates. When a stimulus arrives at a motor unit and it is strong enough, it triggers an action potential, which in turn activates the motor unit. Force is generated by contraction of muscle fibers. (B) Action potential series. If the brain command continues, it triggers a series of action potentials, which keep activating the motor units to produce a sustained force. [2]

To perform any movement some motor units in a muscle or group of muscle have to be activated. For small force, less motor units get activated and for larger force, more motor units get activated. When a prolonged voluntary muscle contraction is sustained, the brain continuously reinforces the descending command.

2.2 Muscle Fatigue

After being activated for a period of time, the activated motor units start to develop fatigue due to factors such as insufficient supply of oxygen and glycogen, increase lactic

acid level in blood and muscle and other reasons of muscle fatigue can be physiology. After reaching critical point the motor units can no longer be activated and hence fatigued. Hence after a long period of time when all the muscles get fatigued and cannot activate any more then the force and movement cannot be continued any more.

Definitions of muscle fatigue

It is difficult both to define and to evaluate human muscle fatigue. According to various authors, the following definitions exists:

1. The failure to maintain a required or expected force. (Edwards, 1977) [11].
2. Any reduction in the force generating capacity. (measured by the maximum voluntary contraction), regardless of the task performed. (Bigland-Ritchie and woods, 1984) [12].
3. A reduction in maximal force generating capacity. (Bigland-Ritchie, 1986) [13].
4. The development of less than expected amount of force as a consequence of muscle activation. (McCully, 1990) [14].
5. Reduction in the maximal force generating capability of the muscle during exercise. (Miller, 1995) [15].
6. Muscle fatigue is a decline in the maximal contractile force of the muscle. (Vollestad, 1997) [16].
7. Fatigue is known to be reflected in the electromyography signal as an increase of its amplitude and a decrease of its characteristics spectral frequencies. (kallenberg, 2007) [17].

The muscular fatigue is combination of physical, chemical and neuron signal changes in the body. The muscle fatigue can be divided in to central and peripheral fatigue depending on signal commands from brain and physical and chemical changes in muscles. The muscle fatigue can be described in short by figure 2.4. Muscle fatigue can occur in two basic mechanisms: (a) central involves proximal motor neurons (mainly in the brain); and (b) peripheral involves within the motor units (i.e., motor neurons, peripheral nerves, motor endplates, muscle fibers). These parts are describes briefly in the next sections.

When muscles run out of energy during either aerobic or anaerobic respiration, the muscle quickly tires and loses its ability to contract. This condition is known as muscle fatigue. It is the decline in ability of a muscle to generate force. There are two main reason of muscle fatigue. First one is the limitations of the nervous system to generate or sustain signals to control muscle [8]. The second reason can be the reduction in the ability of muscle to contract [8]. Fatigue is a feeling of tiredness or exhaustion. It is a need to rest because of lack of energy or strength. Fatigue may result from overwork, poor sleep, worry or lack of exercise. It is a symptom that may be caused by illness, medicine, or medical treatment such as chemotherapy. Anxiety or depression can also cause fatigue.

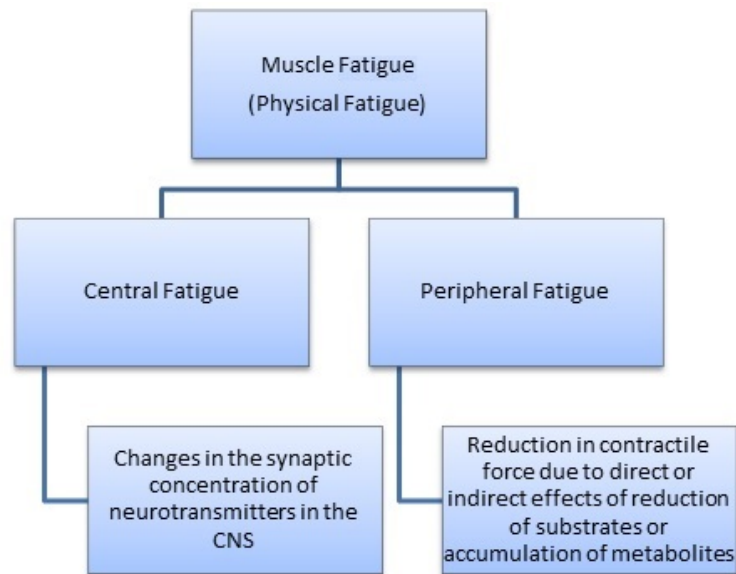


Figure 2.4: Types of Muscle Fatigue [8]

2.2.1 Central fatigue

Central fatigue or Central nervous system fatigue, is a form of fatigue that is associated with changes in the synaptic concentration of neurotransmitters within the central nervous system (CNS; including the brain and spinal cord), affects exercise performance and muscle function. Central fatigue is a reduction in the neural drive or nerve-based motor command to working muscles that results in a decline in the force output. While controlling motor units brain produces serotonin that control muscle contraction. Serotonin is a monoamine neurotransmitter, derived from tryptophan, serotonin is primarily found in the gastrointestinal tract (GI tract), blood platelets, and the central nervous system (CNS) of humans. It is popularly thought to be a contributor to feelings of well-being and happiness [18]. Because of more muscle activity serotonergic pathway contracts which reduces the muscle contraction. It is a form of fatigue that is related with changes in the synaptic concentration of neurotransmitters in the CNS (including the spinal cord and brain). It affects exercise performance and muscle function, and cannot be explained by peripheral factors that affect muscle function.

Central fatigue is generated by an inhibition elicited by nervous impulses from receptors (can be some kind of chemo-receptors) in the fatigued muscles. The inhibition may act on the motor pathways anywhere from the voluntary centers in the brain to the spinal motor neurons. This kind of fatigue should manifest itself by a decrease in the outflow of motor impulses to the muscles. There are several good reasons for assuming this central component is the result of central inhibition called forth by signals from the fatigued muscles. This inhibition, most likely originating in the reticular formation, may itself be inhibited or counteracted by other signals of peripheral or central nervous origin.

2.2.2 Peripheral fatigue

Peripheral fatigue is an inability for the body to supply sufficient energy or other metabolites to the contracting muscles to meet the increased energy demand. This causes contractile dysfunction that causes reduction or lack of ability of a single or group of muscles to do work. This is also known as metabolic fatigue because of reduction in contractile force due to direct or indirect effects of reduction of substrates or accumulation of metabolites in muscle fibers. Peripheral fatigue is a person's inability to maintain their expected level of work intensity. There are many causes of peripheral fatigue like reduction in oxygen supply, ATP (Adenosine Triphosphate) depletion, increase in lactic acid in muscles, chemical imbalance in the muscles. Any one or combination of these can affect your peripheral fatigue [19]. In peripheral muscle fatigue there are at least two different sites where repeated contractions may cause impairment: the "*transmission mechanism*" (neuromuscular junction, muscle membrane, and endoplasmic reticulum), and the "*contractile mechanism*" (muscle filaments). Peripheral muscle fatigue, defined as a transient decrease in a muscle group's capacity for exercise, can be purely peripheral (i.e., located distally to the motor neurons). The mechanical response of the particular or individual active muscle fibers decline with fatigue, a certain reinforcement can be achieved by pumping the innervation frequency and/or the number of motor units which are active. The reasons for the appearance of peripheral muscle fatigue are local changes in the internal conditions of the muscle. These may be biochemical, depletion of substrates such as glycogen, high energy phosphate compounds in the muscle fibers, and acetylcholine in the terminal motor nerve branches, or they may be due to the accumulation of metabolites, such as lactate or electrolytes liberated from the muscles during activity.

One reason of metabolic fatigue can be metabolite depletion. ATP (adenosine triphosphate) and CP (creatine phosphate) levels both decreases with the start of exercise. ATP is the major energy origin for the majority of cellular functions in your body. CP help to maintain ATP levels. As metabolites deplete (especially CP), muscle fatigue start in which causes less energy to fuel muscle contractions. Other reason for muscle fatigue can be metabolite accumulation. To decrease in force, the accumulation of di-hydrogen phosphate ($H_2PO_4^-$) and lactic acid (H^+) can be responsible. An increase in the level of phosphate causes a decrease in the force of muscle contraction. Lactic Acid accumulation also reduces muscle power by moving Calcium ions that are necessary to the process of muscle contraction.

Disturbances in homeostasis can also cause fatigue [20]. This includes dehydration, altitude, heat and cold. When it is very hot or very cold outside, blood is diverted from the working muscles to the skin to release the excess heat or blood is also rerouted from the working muscles to the bodys core to keep more vital organs warm respectively. Both these processes restrict oxygen supply and can cause muscle fatigue.

2.2.3 Conclusions

The interrelationship between central (the brain) and peripheral (muscles, motor nerves etc.) systems are important in the study of fatigue. The study shows how local muscle fatigue is influenced by central nervous factors. Peripheral fatigue is a part of every human worker. Harmful metabolites accumulate with depletion of energy sources and both of these are contributing to the inability to maintain exercise intensity. Cellular changes are not the only things within the muscle that contribute to fatigue, but there is also evidence that the brain has a major effect on human being's perception of fatigue [18, 20]. The human body is a complex network and the connections between the muscular and nervous systems are essential.

The central and peripheral fatigue contribute together in physical fatigue. The contribution varies from 15-30 percent for central fatigue and remaining is peripheral fatigue. A study shows 20% contribution of central fatigue in the reduction of MVC^* , while the intramuscular metabolic milieu ($[H^+]$) was responsible for the remainder of the fatigue [21]. Metabolic milieu ($[H^+]$) controls the mitochondrial activity of the cells in the body. In the current study we are concentrating on physical fatigue in terms of maximum voluntary contraction (MVC) reduction.

* The Maximum Voluntary Contraction (MVC) is a measure of strength. The measure can be a maximal exertion of force reported as force (e.g., lb, kg, Newtons) or as a moment around a joint (e.g., Newton-meters, foot-pounds, kilogram-meters).

2.3 Musculoskeletal Disorders (MSD)

2.3.1 Definition of Musculoskeletal Disorder (MSD)

The definition of MSD is simple and can be written as "Musculoskeletal Disorders or MSDs are injuries and disorders that affect the human body's movement or musculoskeletal system (i.e. muscles, tendons, ligaments, nerves, discs, blood vessels, etc.)" [22].

Musculoskeletal disorders (MSD) consist of physical disabilities. MSD is used to describe conditions that affect the joints, bones, and muscles [23]. The severity of the MSD depends on working conditions. Pain and discomfort can affect everyday work activities. MSD are extremely common in industries and risk increases with age. MSD can affect all main areas of the body, including neck, shoulders, wrists, back (upper and lower), hips, legs, knees, feet [24]. The sign and symptoms of MSD are low back pain, fibromyalgia, gout, osteoarthritis, rheumatoid arthritis and tendinitis [25]. These disorders can create little discomfort leading to debilitating pain. Low back pain is the most common MSD.

According to the study in the USA and Europe, MSDs are a common and severe problem for the people and companies across the United States and Europe [22, 23, 25].

MSD because of work-related injuries are responsible for almost 30% of all workers compensation costs (source: BLS) [26]. Direct cost for MSD compensation costs \$50 millions to U.S. companies (source: CDC) [27] and indirect costs could be up to five times the direct costs (source: OSHA) [28]. The average cost for MSD equate with a direct cost of approximately \$15,000 (source: BLS) [26]. The human and economic costs of MSD are unnecessary and preventable.

2.3.2 The Causes of Work-Related Musculoskeletal Disorder (MSD)

According to Canadian Center for Occupational Health and Safety (CCOHS), “Almost all work requires the use of the arms and hands. Therefore, most work related musculoskeletal disorders (WMSDs) affect the hands, wrists, elbows, neck, and shoulders. Work using the legs can lead to WMSD of the legs, hips, ankles, and feet. Some back problems also result from repetitive activities” [29]. A worker begins to fatigue when he/she is exposed to MSD risk factors. When fatigue overtake their bodys recovery system, they generate a musculoskeletal imbalance. With time, as fatigue continues to overtake recovery and the musculoskeletal imbalance increases, a musculoskeletal disorder develops. These risk factors can be divided into two categories: work-related (ergonomic) risk factors and individual-related risk factors. Over time exposure of human to these risk factors leads to MSD. Some of the major risk factors related to ergonomics and individual are shown in figure 2.5 [22]. The risk factors like excessive force, frequent repetitive tasks, wrong working postures and muscle fatigue are mainly related to ergonomics risk factors. The individual risk factors are mainly because of the poor work practices, poor fitness, less recovery after work, poor health and food habits and working with the injuries.



Figure 2.5: Work-related (Ergonomics) and Individual-related Risk Factors [22]

Recovery plays important role in preventing MSD due to fatigue. When Fatigue overruns recovery of muscles, it leads to MSD. In figure 2.6 [22], we can see the red zone which starts with discomfort due to working conditions and leads to MSD. It also show the command of fatigue over recovery leads the graph towards MSD. The figure shows the

peak head as the initial strength of a human, which reduces with the time while working and after some time it reaches to fatigue then discomfort and finally with pain leads to the loss of function of muscle or MSD. There is a possibility that after muscle fatigue and then recovery, person can have MSD. That is why muscle fatigue is not the only reason of MSD. There are many other parameters which affect MSD.

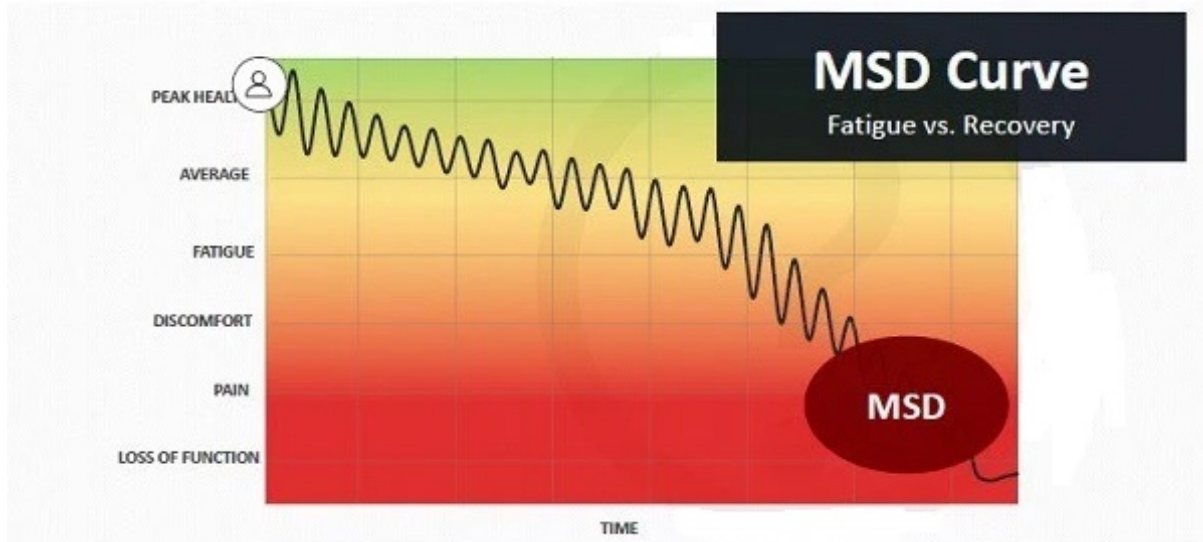


Figure 2.6: MSD curve: Fatigue vs Recovery [22]

2.3.3 Major Injuries due to Work-related Musculoskeletal Disorder

Workplace design plays a crucial role in the development of an MSD. According to CCOHS, “Work-related musculoskeletal disorders (WMSD) are a group of painful disorders of muscles, tendons, and nerves. Carpal tunnel syndrome, tendonitis, thoracic outlet syndrome, and tension neck syndrome are examples” [29].

When a worker does any work, outside his body limitations and capacity, he put his musculoskeletal system at risk. In such kind of situations, evaluation of fatigue and recovery tells us about the incompetency in the workstation design. The evaluation could predict that ergonomic risk factors are eminent, the worker can be at musculoskeletal imbalance risk and a musculoskeletal disorder persist. Punnett and wegman (2004), evident this fact in their work, “Thus there is an international near-consensus that musculoskeletal disorders are causally related to occupational ergonomic stressors, such as repetitive and stereotyped motions, forceful exertions, non-neutral postures, vibration, and combinations of these exposures” [30].

Work-related Musculoskeletal Disorder (WMSD) include three types of injuries [29]:

- **Muscle injury:** During muscle contraction, chemical energy from sugars and by-products such as lactic acid absorb by the blood. Blood flow reduces with long time of muscle contraction. These substances when starts accumulate in muscles

due to lack of absorption by blood. The accumulation of these substances irritates muscles and causes pain. The intensity of this pain depends on the duration of the muscle contractions. It also depends on the amount of time between activities for the muscles to get rid of those irritating substances.

- **Tendon injury:** Tendons are made of many bundles of fibres that attach muscles to bones. Because of repetitive or frequent work activities with awkward working postures occur, tendon problem occur in two main sectors –tendons includes sheaths as shown in figure 2.7 [29], mainly present in the hand and wrist; and tendons without sheaths as shown in figure 2.8 [29], present near the shoulder, forearm, and elbow.

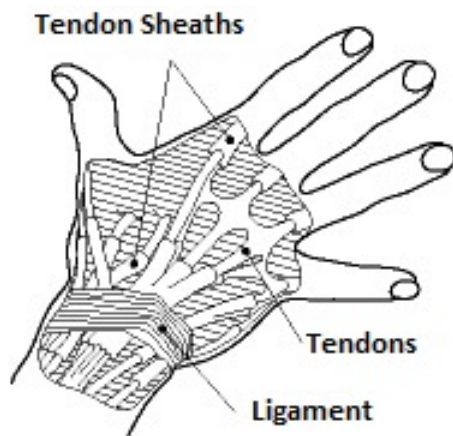


Figure 2.7: Finger tendons and their sheaths [29]

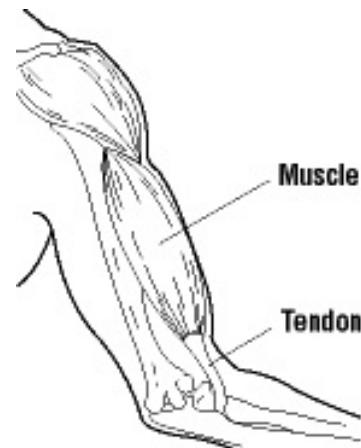


Figure 2.8: Tendon, muscle and bone unit [29]

Repetitive or excessive movement of the hand deteriorate lubrication system which leads to low production and poor quality lubrication fluid. Friction between tendon and sheath increases with the failure of the lubrication system which result into inflammation and swelling of the tendon area. Tendons without sheaths are also prone to repetitive motions and uncomfortable working postures. Fibres of tendons can tear apart with the continuous tension in tendons. This leads to thickening of tendons and tendons become bumpy, causing inflammation.

- **Nerve injury:** Nerves carry signals from the brain to control activities of muscles. Feelings about temperature, pain and touch from the body to the brain are carried by nerves. Nerves are surrounded by muscles, tendons, and ligaments. Nerves get squeeze or compressed with the swollen tissues surrounding nerves because of repetitive movements and uncomfortable postures which causes muscle weakness, sensations of “pins and needles” and numbness.

2.3.4 Primary Risk factors of Work-related MSD

There are various risk factors which can arise from the movement of different body parts of the body. Our study is going to be focused on arm motion so lets take an example of the arm for understanding various risk factors for WMSDs. Work-related MSD arise

from arm and hand movements [29] such as bending, straightening, gripping, holding, twisting, clenching and reaching. In daily life these movements are not harmful in the normal activities. What makes them harmful in work conditions are: a). the continuous repetition in a forceful manner [29], b). the speed of the movements and c). the lack of time for recovery between the movements [29, 31].

Various risk factors are thought to be associated with MSDs [31], include:

1. **Long term highly repetitive tasks:** Many work tasks and cycles are repetitive in nature and because of daily targets their frequency increases. When other risk factors such as high force and uncomfortable working postures combined with long term task repetition can contribute to the formation of MSD. A job is considered highly repetitive if the cycle time is 30 seconds or less [26, 31].
2. **Forceful exertions:** High forces on the human body are necessity of some work tasks. High force requirements increase the muscle forces and associated fatigue which can lead to MSD [28, 31].
3. **Repetitive or sustained uncomfortable postures:** Uncomfortable postures exert more forces on joints and overload the muscles and tendons around the working joint. The mid-range motion of the joint is the most efficient position for the joint movements [29] and repetitive work with sustained period with inadequate recovery time increases the risk of MSD outside the mid-range motion.
4. **Poor Work Practices:** The poor work practices like working in adverse working conditions like severe hot or cold, lack of control on body movements, high pressure also leads to injuries and MSD.

Consequences of the primary risk factors leads to the muscle fatigue and insufficient recovery.

1. **Muscle Fatigue:** Muscle fatigue is also one of the risk factor which causes MSD. The repetitive task for sustained period of time causes muscle fatigue. negligence of muscle fatigue and working above the body capacity can cause MSD.
2. **Insufficient Recovery Time:** After muscle fatigue recovery time is very important to regain the muscle strength. More the repetitive task more recovery time needed after muscle fatigue. less or insufficient recovery time leads to muscles weakness and finally MSD.

Exposure to these workplace risk factors puts workers at a higher level of MSD risk. High task repetition, forceful exertions and repetitive or sustained uncomfortable working postures, fatigue the workers body over their ability to recover and cause musculoskeletal imbalance and eventually MSD.

2.4 Difficulties in the Study of Muscle Fatigue

Muscle is a complex system of the human body. To avoid MSD study of various risk factors is necessary. Muscle fatigue is one of the risk factor cause MSD. In some perspectives, various risk factors have their inclination toward muscle fatigue and recovery.

The study of muscle fatigue is defined on the basis of central and peripheral factors of the muscle. Fatigue is a normal response to physical exertion or stress, but can also be a sign of a physical disorder. Before initiating the study of fatigue it is necessary to define the factor of muscle on which study is based. The factors for study of muscle fatigue can be, changes in the length of muscle, change in the volume of muscle, activity of muscle, intramuscular chemical imbalance or metabolism, and central fatigue factors. All factors have certain contribution in the activity of muscle. That is why the quantification of muscle fatigue is a difficult and complex task. The central and peripheral factors have a particular amount of contribution in physical fatigue [21]. The separate and combined study of central and peripheral fatigue on the basis of certain factors is possible. From the difficulties and limitations described above for the study of muscle fatigue various models have been proposed. All these models are based on different conditions and factors to study muscle fatigue.

2.5 Muscle Fatigue Models

In the field of industrial bio-mechanics, muscle fatigue is defined as “any reduction in the maximal capacity to generate the force and power output”. In industries, mostly repetitive manual tasks lead to work-related MSD problems [32, 33]. Some times people have to work more on the same repetitive task which can be painful and lead to MSD due to muscle fatigue. MSD can cause pain [32, 34, 35] or temporary dysfunction of the affected muscles [31, 36]. Muscle fatigue and uncomfortable working postures can cause drop in the productivity of human. To improve the performance and productivity, improvement in the work environment and ergonomics with the study of muscle fatigue are necessary in order to reduce the chances of MSD [35].

Various muscle fatigue models have been introduced for the analysis of muscle fatigue and recovery. Most of them use different approaches for their study. Some fatigue models worked on static [5, 16, 37] or dynamic motion [1, 2, 6] of the body to analyze the effect of muscle fatigue and recovery on the performance of human. Some models use muscle fiber in analysis and some take joint torque for the analysis according to requirements, hypothesis and type of study [38]. There are many theoretical and experimental models for explaining the effect of fatigue [38]. Validation of various models was done by different approaches.

The motion can be divided into static, quasi-static, and dynamic [39]. A static model is the model in which the acceleration is zero and the object is supposed to be fixed, the joint angles and angular acceleration are constant. There is a state called quasi static, in which speeds are considered slow and acceleration is assumed to be zero. In dynamic models no constraint is specified in the motion and the inertial effect is taken into account. The torque applied to the joint depends on angle, speed and angular acceleration (Fig. 2.9) [40].

Various muscle fatigue models have been proposed [3, 5, 16, 38, 41, 42]. Today, there are few effective models available to predict muscle fatigue [5]. The well known model of muscle was developed by A.V. Hill in 1938 [38], which explains the muscle activation

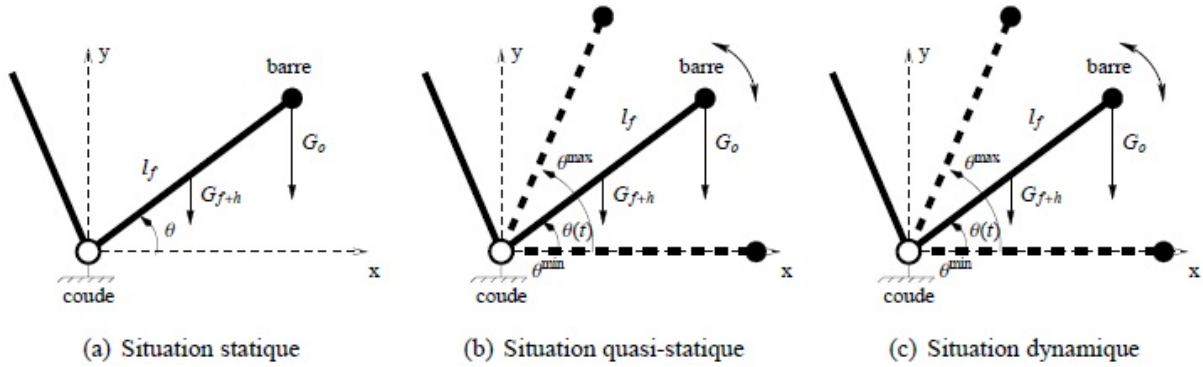


Figure 2.9: Static, Quasi-static and Dynamic Motion [40]

and study of muscle length shortening and dynamic parameters of muscles. Silva [4] used Hill's model to validate muscle model with fatigue using Opensim to simulate and validate his model. Several people worked on the static model of fatigue to study the muscle fatigue and its effect on the muscles and performance of human. L.Ma's fatigue model [5] have experimental validation for fatigue and effect of recovery in arm on static drilling posture. In both the models constant load is applied to the arm end and fatigue is analyzed with respect to time. Some Dynamic fatigue models are also introduced [1, 2, 6]. Ruina Ma has introduced a muscle fatigue model for dynamic motion of the arm [6, 43]. Missenard also explain the effect of fatigue and co-contraction on accuracy [44]. We can say that various fatigue models are defined and used according to the requirements of the study.

In study of muscle fatigue, various techniques are used to observe the behavior of muscle activity [45]: EMG-joint angle relationship [46, 47], twitch interpolation, endurance time [16, 48], joint torque-MVC [6], ultrasonic imaging [49], etc., are some of the ways to study the behavior of muscles.

2.5.1 Muscle Fatigue Models

Attempts were made to mathematically model muscle force that consider many physiological and mechanical factors of the muscle. Muscle length, shortening velocity, neural activation, and muscle architecture [50–58] are some of the factors used to calculate muscle force. Many models calculate the muscle force by summing the activity of each muscle fiber. Fuglevand [59] developed a model based on single motor unit stimulation which can describe the time duration between onset of muscle activation to the peak activation. Later on this model is improved for more accurate stimulation and activation [60].

Sustained muscle activation or contraction make muscle fatigued. Intermittent fatigue and recovery effects in the neuromuscular system affect the force production by muscle [61–64]. However, previous models did not consider the effect of fatigue and recovery time in the models. Hawkins and Hull [65, 66] identified the importance of the fatigue effect in long duration of tasks. They develop a force-time dependence based on

empirical relation to predict muscle force production using indices like endurance time and fatigue rate in fiber based model. But this model could not give satisfactory estimation of muscle forces based on experimental data.

With the advancement of technology, modeling of muscle model starts with the action lines of muscles to represent a 3-D representation of the musculoskeletal system [67–69]. Application of finite element modeling to these models [70, 71] have developed a new approach to determine how muscle force is transferred to the tendons and bone. However, these models are time consuming to run and there are some redundancy in the assumptions made regarding load sharing between muscles. For example if the isometric MVC is being measured at the elbow using a dynamometer then the force produced by the subject can be the average of three measured MVCs with some rest period between each contraction. The muscle model has to determine the contribution in MVC by different muscles or group of muscles: biceps brachii, brachialis, flexor carpi radialis, flexor carpi ulnaris, brachioradialis, and the pronator teres. We cannot individually measure the contribution by each muscles that is why we take a group of muscle as contributor to MVC.

Study on muscle fatigue is going on since late 1800s [72, 73] and research has mainly focused on the fatigue rate in terms of MVC [74–81]. The common methodology for these studies measurement of maximum voluntary contraction at a joint (e.g. ankle, elbow, knee, etc). Three MVCs are measured for each subject. The maximum or in some cases the average of three MVC are used to take a reference value. The subjects perform contractions at some percent of MVC until they are no longer able to produce the required force. The time where the force production can no longer be maintained is defined as the endurance time (ET).

Rhomert in 1960 [74] studied the endurance time of muscles over a range of MVCs. Rhomert found that Endurance time (ET) is a nonlinear function of contraction intensity [74] and response curves have been coined as *Rhomert Curves*. Rohmert discovered a non-linear relation between ones ability to maintain a static force (the onset of fatigue) and the percent of ones maximum voluntary contraction (MVC) they are holding. A formula (equation 2.1) was developed, which calculated time of endurance, based on the percentage of ones MVC they are trying to sustain.

$$T_{(sec)} = -90 + (126/P) - (36/P^2) + (6/P^3) \quad (2.1)$$

Where P is the decimal percentage of maximum force applied. Charting this formula results in what has been referred to as Rohmert’s curve shown in figure 2.10. This curve displays static muscle endurance in time, as a function of the percent of the MVC exerted.

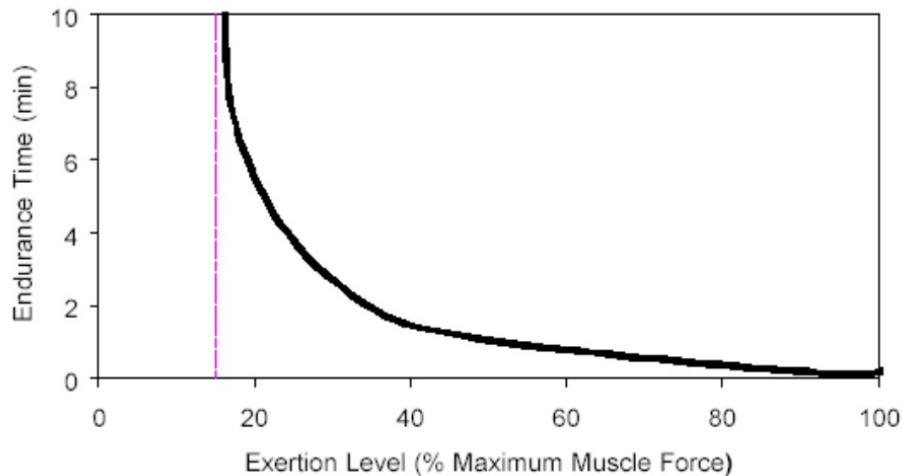


Figure 2.10: Rohmert's curve: endurance time v/s exertion level [74]

From this study, Rohmert deduced that 15% or less of MVC can be sustained indefinitely. The value of this type of analysis is the ability to match equipment or system requirements to the efficiency of human abilities. If a worker's ability to maintain a force is predictable based on a MVC, equipment can be designed to function within these parameters.

Multiple statistical muscle fatigue models have been developed to predict the endurance time (ET) of isometric tasks. These models have been used to validate other isometric regression models that have recently been developed [81,82].

A meta analysis has been done by Frey [83] to curve-fit muscle intensity- endurance time equations for several joint regions, including: ankle, knee, trunk, shoulder, elbow, and wrist/hand. The results of the meta-analysis validate nonlinear time endurance relationship (proposed by Rhomert, 1960), and also found different fatigue rate for each joint region.

Recently, Xia [3,84] gave a digital human muscle fatigue model that use the Modified Denavit-Hartenberg (modified DH) notation system which is the basis of the *SantosTM* kinematic model [85]. However, Xia and Frey-Law, 2008's [3] model has not been validated for dynamic conditions. Various muscle fatigue models are described in more details and classified in three categories according to their work and type of study: Generalized models, Static models and Dynamic models.

Generalized Muscle Fatigue Models

In previous subsections we see the muscle fatigue model for static and dynamic condition for particular kind of posture or movement. Now in this section we will discuss about some generalized muscle fatigue models on the basis of theoretical and mathematical explanation of muscle fatigue observing the effect and correlation with tendons, muscle fibers, muscle length etc.

Researchers have tried to understand muscle, muscle mechanics and anatomy since Leonardo Da Vincis Vitruvian Man around 1490. Researchers tried to study the phenomenon of muscles contraction and force production. Bendall in 1952 [86], first identify the relation between muscle contraction and adenosine triphosphate (ATP). Various models initially focused on the muscles contraction and response models such as the Hill's muscle model to explain the muscle response [38]. Hill proposed three component based model to explain muscle activation and energy generation. The popular state equation applicable to skeletal muscle that has been stimulated to show tetanic contraction. It relates tension to velocity with regard to the internal thermodynamics. The equation is described by the equation 2.2, where F is the tension or load in the muscle, v is the velocity of contraction, F_0 is the maximum isometric tension (or load) generated in the muscle, a is the coefficient of shortening heat, $b = a \cdot v_0 / F_0$, v_0 is the maximum velocity, when $F = 0$.

$$(v + b)(F + a) = b(F_0 + a) \quad (2.2)$$

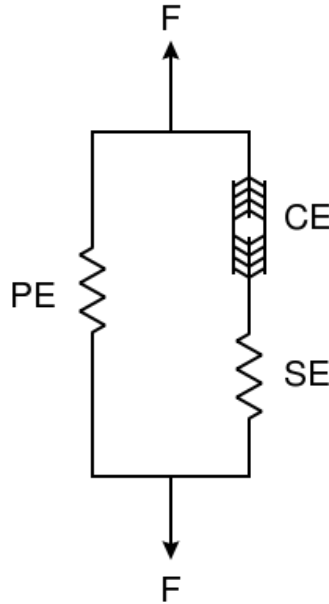


Figure 2.11: Hill elastic muscle model [38]

The three-element Hill muscle model is a representation of the muscle mechanical response. The model is constituted by a contractile element (CE) and two non-linear spring elements, one in series (SE) and another in parallel (PE) as shown in figure 2.11 [38]. The net force-length characteristics of a muscle is a combination of the force-length characteristics of both active and passive elements. The forces in the contractile element, in the series element and in the parallel element, F_{CE} , F_{SE} and F_{PE} , respectively, satisfy equation 2.3.

$$F = F_{PE} + F_{SE} \quad \text{and} \quad F_{CE} = F_{SE} \quad (2.3)$$

On the other hand, the length of muscle L and the lengths L_{CE} , L_{SE} and L_{PE} of

elements from the model satisfy

$$L = L_{PE} \quad \text{and} \quad L = L_{CE} + L_{SE}$$

The total muscle length is considered to be constant, the stretching of the series element can only occur if there is an equal shortening of the contractile element itself.

The parallel element arrangement shows the muscle passive behavior when it is stretched. And the series arrangement represent the tendon and muscle fiber elasticity. This model is helpful for estimating stress and muscle force, but Hills model is not able to model variation in force producing capability over time (i.e., muscle fatigue). This model explains the working of muscle activation and studies the muscle length shortening and dynamic parameters of muscle to understand the behavior of muscle. Various techniques were used to measure the energy of the muscle. Models developed now-a-days use his theory for fatigue study of muscles.

The Visco-elasticity of contracting muscle is considered to be property of two components of muscle. One component is considered to be undamped and elastic while the other is governed by equation 2.4 and corresponding energy relation. In equation 2.4, P represents the force factor (N), v represents velocity of contraction (cm/sec), a is load factor ($gm/sq.cm$) and b is velocity factor ($length/sec$). This shows that large force causes a low energy rate resulting into low speed and forces. The muscles are assumed to be in active state for this model.

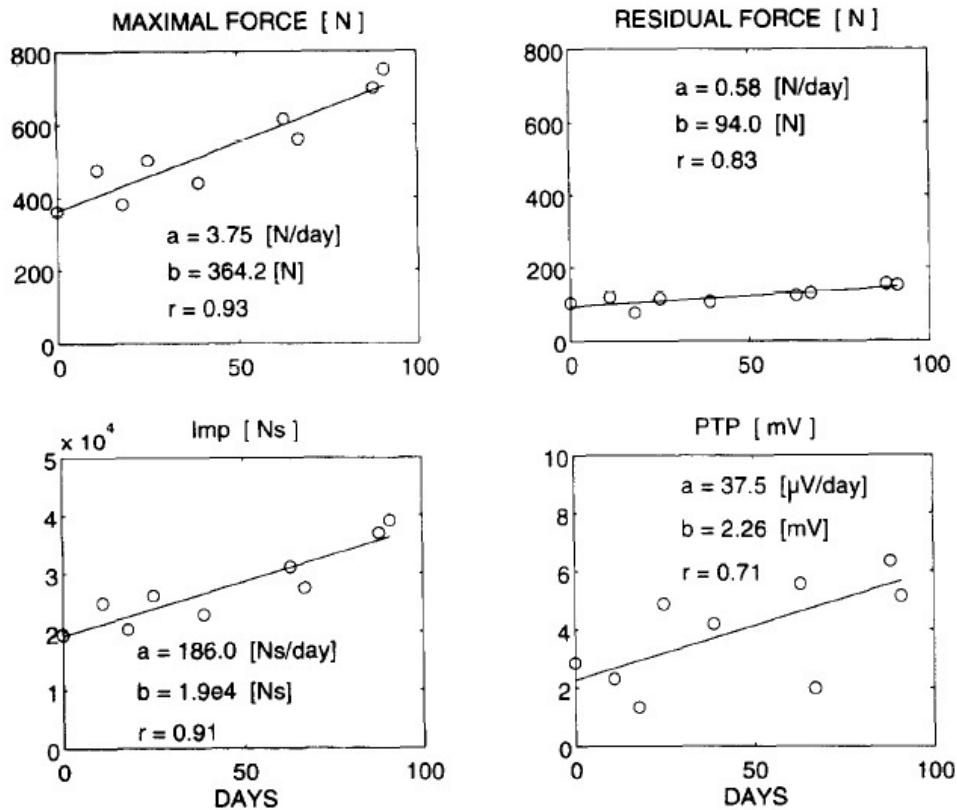


Figure 2.12: Variation of maximal force, residual forces, impulse, and EMG PTP [87]

$$(P + a)(v + b) = Constant. \tag{2.4}$$

There are some models which describe the muscle activation, fatigue and recovery process for muscle. Mizrahi [87] made a study of the work force and electromyographical(EMG) fatigue characteristics of functional electrical stimulation activated paralyzed muscles, both in initially unfatigued state and in reactivated state. The Force and EMG experiments have been performed on two subjects to see the effect of training and recovery on muscles. The variation of maximal force, residual force, impulse, and EMG peak to peak (PTP), over the functional electrical stimulation is shown in figure 2.12 [87]. In this study it was found that there is no significant effect of recovery on fatigue index.

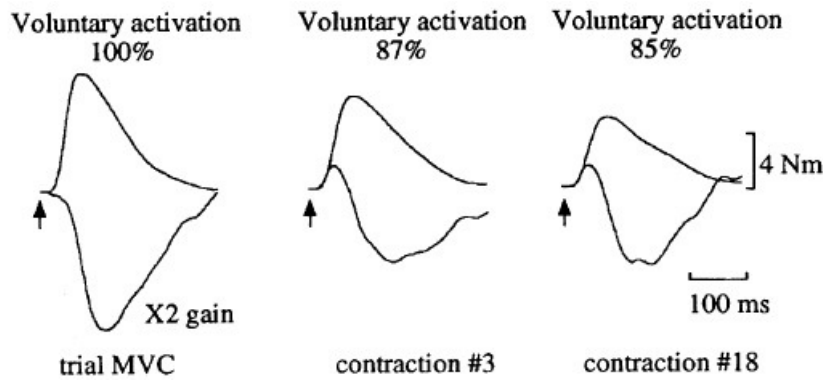


Figure 2.13: Twitch interpolation: Force tracing during fatigue protocol [16]

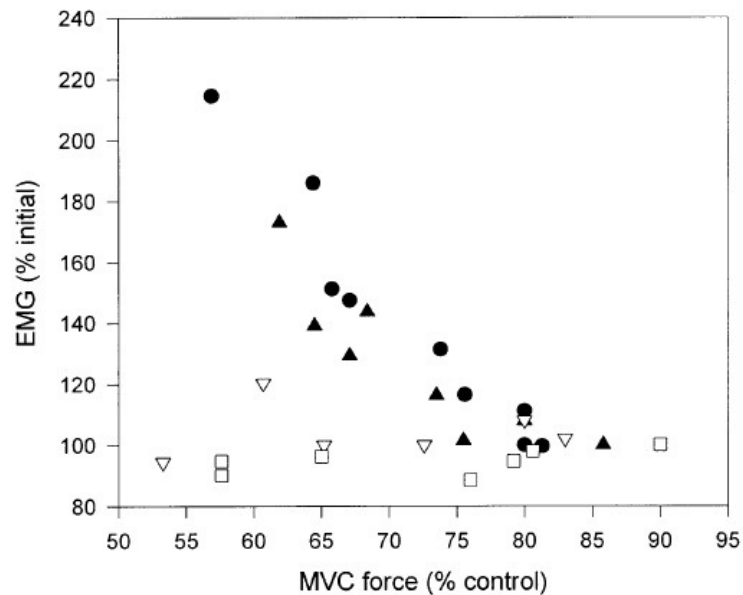
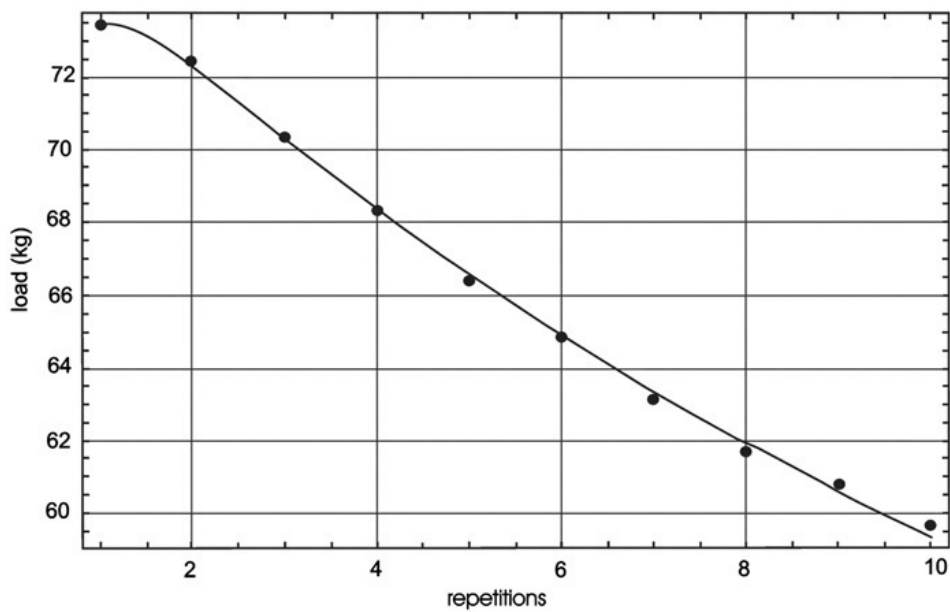


Figure 2.14: EMG amplitude recorded during repetitive quadriceps contraction at 30% of initial MVC in relation to MVC Force [16]

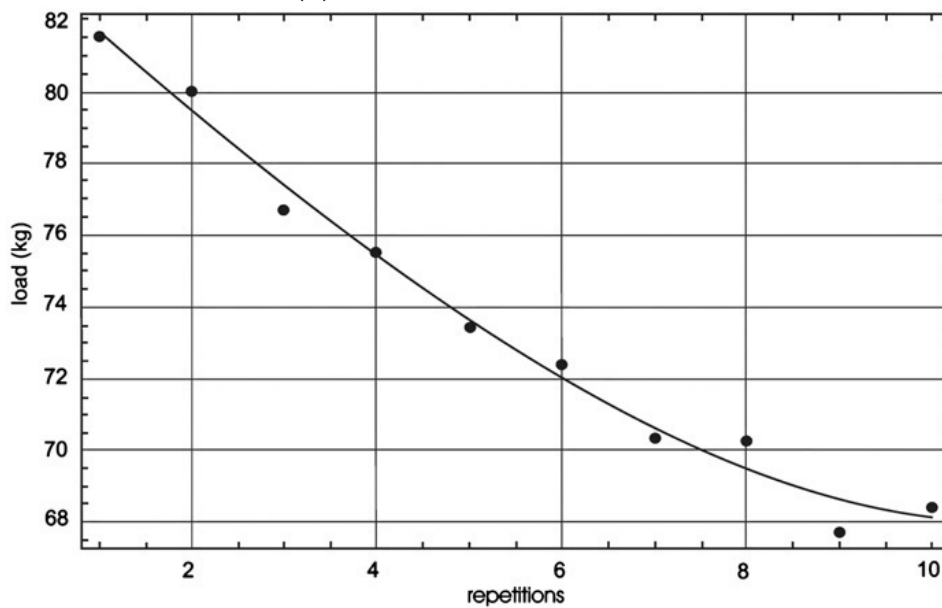
From figure 2.12, we can see the effect training on the muscle activity. The muscle strength which is less in start, increases with the duration of time. Hence muscle capacity and maximal force have direct relation with training time. Some Mechanical and Mathematical model of muscles are described in Syuzev, 2010 [42]. But did not explain much about fatigue and recovery. Various procedures for measurement of muscle fatigue

were explained in Vollestad, 1997 [16].

Twitch interpolation, endurance time and electromyography were considered for measurements. Twitch interpolation method is based on twitch contraction activated either by single or a double electrical stimulus. The twitch superimposed on a given target force level may thus be used to estimate the generated force relative to the current maximal force. In figure 2.13 [16] upper tracing shows the twitch force evoked by a super maximal paired stimuli in the relaxed muscle and the lower tracing shows the force response to the same stimulation during a MVC.



(a) Before Exercise protocol



(b) After Exercise protocol

Figure 2.15: Comparison of number of repetitions to the weight lifted [88]

Endurance time and changes in maximal capacity to generate force or power provide different processes induced in an exercise. But while studying the fatigue with EMG is not the same as expected. Vollestad take the experimental data of the four subjects from previous experiments and compare it. The comparison is based on the calculated MVC force with respect to EMG. The result in figure 2.14 [16] shows that two of the subjects represent an expected rise in EMG as fatigue develops (decrease in MVC), while in two subjects EMG remained unchanged.

During any work and movement some muscles motor units can be in the state active, rest and fatigued. Active motor units means motor units responding for contraction of muscle fibers however rest and fatigued state represent motor units not yet activated and motor units fatigued after activation respectively. In Liu, 2002's [2] model also took the three muscle characteristics, activated, rested and fatigued muscles are taken to explain the muscle activation and fatigue processes. It says that the muscle cannot attain the maximum force generation at the same time the load applied, there will always be some interval between the maximum force generated and the minimum force. It can also be understood by figure 2.16 [2]. The ordinate indicate the percentage proportion of each of the three group of motor units relative to total number of motor unit numbers in the involved muscles. M_A , M_F , M_{UC} represents activated motor units, fatigued motor units and number of motor units in rest state respectively and represented by equations 2.5, 2.6 and 2.7 respectively.

$$\frac{dM_A}{dt} = B.M_{UC} - F.M_A + R.M_F \quad (2.5)$$

$$\frac{dM_F}{dt} = F.M_A - R.M_F \quad (2.6)$$

$$M_{UC}(t) = M_0 - M_A(t) - M_F(t) \quad (2.7)$$

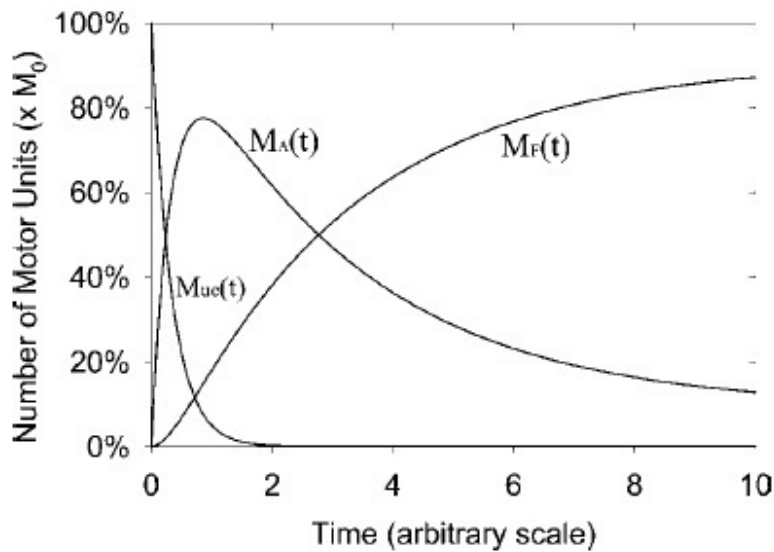


Figure 2.16: Curve for the activation of Motor units [2]

In equations 2.5, 2.6 and 2.7, M_0 represent total number of motor units. In figure 2.16, we see during contraction or muscle or movement, some motor units are activated while some are in the state of rest and fatigue. The state of motor units is represented by their respective notation.

In Gacesa, 2010 [88] the effect of exercise on muscle development and strength is given. The experiment was done on a person for 12 weeks and frequency of five sessions per week. During the experiment resting period of 1 minute was given between each set of the exercise. The results shown in figure 2.15 [88] show that with respect to time the muscle strength increases. The muscle strength increase after the exercise protocol. The load carrying capacity and number of repetitions for particular load also increased.

A theoretical approach for modeling peripheral muscle fatigue and recovery is given in Xia, 2008 [3]. Muscle is considered in one of the three state: resting, activated or fatigued. In this model the characteristic behavior of three basic component, the time course of activated, fatigued and rested muscles and endurance time with respect to four represented isometric loading condition: 100%, 70%, 30% and 10% of MVC shown in figure 2.17. Xia also used equations 2.5, 2.6 and 2.7 proposed by Liu, 2002 [2] for the validation of the model. The model show the recovery effect during the rest period. It also predict the effect of rest period on endurance time.

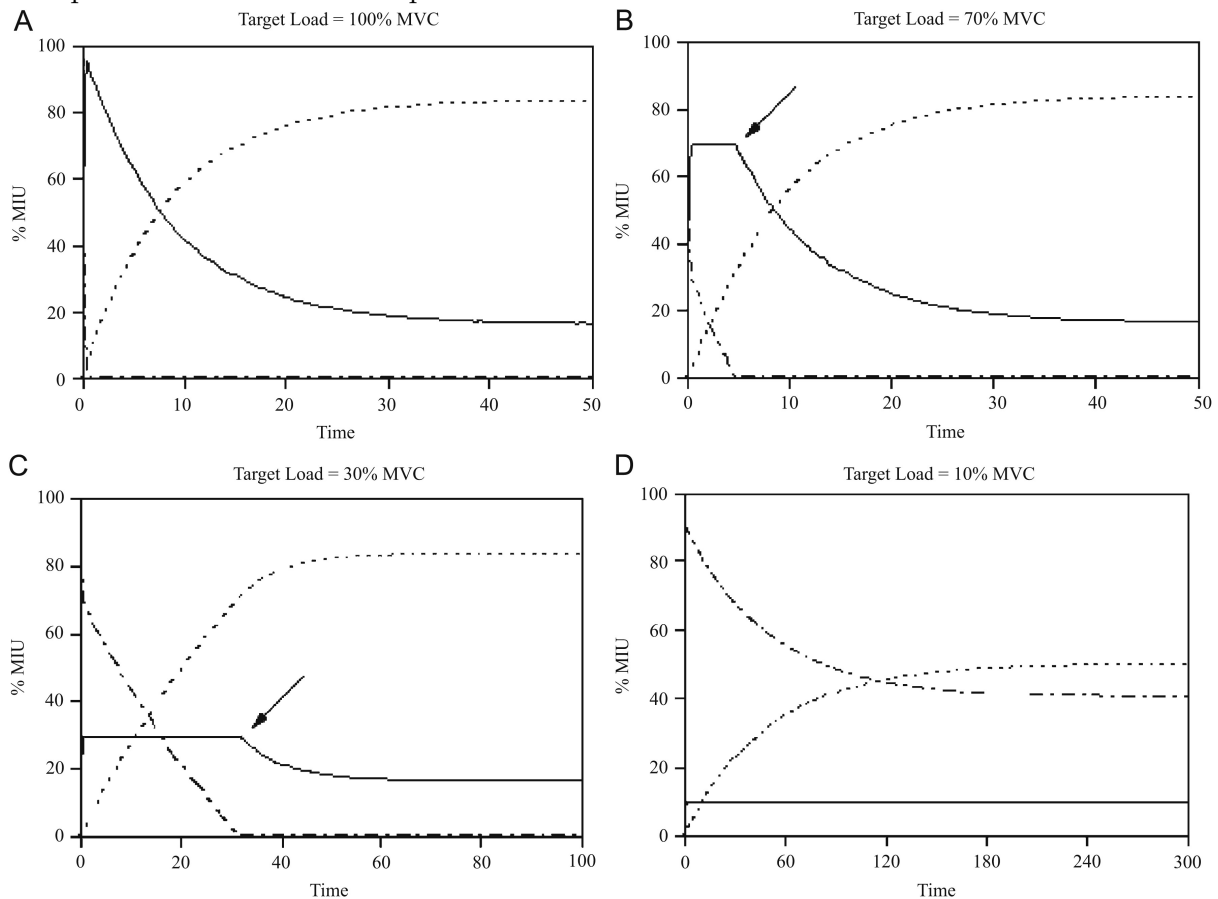


Figure 2.17: Distribution of muscle between compartments for four isometric loading conditions, where active (M_A : solid line), fatigued (M_F : dotted line), and resting (M_R : dotdash line) states are in percent of total motor units (%MU). Arrows demonstrate endurance time, where M_A cannot maintain target load. [3]

Figure 2.17 shows that larger the load less will be the endurance time for the subject. At different level of MVC load the endurance time decreased with the increase in MVC load.

Static Muscle Fatigue Models

L.Ma [5] gave a model for the arm to explain the behavior of fatigue in static condition. The experiment was done on a person carrying a drilling machine. In this posture the static motion muscle fatigue analysis was done. The picture of drilling task with geometrical representation of model is shown in figure 2.18 [5]. The effect of vibration was excluded from the analysis. The force from drilling machine is divided between two arms equally and the analysis was done only on right arm. The result shows that the reduction of elbow and shoulder strength relative to time, more the weight faster the strength will decrease. Figure 2.18b represent the Denavit-Hartenberg (DH) notation methods to describe the geometric structure of the right arm where a_j is angle between axes Z_{j-1} and Z_j around the axes X_{j-1} , d_j is the distance between axes Z_{j-1} and Z_j along the axes X_{j-1} , θ_j is the angle between axes X_{j-1} and X_j around the axes Z_j and r_j is the distance between axes X_{j-1} and X_j along the axes Z_j .

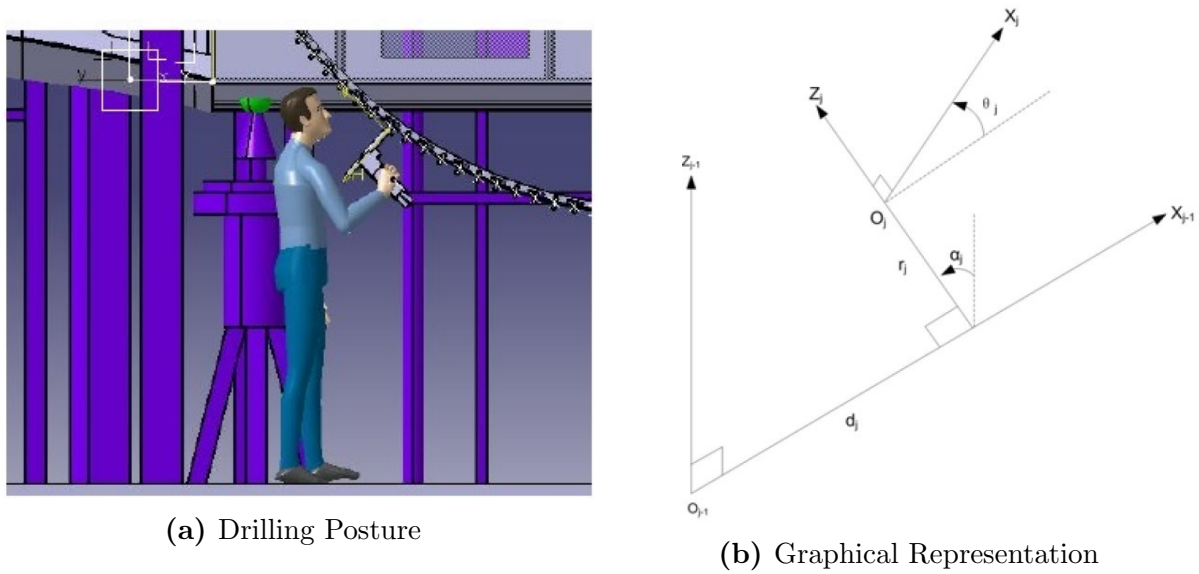


Figure 2.18: L.Ma's muscle fatigue model experimental setup [5]

The basic equation used for validation of fatigue in the static model is shown in equation 2.8. F_{cem} represents the current exertable maximum force, current capacity of muscle or non fatigued motor units of muscles, MVC is maximum voluntary contraction, F_{load} is muscle load at time instant t and k is fatigue ratio equals to 1.

$$\frac{dF_{cem}(t)}{dt} = -k \cdot \frac{F_{cem}(t)}{MVC} F_{load}(t) \quad (2.8)$$

The equation used for study effect of recovery is equation 2.9, where R is the recovery

rate of muscle motor units. The experiments have been performed with drilling machines of $2.5Kg$ and $3.5Kg$. The reduction in the strength with time for static loading condition of elbow and shoulder for different loads are shown in figure 2.19 and 2.20 for elbow and figure 2.21 and 2.22 for shoulder. Result shows that for larger the weight of drilling machine faster is the reduction in strength of both elbow and shoulder while for less weight the reduction in strength is smaller. In figure 2.19 to 2.22, α_e and α_s represents the elbow and shoulder angle respectively. σ_i and σ_j are the standard deviation along strength s_j . Blue horizontal line represent the joint torque, Γ_j .

$$\frac{dF_{cem}(t)}{dt} = R(MVC - F_{cem}) \quad (2.9)$$

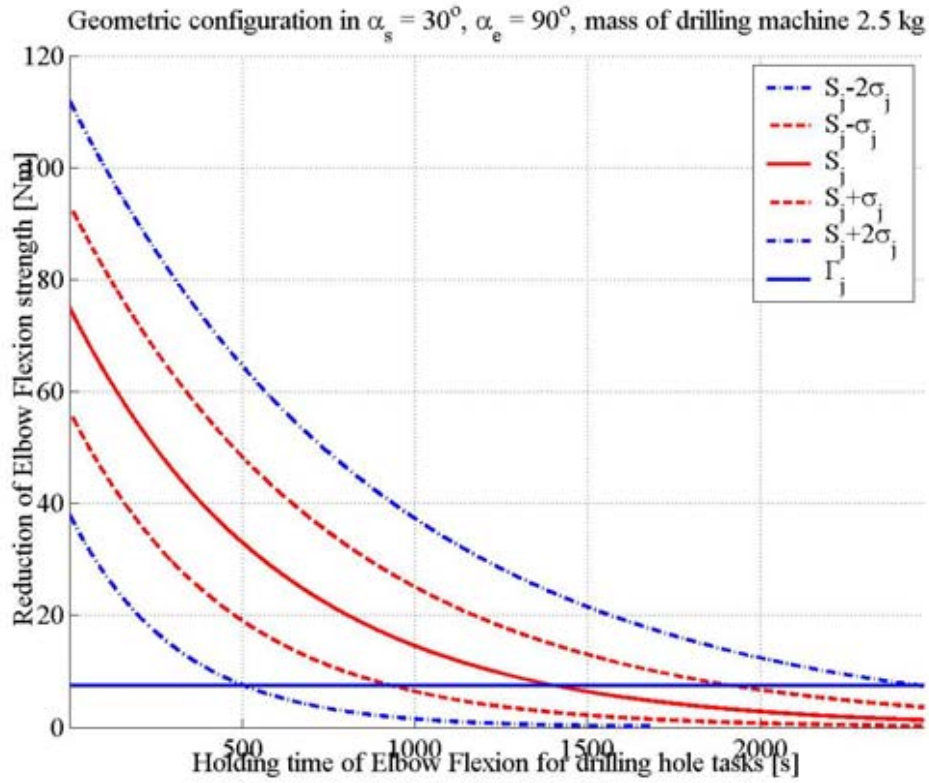


Figure 2.19: Reduction of the elbow strength while holding a drilling machine weighted as 2.5 kg [5]

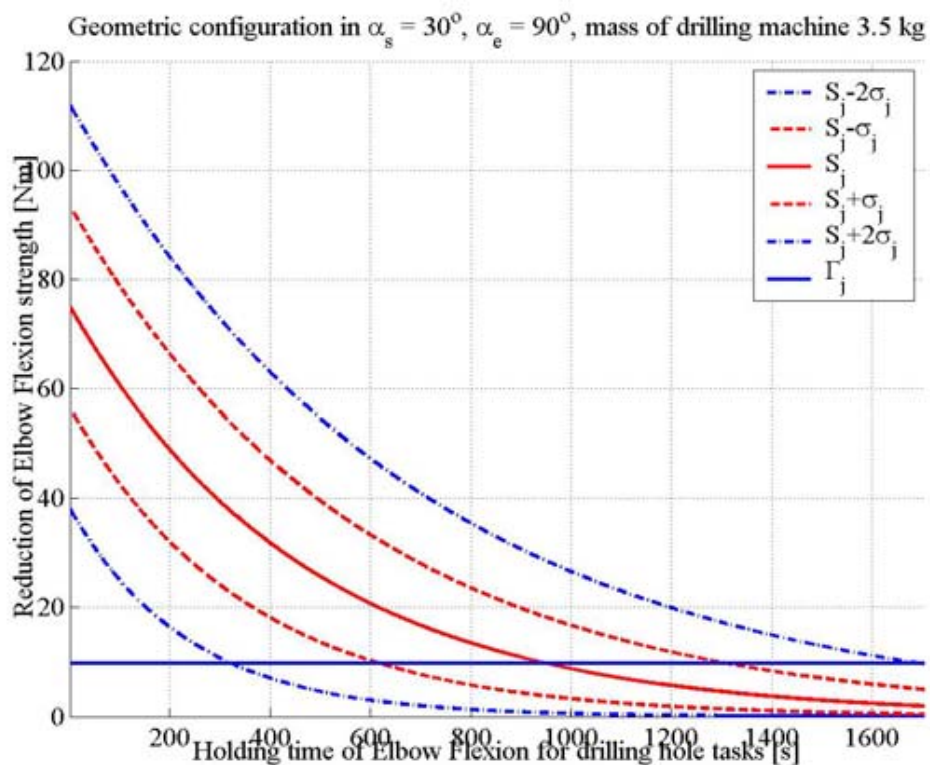


Figure 2.20: Reduction of the elbow strength while holding a drilling machine weighted as 3.5 kg [5]

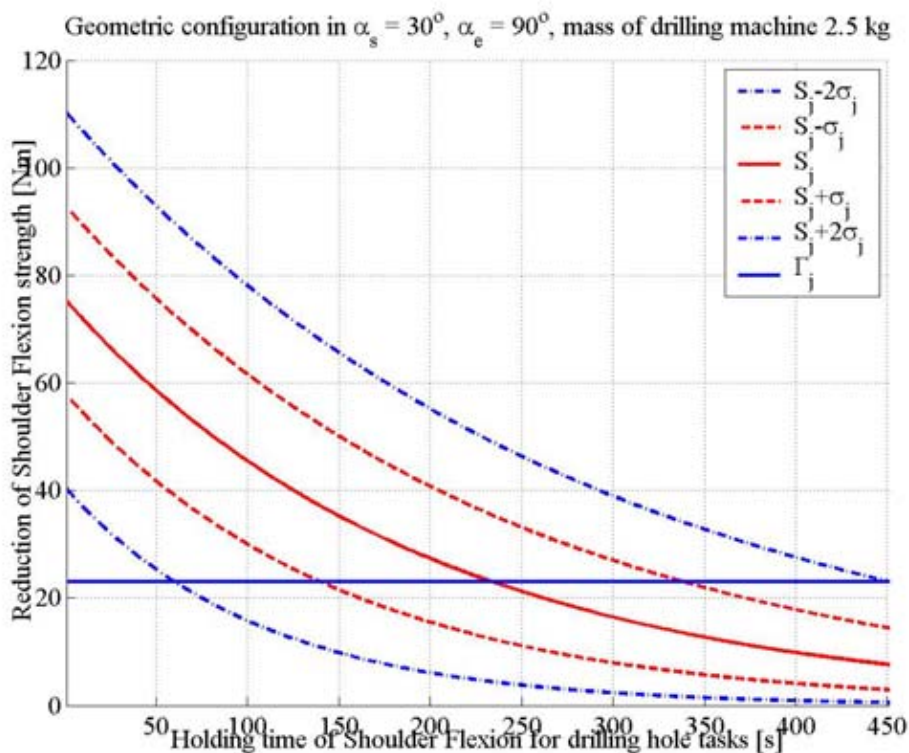


Figure 2.21: Reduction of the shoulder strength while holding a drilling machine weighted as 2.5 kg [5]

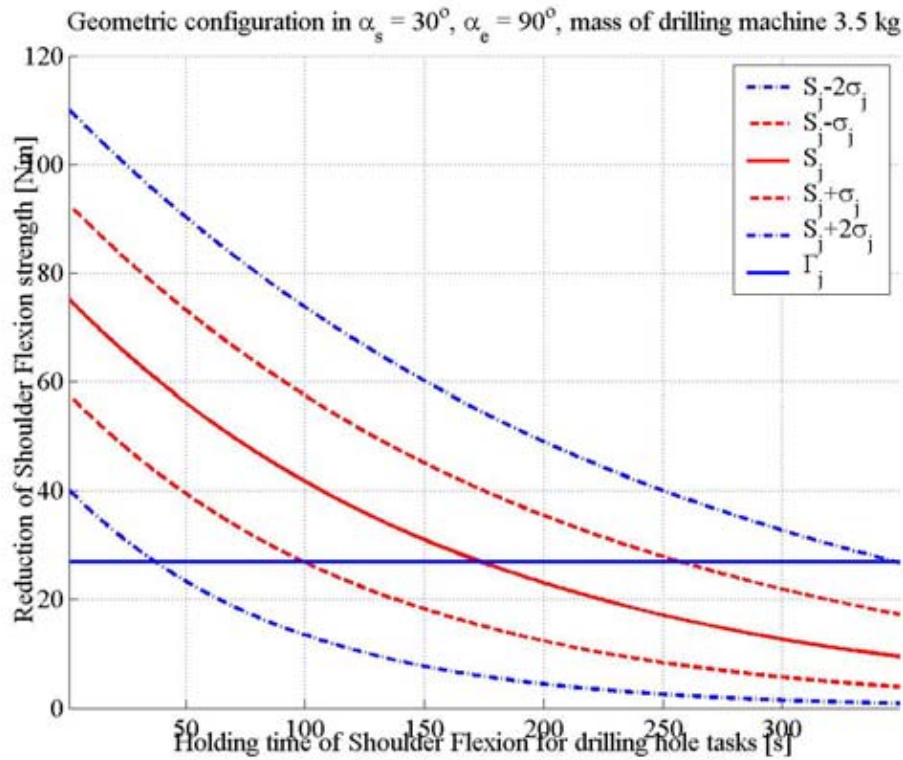
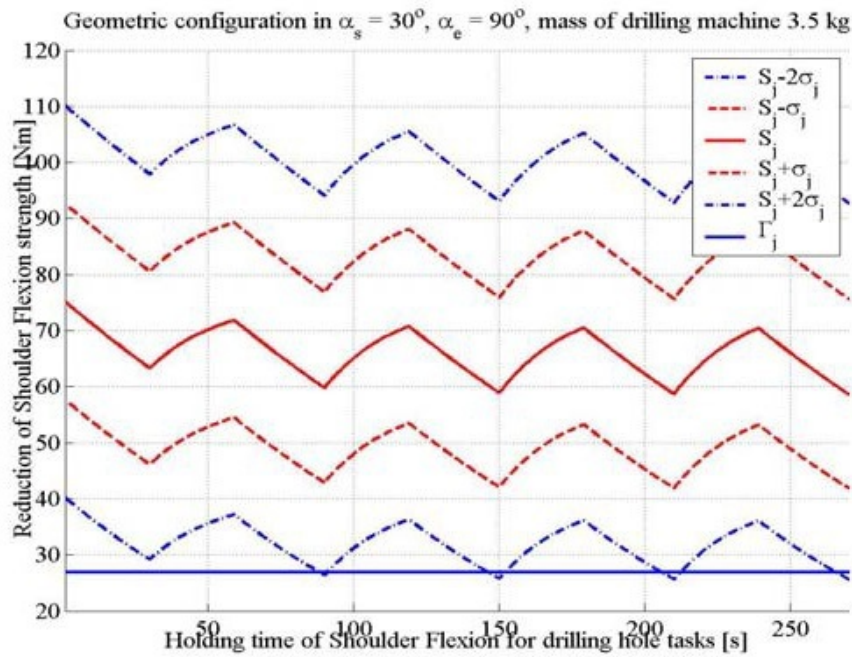
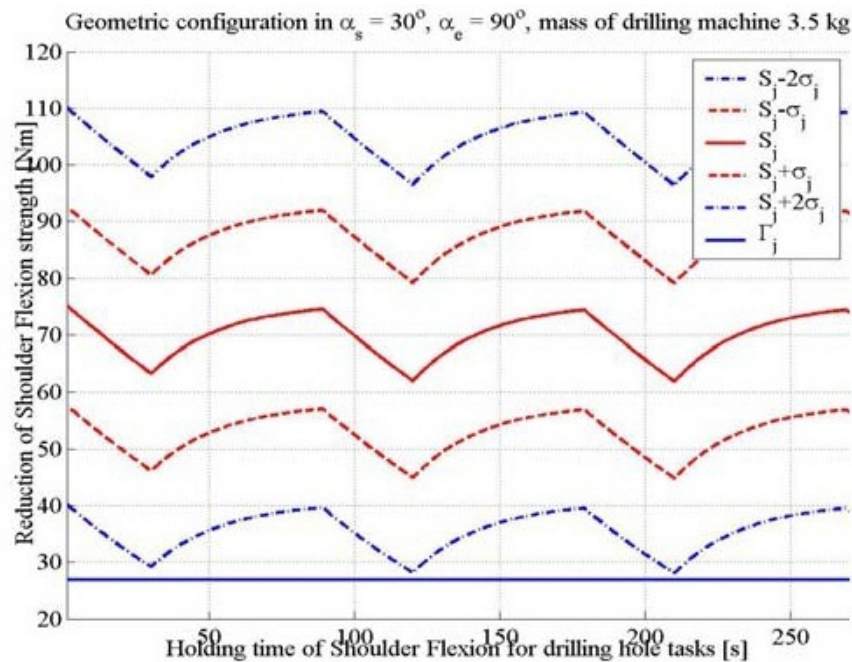


Figure 2.22: Reduction of the shoulder strength while holding a drilling machine weighted as 3.5 kg [5]

It also explains the effect of recovery and shows that larger the recovery time better will be the gain in strength. As shown in figure 2.23 [5] part a shows the recovery for 30 sec and part b shows the recovery for 60 sec which shows that more the recovery time better will be the strength. In both graph comparison we can say that more the recovery time more will be the gain in force output or muscles activation.



(a) Recovery effect in 30 sec [5]



(b) Recovery Effect in 60 sec [5]

Figure 2.23: Recovery effect in L.Ma's Model

In Silva, 2010 [37] Hill Type muscle model is used to validate the muscle model with fatigue. Opensim is used to simulate the model. this model consider five muscles to control the movement of arm. Each group of muscle activated at a time for any movement. The effect of co-activation is neglected. The model is considered to be the static model of human arm, shown in figure 2.24 [37] in which the constant load is applied at the end of arm to see the effect of force on muscle fatigue relative to time. Result shows the decrease in muscle load carrying capacity with respect to time represented in figure 2.25 [37]. The figure represents different muscles and their change in strength with

respect to time for certain amount of load.

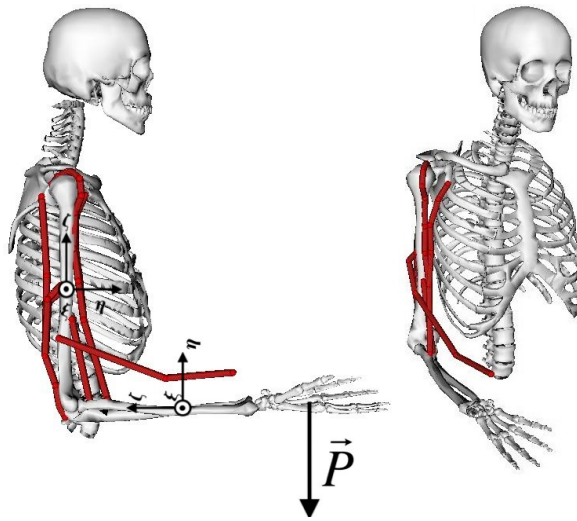


Figure 2.24: Silva's muscle fatigue model in Opensim [37]

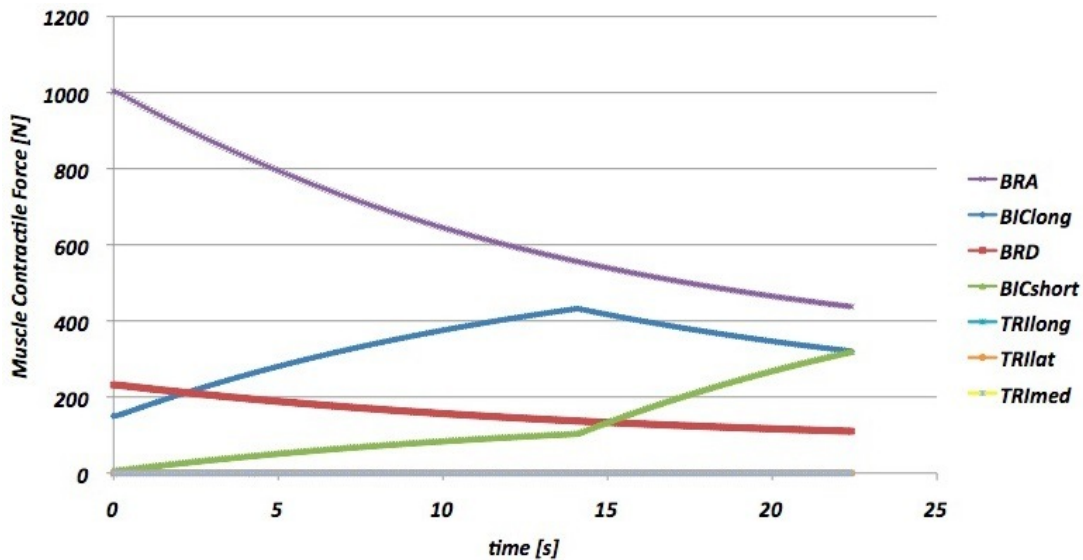


Figure 2.25: Muscle Load Carrying Capacity [37]

Freund [1] gave a bio mechanical model of forearm, consisting of 61 muscle-tendon systems or tendon. This model explains the role of muscle tendon in controlling the movement of forearm. The motion of pressing, grasping and lifting of loads are considered for the analysis figure 2.26 [1]. It explains the co-activation of muscle fibers to control any movement. Simultaneous activity of agonist and antagonist muscles [89] may be observed in this model. In this work it was explained that arms muscles act differently if we carry the load with different postures. For example If the load is carried in hand in downward direction then the weight of the load will try to open the hand and at the same time the muscle will co-contract to not let hand open and hold the weight at the same position.

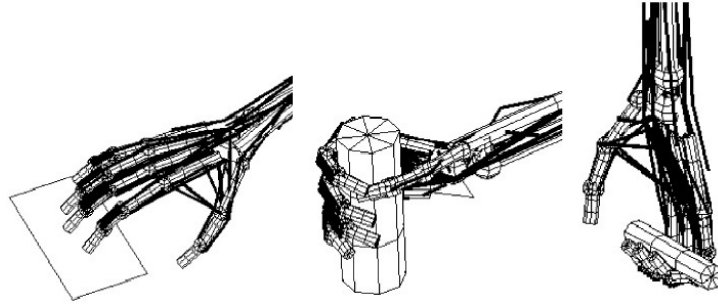


Figure 2.26: Various Posture of Forearm: Pressing, Grasping and Lifting [1]

The static model described above lack in the applicability to the dynamic conditions and some models have many variables which make them less realistic in real conditions. The need of a muscle fatigue model for the dynamic situations is proposed by some of the model described in the next section. One other limitation in this model is that it did not include co-contraction of muscles into consideration.

Dynamic Muscle Fatigue Models

In Fayazi,2013 [90] the optimal pacing time was calculated using a cyclist's fatigue dynamics. Thighs and foreleg muscle are focused during the study. The various parameters of cyclist: Power, force, state of fatigue, velocity, grade force, and elevation was obtained parallel and comparison was made with the obtained data with optimal pace. It was observed that with optimal pace the fatigue is greater as compared to normal riding condition.

In R.Ma, 2012 [6], the dynamic model was presented to analyze the muscle fatigue with recovery. R.Ma also considered two groups of muscles, agonist and antagonist for muscle activity. This model takes fatigue of muscle proportional to joint torque. The study was done on the elbow joint with push pull motion with fixed interval of rest and motion. The graphical representation of arm is in figure 2.27 [6]. In this experiment the torque is measured and compared relative to time and fatigue parameter is observed.

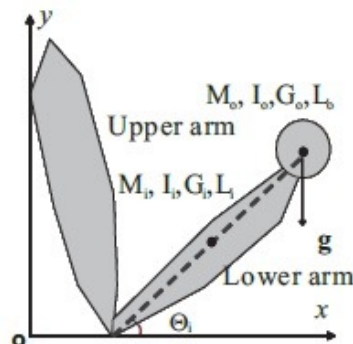


Figure 2.27: Graphical Representation of R.Ma's experimental model [6]

The mathematical equation used in this model for study of fatigue is shown in equation 2.10. Where Γ_{cem} is current exertable maximum joint torque, which represents the torque to apply according to the external load, Γ_{MVC} is maximum voluntary contraction

at joint, $\Gamma_{joint}(t)$ is Joint torque, which represents the torque generated at the joint, and k is fatigue rate.

$$\frac{d\Gamma_{cem}(t)}{dt} = -k \frac{\Gamma_{cem}}{\Gamma_{MVC}} \Gamma_{joint}(t), \quad (2.10)$$

In this model it is shown that more the applied torque, faster will be the reduction of capacity of muscles or increase in fatigue. The fatigue rate k for every person is different.

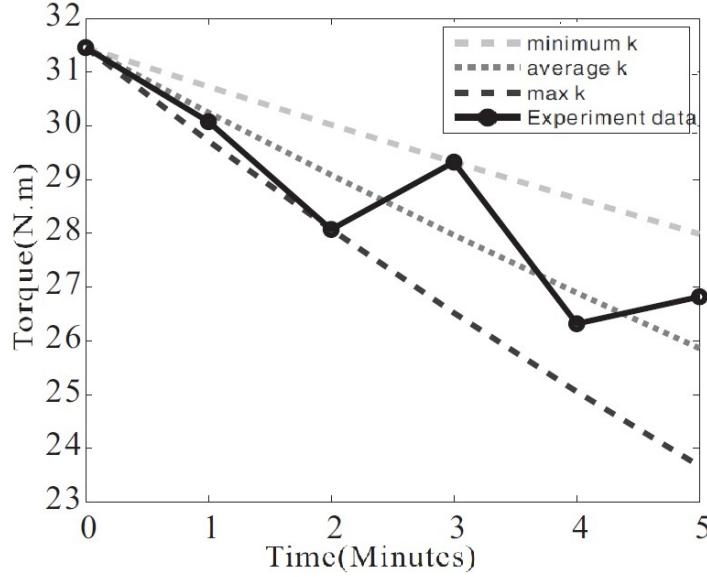


Figure 2.28: R.Ma's theoretical evolution of Γ_{cem} and experiment data using different values of $k_{agonist}$ [6]

For $t = 1, 2, 3, 4, 5$ minutes, the agonistic and antagonistic muscle group fatigue rate were evaluated as follows:

$$k_{agonist} = [0.13, 0.17, 0.07, 0.13, 0.09][min^{-1}]$$

$$k_{antagonist} = [19.56, 28.07, 20.32, 19.86, 18.36][min^{-1}]$$

Figure 2.28 represents the theoretical evolution of Γ_{cem} for different value of k in R.Ma model. The values of K which is obtained from the experiment by putting the experiment data in to the model is 0.87, so $k_{agonist}$ is a realistic value due to the fact that blood circulation is better during dynamic motions. Conversely, $k_{agonist}$ seems to be too high. In fact, due to the co-contraction activities influence, the torque of the antagonistic muscle group is higher than the results computed by the dynamic model. The dark bold line in the figure represents the experimentally calculated values of the torque and dotted lines represents the theoretically calculated values of torque for the values of $k_{antagonist}$ and $k_{agonist}$. This means the experimentally calculated values of torque lies with in the range of theoretical values.

There can be a relation between accuracy of task performed with fatigue rate.

Means more the fatigue less accurate will the task performed. accuracy is depend on the co-contraction of the muscles during the fatigue. For better accuracy good co-contraction can be needed during fatigue. In Missenard, 2008 [44] and Macefield, 2000 [91], the effect of fatigue on accuracy and co-contraction of muscles with their inter-relation has been discussed. The experiment was conducted on seven participants aged between 24 and 34 for two isometric flexions and two isometric extensions. The person had to move his arm in horizontal plane along vertical axis about elbow as shown in figure 2.29 [44]. During motion it is observed that the accuracy of reaching the start point and end point decreases with fatigue and for maintaining the accuracy the co-contraction of muscle pertains. And after some time accuracy again start decreasing with fatigue of supported muscles. The study was done two times as pre-fatigue movement session and post fatigue movement session with different MVC level before each session. Based on EMG signals the index of co-contraction (CI) was estimated by equation 2.11, where EMG_{min} is minimum level of EMG, EMG_{ago} is EMG at agonist pair and EMG_{ant} is EMG at antagonist pair. In (Fig. 2.30) [44] we can see the effect of co-contraction in two movement sessions. The experiment was done on thigh and toe extensors.

$$CI = \frac{\int_{t_o}^{t_f} EMG_{min} \times dt}{\int_{t_o}^{t_f} [EMG_{ago} + EMG_{ant}] \times dt} \times 100 \quad (2.11)$$

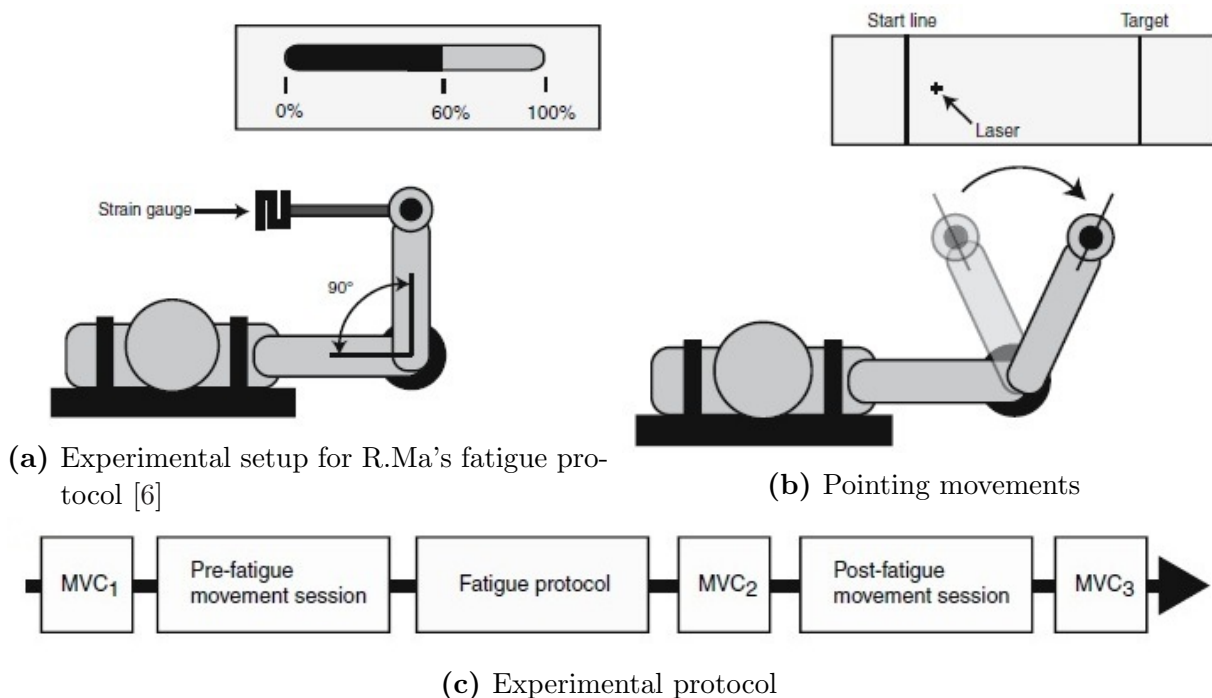


Figure 2.29: Experimental Setup and Protocol [44]

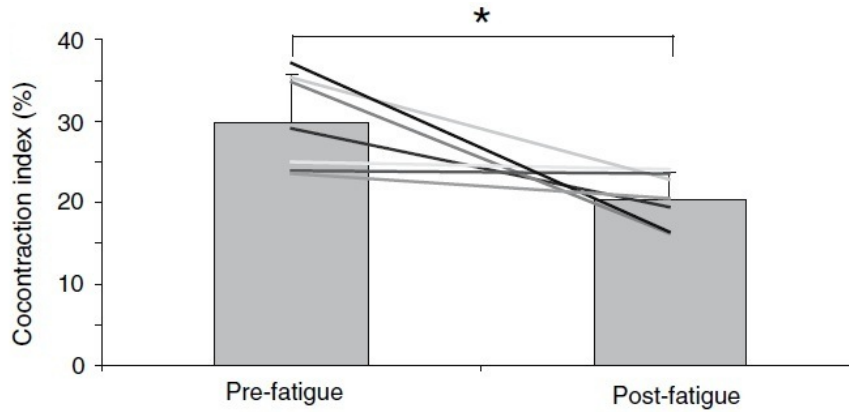


Figure 2.30: Co-contraction Index in two movement session [44]

In figure 2.29, we can see that strain gauge is attached to the hand to measure the force applied by the hand. The subject have to complete the motion in given range from start line to end line. It is observed (see figure 2.30) that co-contraction index percentage reduces after post fatigue condition.

2.5.2 Literature Synthesis

All the models described in section 2.5.1 have some limitations and specific assumptions. The models are compared on the basis of type of model, studied factor (fatigue or recovery), whether the models have experimental validation, assumptions and their limitations. Most of the models lack in including co-contraction or co-activation in their study which make those models less realistic because co-contraction have significant contribution in each motion or activity. The literature synthesis of these models are shown in table 2.1.

In table 2.1, general model means the models which shows the generalized fatigue model not specified to any particular joint, static model are models applicable to static situation and dynamic models are the models applied to dynamic situations. Joint based and fiber based model are the application of models to the motion at the joint and the mechanism of muscles activity like change in length, respectively.

The literature synthesis shown in table 2.1 details assumptions and limitations of various models with the type of their models. Some models have limited experimental validation which means that these models have been explained theoretically but not validated experimentally. Some models have one muscle group activation that means only one group of muscle will be activated for one side of motion neglecting co-contraction of the muscles. Some models limit their study to either peripheral fatigue or central fatigue which make their application limited. In table 2.1, fiber based model means there is no consideration of joint forces to measure the fatigue, which means muscle fatigue is measured by muscles activity and change in physical or chemical properties in muscles with fatigue while joint based model means that some joint is taken into consideration to measure the forces or torque at the joint.

The major limitations of all the models are:

- No co-contraction between the muscles.
- No global dynamic muscle fatigue model.
- No experimental validation of new model.

Co-contraction is a very common phenomenon of the muscle activity. But there is no use of co-contraction in these models. The question arises, whether it is necessary to include co-contraction into this consideration or not? It is required to check the viability of co-contraction factor in the study of muscle fatigue models. Many of the models are mostly limited to application of model either in static or dynamic condition for particular kind of motion. There is a necessity to extend the model for dynamic situation. The dynamic model should be applicable to global joints of the body. Most of the models are not experimentally validated for any static or dynamic situation. The mathematical description of the model can only be validated through real experiments on human being for the real life industrial tasks.

In this thesis we are going to address these major limitations or problems from the previous models to develop a new dynamics muscle fatigue model which will be more realistic and globally applicable to whole body.

Authors	Year	Model	Fatigue	Recovery	Experiment Validation
Mizrahi et al.	1997	Fiber/General	Yes	Partial	Yes/Thigh
Vollestad et al.	1997	Fiber/General	yes	yes	yes
Freund et al.	2000	Fiber/Static	yes	NA	NA/Hand
Liu et al.	2002	Fiber/Dynamic	Yes	Yes	NA/Hand
Xia et al.	2008	Fiber/General	Yes	Yes	NA
L.Ma et al.	2008&2009	Joint/Static	Yes	Yes	Yes/Arm
Gacesa et al.	2010	Fiber/Dynamic	Partial	Partial	Yes/Elbow Extensors
Silva et al.	2011	Joint/Static	Yes	yes	Limited/Arm
R.Ma et al.	2012	Joint/Dynamic	Yes	Yes	Yes/Arm
Fayazi et al.	2013	Fiber/Dynamic	Yes	Yes	Yes/whole body
Assumptions			Limitations		
Mizrahi et al.	Constant recovery,no fatigue effect				No co-contraction
Vollestad et al.	Endurance time change is constant				No co-contraction, experiment validation
Freund et al.	Upper body muscles fix, no co activation				No experimental and mathematical validation
Liu et al.	brain effort constant				No co-contraction and experimental validation
Xia et al.	peripheral fatigue only				limited application
L.Ma et al.	no vibration, external load constant				No co-contraction included
Gacesa et al.	Muscle activity constant				no co-contraction and recovery
Silva et al.	constant muscle area for same force				No co-contraction, limited validation
R.Ma et al.	Constant load, one group of muscle active				No co-contraction
Fayazi et al.	muscle length change is constant				No co contraction

Table 2.1: Literature Synthesis of Fatigue and Recovery Models

2.6 Conclusions

2.6.1 Problematic analysis of muscle fatigue models

Shortcoming in conventional ergonomic tools

The available digital human modeling (DHM) tools cannot evaluate fatigue properly because of lack in dynamic and realistic fatigue model application. That is why new fatigue evaluation tool is needed to be developed and integrated with new more realistic static and dynamic muscle fatigue model.

In most of the tools, the physical capacity is constant in the simulation. The effect of fatigue and recovery with time is not considered properly. Different working postures provide changes in physical status and parameters of fatigue but in most of the tools it is considered to be constant.

Shortcoming in fatigue models

Some fatigue models have been combined with digital human tools to predict the variable physical parameters and strength. For example, Wexler's fatigue model [84] has been incorporated into *SantosTM* [85], and Giat's fatigue model [73] has been developed based on Hill's muscle model [38]. However, these models either have too many variables which makes them less feasible in real situations or too many assumptions which make them less realistic. Most of these models also never integrate co-contraction into fatigue and recovery models. It is necessary to propose a simple new dynamic muscle fatigue model with less variables which can be useful for ergonomics applications.

2.6.2 Solution for fatigue analysis

In order to analyze the physical work in more details and predict muscle fatigue, a new dynamic human muscle fatigue model, including the overall dynamic working process, should be developed. To assess and predict the potential MSD risks objectively it is necessary to analyze muscle fatigue and recovery with taking co-contraction into consideration. This concern became the main content of our research work.

In this thesis, we are going to analyze muscle fatigue for different human gestures. The fatigue and recovery effects are modeled by a new simple dynamic muscle fatigue and recovery model. From macroscopic point of view, temporal parameters, physical factors, and personal factors are taken into consideration. This new model has to be validated theoretically and experimentally both. The experimental case study are done to show the applicability of our model in posture analysis and posture prediction with the analysis of fatigue and recovery. To study muscle activity and co-contraction, electromyography technique is used.

Chapter 3

Dynamic Muscle Fatigue Model

3.1 Theory of Dynamic Muscle Fatigue Model

The dynamic muscle fatigue model will be applicable to the dynamic motion of the human body parts. The motions like push-pull operation of the arm, walking, pronation, supination etc. are examples of the dynamic motion. A dynamic muscle fatigue model is proposed by L.Ma [5,92,93] firstly applied on static drilling task. R.Ma [6,43] developed this model for the dynamic motions like push/pull operation of the arm. However, no consideration about the co-contraction of paired muscles is taken. The co-contraction of the muscles are not included in both the studied models. Any known dynamic muscle fatigue model which includes co-contraction in study is not known till now.

* **Co-contraction:** Simultaneous contraction of both the agonist and antagonist muscle around a joint to hold a stable position is known as co-contraction between the muscle.

***Co-Activation:** It is a phenomenon in which a muscle is activated coordinately with another muscle.

The new dynamic muscle fatigue model which includes co-contraction effect is the extension of R.Ma's Model (without co-contraction). The co-contraction is included as a multiplicative factor to the model. The detailed description of this is given in the next section. Any co-activation and contribution of other body muscles are not included in the current study.

The muscle fatigue can be defined according to the initial MVC, duration of effort, nature of motion (static or dynamic) and the morphological and psychological characteristics of the individual.

3.1.1 Base of the Dynamic Muscle Fatigue Model

R.Ma's dynamic model of muscle fatigue was proposed, based on the considerations static model proposed by Ma et al. (2009) [82]. This model assumes that muscle fatigue is related to the external load, and also to operator fatigue factor. L.Ma's static muscle fatigue model is expressed as follows:

$$\frac{dF_{cem}(t)}{dt} = -k \cdot \frac{F_{cem}(t)}{MVC} \cdot F_{load}(t) \quad (3.1)$$

By solving the differential equation of the first order and considering that $F_{cem}(0) = MVC$, F_{cem} is expressed by:

$$F_{cem}(t) = F_{MVC} \cdot e^{\int_0^t -k \cdot \frac{F_{load}(u)}{F_{MVC}} du} \quad (3.2)$$

If, F_{Load} and F_{MVC} held constant, the model can be simplifies as follows:

$$F_{cem}(t) = F_{MVC} \cdot e^{-kCt}, \quad \text{where } C = \frac{F_{Load}}{F_{MVC}} \quad (3.3)$$

And R.Ma's dynamic muscle fatigue model with joint torque can be written as:

$$\frac{d\Gamma_{cem}(t)}{dt} = -k \frac{\Gamma_{cem}}{\Gamma_{MVC}} \Gamma_{joint}(t) \quad (3.4)$$

and, Γ_{Joint} and Γ_{MVC} held constant, the model can be simplifies as follows:

$$\Gamma_{cem}(t) = \Gamma_{MVC} \cdot e^{-k_{torque}Ct}, \quad \text{where } C = \frac{\Gamma_{Joint}}{\Gamma_{MVC}} \quad (3.5)$$

The various parameters involved in this dynamic model of fatigue are listed in the table 3.1

3.1.2 Hypothesis for New Dynamic Muscle Fatigue Model

In dynamic situation, body posture modified over time. The desired torque at joint is also a function of time.

From robotics vision, we assume that between two joints, the human body is rigid and homogeneous, and we can apply the laws of dynamics. Based on the study by R.Ma [6] and L.Ma [82], we assume the following assumptions. As we know, there are

Elements	Unit	Description
k_{torque}	min^{-1}	Fatigue Parameter, constant
Γ_{MVC}	N.m	Maximum torque that muscle can exert on joint
Γ_{Joint}	N.m	Torque from external load
Γ_{cem}	N.m	Current capacity of the muscle to generate the torque
k	min^{-1}	Fatigue factor
F_{MVC}	N	Maximum force from muscle voluntary contraction
F_{Load}	N	Force generated by muscle because of external load
F_{cem}	N	Current capacity of the muscle to generate the force

Table 3.1: Parameters of muscle fatigue models

two major muscle groups for each joint motion, agonist and antagonist. For push motion, a muscle motivates the motion while antagonist muscle makes the motion accurate and stable. If the motion is reversed, i.e, pull cycles, agonist and antagonist muscles switch their roles. Co-operation of the two muscles is called co-contraction. More the torque imposed on the joint, more will be the rate of muscle fatigue. Similarly, less the muscle's ability, the fatigue will be felt quickly. In this dynamic situation, the torque imposed on the joint is linked to the movement and not only to external loads. We make the following assumptions for a new dynamic muscle fatigue model:

1. Muscle fatigue is directly proportional to the torque applied at the joint (external load).
2. Muscle fatigue is inversely proportional to the maximum capacity (without fatigue) of muscle to generate torque Γ_{MVC} , which represent the maximum torque generated at the joint.
3. From the previous models, we consider that the evolution of Γ_{cem} (variation in torque or capacity of muscle with fatigue) can be represented by a linear differential equation of the first order.
4. Co-contraction factor "n" is introduced in the model as a multiplicative factor with fatigue parameter 'k' (represent the rate of fatigue during the protocol).
5. Co-contraction area for all cycle varies exponentially.
6. No effect of co-activation is considered in the model.
7. There are two group of muscle for each joint, i.e, agonist and antagonist. One group of muscles will be active during one kind of motion and other will be passive with co-contraction in between each activity. The co-contraction will be the common activity during activity of both muscles.

3.1.3 Co-contraction Factor

The co-contraction is the simultaneous contraction of both the agonist and antagonist muscle around a joint to hold a stable position at a time. The co-contraction is generally decreases as the fatigue increase [44]. It signifies that, it could follow some pattern of decrement on co-contraction. In our model we calculate co-contraction from the common activity between the two group of muscles. The co-contraction is the simultaneous contraction of both the agonist and antagonist muscle around a joint to hold

a stable position at a time. Assumptions made for finding co-contraction factor are as follows:

1. The co-contraction is the common activity between the two groups of acting muscles.
2. The co-contraction factor will be the same for each agonist and antagonist activities.

The co-contraction area can be understood by the figure 3.1. This figure is just an example representation of a motion cycle. In this figure, we introduce the common EMG activity between bicep and tricep muscle groups shown by the orange color, which is co-contraction index C_A . The formula for calculating the co-contraction index from EMG activities is given in equation 3.6. The trapezius activity shown along with the two muscles is co-activation.

$$C_A = \text{co-contraction index}$$

$$C_A = \frac{\int_{t_0}^{t_{100}} EMG_{min} \times dt}{\int_{t_0}^{t_{100}} [EMG_{agonist} + EMG_{antagonist}] \times dt} \quad (3.6)$$

where, EMG_{min} is the common area share by the EMG activity of bicep and tricep, $EMG_{agonist}$ and $EMG_{antagonist}$ are the full activities of the bicep and triceps muscle. The activities of the both the muscles are normalized with respect to the normalization value of the activities for the same muscle which can be calculated using the equation 4.1, it is because the absolute value of the external torque is the same for push/pull operation.

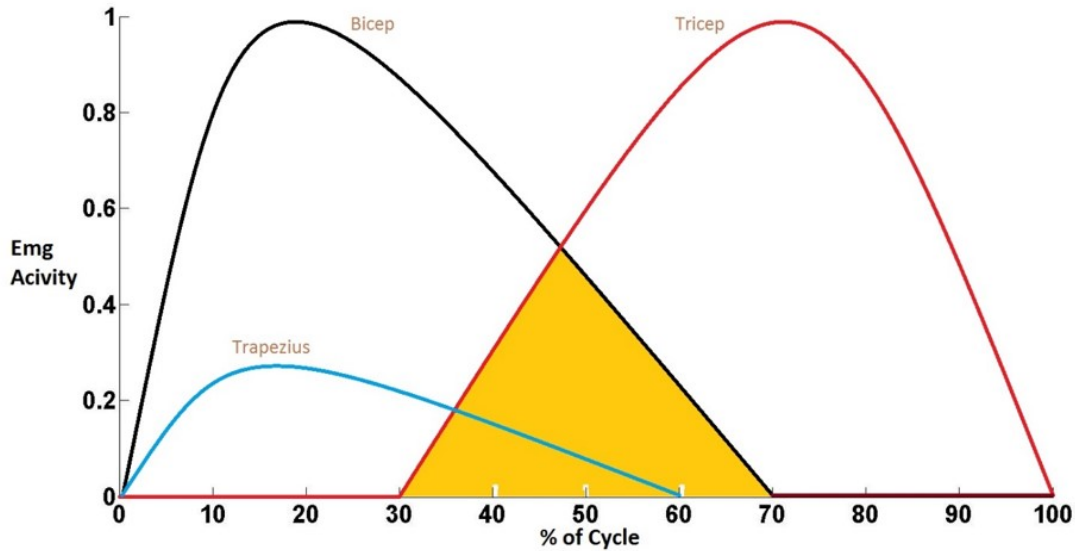


Figure 3.1: A representative plot of EMG activity of bicep, triceps and trapezius normalized with the maximum value of each muscle's activity for one cycle

C_A is a co-contraction index between the two group of muscles or common activities between the two muscle groups. We assume that curve fit for the co-contraction index for each subject will vary exponentially. The curve can vary according to the equation 3.7. This is an assumption for the variation in co-contraction index with respect to muscle fatigue so we have to verify it after experiment.

The curve of co-contraction index C_A can vary according to following equation:

$$C_A = a \cdot \exp b \cdot x \quad (3.7)$$

where, a and b are constant parameters and x is the time of the test.

Now can represent co-contraction factor n as follows:

$$n = \text{co-contraction factor}$$

$$n = 1 + C_A \quad (3.8)$$

The co-contraction factor n will be used in the muscle fatigue model. To see the effect of co-contraction factor in the model we will compare it with other models.

3.2 Proposed Dynamic Model of Muscular Fatigue

We choose the two parameters Γ_{joint} and Γ_{MVC} to build our muscle fatigue model. The hypotheses can then be incorporated into a mathematical model of muscle fatigue which is expressed as follows:

$$\frac{d\Gamma_{cem}(t)}{dt} = -k \cdot n \cdot \frac{\Gamma_{cem}}{\Gamma_{MVC}} \Gamma_{joint}(t) \quad (3.9)$$

where, k is the fatigue factor and n is the co-contraction factor.

If we assume that $\Gamma_{cem}(0) = \Gamma_{MVC} = \text{constant}$ and k is a constant, the integration result of the previous equation is given by the equation 3.10 and further can be written as equation 3.11.

$$\Gamma_{cem}(t) = \Gamma_{MVC} \cdot e^{-\frac{n \cdot k}{\Gamma_{MVC}} \int_0^t \Gamma_{Joint}(u) du} \quad (3.10)$$

$$\Gamma_{cem}(t) = \Gamma_{MVC} \cdot e^{-k \cdot n \cdot Ct}, \quad \text{where } C = \frac{\Gamma_{Joint}}{\Gamma_{MVC}} \quad (3.11)$$

The parameters for this model is expressed in the table 3.2.

Elements	Unit	Description
k	min^{-1}	Fatigue factor, constant
Γ_{MVC}	N.m	Maximum torque that muscle can exert on joint
Γ_{Joint}	N.m	Torque from external load
Γ_{cem}	N.m	Current capacity of the muscle to generate the torque
n	unit less	co-contraction factor

Table 3.2: Parameters of new dynamic muscle fatigue model

The value of Γ_{MVC} is a fixed calculated value for each subject. We assume that Γ_{MVC} is constant for some duration of time. With the help of robotic dynamic model, $\Gamma_{Joint}(u)$ can be modeled as equation 3.12 depending on the joint angle, velocity, acceleration and load.

$$\Gamma_{joint}(u) = \Gamma(u, \theta, \dot{\theta}, \ddot{\theta}) \quad (3.12)$$

Hence equation 3.10 can be written as equation 3.13 as follows:

$$\Gamma_{cem}(t) = \Gamma_{MVC} \cdot e^{-\frac{n \cdot k}{\Gamma_{MVC}} \int_0^t \Gamma(u, \theta, \dot{\theta}, \ddot{\theta}) du} \quad (3.13)$$

Equation 3.13 also represents our dynamic muscle fatigue model for a joint. According to our hypothesis, the motion of joint is driven by a pair of muscles. For each cycle, the Γ_{joint} positive and Γ_{joint} negative sign show joint torque by agonist and antagonist muscles respectively. Hence, there will be two fatigue rate parameters k ($k_{agonist}$ and $k_{antagonist}$) for dynamic motion. However, co-contraction factor will be only one value for dynamic motions.

3.2.1 Maximum Endurance Time (MET)

Endurance (also related to sufferance, resilience, constitution, fortitude, and hardiness) is the ability of an person or muscle to exert itself and remain active for a longer period of time, as well as its ability to resist, withstand, recover from, and have immunity to trauma, wounds, or fatigue. It is usually used in aerobic or anaerobic exercise. The definition of 'long' varies according to the type of exertion—minutes for high intensity anaerobic exercise, hours or days for low intensity aerobic exercise.

Muscular endurance is very important for people playing sports and who have to sustain an activity for long periods of time. Muscular endurance is determined by how well your slow twitch muscle fibers are developed. Muscle endurance is the ability to do some work or task over and over for an extended period of time without getting tired. The time when the force production can no longer be maintained is defined as the endurance time (ET).

The reduction in the maximum exertable force or torque capacity of muscle is one of the hypothesis for the proposed dynamic muscle fatigue model. Maximum endurance time (MET) represents the maximum time during which a static load can be maintained [81]. The MET is generally calculated as the percentage of the maximum voluntary contraction (%MVC) or to the relative force/torque ($\Gamma_{MVC} = \%MVC/100$). MET models are used to predict endurance time of a muscle under static or dynamic conditions or posture.

If we consider $\Gamma_{Load}(t)$ is constant in equation 3.11 for static situation then MET is the duration in which Γ_{cem} drop down to Γ_{Load} . MET can be calculated by equations 3.14 and 3.15.

$$\Gamma_{cem}(t) = \Gamma_{MVC} \cdot e^{-\int_0^t k \cdot n \cdot \frac{\Gamma_{Load}(u)}{\Gamma_{MVC}} du} \quad (3.14)$$

$$t = MET = -\frac{\ln\left(\frac{\Gamma_{Load}(t)}{MVC}\right)}{n \cdot k \cdot \frac{\Gamma_{Load}(t)}{MVC}} = -\frac{\ln(\Gamma_{MVC})}{n \cdot k \cdot \Gamma_{MVC}} \quad (3.15)$$

In the equation 3.15, MET is the time duration for the reduction in muscle strength from initial state to exhaustion. The Γ_{MVC} is the ratio of the external load to the MVC when the external load is constant. We can consider fatigue rate 'k' equal to 1 for best strength condition.

We can classify the dynamic tasks into two categories: periodic tasks (or repetitive) and non-periodic tasks. In the field of ergonomics, repetitive tasks are the main tasks that can lead to musculoskeletal disorders. We therefore, focus our study this type of task. We assume that the generated force can be broken into a periodic task time period 'T'. So for dynamic conditions we will calculate dynamic maximum endurance time (DMET).

For periodic or repeated dynamic motions the dynamic model can be represented by an equation 3.16 [40].

$$\Gamma_{cem}(t) = \Gamma_{MVC} \cdot e^{-\frac{n \cdot k}{\Gamma_{MVC}} \int_0^t \Gamma(u, \theta, \dot{\theta}, \ddot{\theta}) du} \quad (3.16)$$

For $t = T$; Γ_{cem} will be,

$$\Gamma_{cem}(T) = \Gamma_{MVC} \cdot e^{-\frac{n \cdot k}{\Gamma_{MVC}} \int_0^T \Gamma(u, \theta, \dot{\theta}, \ddot{\theta}) du} \quad (3.17)$$

Similarly, for $t = 2T$ or 2 cycles; Γ_{cem} will be,

$$\Gamma_{cem}(2T) = \Gamma_{MVC} \cdot e^{-\frac{n.k}{\Gamma_{MVC}} \int_0^{2T} \Gamma(u, \theta, \dot{\theta}, \ddot{\theta}) du} \quad (3.18)$$

The equation 3.18 can be rephrased as equation 3.19,

$$\begin{aligned} \Gamma_{cem}(2T) &= \Gamma_{MVC} \cdot e^{-\frac{n.k}{\Gamma_{MVC}} 2 \int_0^T \Gamma(u, \theta, \dot{\theta}, \ddot{\theta}) du} \\ &= \Gamma_{MVC} \cdot \left(e^{-\frac{n.k}{\Gamma_{MVC}} \int_0^T \Gamma(u, \theta, \dot{\theta}, \ddot{\theta}) du} \right)^2 \end{aligned} \quad (3.19)$$

To show the value of the function at $t = T$, we introduce the parameter MVC in the formula.

$$\begin{aligned} \Gamma_{cem}(2T) &= \Gamma_{MVC} \cdot \left(\frac{\Gamma_{MVC} \cdot e^{-\frac{n.k}{\Gamma_{MVC}} \int_0^T \Gamma(u, \theta, \dot{\theta}, \ddot{\theta}) du}}{\Gamma_{MVC}} \right)^2 \\ &= \Gamma_{MVC} \cdot \left(\frac{n \cdot \Gamma_{cem}(T)}{\Gamma_{MVC}} \right)^2 \end{aligned} \quad (3.20)$$

Now, we can generalise the formula for $t = mT$, means for m number of cycles by equation 3.21

$$\Gamma_{cem}(mT) = \Gamma_{MVC} \cdot \left(\frac{n \cdot \Gamma_{cem}(T)}{\Gamma_{MVC}} \right)^m, \quad \text{where } m \in N \quad (3.21)$$

$\Gamma_{cem}(mT)$ represent value of muscle strength or capacity after m number of repetitive cycles. As described by R.Ma, the approximate value DMET is used to define the boundary of the DMET [40]. Figure 3.2 illustrates R.Ma's assessment of DMET for fatigue model. In this figure blue repetitive curves shows repetitive cycle, blue dotted line represent external load, and red curve shows the reduction of the value of Γ_{cem} with respect to time. The figure is divided in two zones, i.e, endurance zone and MSD risk zone. The efforts are made in this model to keep the value of Γ_{cem} in safe zone or endurance zone. In endurance zone there are high chances of MSD.

The number of cycles before fatigue limit can be calculated by equation 3.22.

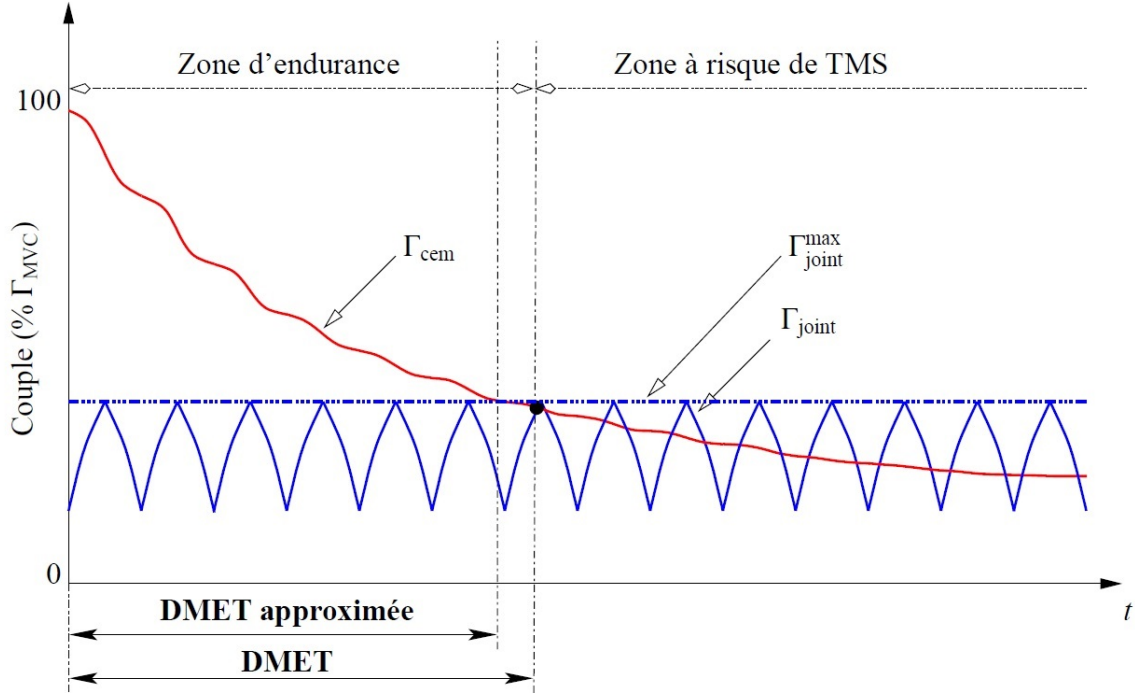


Figure 3.2: The endurance time for dynamic conditions [40]

$$\Gamma_{cem}(mT) = \Gamma_{MVC} \cdot \left(\frac{n \cdot \Gamma_{cem}(T)}{\Gamma_{MVC}} \right)^m = \Gamma_{joint}^{max} \quad (3.22)$$

From R.Ma model Γ_{joint}^{max} is the external load. The value of load will be same for the particular task is direction of motion. For obtaining number of cycles ‘m’, the equation can be modified to equation 3.23:

$$m = \frac{\ln \left(\frac{\Gamma_{joint}^{max}}{\Gamma_{MVC}} \right)}{\ln \left(\frac{n \cdot \Gamma_{cem}(T)}{\Gamma_{MVC}} \right)} \quad (3.23)$$

Hence, approximate value of DMET are as follows:

$$DMET \approx m \cdot T = T \cdot \frac{\ln \left(\frac{\Gamma_{joint}^{max}}{\Gamma_{MVC}} \right)}{\ln \left(\frac{n \cdot \Gamma_{cem}(T)}{\Gamma_{MVC}} \right)} \quad (3.24)$$

The parameters for the equation 3.24 are defined in the table 3.3.

Elements	Unit	Description
T	<i>min</i>	Task period
Γ_{MVC}	N.m	Maximum torque that muscle can exert on joint
Γ_{Joint}^{max}	N.m	Maximum torque from external load
Γ_{cem}	N.m	Current capacity of the muscle to generate the torque
n	unit less	co-contraction factor

Table 3.3: Parameters and variables for DMET calculation equation

3.2.2 Theoretical Validation of Dynamic Model

In the proposed model, there are many parameters. For the theoretical validation of the model we can evaluate the theoretical value of Γ_{cem} by assuming the other parameters from the equation 3.24. In this section we have compared the proposed model with the other models available for MET prediction. The comparison is also done on the basis of the Pearsons correlation ‘r’ and the intraclass correlation ‘ICC’.

The Proposed dynamic muscle fatigue model is based on the reduction of Γ_{cem} during a period of time while doing dynamic motions. To validate dynamic muscle fatigue model with other state of art models we can compare the endurance time evaluation of each model. To compare our dynamic muscle fatigue model with other model we are using Pearsons correlation ‘r’ in equation 3.25 and the intraclass correlation ‘ICC’ in equation 3.26 [48]. Pearson’s correlation ‘r’ gives linear relationship between two models. The closeness between two random variable is represented by ICC. The range of ‘r’ varies between 0 to 1. The value nearest to 1 represents better linear relation. The closeness of ICC value to 1 gives better similarity between the models. In equation 3.26, $MS_{between}$ represents the mean square between different MET or DMET values at different Γ_{MVC} levels while MS_{within} represents the mean square within MET or DMET values at the same Γ_{MVC} level. The number of models to be compared is represented by p, since we are comparing 2 models at the same time so the value of p is 2.

$$r = \frac{\sum_n (A_n - \bar{A})(B_n - \bar{B})}{\sqrt{\sum_n (A_n - \bar{A})^2 \cdot \sum_n (B_n - \bar{B})^2}} \quad (3.25)$$

In equation 3.25, A_n and B_n represent value of endurance time for model number one and two respectively. Similarly, \bar{A} and \bar{B} represent average value of endurance time for model number one and two respectively. In equation 3.26, $MS_{between}$ is polled variance between the subjects and MS_{within} represent trait variance between the subjects.

$$ICC = \frac{MS_{between} - MS_{within}}{MS_{between} + (p - 1)MS_{within}} \quad (3.26)$$

By solving equation 3.24 by considering $\Gamma_{cem}(T) = \Gamma_{Joint}^{max} = \Gamma_{Load}$ with physical

and mechanical parameters of motion using method described by R.Ma [40], DMET can be rewritten as:

$$DMET = -\frac{\ln\left(\frac{\Gamma_{joint}^{max}}{\Gamma_{MVC}}\right)}{n.d.k.\left(\frac{\Gamma_{joint}^{max}}{\Gamma_{MVC}}\right)} \quad (3.27)$$

Also, $f_{MVC} = \frac{\Gamma_{joint}^{max}}{\Gamma_{MVC}} = \frac{\Gamma_{Load}}{\Gamma_{MVC}}$, DMET will be:

$$DMET = -\frac{1}{n.d.k} \cdot \frac{\ln(f_{MVC})}{f_{MVC}}, \quad \text{where } 0 < d \leq 1 \quad (3.28)$$

MET models usually involve the relative strength or the percentage of maximal voluntary contraction, ($F_{MVC} = F/MVC$), in their calculation. In the particular case of the static model or quasi-static [82], the parameter F represents the external load F_{Load} so in this condition, $F_{MVC} = F_{Load}/F_{MVC}$. In our case, this external load is time dependent and replaced by Γ_{joint} . We use the maximum value as a reference. The relative torque or percent of maximum voluntary contraction is then defined as $f_{MVC} = \Gamma_{Load}/\Gamma_{MVC}$. In order to compare our model DMET with existing models of MET, we want to intervene f_{MVC} the parameter in our calculation of DMET by using equation 3.28. Here, d represents dynamic parameter of motions. The value of d varies from 0 to 1.

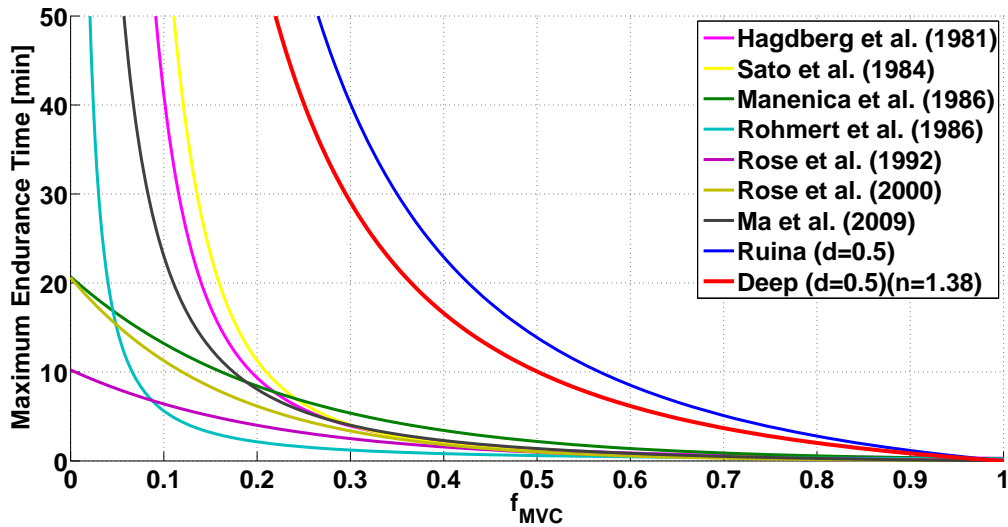


Figure 3.3: Comparison of maximum endurance time of our model with various models

The fatigue parameter k is involved in L.Ma's MET model [82], R.Ma's DMET model [6] and our DMET model which also include co-contraction factor. For comparison between these models we assume $k = 1$. Figure 3.3 represents change in maximum endurance time different with respect to f_{MVC} in model with $d = 0.5$. In reality, when people perform work, static posture is mostly more tiring than the dynamic movement.

Because in a dynamic work, there is better transport of blood in muscles which causes transfer of oxygen to the cells and thereby brings more energy to the muscles. This is one of the reasons why people feel less fatigue in dynamic situations as compare to static situations. In figure 3.3, we see that for any particular value of f_{MVC} , the endurance time is different for each model. For example if we take load as 30% of MVC i.e., $f_{MVC} = 0.3$ then we can see that except R.Ma model and D.seth model, all other models which are static models have very less endurance time in comparison the proposed dynamic model. The DMET for Proposed model is less than the R.Ma model because we have include the effect of co-contraction in the model.

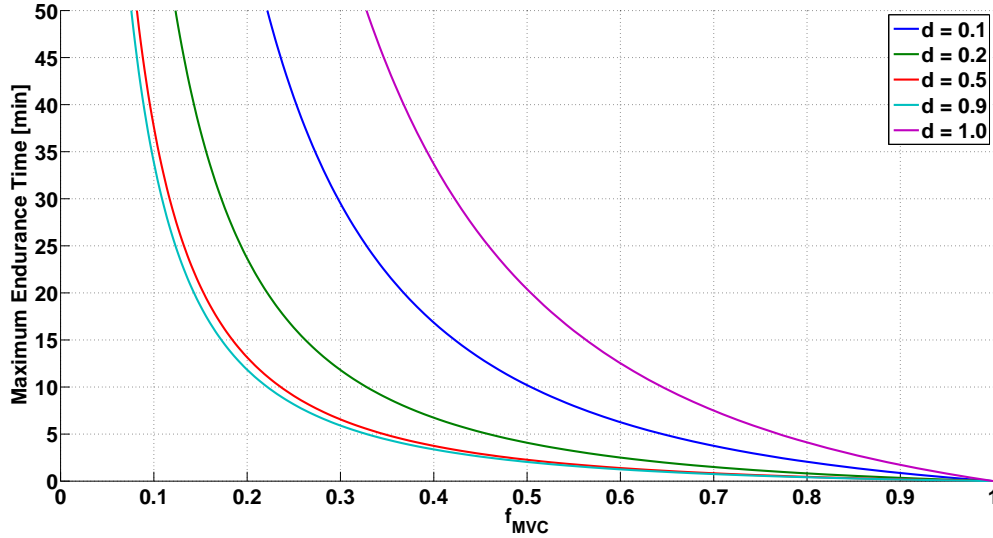


Figure 3.4: Comparison of maximum endurance time for different value of ‘ d ’ for our model

In the model of DMET, we introduce a parameter representing the dynamic factor ‘ d ’. The value varies from 0 to 1. DMET variation according to different values of d are shown in figure 3.4. More we reach towards $d = 0$, more maximum endurance time become more important. It means less number of repetitive cycles or static condition leads to less fatigue and more endurance time. To compare in better way we can compare our models with other models by Pearson’s coefficient and ICC using equation 3.25 and equation 3.26 respectively. In figure 3.4, we see that for any particular value of f_{MVC} , the endurance time is different for the proposed model. For example if we take load as 30% of MVC i.e., $f_{MVC} = 0.3$ then we can see that the model is giving different value of DMET at different value of ‘ d ’. The DMET for Proposed model is less for mode value of d and more for less value of d toward 0.

In this section, we compare our DMET approach with other MET approaches. The parameter ‘ d ’ involved in the DMET model is between 0 and 1 and depends on the magnitude and speed of the movement. To compare our DMET model with MET model proposed by L.Ma and R.Ma, we take $k = 1$ and $n = 1.38$ (from experiment calculation explained in chapter 5) as a reference. We compare our model for various values of d ($d \in 0.1, 0.5, 0.9, 1$) in order to analyze the variation of the parameters ‘ r ’ and ‘ICC’ correlations for different dynamic situations. The variation of endurance time for different value of d are shown in figures 3.5 to 3.8 and corresponding correlation coefficients are shown in tables 3.4 to 3.7. We can observe in figure 3.5, that at $d = 0.1$, all models except R.Ma model are very far from the the red dotted line which represent the proposed model.

It is because other models are static and our model is dynamic, that is why Ruina model which is dynamic is nearer to the proposed model. And at the same time if we observe in figure 3.8, that at $d = 1$, all models came closer to the red dotted line which represent the proposed model. It is because other models are static and our model is dynamic, but at $d = 1$ R.Ma model and the proposed model also act as static model. And in the same figure we can observe that R.Ma model and L.Ma model become same line, it is because at $d = 1$ both models act in same way or become similar. We can observe from figures 3.5 to 3.8 that from $d = 0.1$ to $d = 1$ the model are getting closer when we move from 0 to 1 for the value of d , i.e., when we mode from dynamic to the static condition.

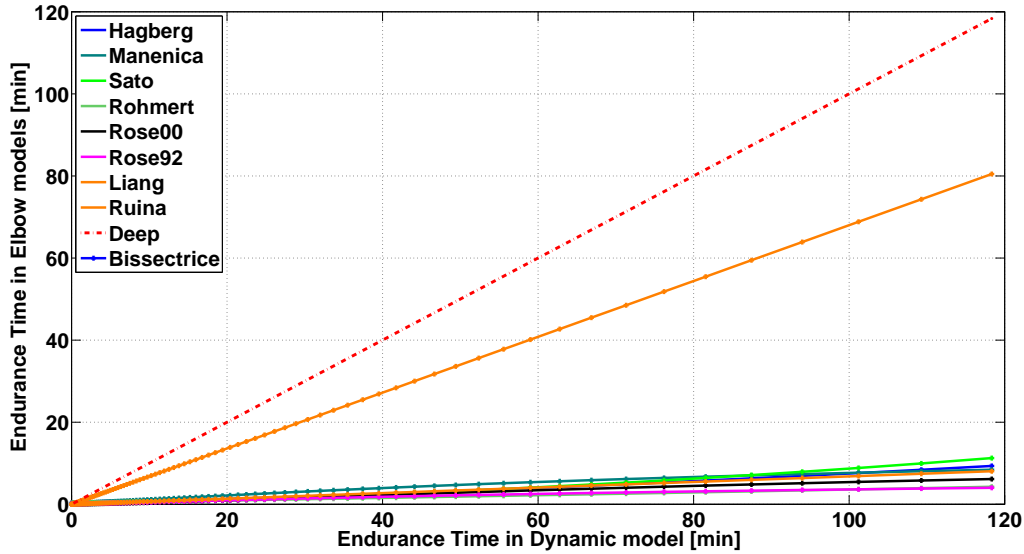


Figure 3.5: Evolution of ICC coefficient for different value of f_{MVC} , $d = 0.1$

Model	Model's Equation	r	ICC
Hagberg (1981) [76, 94]	$MET = 0.298(f_{MVC})^{-2.14}$	0.9930	-0.1277
Sato (1984) [95]	$MET = 0.195(f_{MVC})^{-2.52}$	0.9827	-0.1104
Manenica (1986) [96]	$MET = 20.6972\exp(-4.5f_{MVC})$	0.9915	-0.1065
Rohmert (1986) [97]	$MET = 0.2285(f_{MVC})^{-1.391}$	0.9996	-0.1970
Rose (1992) [98]	$MET = 10.23\exp(-4.69f_{MVC})$	0.9985	-0.1911
Rose (2000) [79]	$MET = 20.6\exp(-6.04f_{MVC})$	0.9931	-0.1589
L.Ma (2009) [82]	$MET = -\ln(f_{MVC})/k \cdot f_{MVC}$ (K=1)	1	-0.1331
R.Ma (2012) [6]	$MET = -\ln(\Gamma_{MVC})/d \cdot k \cdot \Gamma_{MVC}$ (K=1)	1	0.8828

Table 3.4: Determination of correlation coefficients 'r' and 'ICC' with $d=0.1$.

Discussion and Conclusion

If we compare correlation coefficient in tables 3.4, 3.5, 3.6 and 3.7 and figures 3.5, 3.6, 3.7 and 3.8, it is clear that endurance time calculation by our model and other models have high degree of linear relationship. Indeed, all the coefficients of Pearson (r) exceed 0.9. At the same time, we can notice that the value of coefficient r is independent of the parameter. Indeed, the calculation of r gives same value regardless of the value of d . We

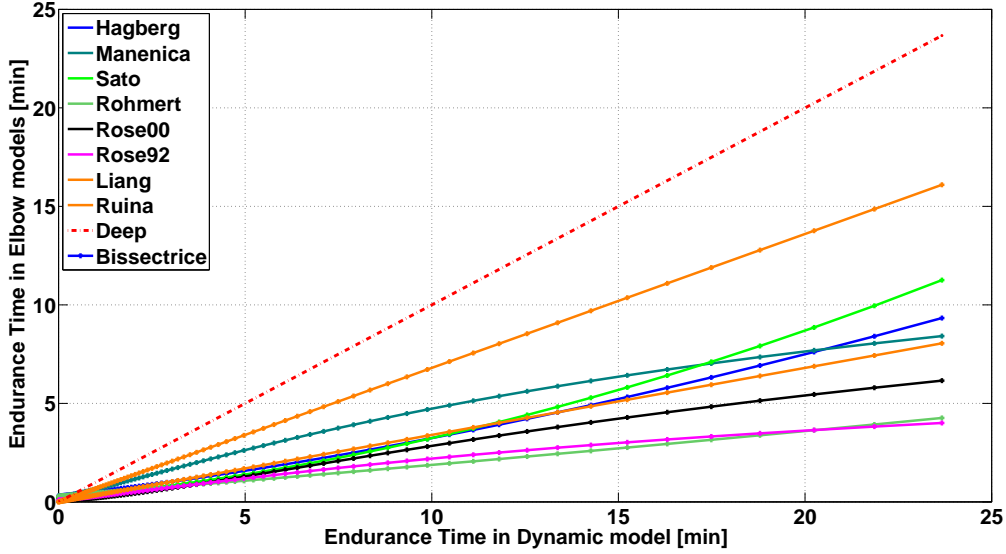


Figure 3.6: Evolution of ICC coefficient for different value of f_{MVC} , $d = 0.5$

Model	Model's Equation	r	ICC
Hagberg (1981) [76, 94]	$MET = 0.298(f_{MVC})^{-2.14}$	0.9930	0.4382
Sato (1984) [95]	$MET = 0.195(f_{MVC})^{-2.52}$	0.9827	0.5073
Manenica (1986) [96]	$MET = 20.6972\exp(-4.5f_{MVC})$	0.9915	0.5403
Rohmert (1986) [97]	$MET = 0.2285(f_{MVC})^{-1.391}$	0.9996	0.0887
Rose (1992) [98]	$MET = 10.23\exp(-4.69f_{MVC})$	0.9985	0.1192
Rose (2000) [79]	$MET = 20.6\exp(-6.04f_{MVC})$	0.9931	0.2838
L.Ma (2009) [82]	$MET = -\ln(f_{MVC})/k \cdot f_{MVC}$ (K=1)	1	0.4121
R.Ma (2012) [6]	$MET = -\ln(\Gamma_{MVC})/d \cdot k \cdot \Gamma_{MVC}$ (K=1)	1	0.8828

Table 3.5: Determination of correlation coefficients 'r' and 'ICC' with $d=0.5$.

can conclude that the linear correlation is unaffected by torque variations but only by its maximum value.

The analysis of the tables 3.4, 3.5, 3.6 and 3.7 shows that the value of d will affect the ICC coefficient. More the d value tends to 1, the more the ICC coefficient tends to move toward 1 and hence at the same time proposed endurance time model is more similar to other models. We can also notice that when $d = 1$, the values of r and ICC coefficients between our model and the one proposed by L.Ma and R.Ma are equal. This signifies that in this situation, the two models are similar.

3.2.3 Position profile for dynamic situation

In the dynamic situation for position, we have defined the motion in repetitive cycles with intervals of 3 seconds. The path is defined with polynomial function, 'P' of third

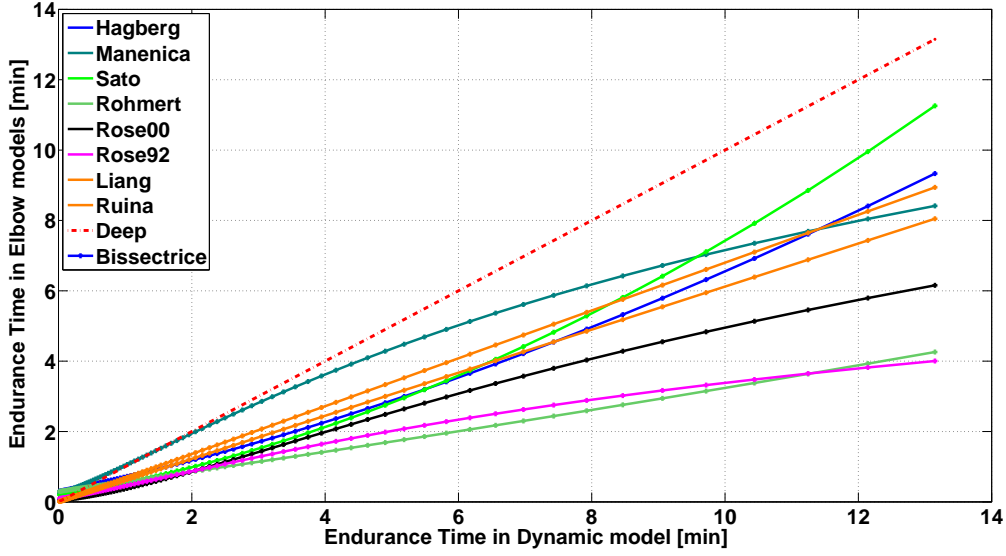


Figure 3.7: Evolution of ICC coefficient for different value of f_{MVC} , $d = 0.9$

Model	Model's Equation	r	ICC
Hagberg (1981) [76, 94]	$MET = 0.298(f_{MVC})^{-2.14}$	0.9930	-0.8430
Sato (1984) [95]	$MET = 0.195(f_{MVC})^{-2.52}$	0.9827	0.8884
Manenica (1986) [96]	$MET = 20.6972\exp(-4.5f_{MVC})$	0.9915	0.9236
Rohmert (1986) [97]	$MET = 0.2285(f_{MVC})^{-1.391}$	0.9996	0.3782
Rose (1992) [98]	$MET = 10.23\exp(-4.69f_{MVC})$	0.9985	0.4269
Rose (2000) [79]	$MET = 20.6\exp(-6.04f_{MVC})$	0.9931	0.6675
L.Ma (2009) [82]	$MET = -\ln(f_{MVC})/k \cdot f_{MVC}$ (K=1)	1	0.8191
R.Ma (2012) [6]	$MET = -\ln(\Gamma_{MVC})/d \cdot k \cdot \Gamma_{MVC}$ (K=1)	1	0.8828

Table 3.6: Determination of correlation coefficients 'r' and 'ICC' with $d=0.9$.

order [40], see equation 3.29. This equation is just showing the position profile.

$$P = a_0 + a_1t + a_2t^2 + a_3t^3 \quad (3.29)$$

where a_1 , a_2 and a_3 are the coefficient define according to the following conditions:

$$\theta(0) = \theta^{initial}, \dot{\theta}(0) = 0, \quad \theta(t_f) = \theta^{end}, \dot{\theta}(t_f) = 0 \quad (3.30)$$

$\theta(0)$ is initial angle and $\theta(f)$ is final angle of motion. The position can be determined by equation 3.31. $\theta(t)$ is angle at time period t.

$$\theta(t) = \theta^{initial} + r(t) \cdot (\theta^{end} - \theta^{initial}) = 0, 0 \leq t \leq t_f \quad (3.31)$$

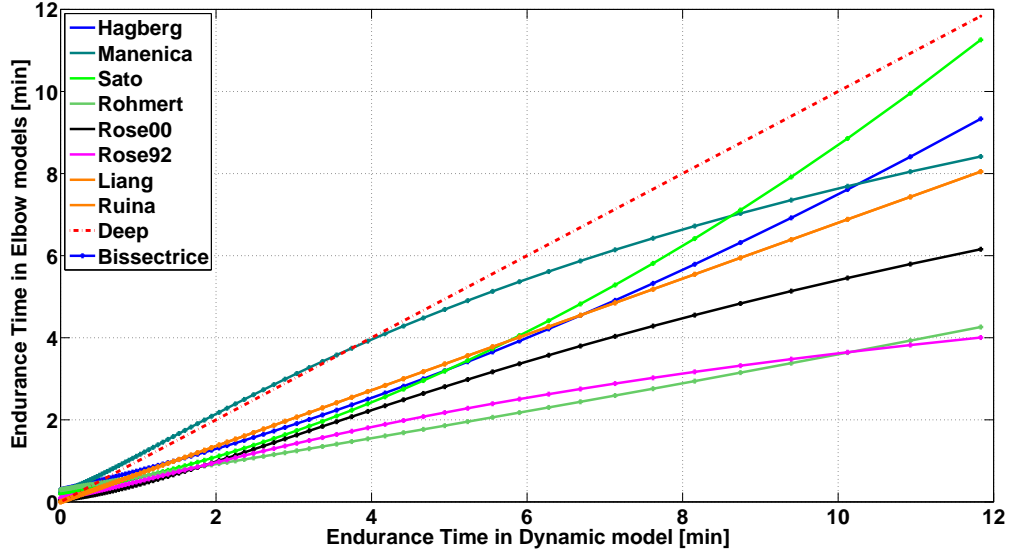


Figure 3.8: Evolution of ICC coefficient for different value of f_{MVC} , $d = 1$

Model	Model's Equation	r	ICC
Hagberg (1981) [76, 94]	$MET = 0.298(f_{MVC})^{-2.14}$	0.9930	0.9020
Sato (1984) [95]	$MET = 0.195(f_{MVC})^{-2.52}$	0.9827	0.9336
Manenica (1986) [96]	$MET = 20.6972\exp(-4.5f_{MVC})$	0.9915	0.9612
Rohmert (1986) [97]	$MET = 0.2285(f_{MVC})^{-1.391}$	0.9996	0.4464
Rose (1992) [98]	$MET = 10.23\exp(-4.69f_{MVC})$	0.9985	0.4977
Rose (2000) [79]	$MET = 20.6\exp(-6.04f_{MVC})$	0.9931	0.7419
L.Ma (2009) [82]	$MET = -\ln(f_{MVC})/k \cdot f_{MVC}$ (K=1)	1	0.8828
R.Ma (2012) [6]	$MET = -\ln(\Gamma_{MVC})/d \cdot k \cdot \Gamma_{MVC}$ (K=1)	1	0.8828

Table 3.7: Determination of correlation coefficients 'r' and 'ICC' with $d=1$.

where, $r(t)$ can be interpolate as follows:

$$r(t) = 3\left(\frac{t}{t_f}\right)^2 - 2\left(\frac{t}{t_f}\right)^3 \quad (3.32)$$

Here, t is current time and t_f is final time period.

3.3 Recovery Model

3.3.1 Introduction

Definition of recovery: "Increase of the functional capacity of an organ or organism, of which the functional capacity was reduced as a result of fatigue; recovery occurs by

ending, reducing or changing the action which results in reduction of the functional capacity of an organ or of an organism” [99]. Insufficient amount of recovery of muscles may lead to increased risk of MSD [36].

Muscle fatigue can result from over training, exercising your muscles beyond their current endurance level or improper nutrition. After exercise, muscles might ache, and person might find it hard to do anything beside crash in a chair from the fatigue he feel. Recovery is an important parameter to avoid MSD after muscle fatigue prevails.

The fatigue rate and recovery rate are often use to understand fatigue level and to depict the process of fatigue and recovery, respectively [99]. The fatigue rate represents the rate with which the fatigue occur while the recovery rate represents the rate of regaining the muscle strength after fatigue.

For physical fatigue, fatigue rate can be measures using physical factors such as heart rate [100], force [101] or EMG data [102]. According to physiological factors many empirical models have been developed. Recovery can also be measured as force strength [11, 103–105]. In some cases endurance time is used as recovery factor [106]. Various physical factors [99, 107] as well as power spectral analysis of surface EMG [108], were used as parameters for modeling recovery processes. All parameters normally suggest that recovery is totally depend on time. However, speed of recovery for different parameters it might be different. Study suggest that muscle strength regain in much faster rate in comparison to muscle endurance [108] and recovery of the power spectrum of EMG is better then both muscle endurance and muscle strength. Different Physiological parameters are described in Winkel, 1996 [109]. There are many discrepancies in different parameters. But individual differences are not considered appropriately to design a physical task.

3.3.2 Recovery Model

We have proposed a muscle fatigue model in form of negative exponential function. The model includes individual recovery rate to model recovery strength. The recovery model is derived from the L.Ma’s work [48, 110]. The exponential recovery variation is shown by L.Ma’s model in figure 3.9. The figure shows the variation of recovery curves with respect to the different recovery rates. The model is described by an equation 3.33. Recovery is measured in the form of force $F_{cem}(t)$ or joint torque strength $\Gamma_{cem}(t)$. $F_{cem}(t)$ or $\Gamma_{cem}(t)$ represents the current exorable maximum force at a given time. Fatigue and recovery shows the change in strength. $F_{max}(t)$ or $\Gamma_{max}(t)$ represents maximum voluntary contraction for given posture without fatigue. Recovery rate ‘R’ represents the recovery speed of a muscle group after fatigue. It is assumed that recovery rate remains stable for a certain period according to physical conditions. The differential equation shows that the recovered strength is proportional to the fatigue ($\Gamma_{max} - \Gamma_{cem}(t)$).

$$\frac{d\Gamma_{cem}(t)}{dt} = R(\Gamma_{max} - \Gamma_{cem}(t)) \quad (3.33)$$

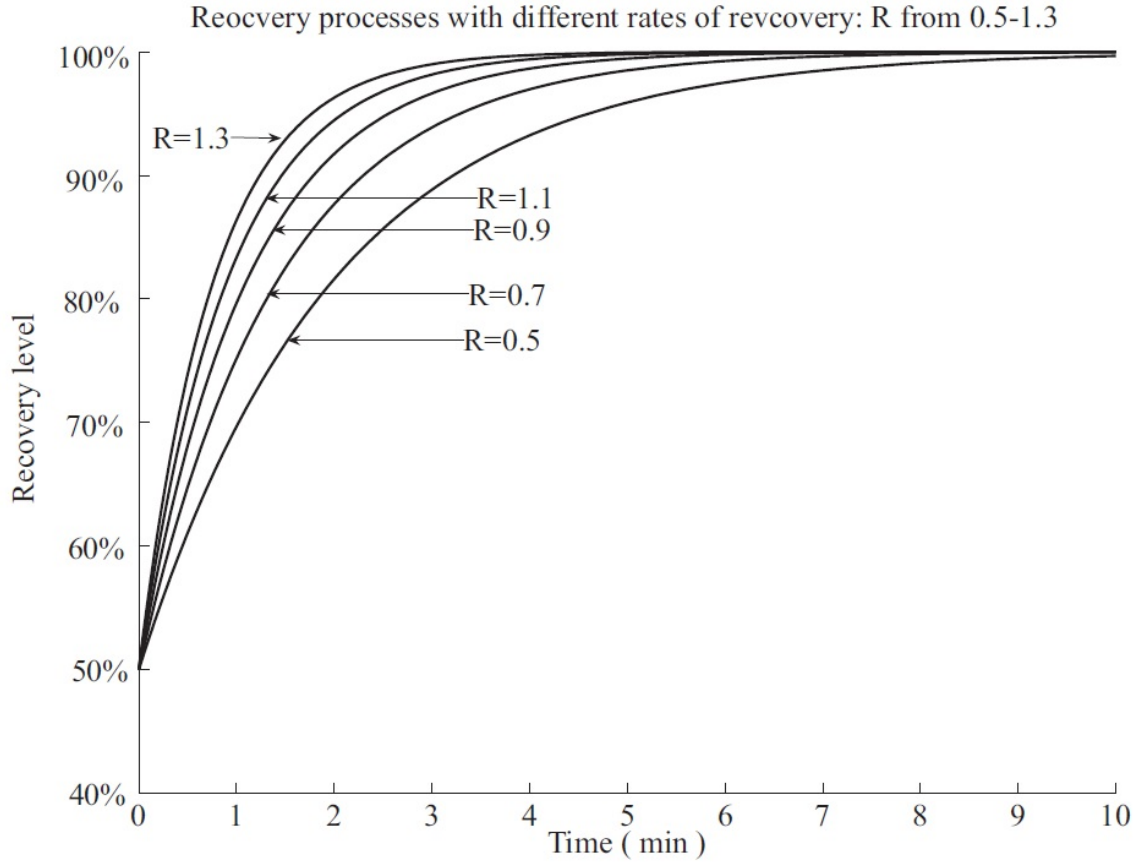


Figure 3.9: Recovery curves of a joint under different rates of recovery

The fatigue level can be expressed as a percentage ($\frac{\Gamma_{cem}(0)}{\Gamma_{max}} \times 100\%$), where $\Gamma_{cem}(0)$ is the initial strength at beginning of the recovery and Γ_{max} is maximum strength. Similarly for the time instant t , fatigue level can be the ratio of recovery strength till time t to the Γ_{max} , expressed as ($\frac{\Gamma_{cem}(t)}{\Gamma_{max}} \times 100\%$). The recovery strength after time duration t can be represented by the integral of equation 3.33 and represented by equation 3.34.

$$\Gamma_{cem}(t) = \Gamma_{max} + (\Gamma_{cem}(0) - \Gamma_{max})e^{-Rt} = \Gamma_{cem}(0) + (\Gamma_{max} - \Gamma_{cem}(0))(1 - e^{-Rt}) \quad (3.34)$$

According to the muscle physiology, there are two type of motor unit to control muscle: Type I and Type II. These motors units are responsible for muscle contraction and force generation. Type I is more fatigue resistant in comparison to Type II. These motor units do not work until get recovered from fatigue. The fatigue level of motor unit is represented by $(\Gamma_{max} - \Gamma_{cem}(0))$. Total fatigue is composition of fatigue from both type of motor units. But it is difficult to determine the exact proportion of fatigue from each type, we are considering constant fatigue from combining each unit and and one recovery parameter 'R' is used to represent the overall fatigue.

3.3.3 State of the art recovery models

The recovery model is explained above. In this section we explained other state of the art models to observe their limitation and application.

Elfving's EMG Model

In this model recovery of the median frequency of the EMG power spectrum has been studied on the healthy 55 subjects, and reference data is calculated by equation 3.35 [108]. The exponential dependence was obtained by root means square of determination, $r^2 = 0.98$.

$$f = f_e + (f_i - f_e) \left(1 - \exp\left(-\frac{1}{\tau}\right) \right) \quad (3.35)$$

where, τ (minutes) is the relaxation time constant, f_i and f_e are frequency at the start and end of fatigue contraction respectively.

The experimental results shows very good correlation of mean recovery data with the exponential model. The individual recovery analysis can be possible with regression. In this model, further experimental data for strength recovery for shoulder and torso extension also demonstrate exponential dependence of the recovery.

Liu's Motor Unit Model

Liu's model provide a general approach to analyze fatigue with motor unit pattern [2]. If we assume that during recovery period signals from central nervous system are zero hence no motor unit activity then recovery can be predicted by equation 3.36 based on Liu's model. In this equation M_F represents total number of fatigued unit and can be replaced by $(F_{max} - F_{cem})$, as this also represents fatigued motor unit in muscles and can be written as equation 3.37

$$\frac{dM_F}{dt} = -RM_F \quad (3.36)$$

$$\frac{d(F_{max} - F_{cem})}{dt} = -R(F_{max} - F_{cem}) \quad (3.37)$$

After simplifying equation 3.37 can be written as equation 3.38, considering F_{max}

as constant.

$$\frac{dF_{cem}}{dt} = R(F_{max} - F_{cem}) \quad (3.38)$$

This model was validated through experiments for maximum gripping strength. R ranges from 0.0042 to 0.0125 s^{-1} after fitting result data. In this experiment, recovery process was occurred simultaneously with fatigue process. The variation in recovery rate represents differences of physical parameters between different subjects.

conclusion

These models used different physical parameters to validate and analyze recovery. Wood's work cycle model [103] use fatigue model to predict fatigue strength and later use same model to identify recovery. This model used to predict amount of grip strength recovered for grip jobs. It shows during recovery the worker do not use grip force. Although different physical parameters were taken in different model but all of them have common exponential function to predict recovery. This exponential behavior of recovery can also be seen in the new model. The recovery model does not depend on any kind of motion hence it can be applicable to either static or dynamic situations

3.4 Summary

From all the description in this chapter now we have a new dynamic muscle fatigue model which includes co-contraction into consideration. This model can contribute to study the muscle fatigue in the dynamic situation for different kind of simple postures and motions at different joints. For the application of this model to complex motion and at different joints for simple motion there may be requirement of some modification in the model according to the situation. Similarly for the DMET model the modification may required according to the type of motion and joint. The co-contraction in this model is evaluated for two group of muscles activation, but if more that two group of muscle are involved like in complex motion then we have to modify the method to evaluate the co-contraction index. But when we move to the recovery part we can see that this model is applicable for both dynamic and static situation as for recovery we don't need any kind of motion, it is a rest process. The variation in recovery model may be needed according to the situation like complex real life. The validity of model is not yet verified for different kind of motions ot task. In this thesis we will validate this model after push-pill activity for the arm.

A new dynamic muscle fatigue model with a co-contraction factor has been introduced. The dynamic maximum endurance time model has also been introduced to measure DMET for each subject in dynamic conditions. The theoretical validation of dynamic muscle fatigue model on the basis of DMET model has also been done and

compared with other models. A recovery model is also introduced and compared with other existing models. The experimental validation and analysis of these models will be in upcoming chapters.

Chapter 4

Methodology: Experiment and Description

4.1 Introduction

4.1.1 Description

The muscle fatigue model described in chapter 3 needs to be validated experimentally. Before going for experimental validation we have to define the different dynamic postures and experiment protocol for the study of muscle fatigue and recovery. Electromyography (EMG) technique has been used for the analysis of muscle activity. The analysis of the muscle activity during the motion helps us in understanding the behavior of muscles for different types of motions. The assumptions, methodology and analysis of the EMG data as well as the findings which support the co-contraction factor and the description of the experiments performed for push-pull motion of the human arm are also described in this chapter. The results and findings will be discussed in detail in the next chapter.

4.1.2 Push-Pull operation and muscles activity

The push-pull motion of the arm is the extension and flexion motion about the elbow respectively as shown in figure 4.1. The plane of the motion is vertical plane (sagittal plane). The Push pull activity with the muscle activation is shown in figure 4.2. In our model, the effect of co-contraction and delay (rest) between each cycles were introduced to analyze the activities in a realistic way. Here the push motion is considered as the motion of the arm away from the body while pull motion is the motion of the arm towards the body.

According to the hypothesis made in the previous chapter, when the biceps muscles

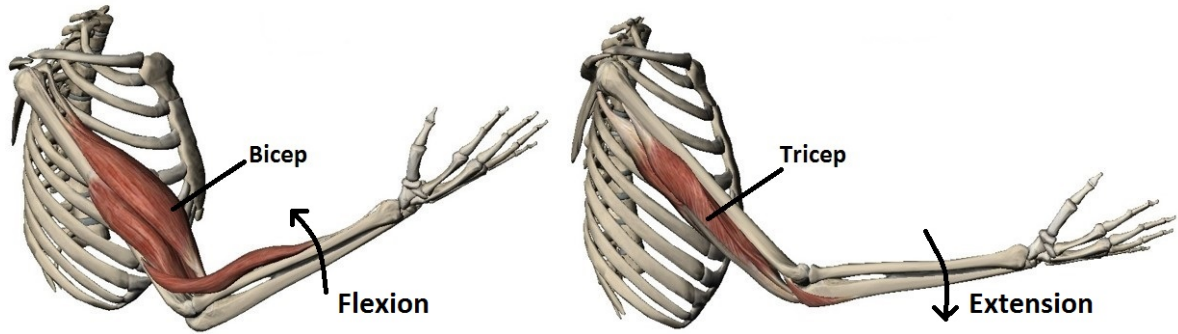


Figure 4.1: Arm movement - flexion and extension

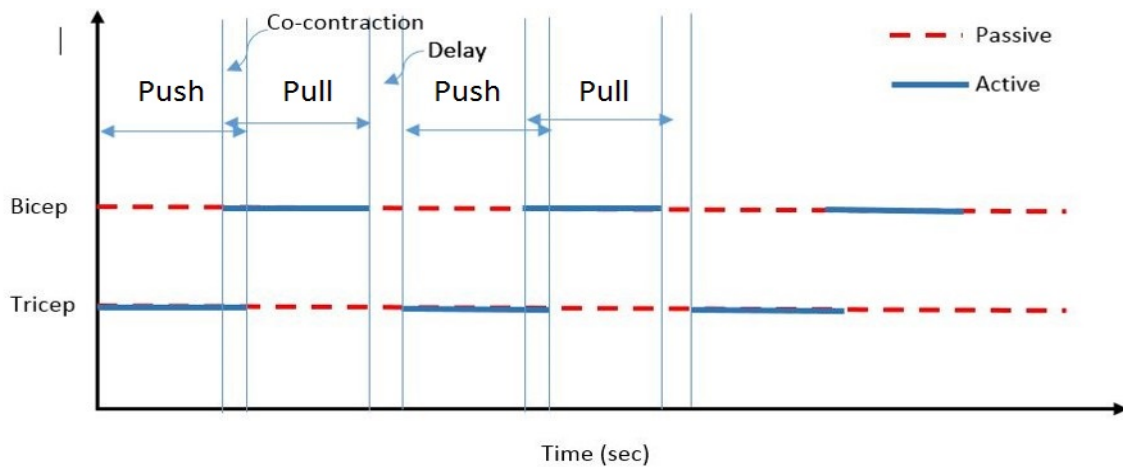


Figure 4.2: Push-Pull Motion and Muscles Activity

are active triceps should be passive and vice versa. But in between the activities of the biceps and the triceps there will be some co-contraction which will affect the accuracy [44] and reach-ability of the motion. The co-contraction is just after the consecutive activity of biceps and triceps i.e., flexion. But it is not prominent after the activity of triceps and biceps (extension) because there is a long delay between these activities after completion of one cycle as shown in figure 4.2. There are some activity during delay period from both the muscles but we are neglecting those activities.

4.2 Experiment Set-Up

4.2.1 Main objectives

The aim of this experiment is to study the muscle fatigue and recovery parameters for arm flexion and extension in the vertical plane. The analysis of the activities of the biceps, triceps and trapezius muscle were done using EMG data. The co-contraction between the bicep and tricep muscles is included in the model as co-contraction factor.

The main part of this experiment is to validate the dynamic muscle fatigue model with the MVC data and newly introduced co-contraction factor.

4.2.2 Experiment protocol

The protocol is divided into the following ten steps:

1. Warm up exercise for 2 minutes to relax the muscles.
2. Initial MVC measurement for no fatigued condition. Always during the experiments, MVC measurement just take 2-3 second, in this duration subject have to apply full effort.
3. A biodex (REF) system is used to perform flexion and extension in isotonic mode in vertical plane (figure 4.3).
4. The motion range is about seventy degrees and it is from -20° to 50° as shown in figure. 4.3.

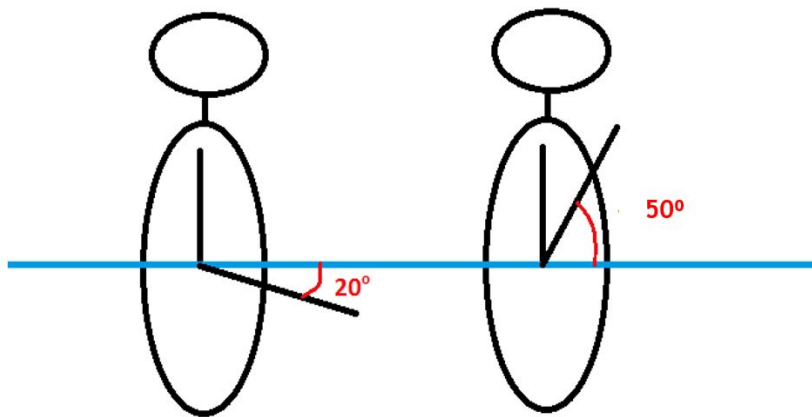


Figure 4.3: Arm movement range while flexion and extension in vertical plane

5. The subject has to complete 20 cycles in one minute of the fatigue protocol.
6. Each cycle (flexion + extension) should be completed within 3 seconds.
7. External load was 20% of initial MVC. MVC was calculated between each fatigue protocol.
8. MVC Was measured at 0° arm position in sagittal plane during the experiments. At the time of MVC measurement the hand will be fixed at 0° position for isometric condition to measure the MVC.
9. The test protocol repetition continues till exhaustion. MVC is measured for push and pull phase separately.
10. After exhaustion MVC will be measured at duration of 1 minute, 5 minutes and 15 minutes respectively from the end of protocol. This MVC will be used to analyze the recovery after fatigue in 15 minutes.

The Experiment protocol can easily be understood by the flowchart shown in the figure 4.4.

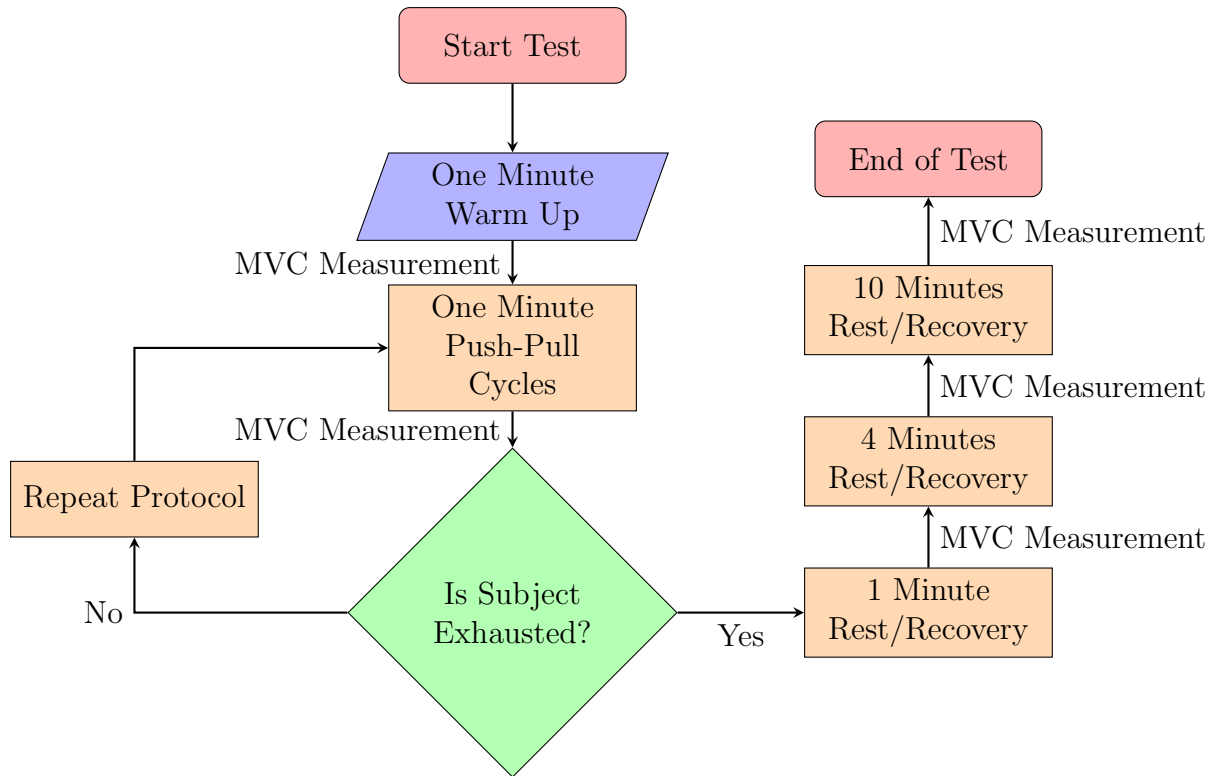


Figure 4.4: Flow chart for experiment protocol

4.2.3 Subject's description

Ten subjects, aged 24 ± 4 years old, performed the protocol. All the subjects were sportive. The subjects were physically fit and had no injuries in the upper limb. The subjects characteristics for age, weight, height, upper arm and fore arm length are given in table 4.1, as well as their sport activities.

Subject	Age	Weight	Height	Upper arm	Forearm	Sports
1	28	89 kg	185 cm	29 cm	26.5 cm	Running
2	24	80.2 kg	183.5 cm	31.5 cm	28 cm	Musculation
3	20	69.8 kg	180.1 cm	30 cm	29.5 cm	Handball
4	20	80.9 kg	177 cm	29.8 cm	29 cm	Handball
5	21	62.2 kg	172.8 cm	29.2 cm	26.5 cm	Tennis
6	25	61.1 kg	164.8 cm	26 cm	24.5 cm	Rugby
7	26	74 kg	176 cm	28.5 cm	27 cm	Tennis
8	27	66 kg	181 cm	29.5 cm	26.5 cm	wall climb
9	23	66.3 kg	164 cm	27 cm	25.5 cm	Swimming
10	26	85 kg	184 cm	29 cm	26.5 cm	Football

Table 4.1: Subject's anthropometric data and description

4.2.4 Equipment required for experiment and data acquisition

To prepare an experiment setup, we need various equipments. The experiments are performed on the a Biodex system 3 research (Biodex medical,shirley, NY) isokinetic

dynamo-meter. The biodex system and dynamometer are shown in figures 4.5 and 4.6 respectively.



Figure 4.6: Experiment dynamometer

For the acquisition of EMG activity from the bicep, tricep and trapezius muscles, the surface electrodes are fixed on the skin at middle and parallel to the muscles (figure 4.7). The position of the electrodes on the arm are shown in figure 4.7. Prior to electrodes

placement, the skin was shaved and cleaned with alcohol in order to reduce electrode-skin impedance below $55\text{ k}\Omega$. Then, EMG signals were preamplified (gain = 600, bandwidth 6–400 Hz) and sampled at 2000 Hz.

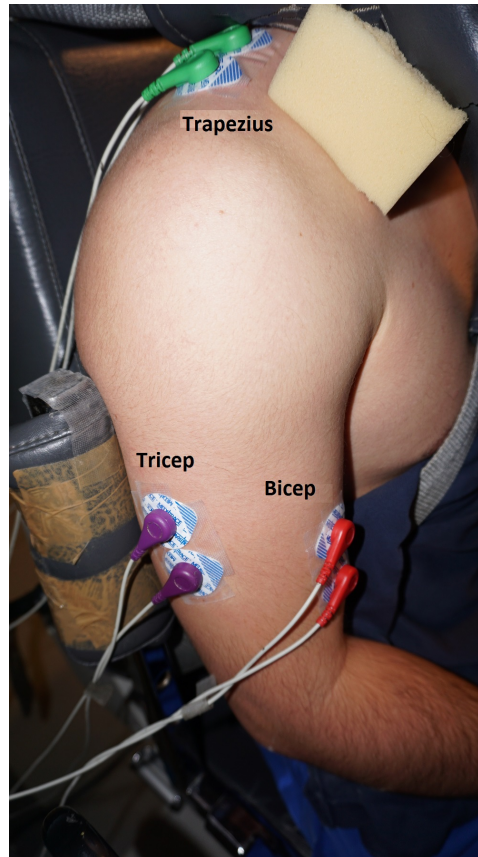


Figure 4.7: Position of the EMG electrodes on the arm

To record all the data, mechanical or EMG data, PowerLab data acquisition system is used with Labchart software to visualize and convert the data from raw to digital readable format. After data acquisition the raw data is drafted into a text file for further process on Matlab.

Every time before starting a new experiment, setup has to be arranged according to protocol and a calibration of dynamo-meter is necessary. The inputs to Powerlab data acquisition system are EMG data from bicep, tricep and trapezius muscles and velocity, position and torque generated by subject.

Procedure

The subjects should not move body during the test. To avoid this we can fix the subjects with soft belts to the experiment chair as shown in figure 4.8. The exercise were performed using the right arm. The experiment setup with a subject is shown in figure 4.9. The maximum MVC for the unfatigued subject were calculated before starting the test. This MVC value will be a reference for all the further analysis for the same subject. To set the torque on the handle, the value of torque will be 20% of initial maximum MVC, which means when subjects will apply torque more than or equal to the



Figure 4.8: Experiment setup without any subject

fixed value then only they will be able to move the handle. Here subject have to perform repetitive 20 cycles of push and pull at fixed load which is 20 % of maximum capacity of subject. Hence subject can do repetitions easily until he reached the level of fatigue or endurance time. If subject fail to apply this effort then it may possible that subject will not complete the cycle or return it from middle. But the experiment was in proper vision and subjects were always motivated to complete all the cycles. Each MVC measurement required just 2-3 seconds for both flexion and extension phases. After measuring the MVC value the subjects take rest for ten seconds (which is also the time to prepare the device again for next protocol) before starting the fatigue protocol. In this experiment, subjects should complete 20 repetitions or cycles in one minute as mentioned in the experiment protocol. After each one minute protocol MVC have to be measured again. If the subject is exhausted or not able to do further exercise after certain repetitions of fatigue protocol, then the test should be stopped. After ending the fatigue test MVC was measured. During the fatigue protocol the subject should always be motivated to go up to exhaustion but until comfortable limit so that the subject can not hurt muscles if he go beyond limit.

The EMG is measure to three muscle group for biceps, triceps and trapezius. For flexion and Eextension motion of the arm biceps and triceps muscle are mainly responsible. But at the time of some motion or fatigue it is possible that other muscles come into effect to support the motion that is why we have also included trapezius muscle as test to see the activity of third muscle during flexion and extension.

4.3 Data processing and analysis

All the data were processed using standardized Matlab program. The processing includes filtering the raw data, normalization, cycles separation, and frequency analysis. The step by step procedure for all this are as follows:



Figure 4.9: Experiment setup with one subject

4.3.1 Filtering the raw EMG data

1. The data processing was performed using standardized Matlab scripts (User Guide - The Mathworks, Natick, USA). A Butterworth second order low pass (10 Hz) was applied to the raw mechanical signals (torque, position and velocity) from flexion and extension movements of the arm (Appendix .1 and .2). The raw data contains some noise at low frequency. To remove these noise we filter the signal at low frequency.
2. Remove DC offset from the signal by detrend the raw EMG data and mechanical data (position, velocity and torque). Normally, we want the signal to have a middle point at zero to allow a maximum dynamic range. Offset refers to adjust data to zero point. While DC offset can be useful in the control of some kinds of synthesis (such as Amplitude Modulation or Frequency Modulation). Detrending means, removing offsets, or linear trends from regularly sampled time-domain input-output data signals. This data processing operation helps us estimate more accurate linear models because linear models cannot capture arbitrary differences between the input and output signal levels (Matlab code for this process is explained in appendix .2). Detrend computes the least-squares fit of a straight line (or composite line for piecewise linear trends) to the data and subtracts the resulting function from the data.
3. Rectify the EMG signal. This will help in easy filtering of the raw data (Appendix .2).
4. Apply Butterworth band pass filter of 20-400 Hz to the rectified raw EMG data (Appendix .3).
5. Normalize the filtered data for all the EMG activity with respect to the normalization value of each signal calculated equation 4.1. Normalization means dividing the

whole data for each index with its reference value or normalization value.

The output data after filtering are shown in the figure 4.10 is normalized based on the value described in equation 4.1 . Bicep, Tricep, Trapezius, and Torque are normalised, position has units in degrees in which range is varying between -20 to 50 degrees and velocity has units in degrees/sec.

All the cycles are normalized according to the equation:

$$value_{Normalization} = value_{mean}^{max} + 3\sigma \quad (4.1)$$

- $value_{Normalization}$: Reference value for data normalization.
- $value_{mean}^{max}$: Maximum value of standard deviation along the mean.
- σ : Standard deviation

For each case there are some cycles which reached abnormal maximum and minimum, so to avoid those cycles from comparison we select the normalization value calculated on the basis of the equation 4.1.

4.3.2 Extraction of the relevant data

Different number of cycles(push+pull) were completed by the ten subjects in the fatigue test. Time scale normalization is needed for the comparison of the repetitions of cycles, like in highly standardized movement patterns or repetition cycles, such as normal gait or iso-kinetics knee. Not being robots, it is difficult for normal subjects to really reproduce a movement a second time. As an example, figure 4.10 illustrates a sample of filtered data for 5 push-pull cycles. The three first lines represents EMG signals for biceps, triceps and trapezius respectively, line 4 to 6 are joint torque, position and velocity respectively. We can observe on this figure that the delay between two biceps EMG activity. We can also observe that some period show no EMG signals at all.

The filtered EMG data for all the muscles without removing any delay for the five cycles is shown in figure. 4.10. In this figure we can see that there is a delay between each activity, means after completing each cycle subject stop for some duration and then again start the cycle.

In case of isometric contraction there is no movement at that time we compare cycles on the basis of EMG. But in dynamic conditions there will be movement of the body part so cycles can be compared on the basis of actual mechanical data.

We are comparing the cycles with having some activity that is why delay or rest in which there is no activity is not taken in for the study. But this will not have significant

effect on the result. It is done just to analyze the required part of the data.

Here we break the cycle in two parts; (i) Push phase and (ii) Pull phase separately, so that we can compare the activity of all the three muscles in the both phases in figure 4.10, the muscle activity is characterized by the biceps EMG activity, the pull is characterized by the triceps EMG, we can see an absence of EMG activity between one pull phase and the following push phase (figure 4.10), denoted by area 'b'. Another resting period can be observed on the joint torque, position and velocity data denoted by 'a' in figure 4.10: all these data are zero in this phase, even though EMG signals show activity. For doing this we have two options to separate the push and pull phases: one is according to EMG activity of biceps and tricep (zero EMG signal = no muscle activity) and another one is according to the change in mechanical data of joint position or velocity (zero joint torque, velocity and position = no movement). The assumption made for separating the cycles are as follows:

- Separation of push-pull cycles in two parts.
- Neglecting the delay or rest in between each cycle (no motion and no activity).
- For cycles separation and comparison on the basis of EMG, all the push cycles will start at onset of EMG signal and end at offset of EMG signal of Biceps. Similarly, all the pull cycles will start at onset of EMG signal and end at offset of EMG signal of Triceps.
- For cycles separation on the basis of joint velocity, the positive part of the velocity is considered as push phase and the negative part is considered as pull phase.

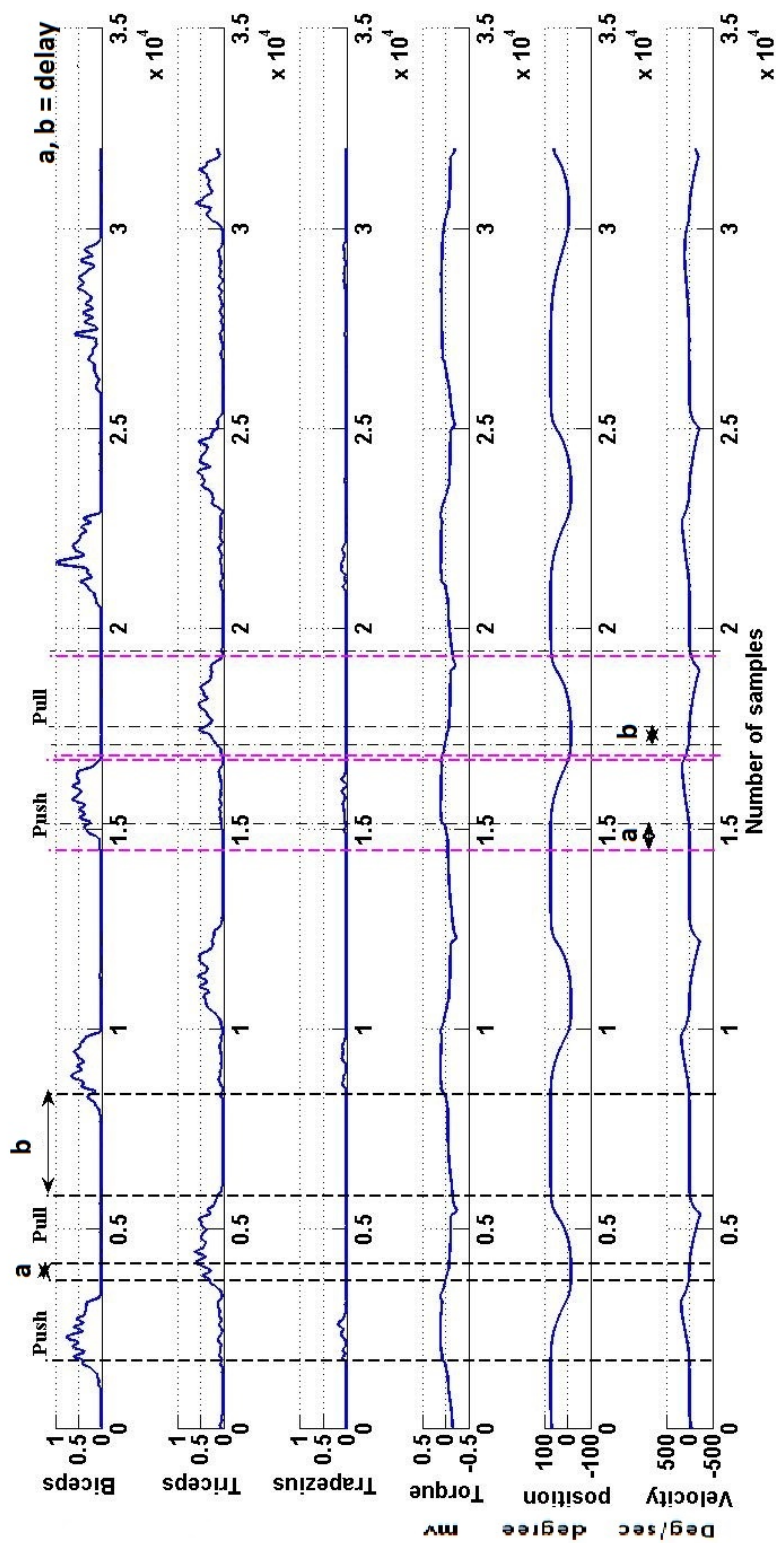


Figure 4.10: Experimental filtered data for five cycles

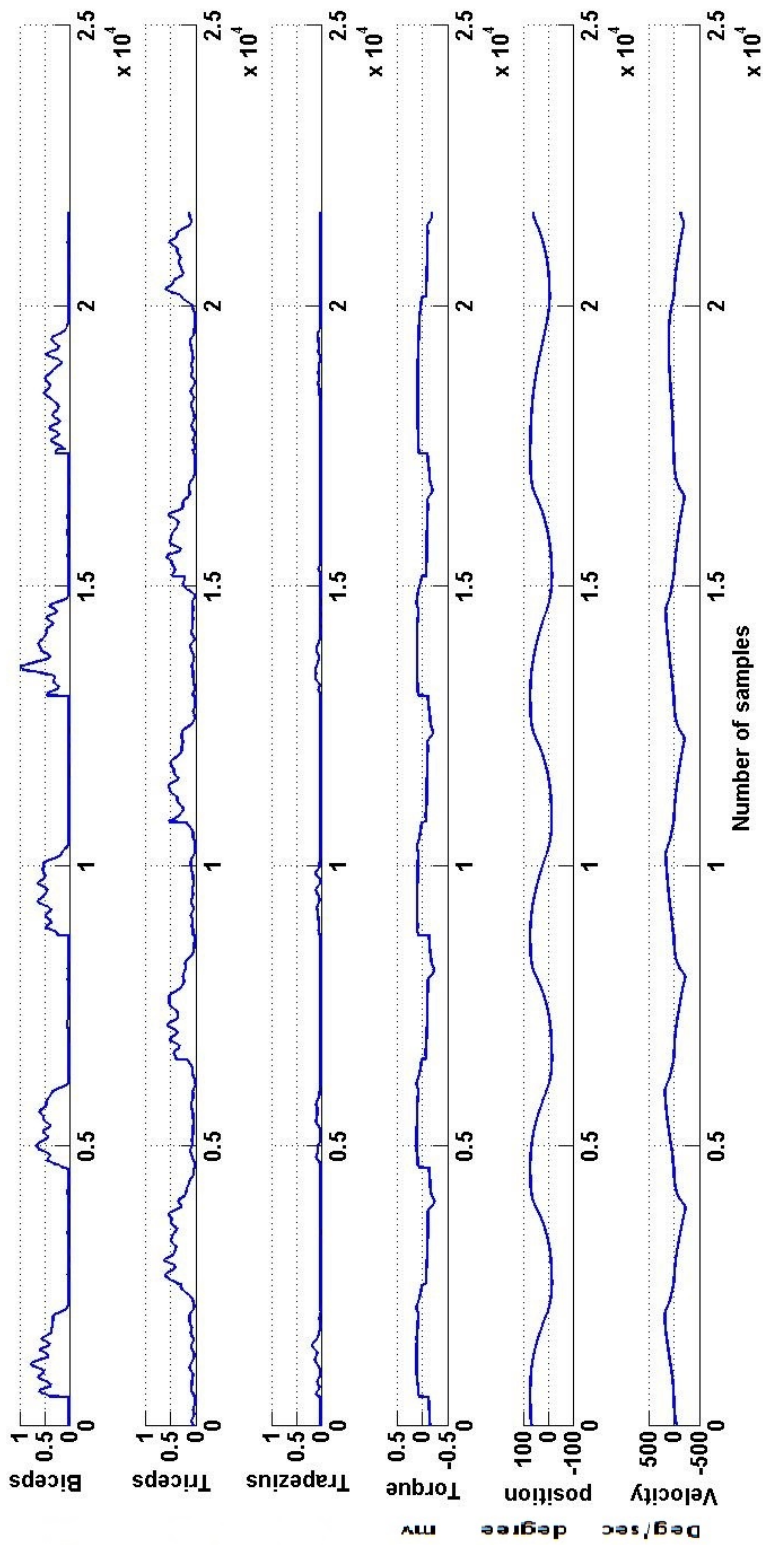


Figure 4.11: Experimental filtered data for five cycles after removing the delay

EMG onset is the approximate point where the muscle activity starts and Offset is the point where the same ends.

According to these assumptions, we have removed the delays due to no motion or no EMG activity between all the cycles.

In figure 4.10 (raw filtered EMG data) the vertical black dotted line is showing the lines the joint velocity where signal is zero. Similarly, pink dotted lines passing from the onset and offset of EMG activity of biceps and triceps. 'a' and 'b' representing the EMG delay between the cycles. If we remove the delay between the cycles on the basis of the velocity changes, then the processed data will be smoother for velocity and position as shown in figure 4.11. This figure is the outcome of plotting the data of all the five cycles again after removing the data corresponding to zero joint velocity. The method is explained in next section 4.3.3 and compared with the cycles separation on the basis of EMG Onset and Offset Values.

The plot for velocity and position is smoother in figure 4.11 in comparison to figure 4.10, while the plot for EMG for Biceps, Triceps, Trapezius, and torque do not have same start and end points for each cycle. It is because the EMG activity for both push and pull is for longer duration than the velocity or position change.

4.3.3 Time-scale normalization of the cycles

It is impossible to precisely repeat the duration of a repetition in human locomotion. Any averaging of such repetitions requires a time normalized format. Any averaging of such repetitions requires a time normalized format. The most popular concept, originally developed for gait analysis, separates all repetition within a given sequence into an equal amount of periods and calculates the mean value of each period. The original (mili-) second time scale is converted to percent of cycle ranging from 0 to 100%. Usually a segmentation of 100 (1 data point at each 1% step) is used. The simple method to normalize the EMG data on the basis of time scale and separation of each cycle the method is explained below:

- First data point, $T_0 = 0\%$ of the cycle = $(\frac{Time\ for\ first\ sample}{Total\ time\ for\ whole\ cycle})$
- Last data point, $T_{100} = 100\%$ of the cycle = $(\frac{Time\ for\ last\ sample}{Total\ time\ for\ whole\ cycle}) * 100$
- Interpolate all the data points with piecewise linear interpolation to 100 points for whole cycle. (Appendix .5)

The plot of each cycle on the same normalized scale would give the better comparison between each cycle. Based on the steps described above each repetition cycle is averaged to a mean curve. According to the filtered data shown in figure 4.10, we can do time scale normalization and cycle separation on the basis of two methods:

1. EMG Onset and Offset
2. Velocity variation

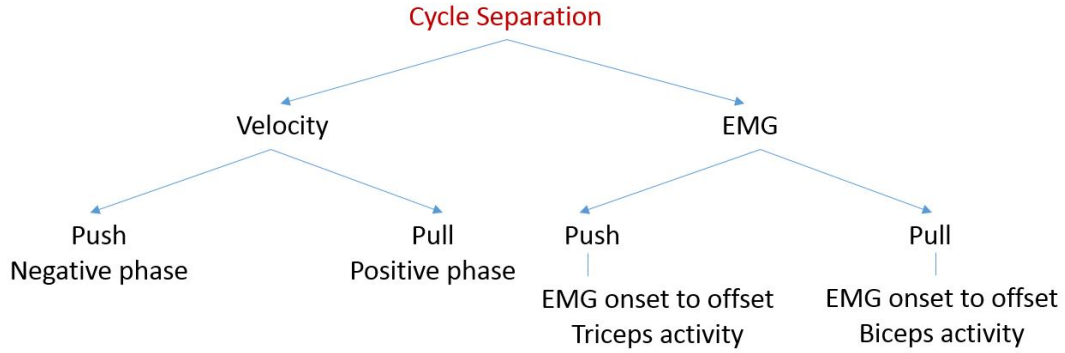


Figure 4.12: Cycle Separation

Here we will do time scale normalization on the basis of both methods and we will select a method best suitable for further data analysis. The figure 4.12 represent the same in a graphical way.

Cycles separation on the basis of EMG

If we separate the cycles and remove the delay on the basis of EMG activity (EMG onset and offset) as shown in figure 4.13 and figure 4.14. We have taken a small part from figure 4.13 to expand it in bigger form and understand it minutely. In figure 4.14 we can see that the data from first red dotted line to the second red dotted line is pull and from second red dotted line to third one is push. Here we can see that both push and pull have some co-contraction in between because of overlapping of some data of biceps and triceps. These plot is for one cycles. In this figure we can see that we have proper start and end for EMG data but for the mechanical data (torque, position and velocity) we have the data in which there is no actual mechanical work or motion. In figures /reffig:emgemplm to /refemgvelplm we can observe plots with standard deviation range which shows that the most cycles are near the standard deviation around the mean. In these figures we can also observe that EMG data have smooth plot but in mechanical data of position, velocity and torque have missing data in the start and end of the cycle. This is because of the cycle separation on the basis of EMG. The cycles separation on the basis of EMG Onset and Offset activity are done as follows:

- $\text{cycle}(\text{start}) = \text{data}^{\text{onset}}$
- $\text{cycle}(\text{end}) = \text{data}^{\text{offset}}$
- First data point = cycle(start), $T_0 = 0\%$ of the cycle = $\left(\frac{\text{Time for first sample}}{\text{Total time for whole cycle}}\right)$
- Last data point = cycle(end), $T_{100} = 100\%$ of the cycle = $\left(\frac{\text{Time for last sample}}{\text{Total time for whole cycle}}\right) * 100$
- Interpolate all the data points with piecewise linear interpolation for 100 points for whole cycle. (Appendix .5)

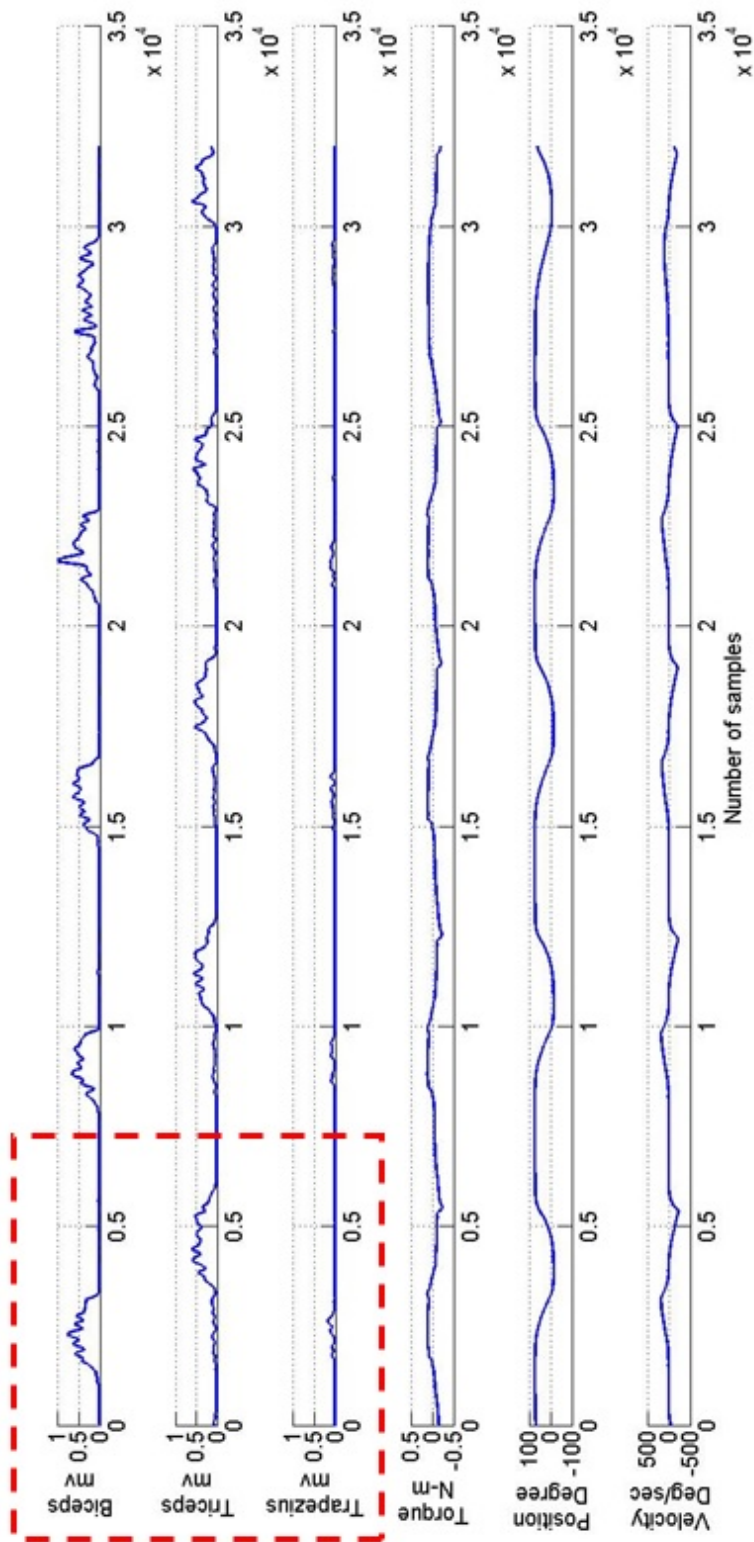


Figure 4.13: Study part from raw data

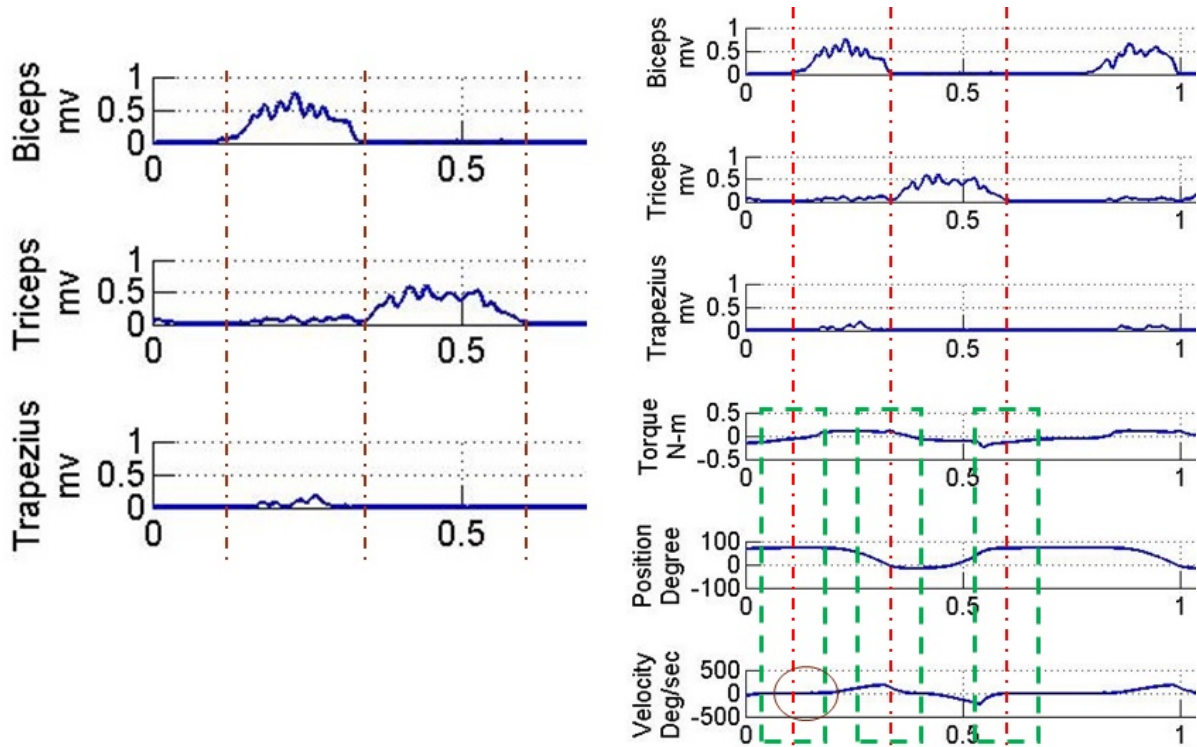


Figure 4.14: Separation on the basis of EMG

The above procedure is repeated for the biceps and triceps muscles separately, so that we can extract push part of the cycle and pull part of the cycles in separate comparable platform. The procedure of this part to compare the cycles on the basis of EMG (Appendix .6).

The cycles separated on the basis of EMG are shown in figure 4.15 for EMG activities of all the muscles in push phase and figure 4.16 shows the mechanical data for velocity, position and torque with the maximum and minimum region of reach for the cycles and the single mean cycle of all the cycle.

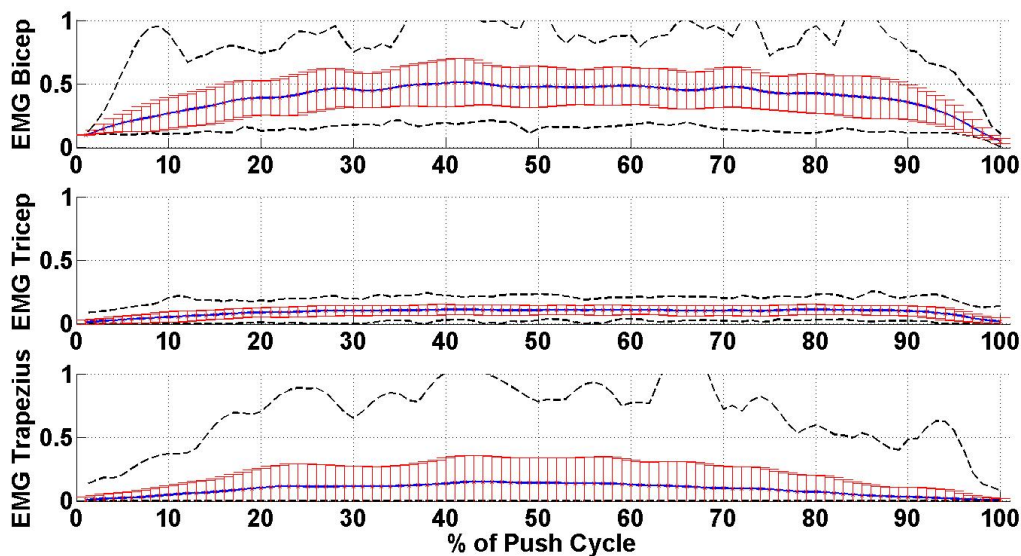


Figure 4.15: Mean and Standard deviation plot for EMG data of Bicep, Tricep and Trapezius in Push Phase of the cycles based on EMG onset and Offset

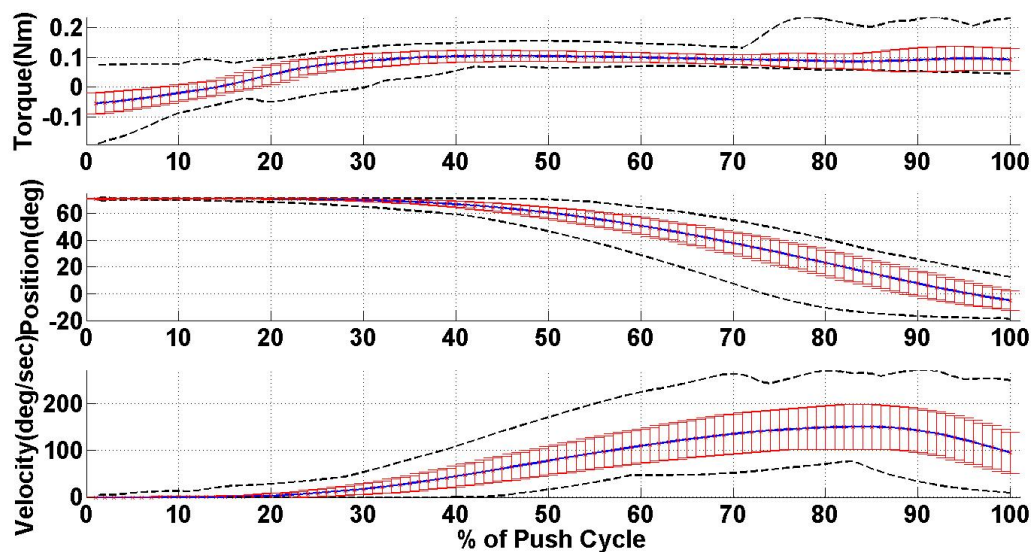


Figure 4.16: Mean and Standard deviation plot for velocity, position and torque in Push Phase of the cycles based on EMG onset and Offset

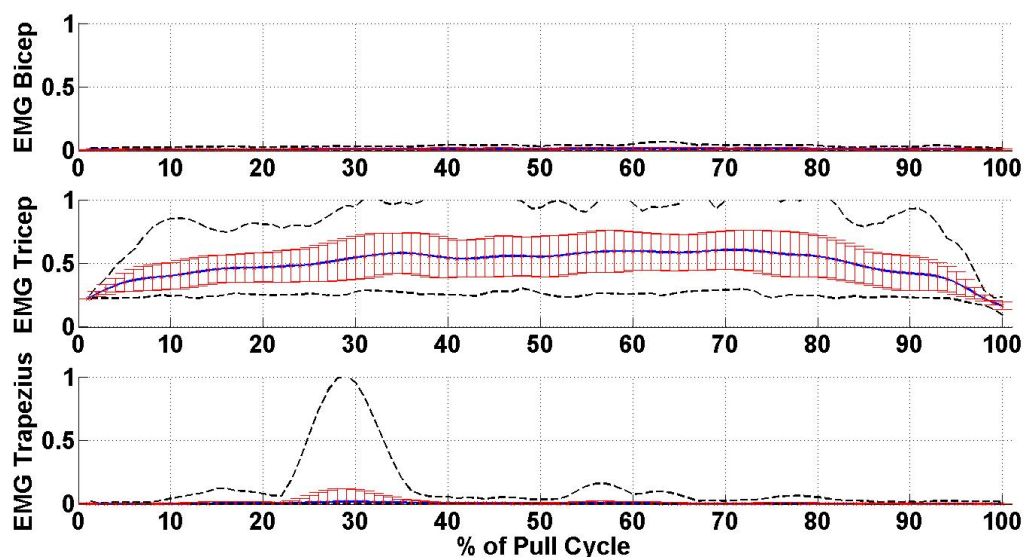


Figure 4.17: Mean and Standard deviation plot for EMG data of Bicep, Tricep and Trapezius in Pull Phase of the cycles based on EMG onset and Offset

Note:- Curves representation in figures 4.15 to 4.24:

--- Blue color curves show mean EMG activity.

I Red bar plotted on blue curves show the standard deviation of all the EMG activities along the mean.

– Black dotted curves show the maximum and minimum reach from the EMG activities.

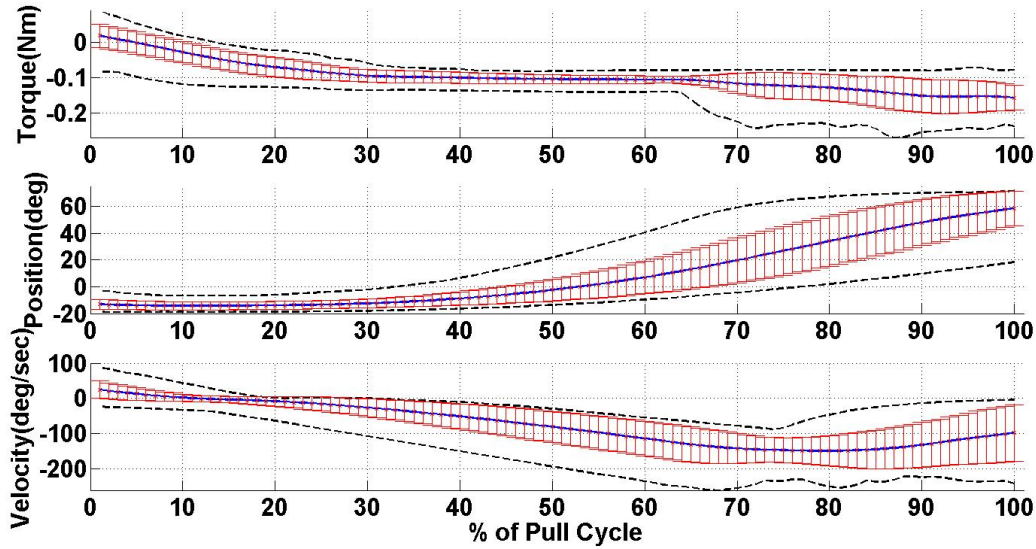


Figure 4.18: Mean and Standard deviation plot for velocity, position and torque in Push Phase of the cycles based on EMG onset and Offset

The red bars shows the standard deviation of all cycles plotted. And for pull part cycles are shown in figure 4.17 for EMG activities of all the muscles and figure 4.18 shows the mechanical data for velocity, position and torque with the maximum and minimum region of reach for the cycles and the single mean cycle of all the cycle. There are 2000 cycles plotted on the same graph for ten subjects. All the cycles are compared on the same scale. In these figures we observed that we are losing some of the mechanical data from torque, position and velocity. The advantage of this approach is that we can have very good EMG activity without any loss but at the same time we have very big disadvantage that we will lose the mechanical data, which is actually showing the real movement during the fatigue protocol.

Cycles separation on the basis of velocity

The cycles separated on the basis of velocity are on the basis of positive and negative phase of the velocity as shown in figure 4.19 and figure 4.20. We have taken a small part from figure 4.19 to expand it in bigger form and understand it minutely. In figure 4.20 we can see that the data from first red dotted line to the second red dotted line is pull and from second red dotted line to third one is push. Here we can see that both push and pull have some co-contraction in between because of overlapping of some data of biceps and triceps. These plot is for one cycles. In this figure we can see that we have proper start and end for the mechanical data (torque, position and velocity) without much loss but in EMG data we lose some of the data in start or end of EMG of biceps and triceps. In figures [/reffig:velemgphm](#) to [/refvelvelplm](#) we can observe plots with standard deviation range which shows that the most cycles are near the standard deviation around the mean. In these figures we can also observe that velocity data have smooth plot but in EMG data have missing data in the start and end of the cycle. This is because of the cycle separation on the basis of velocity.

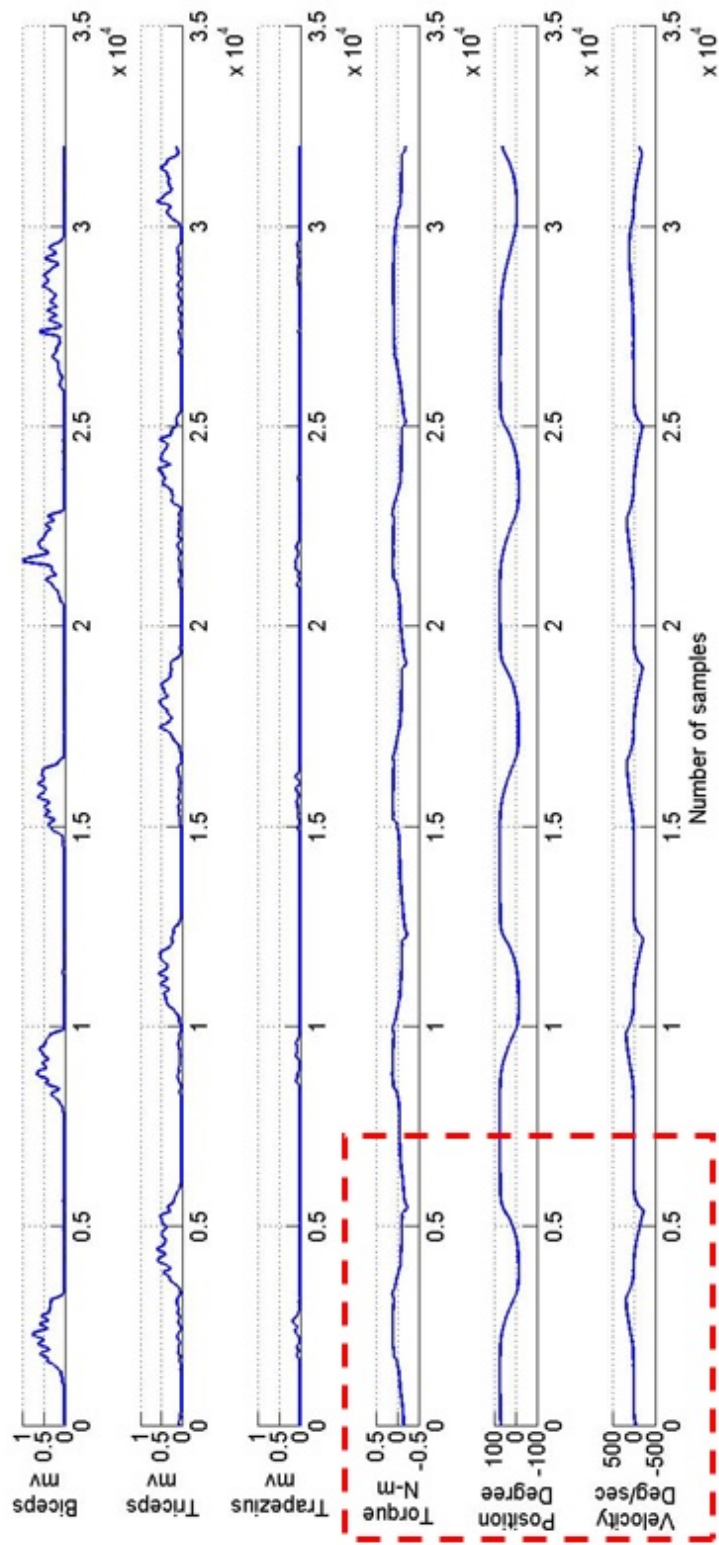


Figure 4.19: Study part from raw data

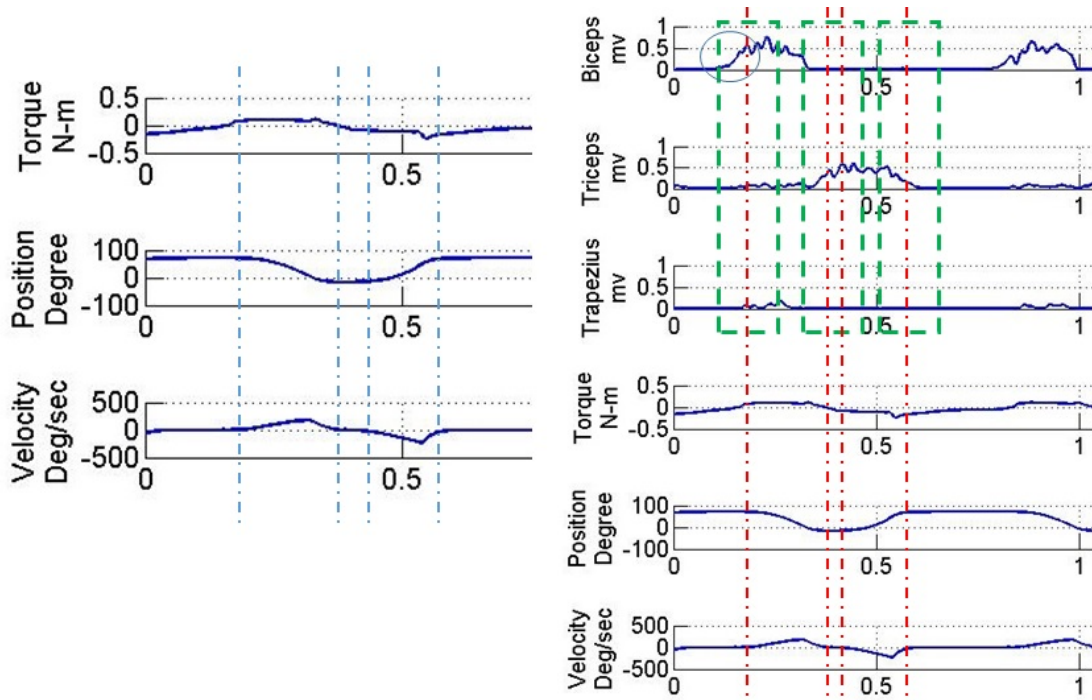


Figure 4.20: Separation on the basis of velocity

Here graphs plotted for push part are shown in figure 4.21 for EMG activities of all the muscles and figure 4.22 shows the mechanical data for velocity, position and torque with the maximum and minimum region of reach for the cycles and the single mean cycle of all the cycle. The red bars shows the standard deviation of all cycles plotted. And for pull part cycles are shown in figure 4.23 for EMG activities of all the muscles and figure 4.24 shows the mechanical data for velocity, position and torque with the maximum and minimum region of reach for the cycles and the single mean cycle of all the cycle. There are 2000 cycles plotted on the same graph with standard deviation range which shows that the most cycles are near the standard deviation around the mean.

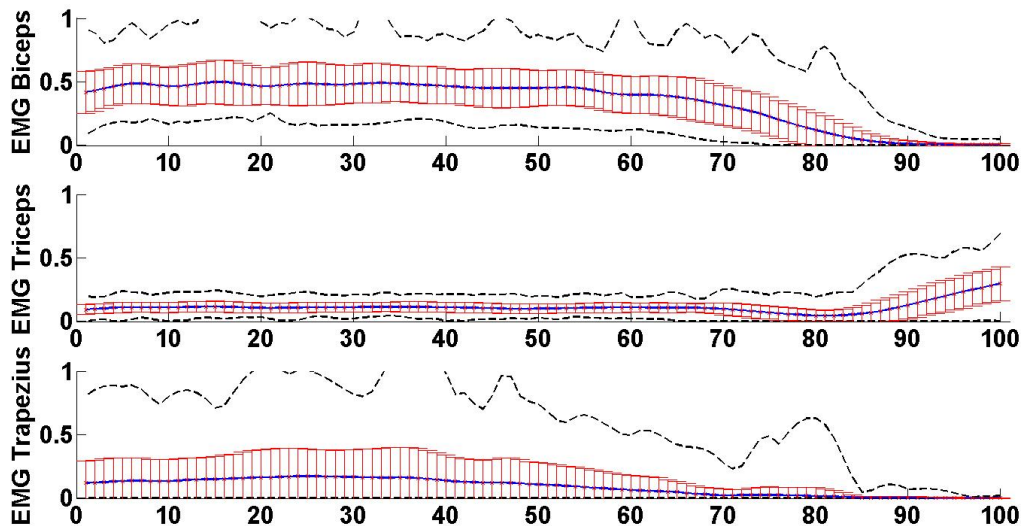


Figure 4.21: Mean and Standard deviation plot for EMG data of Bicep, Tricep and Trapezius in Push Phase cycle based on velocity change

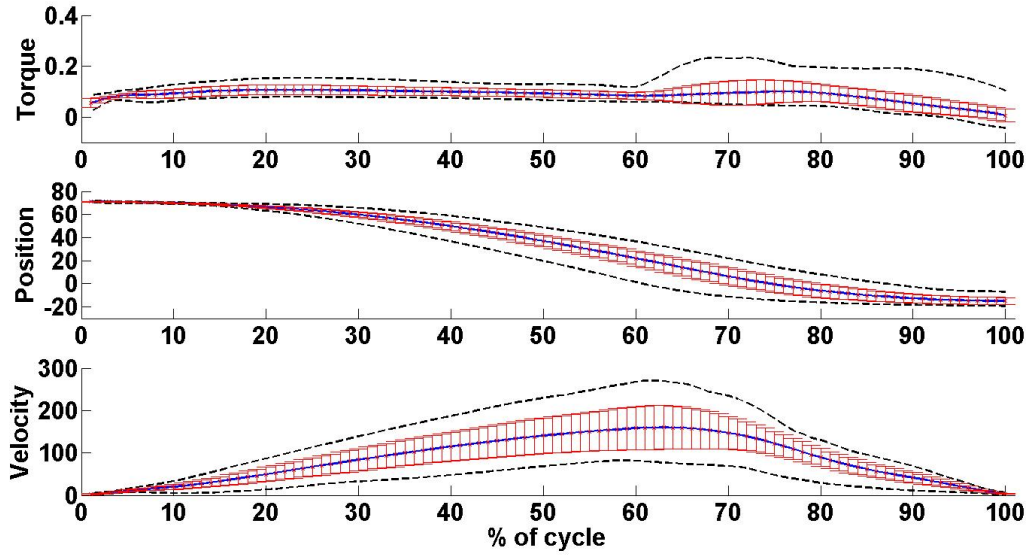


Figure 4.22: Mean and Standard deviation plot for EMG data of Bicep, Tricep and Trapezius in Pull Phase cycle based on velocity change

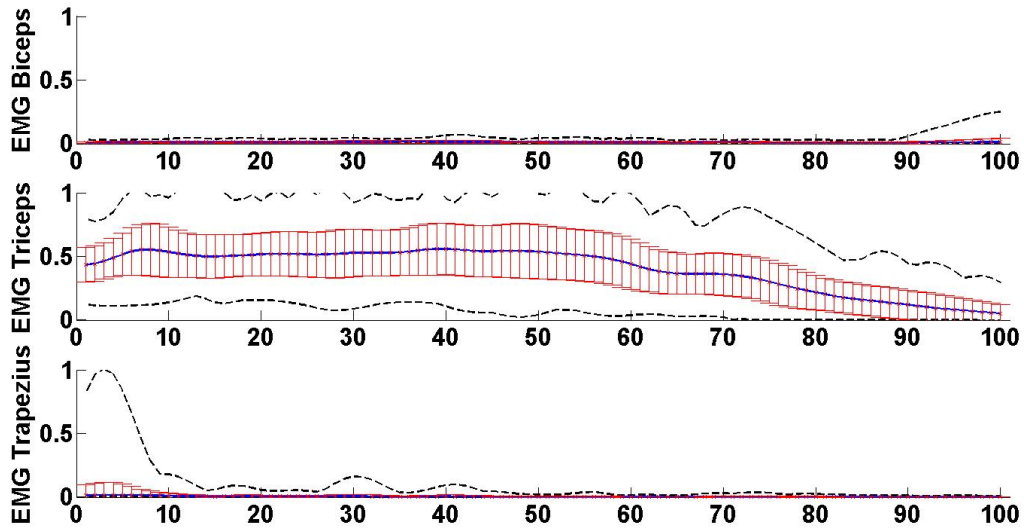


Figure 4.23: Mean and Standard deviation plot for velocity, position and torque in Push Phase cycle based on velocity change

The cycles separation on the basis of velocity are done as follows:

- velocity(Positive) = v_+^{first} to v_+^{second} = push cycle
- velocity(Negative) = v_-^{first} to v_-^{second} = pull cycle
- First data point = velocity(Positive), $T_0 = 0\%$ of the cycle = $(\frac{Time\ for\ first\ sample}{Total\ time\ for\ whole\ cycle})$
- Last data point = velocity(Negative), $T_{100} = 100\%$ of the cycle = $(\frac{Time\ for\ last\ sample}{Total\ time\ for\ whole\ cycle}) * 100$
- Interpolate all the data points with piecewise linear interpolation for 100 points for whole cycle.(Appendix .5)

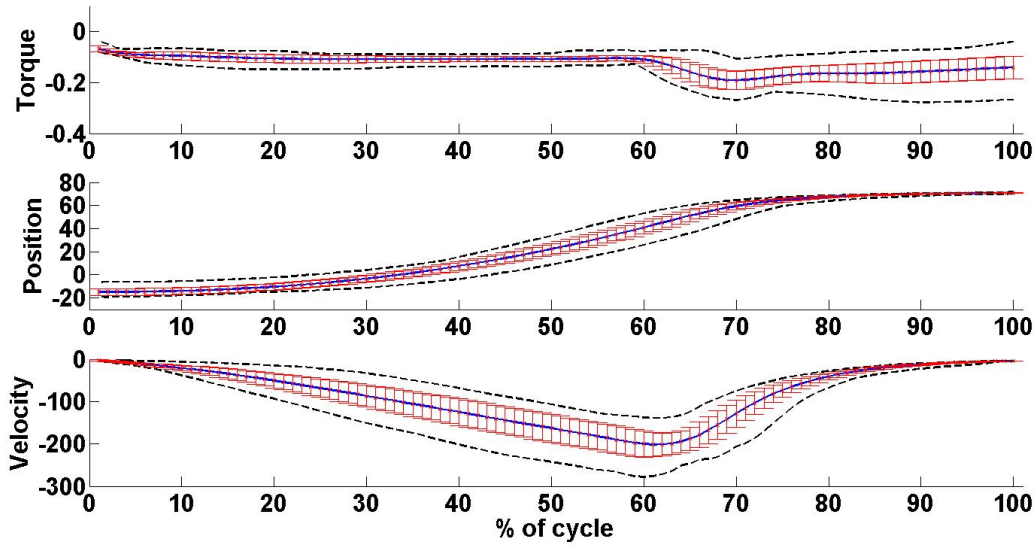


Figure 4.24: Mean and Standard deviation plot for velocity, position and torque in Push Phase cycle based on velocity change

v_+^{first} and v_+^{second} are respectively the start and the end point of positive velocity part of each cycle and v_-^{first} and v_-^{second} are respectively the start and the end point of negative velocity part of each cycle. The procedure of this part to compare the cycles on the basis of velocity is given in appendix .4. But in this case the mechanical data plots show that the position and velocity have some delay in the plots and not able to complete the push or pull cycles properly. This shows that this type of cycles separation is not good for mechanical data, which is the actual identification of push and pull phases in the cycle.

There are 2000 cycles plotted on the same graph for ten subjects. All the cycles are compared on the same scale. In these figures we observed that we are losing some of the EMG data in start or end of muscles activity. The advantage of this approach is that we can have very good mechanical activity without any loss but at the same time we have disadvantage that we will lose the EMG data, which is actually not showing the actual movement during the fatigue protocol.

4.3.4 Analysis and Conclusions

On comparing the figure 4.15 and figure 4.17, respectively with the figure 4.21 and figure 4.23, we can see that the EMG plots in figure 4.15 and figure 4.17 are more consistent at end points as compared to the other two plots in both push and pull cases. This shows that cycles separation on the basis of EMG gives consistent plots for EMG activity in comparison to velocity based cycles separation. On the other hand on comparing the figure 4.16 and figure 4.18, respectively with the figure 4.22 and figure 4.24, we can see that the mechanical data plots in figure 4.22 and figure 4.24 are more consistent at end points as compared to the other two plots in both push and pull cases. But in this case the EMG data plots are not consistent at the start and end in both push and pull cases.

This is mostly because we have remove the delay part from the analysis which also consist some EMG data. We can observe that the velocity based comparison and separation of cycles are more precise from the above analysis. So we made all the further analysis and results based on this method.

4.4 Summary

For the validation of proposed muscle fatigue and recovery model, experiment setup and protocol is defined. The experiments were performed on ten subjects for flexion-extension (push-pull) motion of the arm. The aim of these experiments was also to study the effect of co-contraction and effect of muscle fatigue on the EMG activity of muscles. The method to process and analyze the experimental raw EMG data is described, which includes filtering of the raw EMG data, extraction of requisite data and time-scale normalization of cycles. Two methods for the separation of cycles are proposed, i.e, EMG Onset and Offset basis and velocity change basis. Out of these two methods cycles separation on the basis of velocity found to be more precise for the purpose and will be applied further for any analysis of the cycles. The results and discussion for all the findings are given in next chapter.

Chapter 5

Results and Discussion

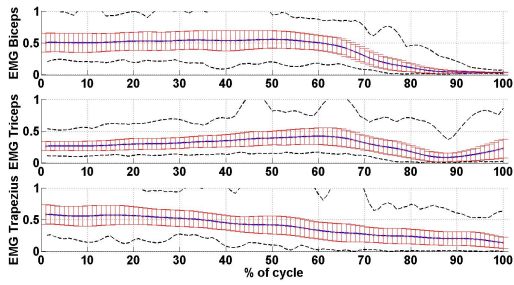
5.1 EMG Data Analysis

5.1.1 Description

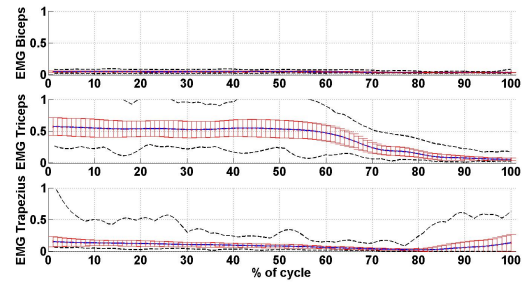
The EMG data for each subject is filtered and normalized on time scale to compare push and pull phases of cycles. The number of cycles compared for each subject is shown in table 5.1. The phases are separated from the main cycle on the basis of the change in velocity. The EMG plots for Biceps, Triceps, Trapezius muscle are shown in figure 5.1. The mechanical data for joint position, velocity and torque are shown in figure 5.2. Both Biceps and Triceps have three muscles, i.e, for biceps muscles are Biceps brachii short, Biceps brachii long and Brachialis, similarly for Triceps muscles are Triceps long, Triceps lateral and Triceps medial. In Our experiment we just put sensors on Biceps brachii long and Triceps lateral. Hence in whole thesis term Triceps represents triceps lateral and Biceps represents Biceps long. We did this because we just want to confirm the usefulness of co-contraction between muscles for our model. For Triceps we did not put sensor on Triceps long because it was covered by the back support.

Subject Number	Number of cycles	Total Test Duration (Minutes)
1	340	17
2	220	11
3	120	6
4	100	5
5	160	8
6	420	21
7	640	32
8	160	8
9	140	7
10	60	3

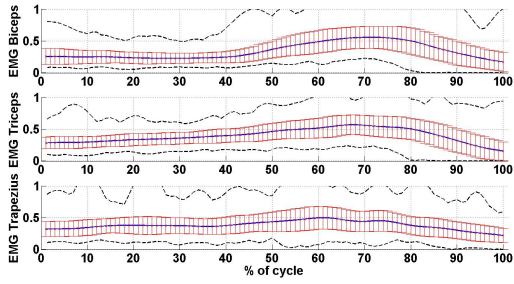
Table 5.1: Number of cycles completed by each subject



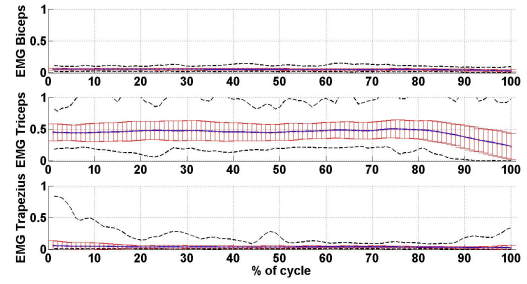
(a) Flexion in the subject 1



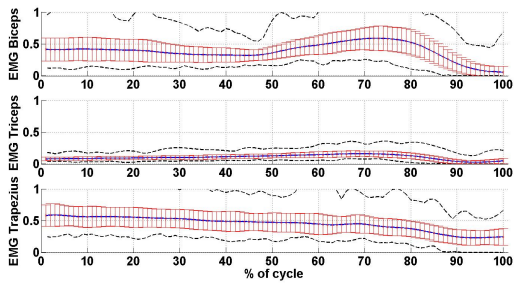
(b) Extension in the subject 1



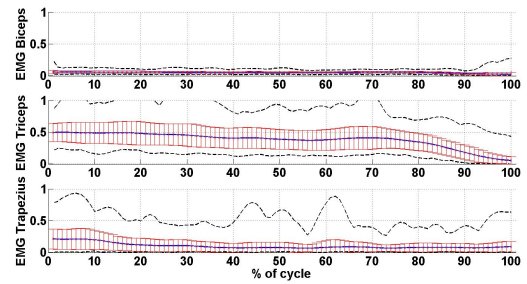
(c) Flexion in the subject 2



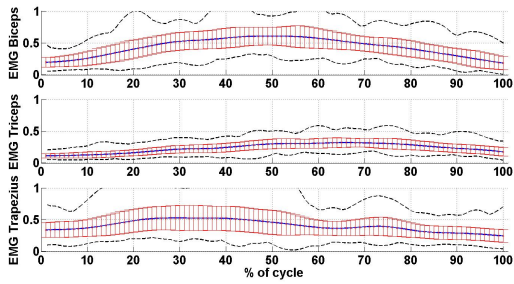
(d) Extension in the subject 2



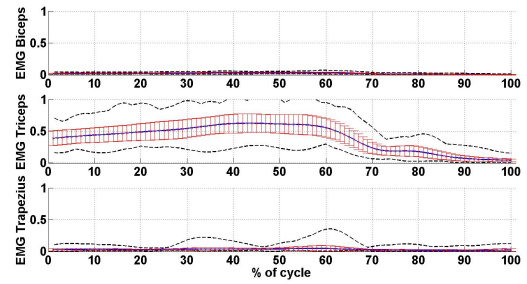
(e) Flexion in the subject 3



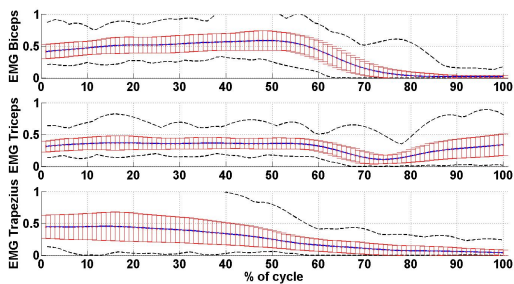
(f) Extension in the subject 3



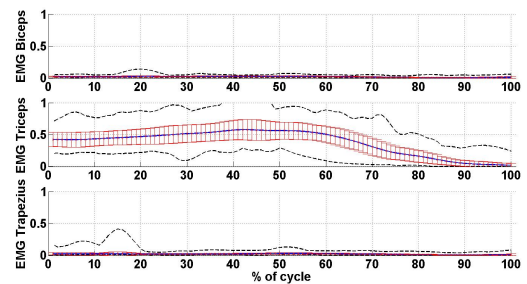
(g) Flexion in the subject 4



(h) Extension in the subject 4



(i) Flexion in the subject 5



(j) Extension in the subject 5

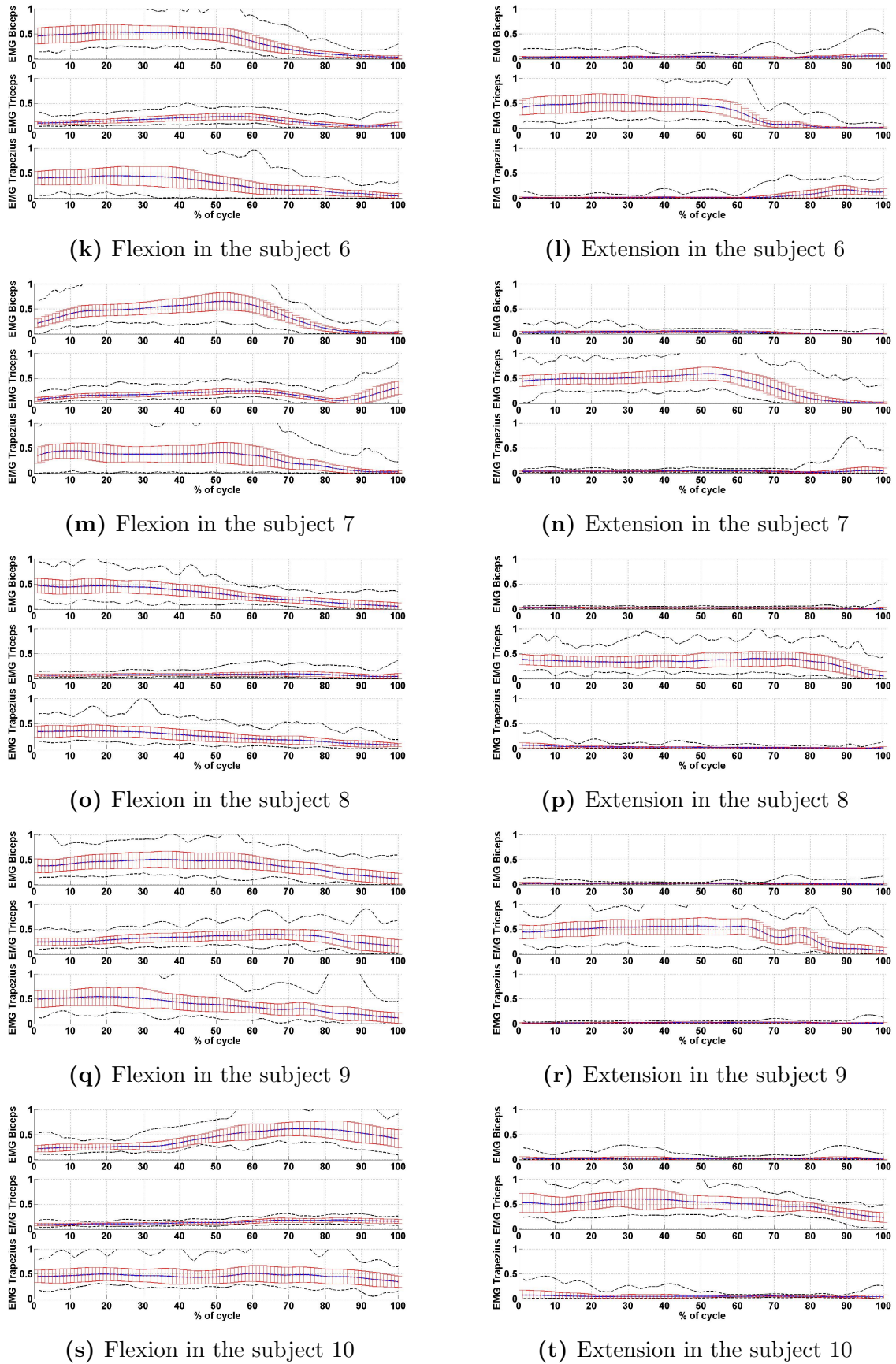
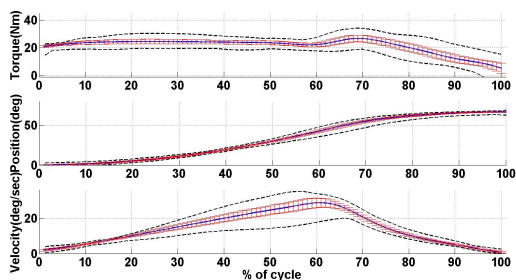
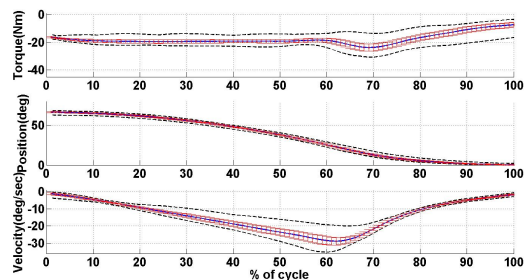


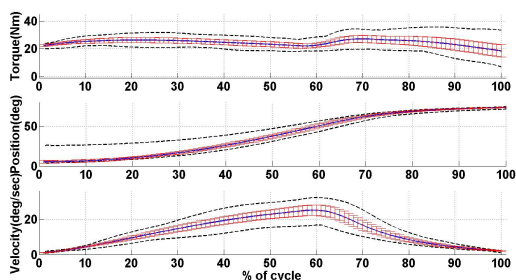
Figure 5.1: Mean and Standard deviation plots for EMG data of Bicep, Triceps and Trapezius for each subjects



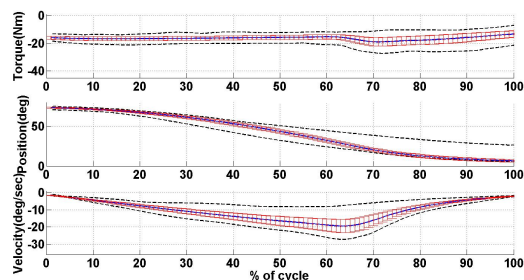
(a) Flexion in the subject 1



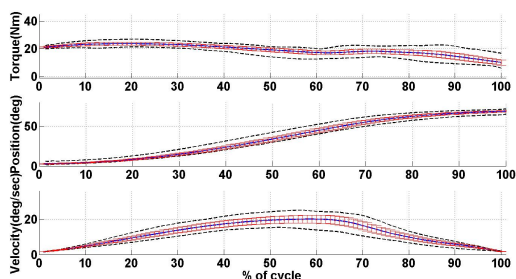
(b) Extension in the subject 1



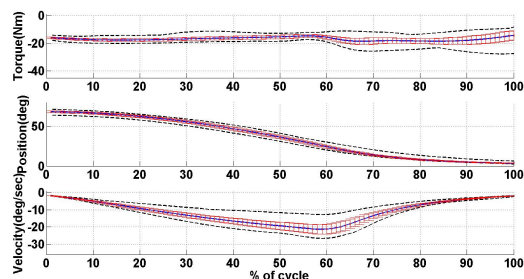
(c) Flexion in the subject 2



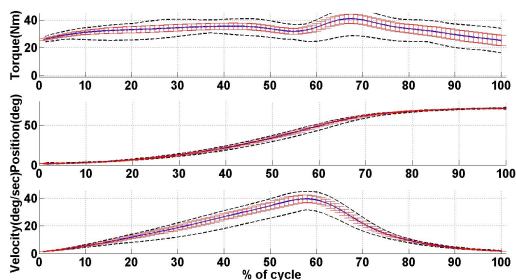
(d) Extension in the subject 2



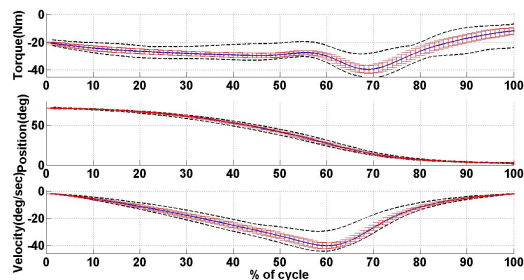
(e) Flexion in the subject 3



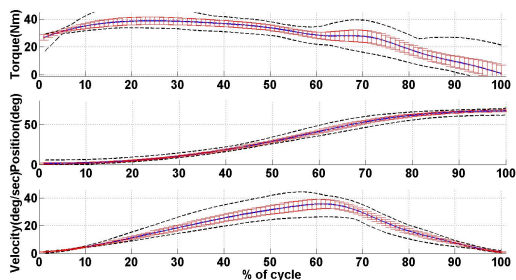
(f) Extension in the subject 3



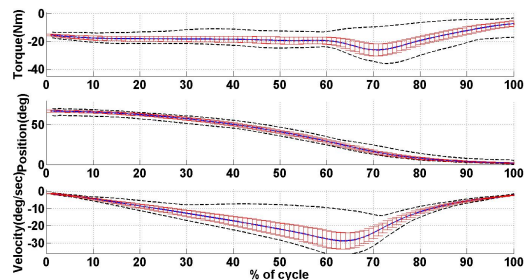
(g) Flexion in the subject 4



(h) Extension in the subject 4



(i) Flexion in the subject 5



(j) Extension in the subject 5

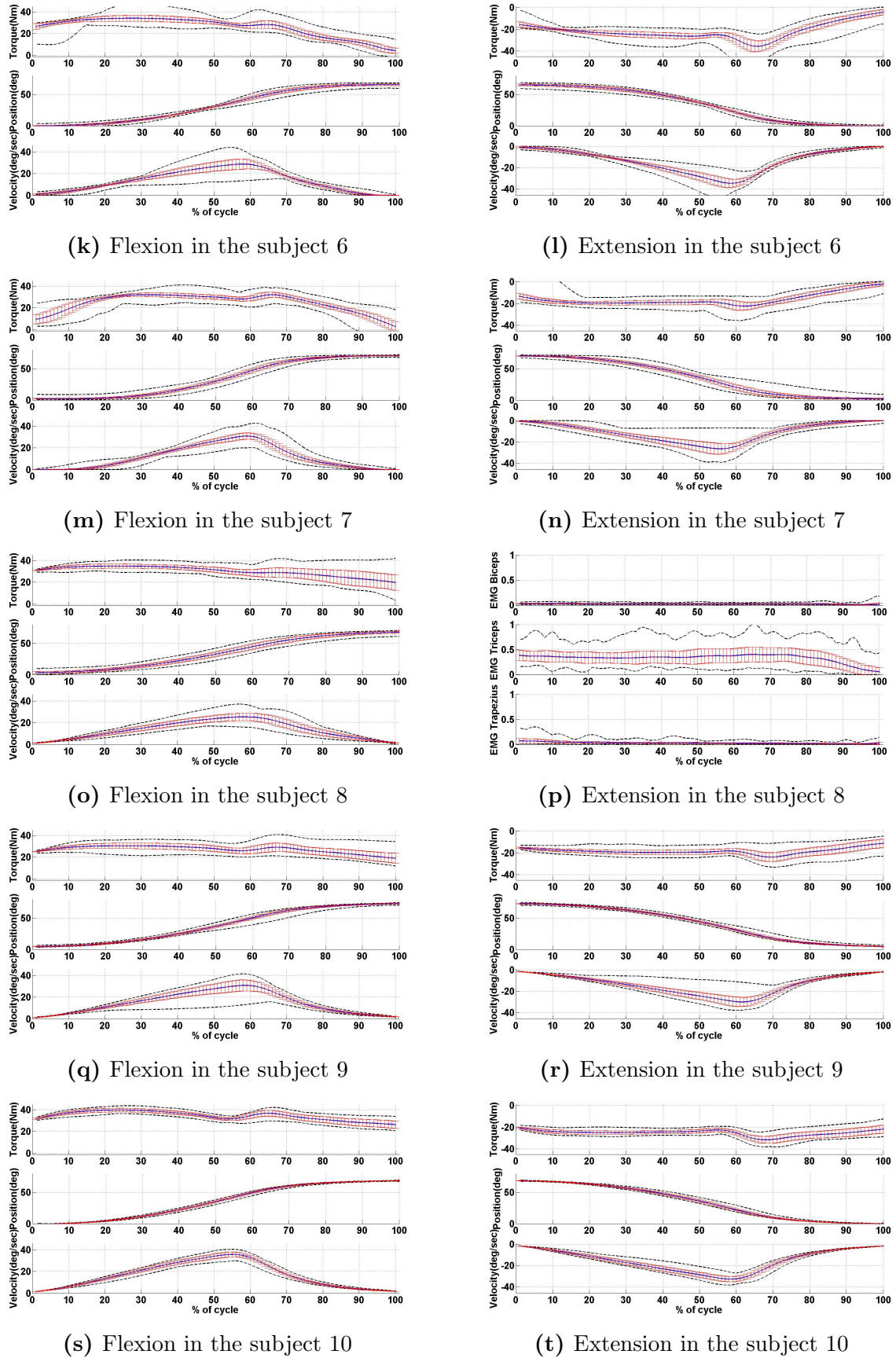


Figure 5.2: Mean and Standard deviation plots for velocity, position and torque for each subjects

It is obvious that if we put sensor on all the muscles we could have better data for co-contraction but because we have joint based model that is why we prefer just one muscle activity in place of group of muscles.

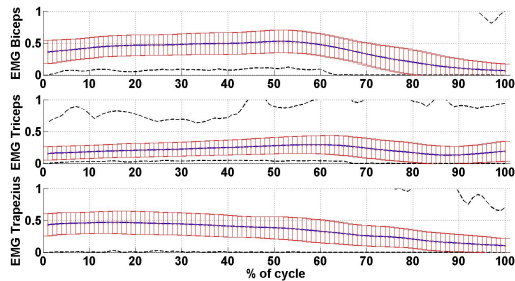
For figures 5.1, 5.2, 5.3 and 5.4 representations are as follows:

--- Blue colors curve show mean EMG activity.

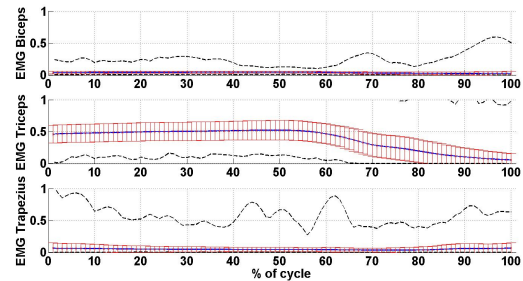
I Red bar plots on blue curve shows the standard deviation of all the EMG activities along the mean.

– Black dotted curves shows the maximum and minimum reaches from the EMG activities.

The total number of cycles compared for all the ten subjects are 2000 cycles in figures 5.3 and 5.4. The collective EMG plots for Biceps, Triceps and Trapezius muscle are shown in figure 5.3a and figure 5.3b for all the ten subjects and the collective comparison for the mechanical data position, velocity and torque are shown in figure 5.4b and figure 5.4a for all the ten subjects.

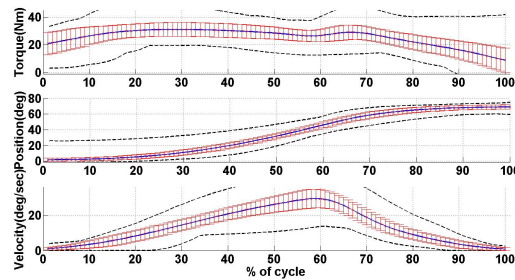


(a) Flexion/pull phase for all ten subjects

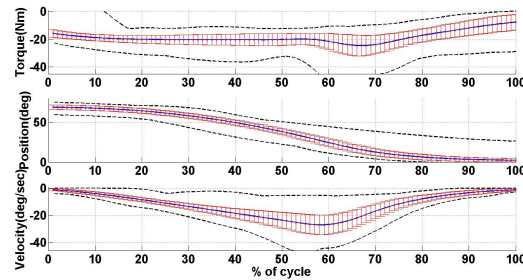


(b) Extension/push Phase for all ten subjects

Figure 5.3: Mean and Standard deviation plots for EMG data of Bicep, Triceps and Trapezius for all subject



(a) Flexion/pull Phase for all ten subjects



(b) Extension/push Phase for all ten subjects

Figure 5.4: Mean and Standard deviation plots for joint velocity, position and torque for all subjects

In figures 5.1, 5.2, 5.3 and 5.4 it is observed that because of the cycle separation on the basis of velocity we lose some data of EMG from the activity of triceps and biceps. It may be interesting to see if we can provide some factor to compensate this loss of activities. That can help us in analyzing the EMG activity and co-contraction in better way.

5.1.2 Analysis and Discussion

After processing the EMG data of all the muscle groups from figures 5.1, 5.2, 5.3 and 5.4 we can observe that when the biceps are active during flexion phase there are always some activities from the triceps and on the other hand when triceps are active during pull phase the biceps are almost passive or activities are very near to zero. But in our assumptions we said that when one group of muscle is active, other will be passive. We can also observe the co-activation of Trapezius muscle with the activation of biceps. During flexion phase, the activation of triceps with the biceps is co-contraction between two muscles. We can observe that the activities of biceps and triceps during flexion and extension are according to our hypothesis for extension phase while for flexion phase it is not true. This is because during flexion the subject have to work against gravity while during extension the gravity supports the motion. In our experiment we did compensate the effect of gravity by increasing the load for extension equivalent to the load by handle alone at zero position but still this does not show much effect. For study we divided this co-contraction equally between both the phases for further simplification and analysis of co-contraction factor. In these results we see the activity from the trapezius which is intermittent in many cycles, but we can also analyze the activity from other nearby muscles which can help in motion at the time of fatigue. We see the co-contraction with the triceps muscle at the time of flexion. It will be better to analyze more muscle of the triceps and biceps.

5.2 Results and analysis for co-contraction

5.2.1 Description

The co-contraction index calculated by using equation 3.6 is fitted with the exponential equation 3.7 in section 3.1.3. The figure 5.6 shows the fitted graphs for the co-contraction index for test cycles of all ten subjects. Figure 5.6 blue dots show the contraction index during each extension-flexion phase and red curve shows the exponential fit for the co-contraction index but in every case this fit looks almost linear in behavior. In figure 5.6 we observe the decreasing slope of C_A for almost 9 subjects, see figures 5.6a to figure 5.6j, the co-contraction index for activity between the muscles reduces as the fatigue test proceed or the muscles gets fatigued. We can notice that only the subject number 8 in figure 5.6h has increasing slope for the co-contraction area. This behavior can be associated with his sport activity which is wall climbing and very different from other subjects as shown in table 4.1. But still for 9 subjects we are getting the upward slope. By the equation 3.6 and 3.8 we can find n_i (co-contraction factor for each subject) as shown in table 5.2, where i is the subject number:

n_i	n_1	n_2	n_3	n_4	n_5	n_6	n_7	n_8	n_9	n_{10}
<i>Mean co-contraction factor</i>	1.4	1.45	1.33	1.4	1.41	1.35	1.36	1.26	1.5	1.3

Table 5.2: Co-contraction factor for each subject

The residual of co-contraction index data from all the subjects is shown in figure 5.5. The plot of the residual of C_A shows that all the data lie within the deviation range of ± 5 from the zero line. Hence distribution is good for 10 human subjects.

Normally the residual should be near to zero line in mechanical experiments for better accuracy. But in this case because of human unpredictable data behavior the 5% variation is quite satisfactory.

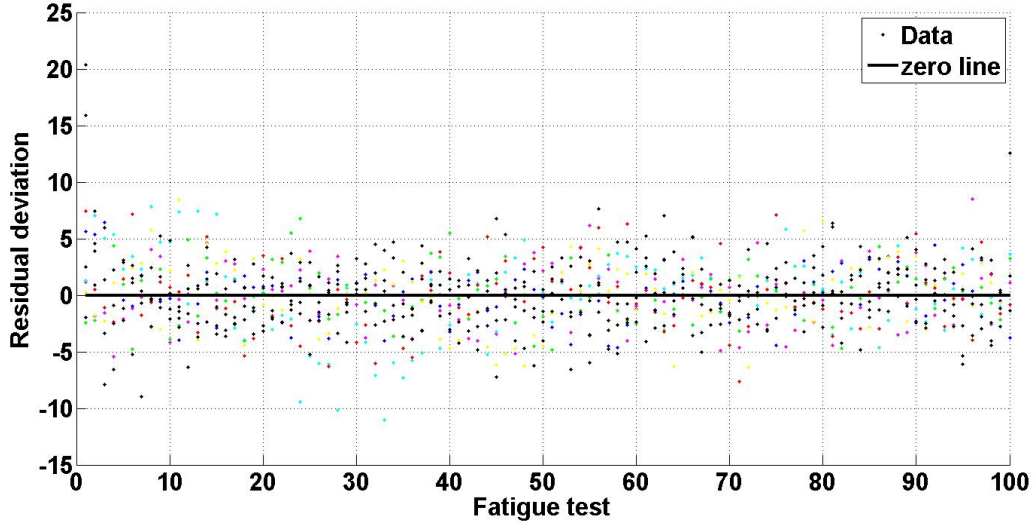


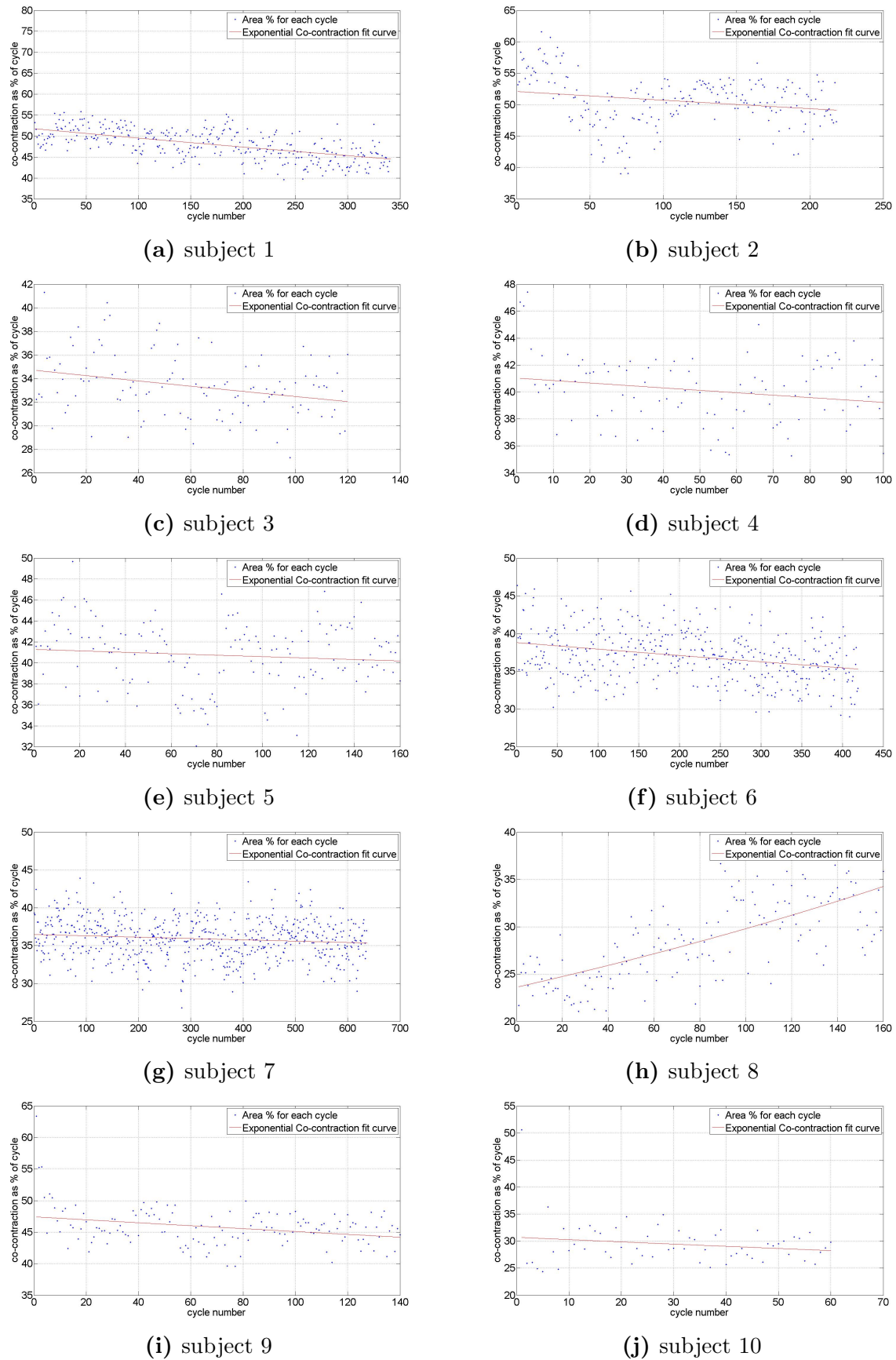
Figure 5.5: Residual analysis for co-contraction index data

The sigma distribution of co-contraction index as shown in figure 5.7 gives the mean value of co-contraction for general public. The sigma distribution is in the range of $[-2\sigma$ to $+2\sigma]$. The mean average value of C_A for all subjects is approximately 0.38 calculated by using equation 5.1 with $\sigma = 0.0791$. The sigma distribution is done using the equation described above the figure 5.7.

$$f(x) = \frac{e^{-(x-\mu)^2/(2\sigma^2)}}{\sigma\sqrt{2\pi}} \quad (5.1)$$

As we know the mean co-contraction index value for the 10 subjects (table 5.2), we can also find the normalized mean value of co-contraction factor. This value will be used further to analyze and validate muscle fatigue model. Here, we are using the value of C_A calculated by equation 5.1.

$$\begin{aligned} C_A &\approx 0.38 \\ n &= 1 + C_A \\ n &= 1 + 0.38 = 1.38 \end{aligned}$$

**Figure 5.6:** Exponential curve fit for the co-contraction area

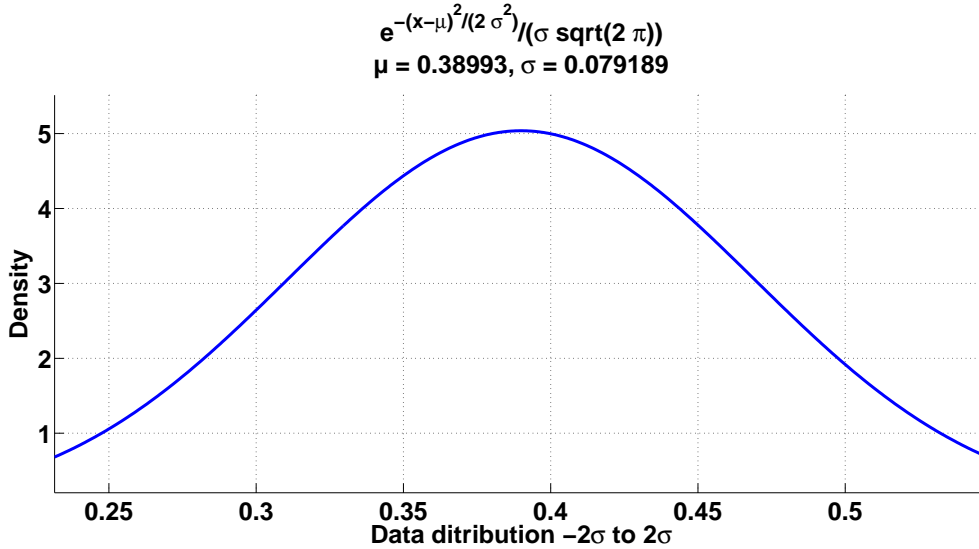


Figure 5.7: Sigma distribution for co-contraction index

5.2.2 Co-contraction Analysis

The co-contraction index in figure 5.6 shows the decreasing slope for 9 out of 10 subjects which is satisfactory which means for 90 % of population co-contraction index behave in this way. The value of co-contraction index signifies the participation of co-contraction between two muscles activity during the protocol. The value is 1.38 which means during whole activity co-contraction contribution is 38% of whole activity. Hence by including co-contraction in the dynamic muscle fatigue model will make it more realistic. The co-contraction factor value of $n = 1.38$ represent part of the whole cycle excluding co-contraction effect. If we include this into model it means that the model will predict the fatigue with excluding the forces or activity because of co-contraction and hence represent the real forces produced by muscles without being contracted at the same time. The co-contraction signifies varies characteristics with fatigue with linear and downward slope see figures 5.6a to 5.6j. The effect of co-contraction will be visible in upcoming sections.

5.3 Experimental Validation of Muscle Fatigue Model

The MVC values are measured between each protocol of one minute (Appendix .7). We can see in most of the cases MVC decreases as fatigue increases. The MVC is same as Γ_{cem} used in our model. The theoretical and experimental evolution of Γ_{cem} is on the basis of k (fatigue rate) using equation 3.11 and equation 3.15 and calculated, n_i and $C = 0.2$. The evolution of Γ_{cem} extension for fatigue parameter ‘ k ’ is shown in figure 5.8, 5.10, 5.12, 5.14, 5.16, 5.18, 5.20, 5.22, 5.24 and 5.26. Similarly the evolution of Γ_{cem} flexion is shown in figure 5.9, 5.11, 5.13, 5.15, 5.17, 5.19, 5.21, 5.23, 5.25 and 5.27. In these figures blue line show the MVC measured for flexion and extension after each test protocol of 1 minute. The MVC values measured are $\Gamma_{cem}(t)$, used in calculating

fatigue rate ‘ k ’ using equation 5.2.

$$k = \frac{-1}{n.Ct} \cdot \ln \left(\frac{\Gamma_{cem}(t)}{\Gamma_{MVC}} \right) \quad (5.2)$$

The theoretical Γ_{cem} is then calculated with respect to minimum, maximum and average value of fatigue rate using equation 3.11. The theoretical and experimental evolution of Γ_{cem} shows that the experimental values are well fit with in the theoretical model. The co-contraction factor have significant effect on the model. The minimum, maximum and average value of ‘ k ’ for each subject are shown in table 5.3. The red, pink and black dotted curves in figures 5.8 - 5.27 represent theoretical Γ_{cem} , calculated from minimum, maximum and average value of fatigue rate ‘ k ’ respectively.

Subject	$k_{extension}$			$k_{flexion}$		
	Minimum	Maximum	Average	Minimum	Maximum	Average
1	-0.0718	0.0605	0.0069	0.0420	0.3943	0.1182
2	0.1389	0.3606	0.2161	0.0923	0.2170	0.1341
3	0.3116	0.6096	0.4791	0.2073	0.4029	0.2821
4	0.9171	1.8485	1.2438	0.8476	1.4332	1.0815
5	0.2152	0.5310	0.3468	0.2607	0.5231	0.3657
6	0.0428	0.3305	0.1403	0.1268	0.6086	0.2542
7	0.0103	0.1423	0.0376	0.0036	0.1009	0.0436
8	0.1103	0.2972	0.1944	0.1196	0.6325	0.3636
9	0.2116	0.7701	0.5601	0.0884	0.4942	0.2876
10	0.2329	0.2883	0.2606	0.1687	0.3443	0.2565

Table 5.3: Experimentally calculated values of ‘ k ’ for flexion and extension motion

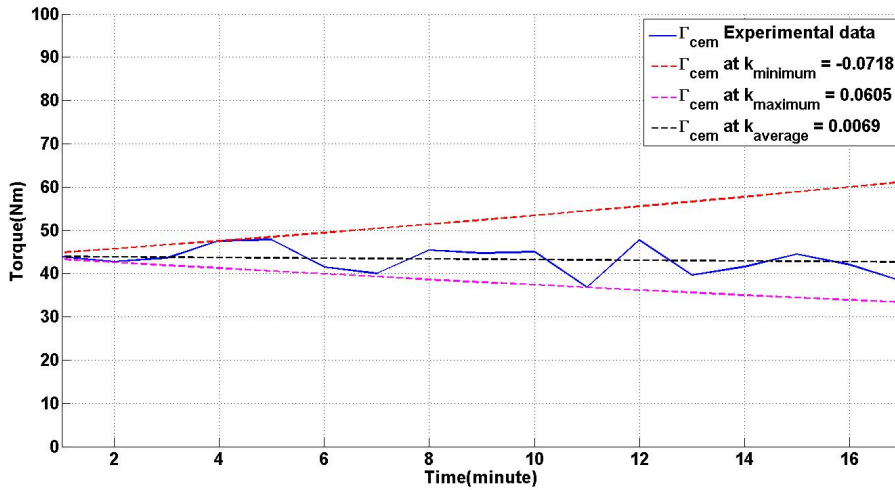


Figure 5.8: Γ_{cem} evaluation for extension phase in the subject 1

The experimentally calculated value of $\Gamma_{cem}(t)$ is mostly in the range of theoretical Γ_{cem} , which validate our muscle fatigue model. The fatigue rate increased with the input Γ_{cem} , which show the significant effect of co-

contraction factor in the fatigue model. In Place of particular value of n for each subject we are going to use $n = 1.38$ as calculated in previous section. The experimentally calculated values of MVC for each subject in Appendix .7.

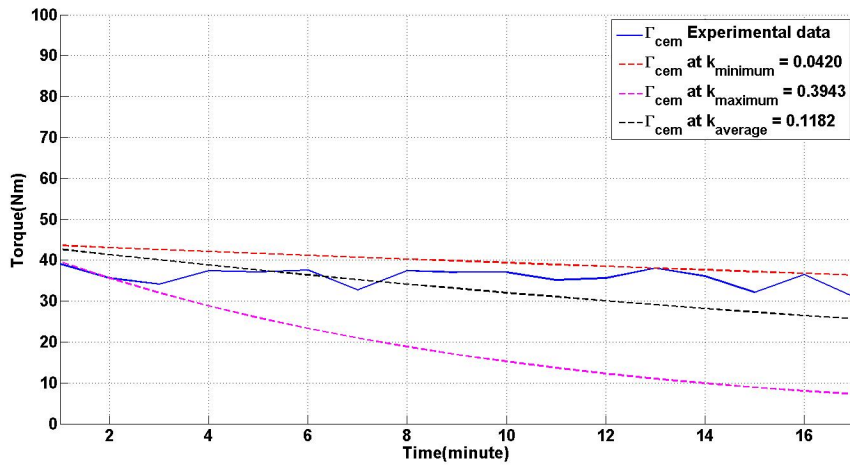


Figure 5.9: Γ_{cem} evaluation for flexion phase in the subject 1

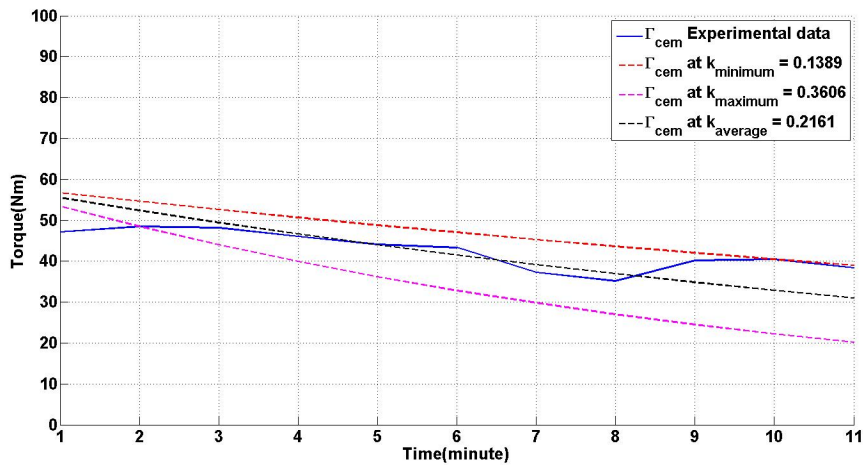


Figure 5.10: Γ_{cem} evaluation for extension phase in the subject 2

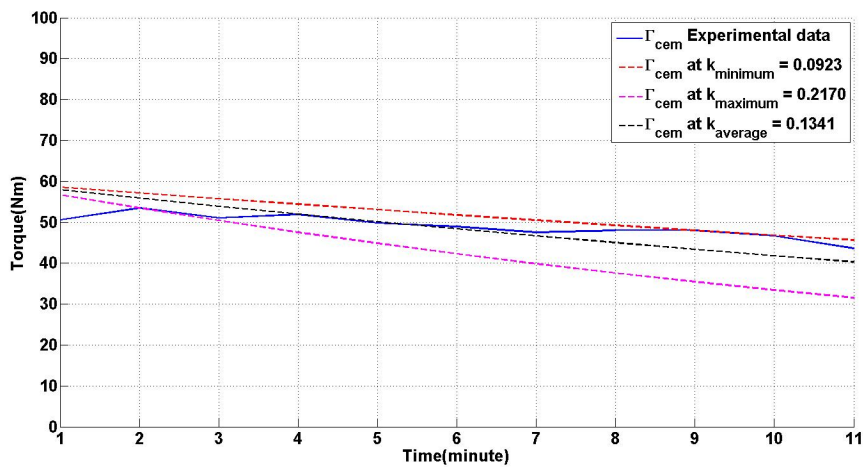


Figure 5.11: Γ_{cem} evaluation for flexion phase in the subject 2

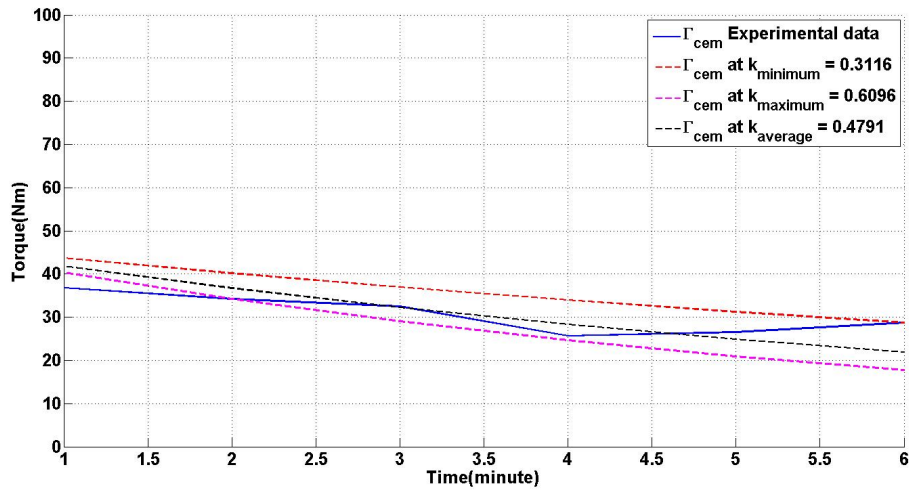


Figure 5.12: Γ_{cem} evaluation for extension phase in the subject 3

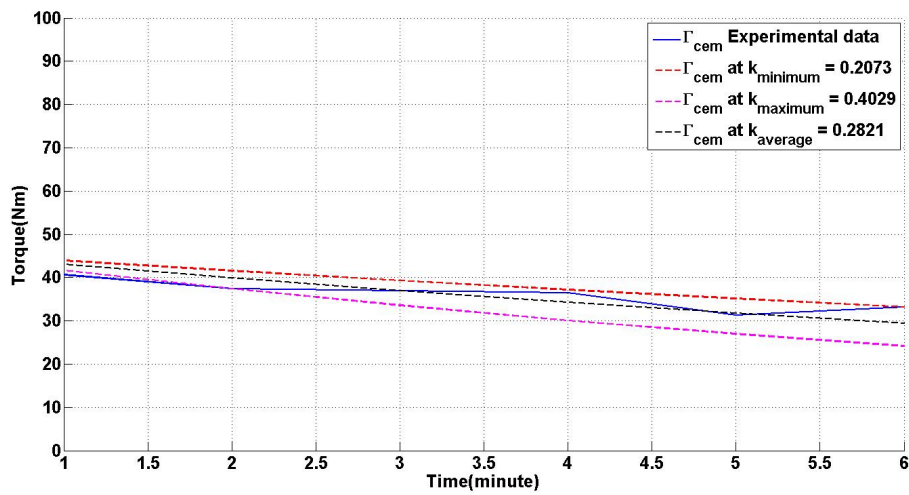


Figure 5.13: Γ_{cem} evaluation for flexion phase in the subject 3

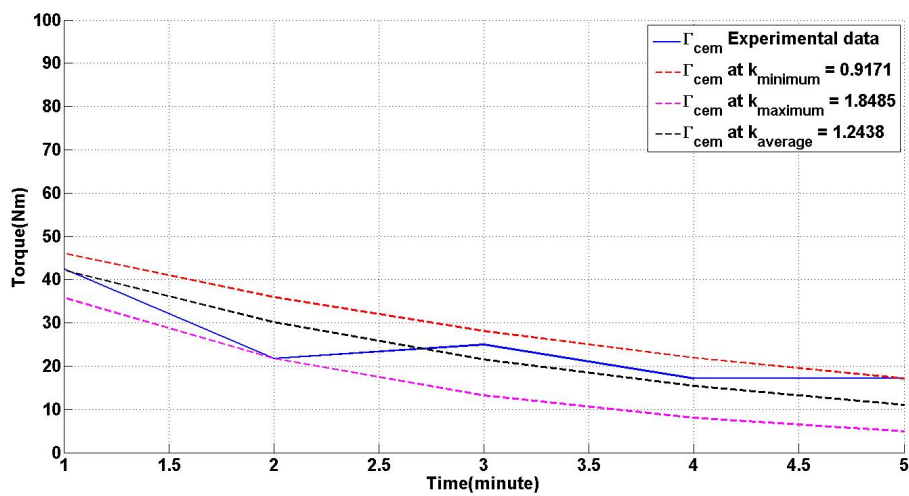


Figure 5.14: Γ_{cem} evaluation for extension phase in the subject 4

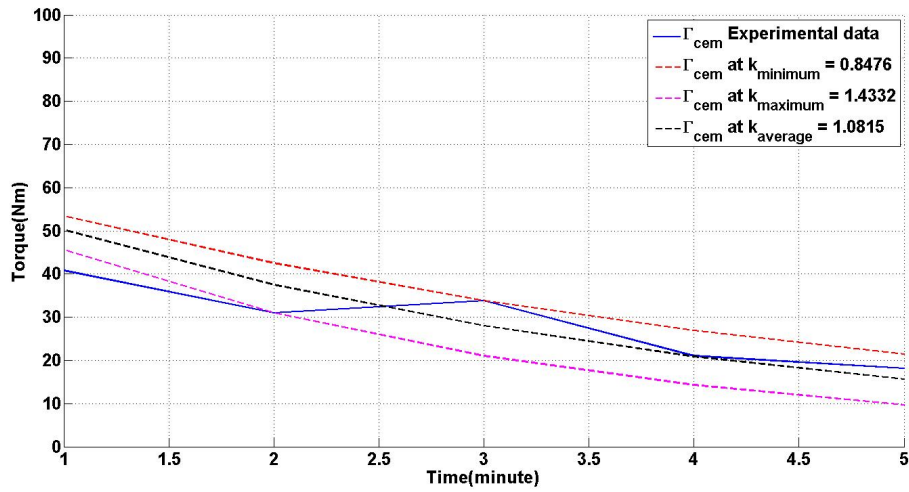


Figure 5.15: Γ_{cem} evaluation for flexion phase in the subject 4

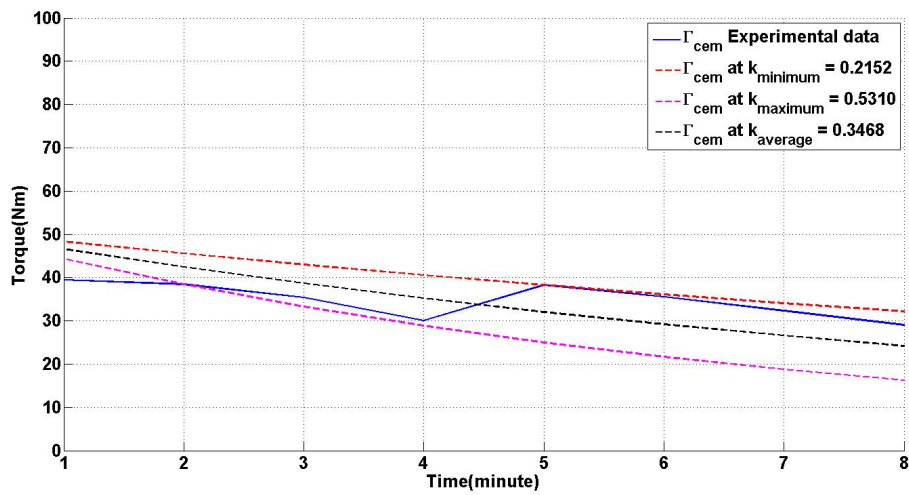


Figure 5.16: Γ_{cem} evaluation for extension phase in the subject 5

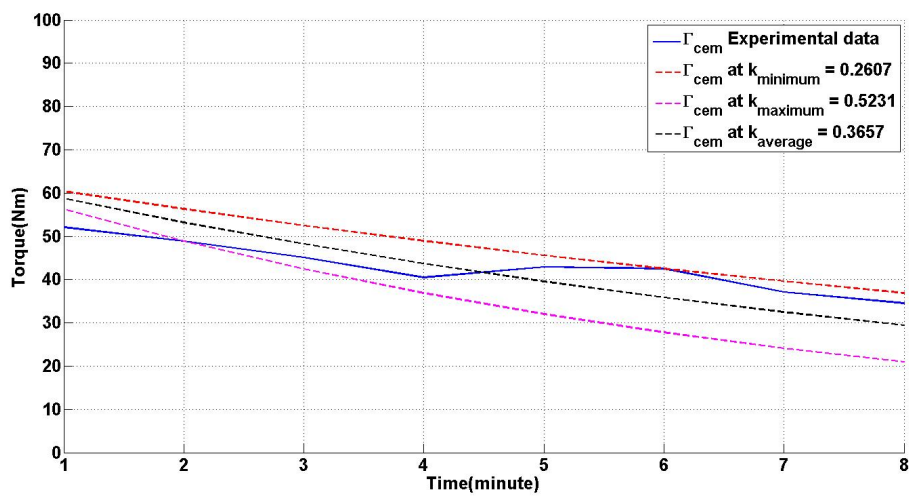


Figure 5.17: Γ_{cem} evaluation for flexion phase in the subject 5

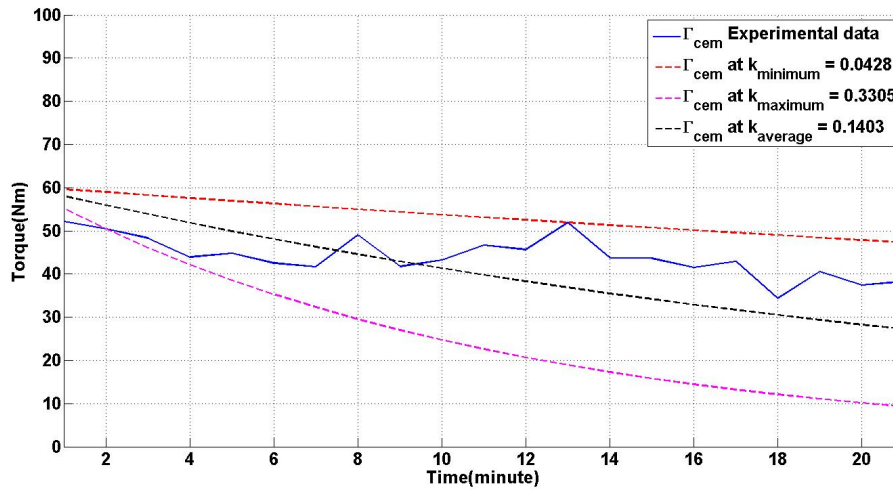


Figure 5.18: Γ_{cem} evaluation for extension phase in the subject 6

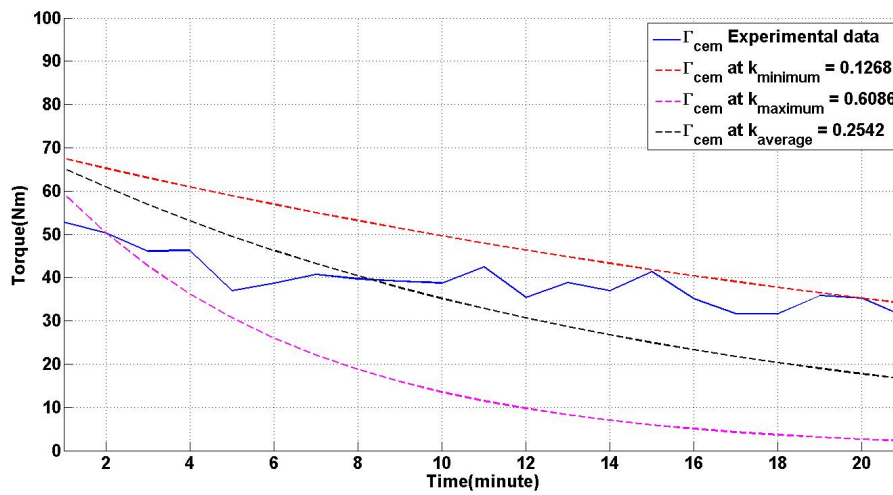


Figure 5.19: Γ_{cem} evaluation for flexion phase in the subject 6

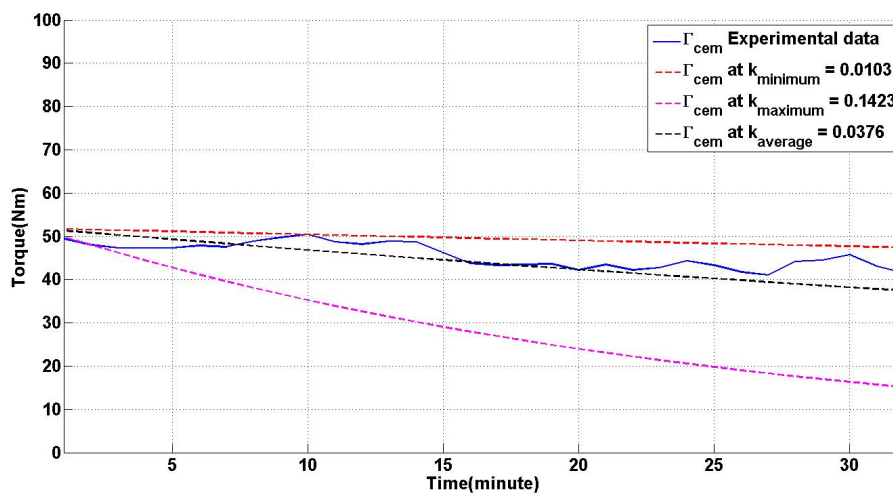


Figure 5.20: Γ_{cem} evaluation for extension phase in the subject 7

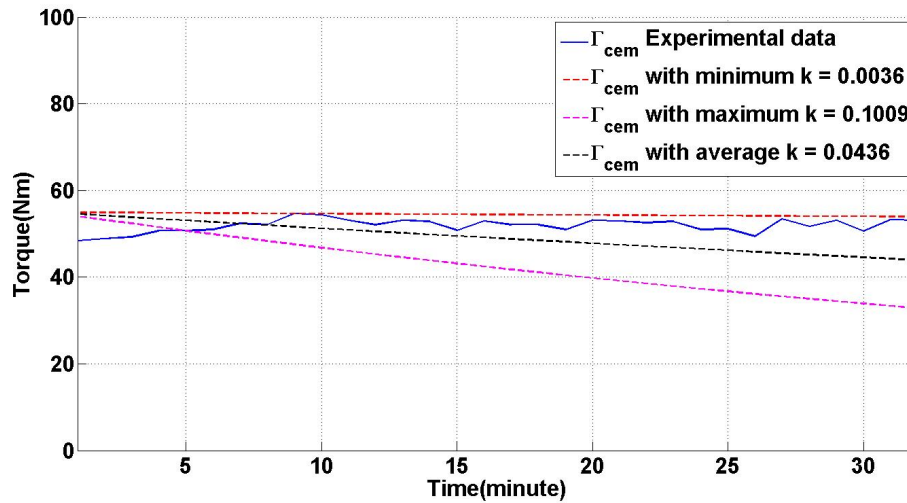


Figure 5.21: Γ_{cem} evaluation for flexion phase in the subject 7

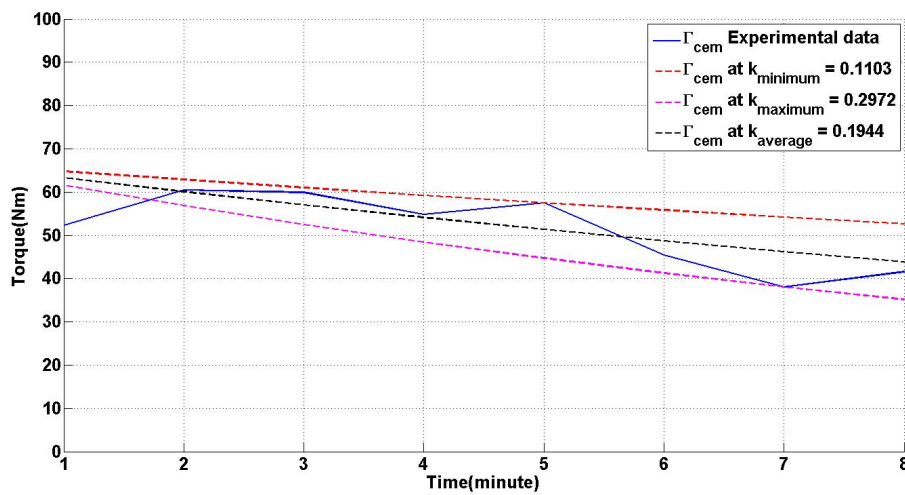


Figure 5.22: Γ_{cem} evaluation for extension phase in the subject 8

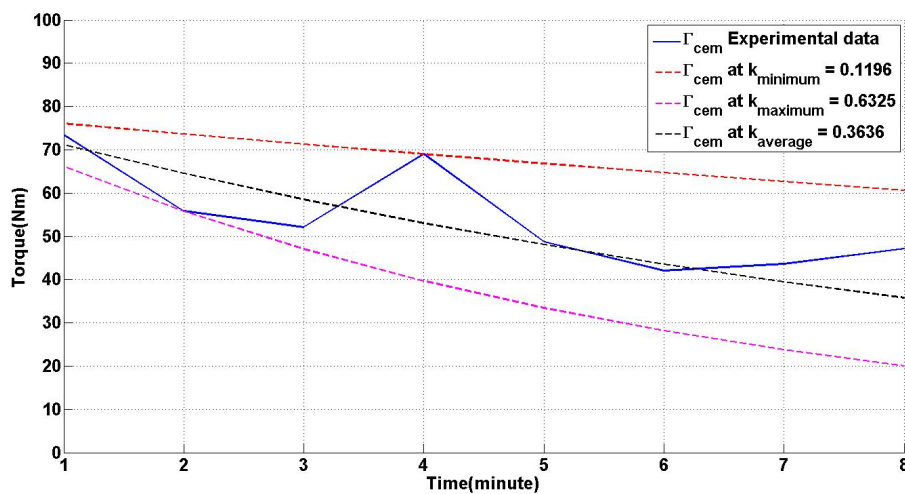
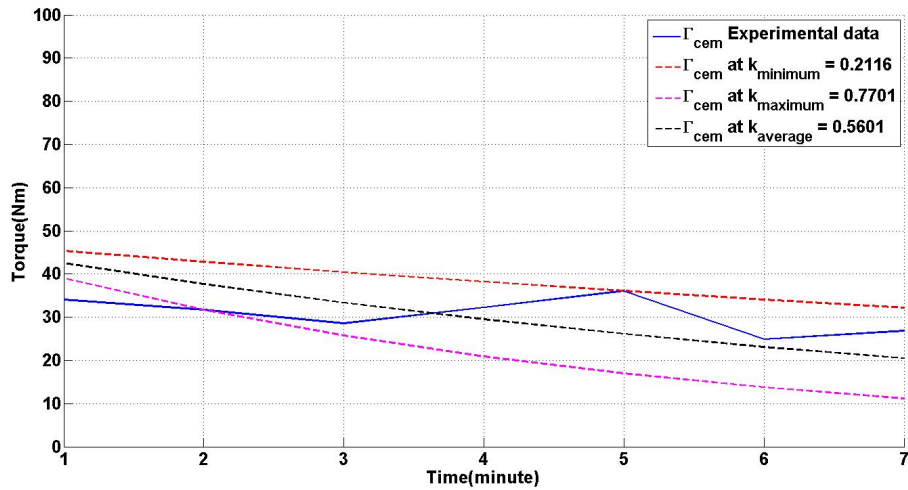
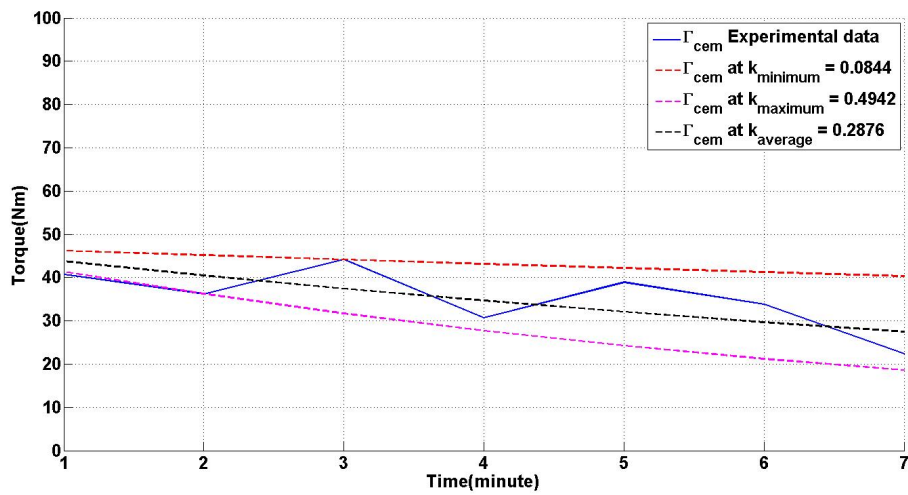
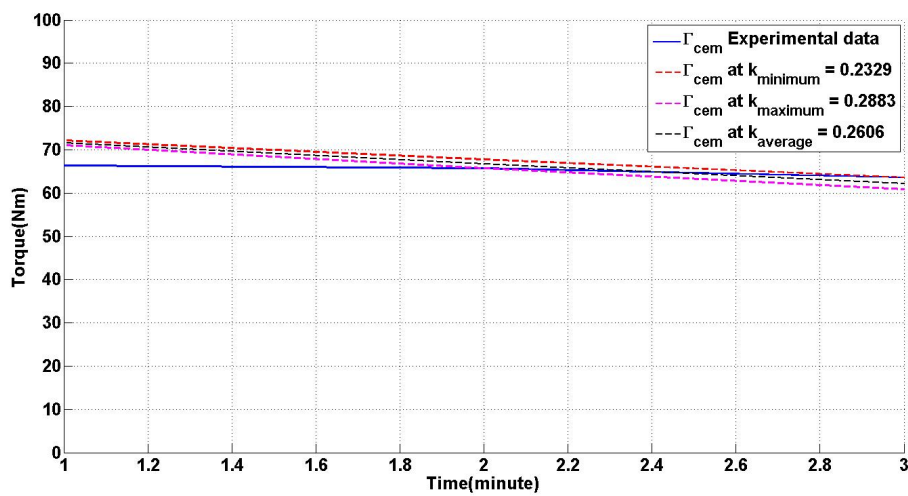


Figure 5.23: Γ_{cem} evaluation for flexion phase in the subject 8

Figure 5.24: Γ_{cem} evaluation for extension phase in the subject 9Figure 5.25: Γ_{cem} evaluation for flexion phase in the subject 9Figure 5.26: Γ_{cem} evaluation for extension phase in the subject 10

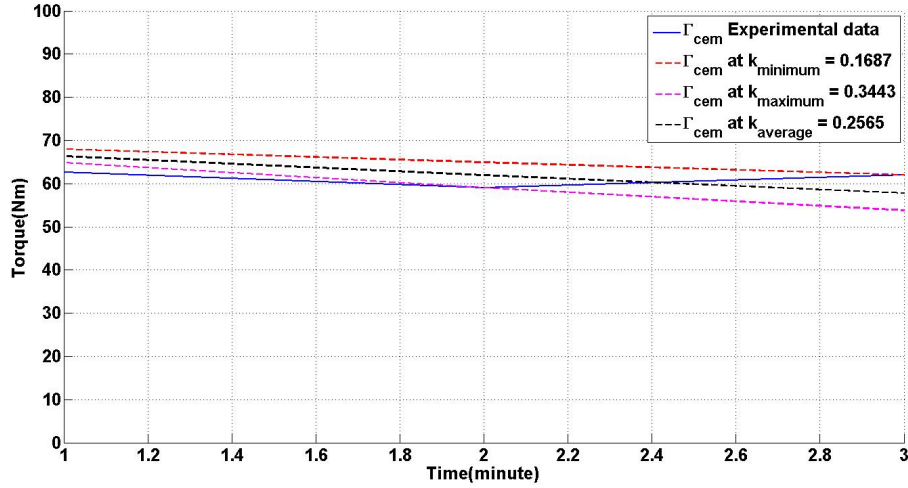


Figure 5.27: Γ_{cem} evaluation for flexion phase in the subject 10

5.4 Experimental Validation of Recovery Model

The recovery model is described in chapter 3 with equation 3.33. The model can be simplified to equation 5.3 to find the value of recovery factor R for each subject. It is suppose that recovery rate ‘R’ keeps model constant for certain period of time.

The individual recovery rate (R) was proposed in the recovery model to characterize individual differences between different subjects. The recovery rates demonstrate the speed in which the subjects return to their original maximum muscle strength. In this study, individual recovery rate of each subject was determined mathematically after fitting the measured recovered strengths to the recovery model.

Recovery capacity w.r.t., time is proportional to fatigue part ($MVC \Gamma_{cem}(t)$). After finding the value of R we will use equation 3.34 to find theoretical values of $\Gamma_{cem}(t)$.

$$R = \frac{\log \left(\frac{MVC - \Gamma_{cem}}{MVC - \Gamma_{cem}(t)} \right)}{t} \quad (5.3)$$

During Experiment we have measures MVC three times. First after one minute, second after five minutes and third after fifteen minutes of completing the fatigue protocol. These values are the experimental data to validate recovery model. During the rest the subject was sitting normally and not holding the handle. MVC measurement was done once between each rest period. During the MVC measurement there is a possibility that subject can have fatigue. The recovery model is valid for both static and dynamic conditions because during recovery subject do not need to perform any kind of motion. We used this data in equation 5.3 to calculate different values of R for flexion’s and extension’ muscles recovery.

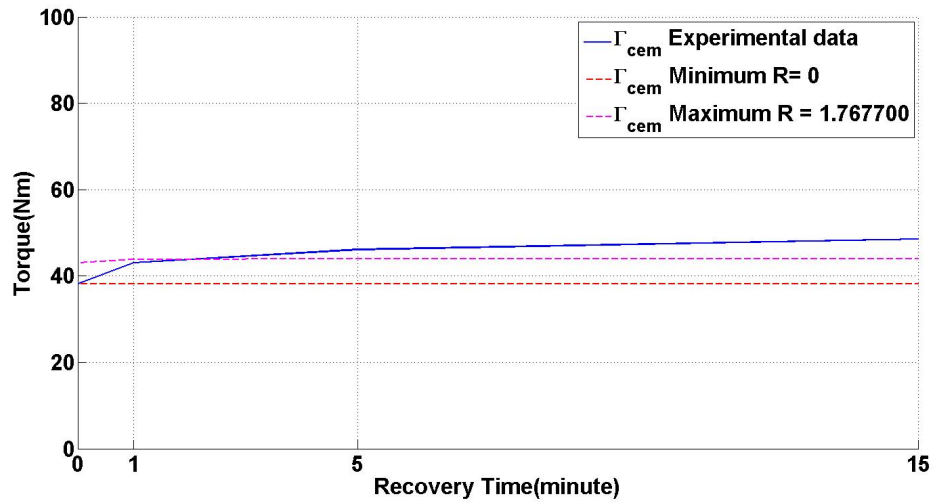


Figure 5.28: Γ_{cem} evaluation for recovery in extension phase in the subject 1

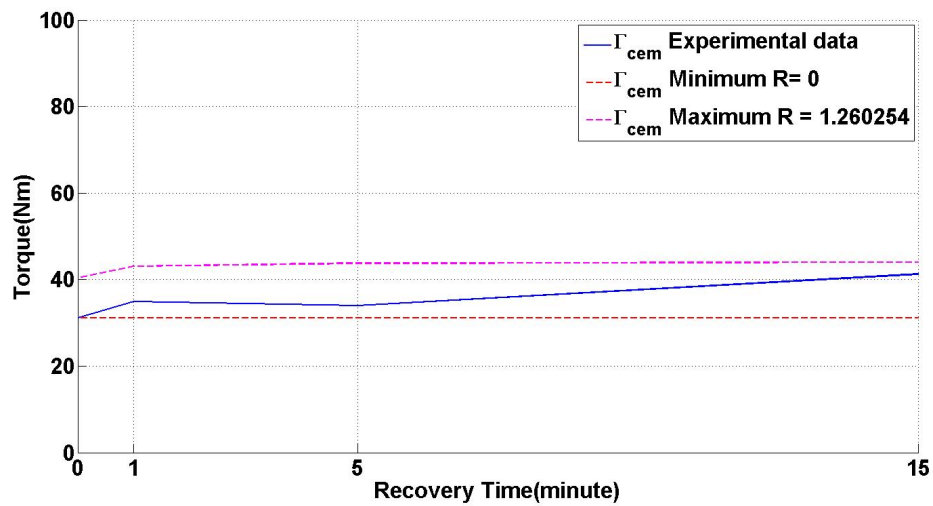


Figure 5.29: Γ_{cem} evaluation for recovery in flexion phase in the subject 1

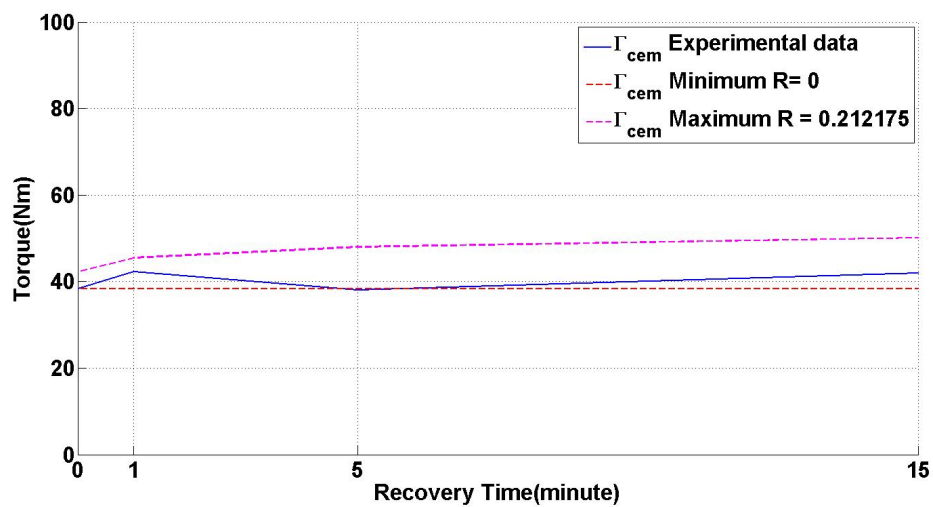


Figure 5.30: Γ_{cem} evaluation for recovery in extension phase in the subject 2

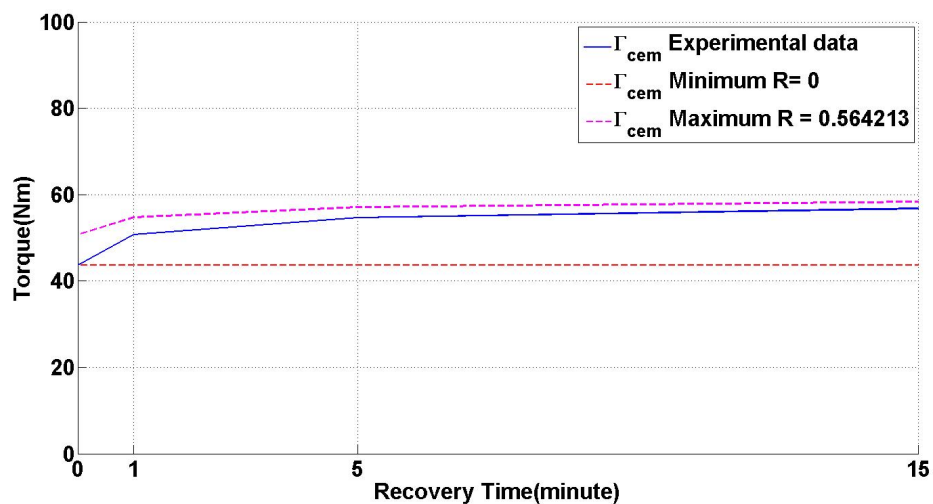


Figure 5.31: Γ_{cem} evaluation for recovery in flexion phase in the subject 2

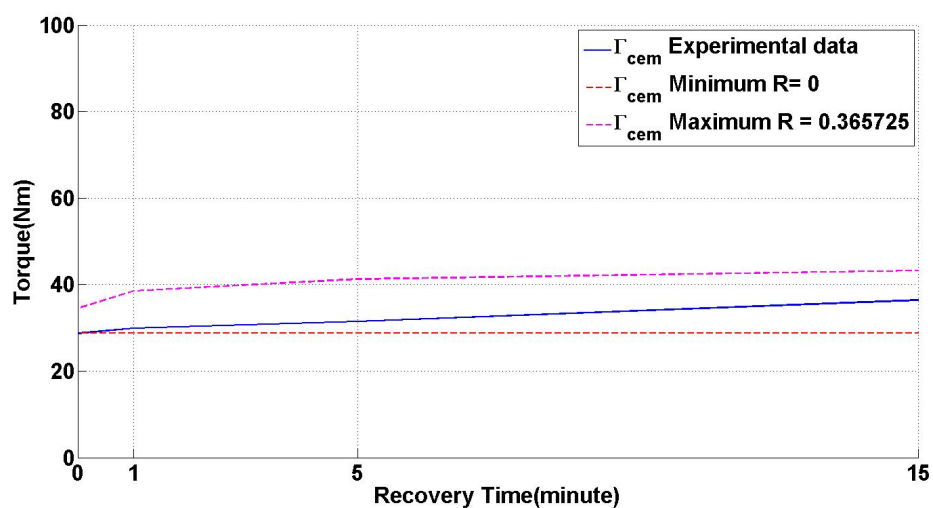


Figure 5.32: Γ_{cem} evaluation for recovery in extension phase in the subject 3

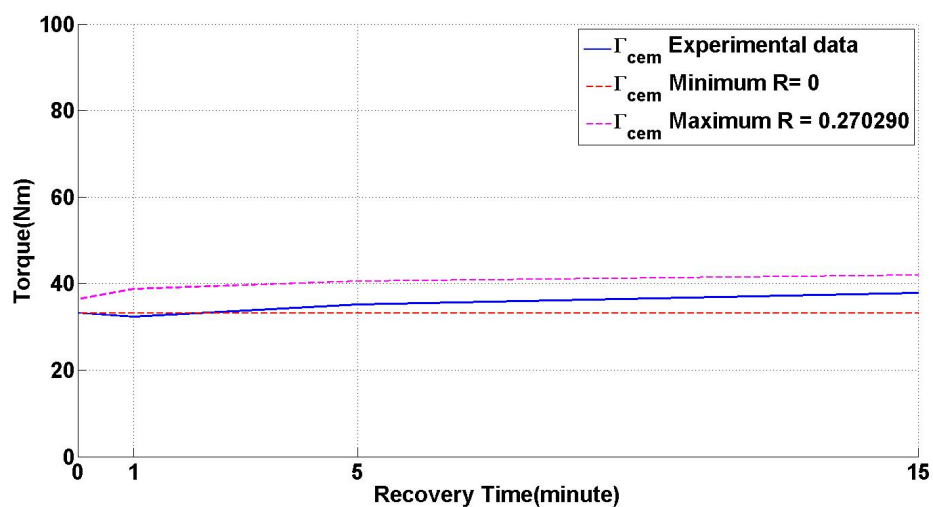


Figure 5.33: Γ_{cem} evaluation for recovery in flexion phase in the subject 3

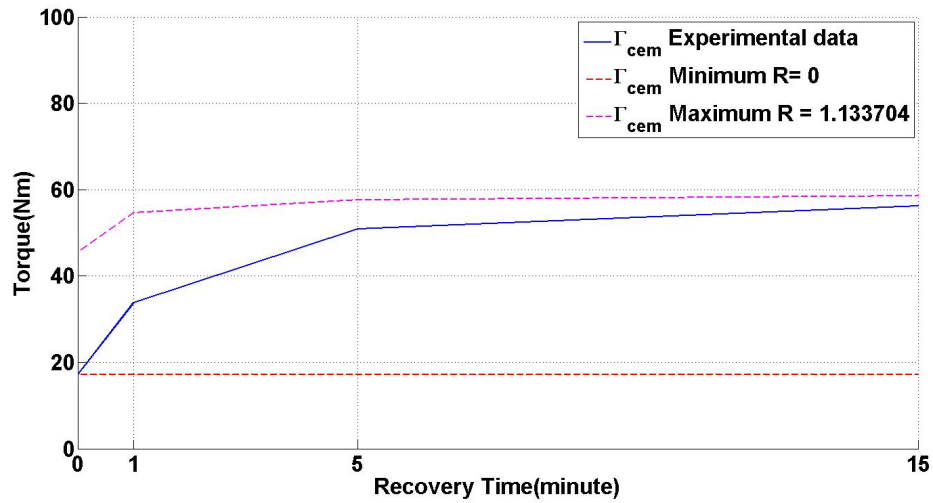


Figure 5.34: Γ_{cem} evaluation for recovery in extension phase in the subject 4

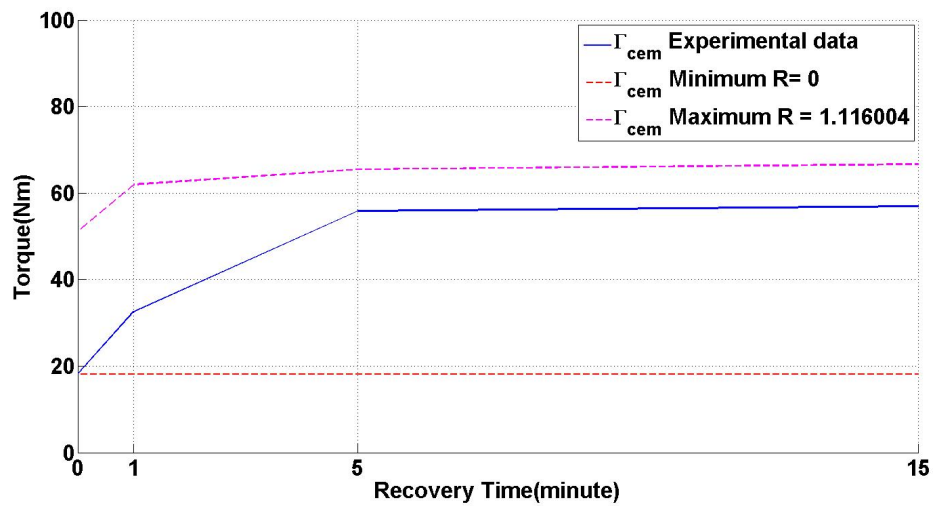


Figure 5.35: Γ_{cem} evaluation for recovery in flexion phase in the subject 4

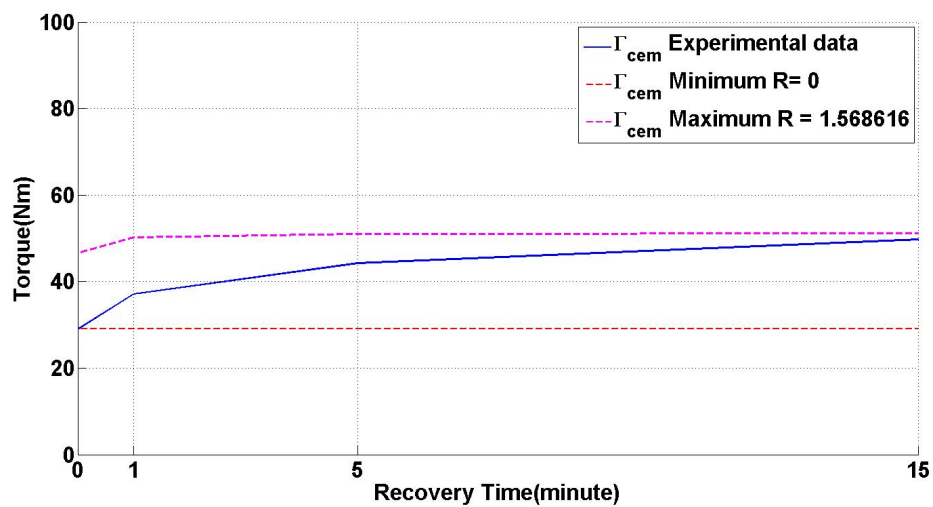


Figure 5.36: Γ_{cem} evaluation for recovery in extension phase in the subject 5

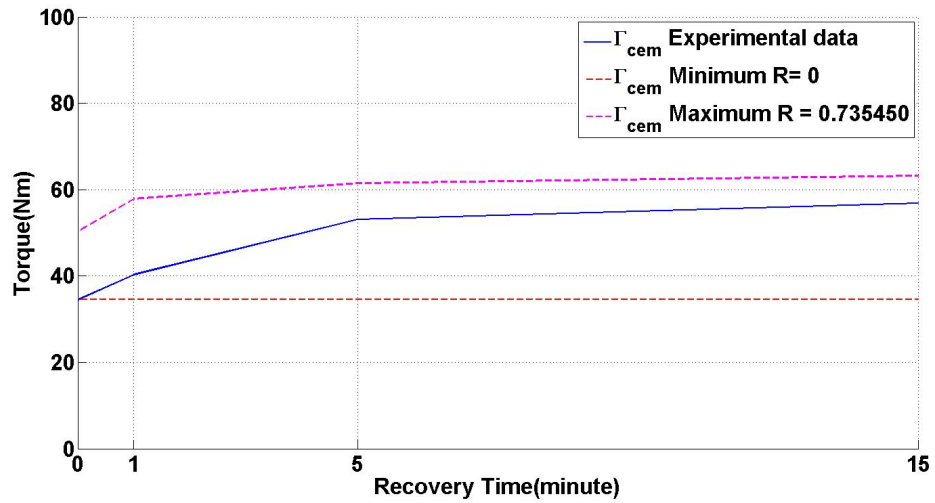


Figure 5.37: Γ_{cem} evaluation for recovery in flexion phase in the subject 5

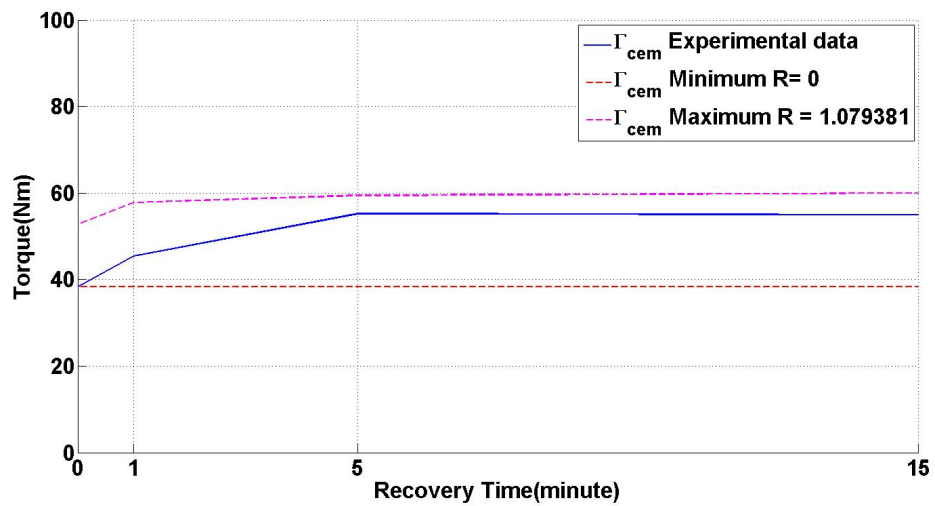


Figure 5.38: Γ_{cem} evaluation for recovery in extension phase in the subject 6

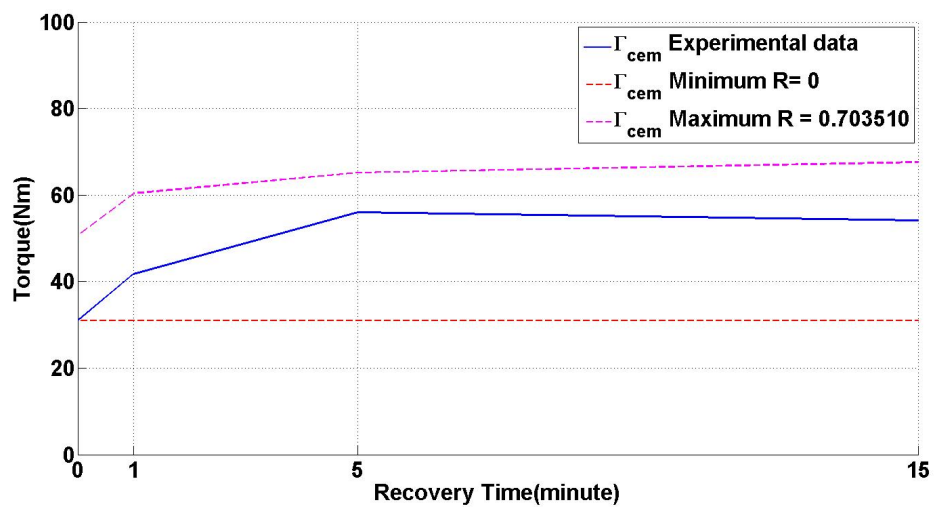


Figure 5.39: Γ_{cem} evaluation for recovery in flexion phase in the subject 6

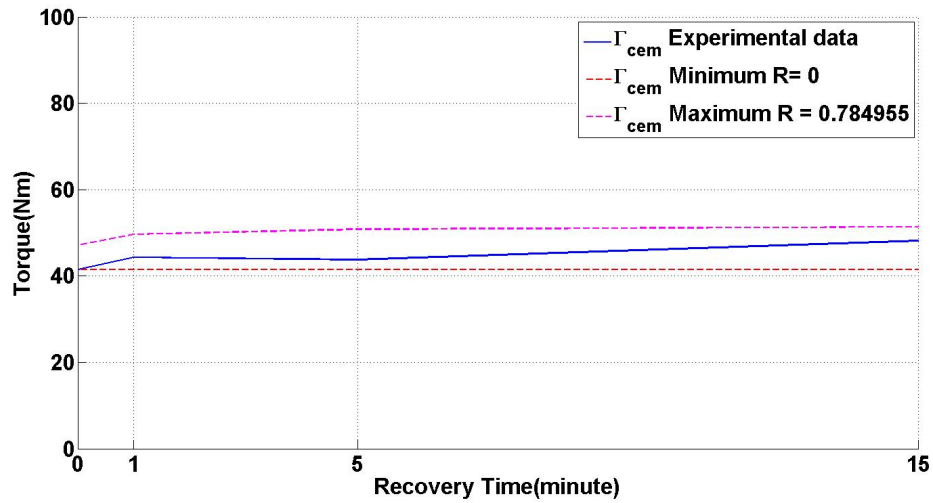


Figure 5.40: Γ_{cem} evaluation for recovery in extension phase in the subject 7

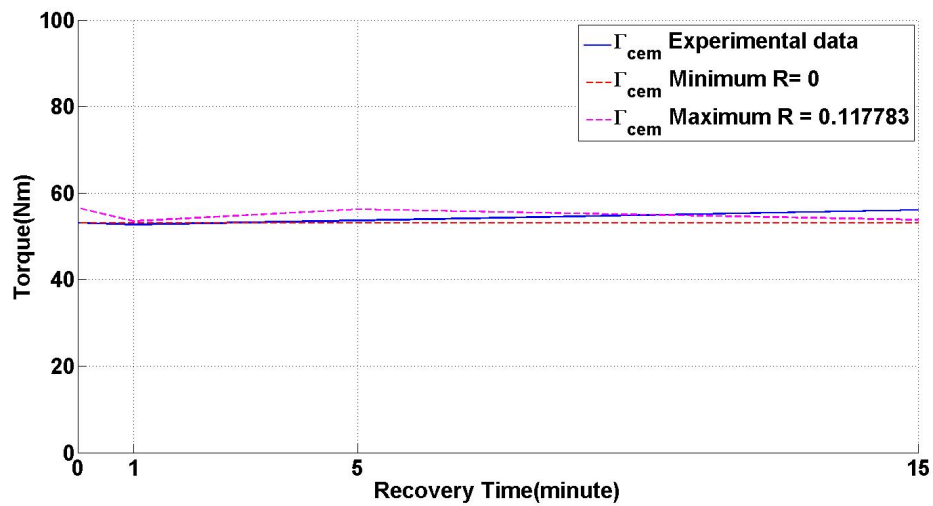


Figure 5.41: Γ_{cem} evaluation for recovery in flexion phase in the subject 7

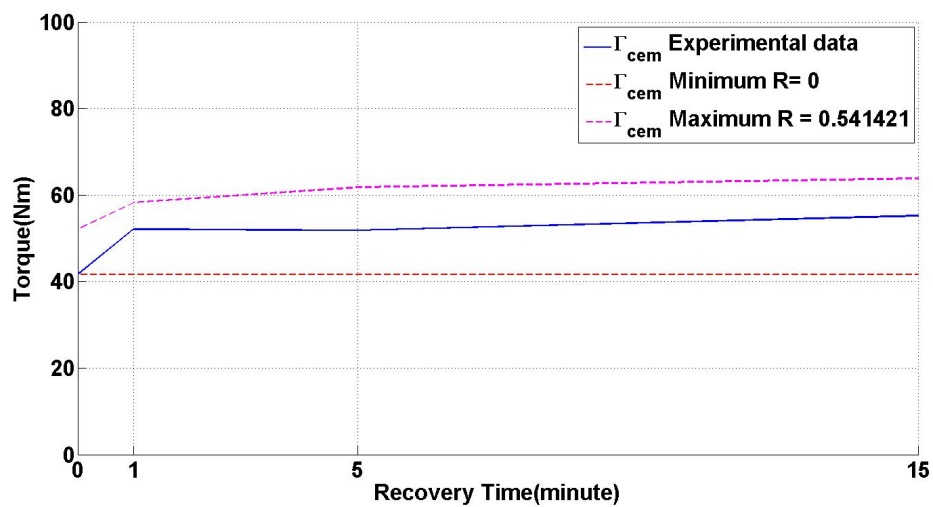


Figure 5.42: Γ_{cem} evaluation for recovery in extension phase in the subject 8

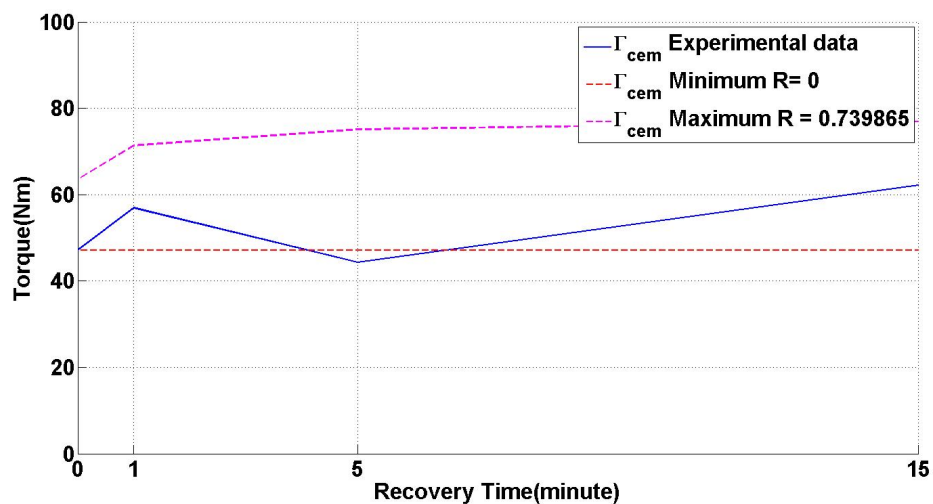


Figure 5.43: Γ_{cem} evaluation for recovery in flexion phase in the subject 8

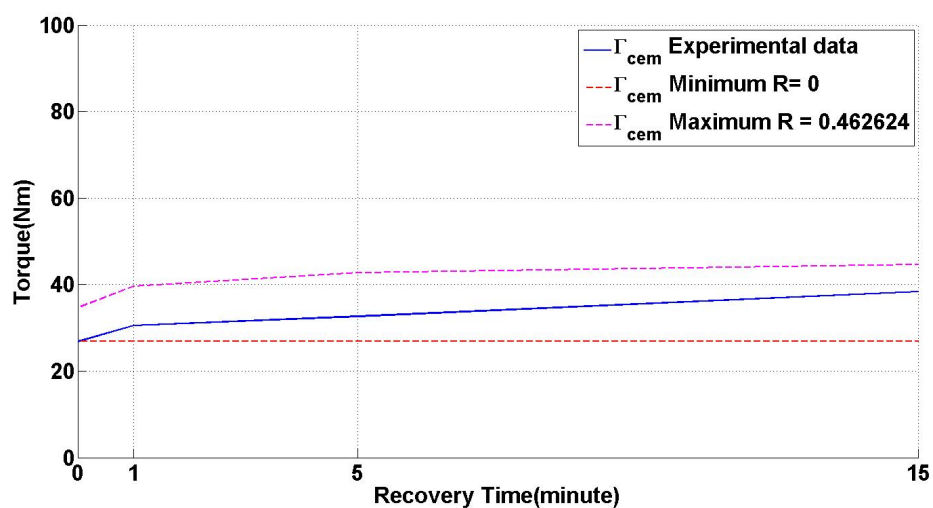


Figure 5.44: Γ_{cem} evaluation for recovery in extension phase in the subject 9

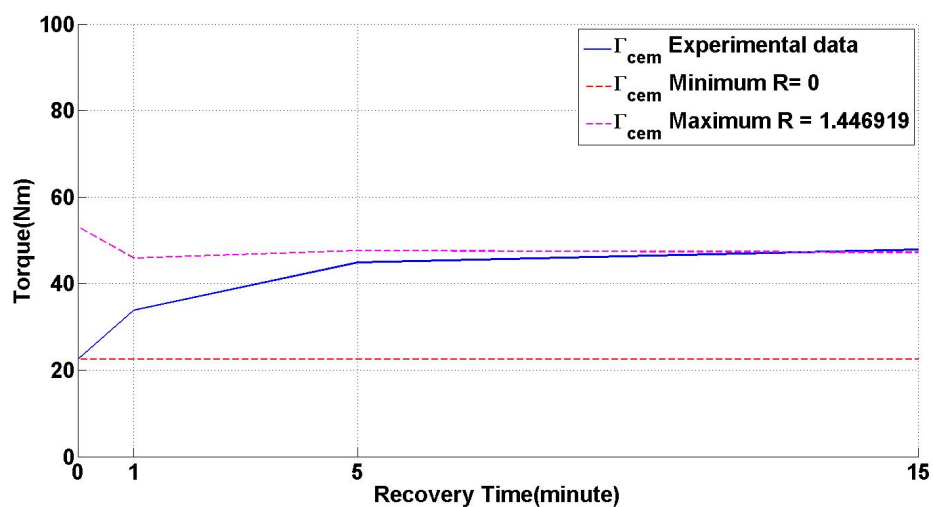


Figure 5.45: Γ_{cem} evaluation for recovery in flexion phase in the subject 9

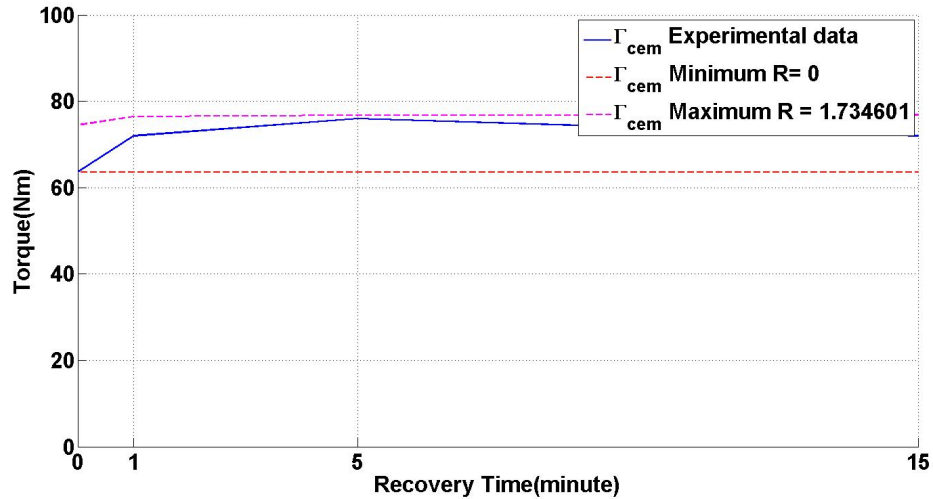


Figure 5.46: Γ_{cem} evaluation for recovery in extension phase in the subject 10

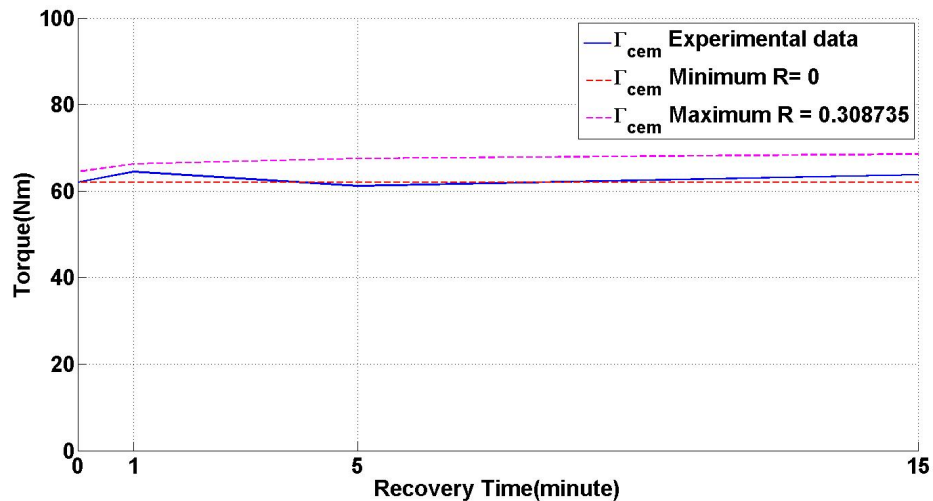


Figure 5.47: Γ_{cem} evaluation for recovery in flexion phase in the subject 10

Theoretically R cannot be negative that's why we take $R_{minimum}$ as 0 and $R_{maximum}$ is calculated value from the equation 5.3. The calculated value of R are used to calculate theoretical values of $\Gamma_{cem}(t)$. Then the theoretical values of $\Gamma_{cem}(t)$ and experimental values of MVC for recovery are plotted on the same graph for flexion and extension separately. We can see that for most of the cases the the experimental values are in the range of theoretically calculated values. Hence our recovery model is well fitted for this posture.

5.4.1 Recovery Analysis

Recovery model validated through experimental data for all subject have close validation with the theoretical data because in most of the cases the experiment values are within the limits of theoretical values. The drop in Γ_{cem} during recovery gives negative value of R because during experiment sometimes there is no recovery in strength of the

subject and that is why the value of Γ_{cem} decreases. The drop in recovery is also possible because of the fatigue occur from MVC measurement after rest. During recovery it is expected that the subject will recover its strength and hence if we measure Γ_{cem} after that it should increase. Hence theoretically it can not be negative, so we take 0 as the minimum value. The drop in recovery MVC for some subjects shows the intermittent behavior of human muscles to get recover. In some cases we can see the recovery is above 100% but in some cases it is below 100%. It is because subject didnt perform Maximum exertion in starting of the test when we measure the maximum MVC.

5.4.2 Limitations

There are several limitations to the validation of the recovery model. These limitations can be further classified into theoretical limitations and experimental limitations. The recovery model is relatively simple, and only one parameter (R) is used to represent the average recovery property of both muscle fibers. Muscle fibers can be roughly classified into two types: type I and type II. Each muscle fiber has its specific fatigue resistance and recovery attributes. However, the differences between the two types of muscle fibers were not considered in the development of the model. This assumption might cause certain shortcomings for the application of this recovery model. In the experimental design, an exponential relationship was found in the different empirical models from the literature.

It is not easy to obtain the original data from those studies to validate the model with the original data. In addition, only the hand grip strength and shoulder joint strength were selected and validated within a limited range of relative forces. It is necessary to investigate the applicability of the recovery model to other muscle groups and other relative force levels. The recovered strength measurement is still difficult under the protocol in this study. A better approach should be designed appropriately to measure the strength without interrupting the recovery process.

5.4.3 Conclusions

A theoretical model was proposed and validated in this study. The fatigue level and muscle strength were incorporated in this model to reveal the macro appearance of the overall behavior of micro motor units. The recovery rate (R) was used in this model to account for the individual fatigue attributes. Through theoretical and experimental validation, it was found that this recovery model was able to interpret the empirical exponential recovery relationship. Substantial differences between the calculated recovery rates were found as well. These differences imply that recovery rate has both individual dependency and muscle group dependency.

5.5 Position plot comparison with old model

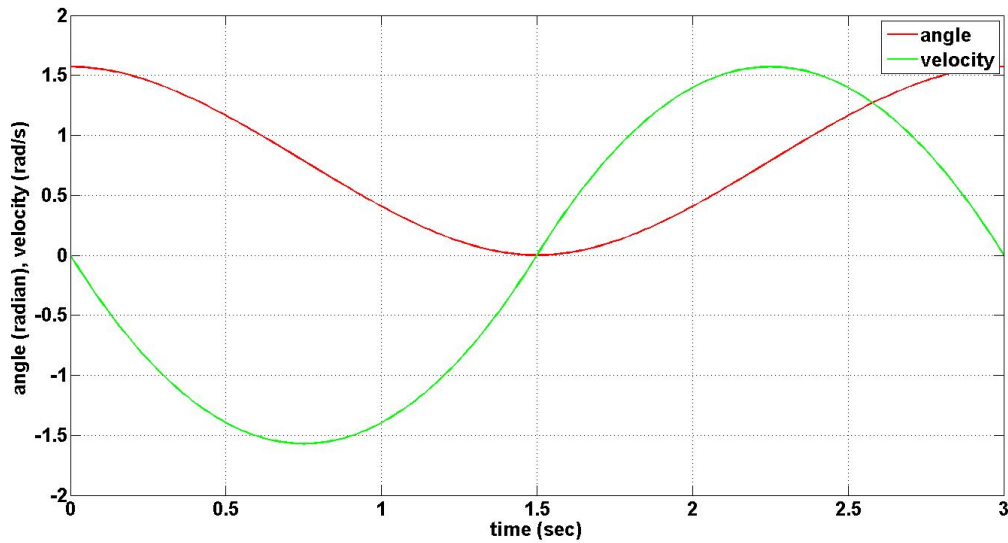


Figure 5.48: Predictive velocity and position profile for dynamic cycles in R.Ma's model

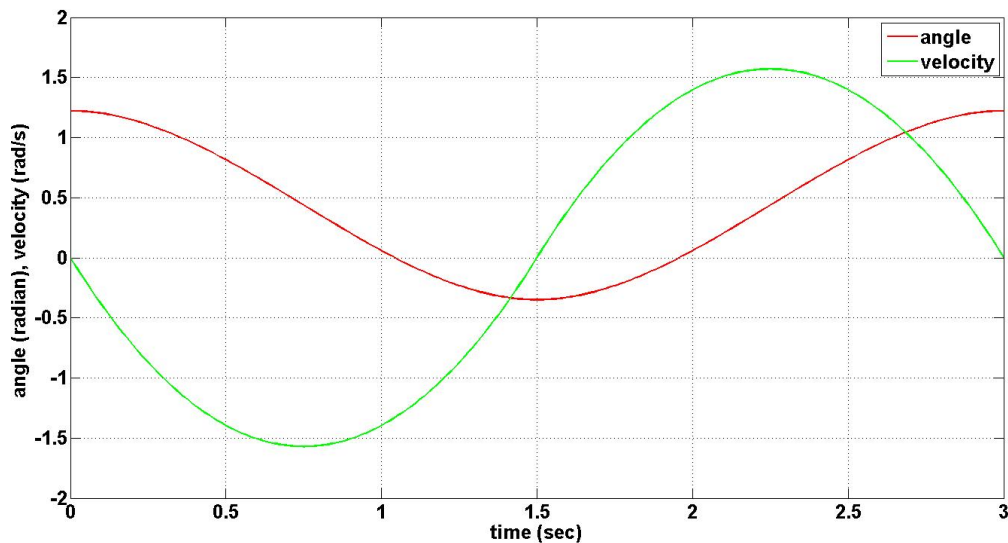


Figure 5.49: Predictive velocity and position profile for dynamic cycles in Experiment

In Ruina's model arm motion is in between 0 to 90 degree position. The predictive velocity and position behavior is shown in figure 5.48 In the current experiment, motion is in between -20 to 50 degrees. the velocity and position predictive profile is shown in figure 5.49 The comparison shows almost the same behavior of velocity and position. The comparison is on the basis of the model described in section 3.2.3. Position is shown in radians and velocity in radians/second.

5.6 DMET Analysis

5.6.1 Discussion

The dynamic maximum endurance time has been calculated for each subject according to their respective fatigue rate values. The fatigue rate for flexion and extension phases are different in each subject that is why we have selected the maximum average value of k from both the phases because more the fatigue rate we choose for work design with our model more safer will the endurance time for the subject. In DMET calculations, ' d ' represents the dynamic factor ranging between 0.1 to 1. Larger the value of d more static will be the model and smaller value of d represent more dynamic model.

The comparison between the DMET calculated for the same subject by R.Ma's model and MET calculated by L.Ma's model has been done. The DMET comparison is shown in table 5.4. The DMET is calculated using the equation 3.24 for proposed model. MET calculation for L.Ma is done by the same equation with $n = 1$ and $d = 1$ because this model is for static conditions and without co-contraction. The DMET calculation for R.Ma is also done by the same equation with $n = 1$ and $d = 0.5$ because there is no co-contraction included and dynamic factor is for medium dynamic motion. For Seth's model, the DMET is calculated with the parameters, $n = 1.38$ and $d = 0.5$. The percentage difference between the DMET calculated from Seth's model and experiment test duration is also presented in table 5.4. The DMET is calculated for each subject on the basis of their maximum fatigue rate ' k ' so that the DMET calculated can be safer to subjects. According to fatigue experiment protocol for each subject load was 20% of MVC. The values of load for each subject corresponding to their maximum MVC values are also presented in table 5.4.

Subject Number	Test Duration (Minutes)	MET-L.Ma (Minutes)	DMET-R.Ma (Minutes)	DMET-Seth (Minutes)	% difference	K Value	Load (N.m)	MVC (N.m)
1	17	11.2	22.4	17.8	4.5	0.3043	8.8	44.02
2	11	9.69	19.38	14.35	23.3	0.3606	11.6	58.58
3	6	5.73	11.46	8.5	29.4	0.6096	9.5	47.5
4	5	3.46	6.920	5.12	2.3	1.08	11.8	59.01
5	8	6.59	13.18	9.76	18.03	0.53	10.2	51.1
6	21	14.5	29.12	21.57	2.6	0.25	12.06	60.32
7	32	24.96	49.92	36.98	13.43	0.14	10.36	51.8
8	8	5.8	11.64	8.62	7.19	0.61	13.34	66.7
9	7	4.9	9.98	7.4	5.4	0.7	9.5	47.9
10	3	10.2	20.5	15.2	80.2	0.34	15.36	76.82

Table 5.4: Maximum Endurance Time Comparison

The DMET is also predicted for Seth's model (deep's model) keeping the value of co-contraction factor $n = 1.38$ at different values of d . The value of k is also kept constant is calculated from the average of the values of k given for all the subjects. The average value of k is 0.41. The value is just to predict the DMET behavior at average value of k . The DMET predicted at $d = 0.1, 0.5, 0.9, 1$ are shown in figures 5.50, 5.51, 5.52 and 5.53 respectively. These figures represent the maximum endurance time for subject with respect to the value of load f_{MVC} , which is the ratio of external load to the maximum capacity or MVC of a subject. From figures 5.50 to 5.53, we observe that larger the value of d better will be the maximum endurance time that is why at $d = 0.1$ the maximum endurance time with respect to each value of f_{MVC} is larger in comparison to the maximum endurance time at other values of d . At $d = 1$ all the models acts like static model. That is why in figure 5.53, maximum endurance time curve for R.Ma's and L.Ma's model are overlapping on the other hand because of $n = 1.38$ only in Seth's model the curve is away from the both other models.

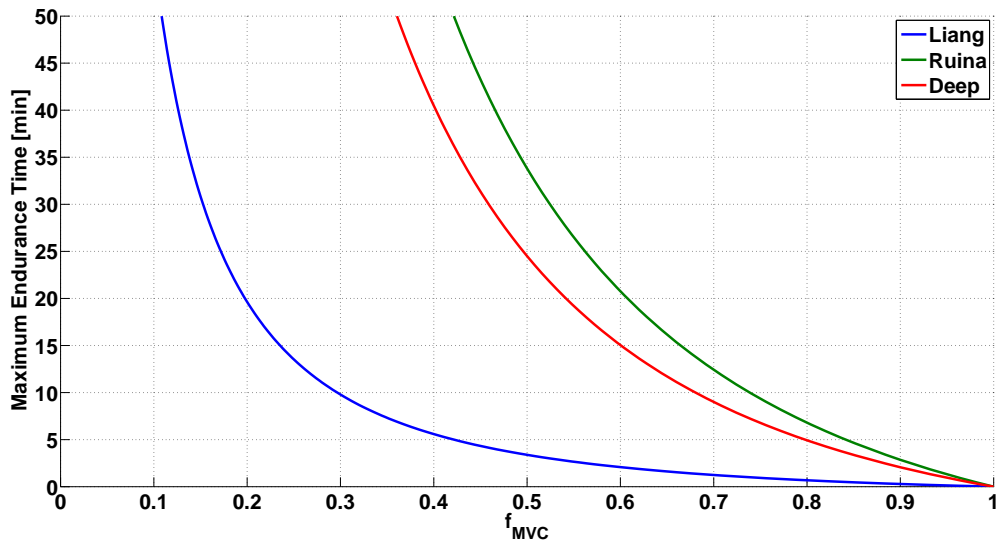


Figure 5.50: DMET prediction at $d = 0.1, k = 0.41$

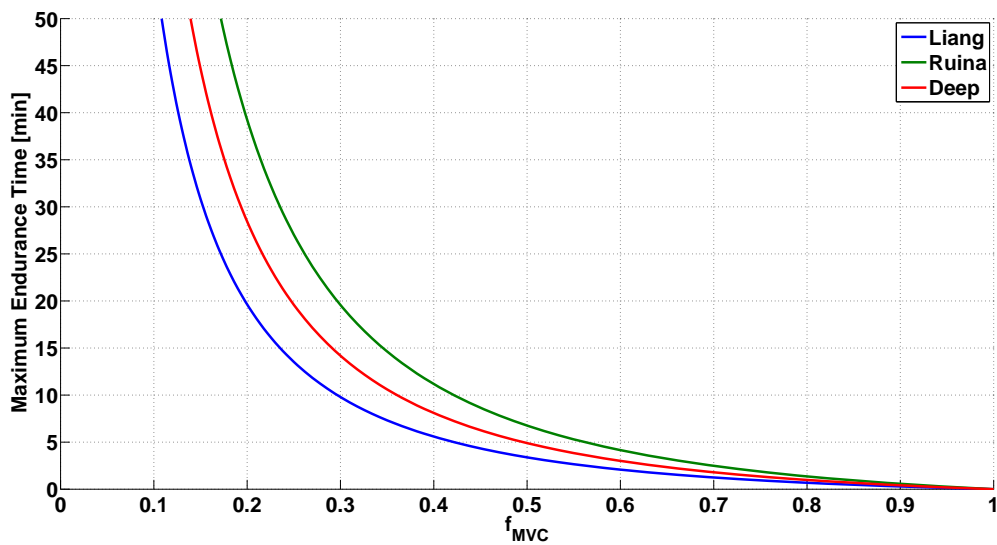


Figure 5.51: DMET prediction at $d = 0.5, k = 0.41$

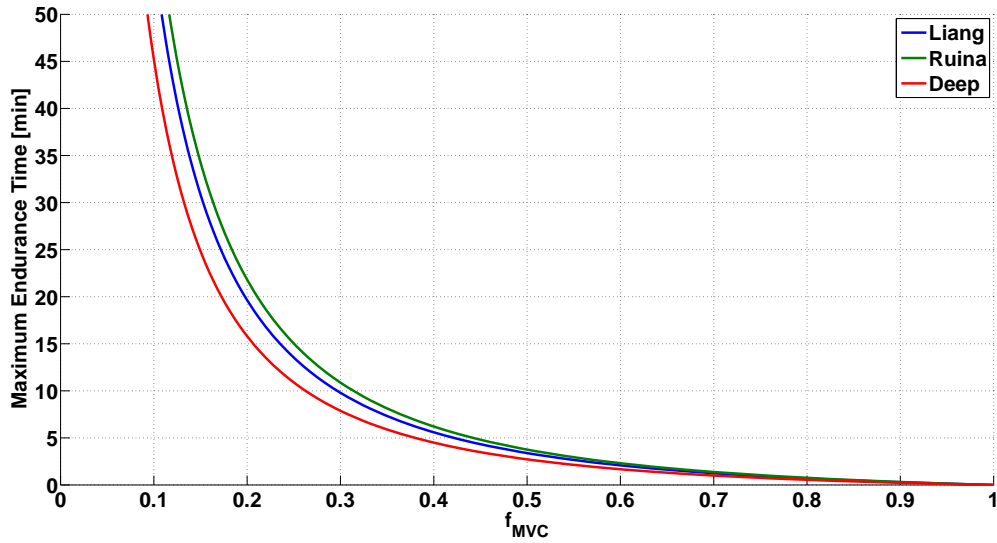


Figure 5.52: DMET prediction at $d = 0.9$, $k = 0.41$

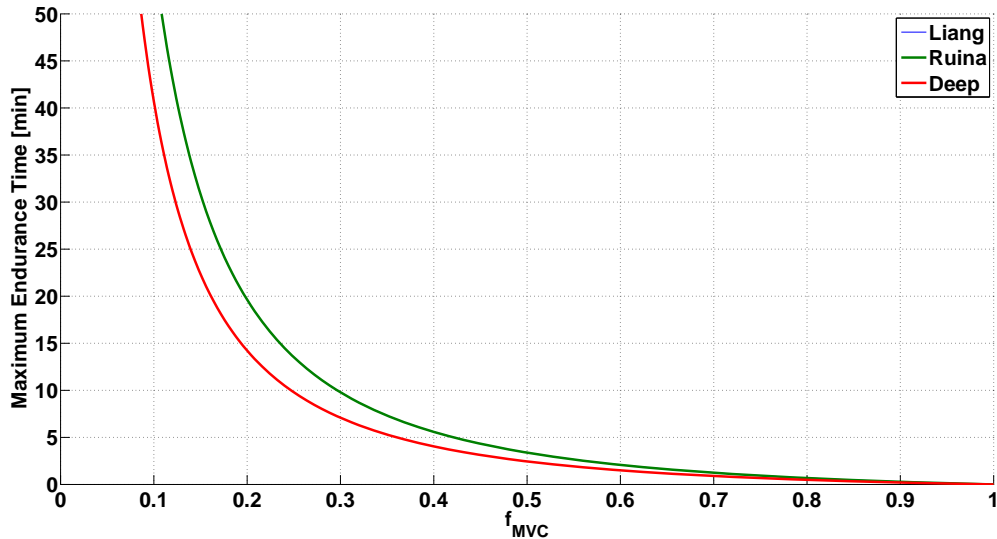


Figure 5.53: DMET prediction at $d = 1$, $k = 0.41$

5.6.2 Analysis

From the DMET calculation for each subject, we can see that the DMET calculated for each subject is more than the experimented value. It is because DMET is the maximum limit of any human and we did the experiment for each subject up to comfortable exhaustion level. Comfortable exhaustion level means the level at which the subjects want to stop the experiment because of fatigue. It may be possible in this cases that the subject do not reach their maximum limit but they stop the test, for example, subject number 10, we can see in table 5.4 that he stopped the test after 3 minutes means after completing 60 cycles but the maximum endurance time is much larger than the experiment duration. For subject 10 the percentage difference between the DMET calculated

by Seth's model and experiment duration is 80.2%, which is much higher in comparison to other subjects. L.Ma's model is static model, that is why maximum endurance time calculated is less than the experimented value. R.Ma's model gives more DMET for each subject in comparison to Seth's DMET model which is much near to the experimental values. The DMET calculated by Seth's model is 25.9% less than the DMET calculated by R.Ma's model. The DMET calculated for Seth's model is less because we have introduced the co-contraction factor into the model. This gives more approximate value to the experimental data. So the work design according to this model will be more safer in comparison to R.Ma's Model, L.Ma's model is for static posture hence, it may not be real to compare for dynamic situations.

Chapter 6

Conclusions and Future perspectives

6.1 Conclusions

The main focus of this thesis is to deal with the dynamic posture and motion analysis of the human body. The thesis contributes in defining the human muscle fatigue status and its mathematical description, then develop a dynamic muscle fatigue and recovery model for ergonomic applications, posture analysis and posture prediction with consideration of fatigue effect.

In Chapter 1, we present a general concept and motivation for the project with a summary of all the chapters in the thesis. In Chapter 2, we discuss about the behavior of a muscle, its activity and related problems. The importance of muscle fatigue in MSD and the limitations in the study of muscle fatigue and recovery analysis like proper quantification of fatigue, co-contraction analysis in other models, lack of recovery parameters and lack of more realistic muscle fatigue model lead us to the study of muscle fatigue models. From the literature survey of various muscle fatigue models we came to know that most of the models have different parameters or factors to represent muscle fatigue. Some models just gave the theoretical description of muscle fatigue activity without validating it. The limitations in these models are shortcoming in the applicability of the models with the ergonomics tool, too many variables include studying muscle fatigue and no consideration of co-contraction give us a direction to develop a new dynamic muscle fatigue model with inclusion of co-contraction factor.

In chapter 3, first we have introduced the base of a new dynamic muscle fatigue model with its hypothesis. The details of the co-contraction factor are given in this chapter. We define the co-contraction factor as a representation of common muscle activity between two groups of muscle. Then we defined a proposed dynamic muscle fatigue model. The model is then compared with other muscle fatigue models and validated theoretically. After the analysis and validation of the dynamic fatigue model, we find that muscle fatigue follows the exponential curve for each subject and inclusion of co-contraction factor reduces the strength prediction for the model in comparison to other models. The recovery model is also introduced in this chapter to understand the effect of

recovery. The recovery model is then compared theoretically with other recovery models. In this comparison we find that recovery model also follows a negative exponential curve for the muscle strength recovery like other models. Dynamic maximum endurance time has also been described in this chapter with a theoretical comparison to other models.

In chapter 4, an experiment for the validation of muscle fatigue and recovery model is described. The experiment protocol has been defined and then setup for experiment is done accordingly. The experiment has the main objective to study muscle fatigue and recovery model by predicting changes in the muscle strength with the factors like fatigue rate, recovery rate, co-contraction index, and dynamic variability factor. The experiments were performed on 10 sportive male subjects according to the experimental protocol. All the data collected from the experiment are then processed and analyzed for further study. After the analysis we found that most of the hypothesis and assumptions made in the previous chapters are supported by the experiment outcome. The behavior and pattern of muscle activity, co-contraction between muscle and muscle strength reduction are approximately the same as expected. The experiment gives the direct measurement of muscle strength of the arm of a human subject with dynamic push-pull motion in a vertical plane. It also measures the changes in strength during recovery phase. These measurements helped us in validating muscle fatigue and recovery model in chapter 5. All the analysis related to co-contraction and validation of muscle fatigue and recovery model are done in chapter 5. The results show that co-contraction index reduces as the muscle fatigue increases. The reduction in co-contraction index does not mean the reduction in co-contraction itself. The co-contraction factor calculated after analysis is normalized to one value that is 1.38. For the validation of dynamic muscle fatigue model, we use muscle strength measuring during the fatigue protocol for each subject. The experimentally calculated values of Γ_{cem} lie well within the range of theoretically calculated values which shows the prediction of the muscle fatigue model is good. The same data is used to validate the recovery model. In both validations, we found that muscle fatigue and recovery have exponential variation in experimentally measured values. The DMET model validation shows that there is a reduction of endurance time in comparison with R.Ma model and in some cases with L.Ma model. The DMET comparison shows that there is 25.9% reduction in maximum endurance time in Seth's model as compared to R.Ma model. When we compare the DMET at dynamic variation factor, $d = 0.5$, it shows that the time taken by different subject to complete the fatigue protocols during the experiment are near to the DMET values calculated for maximum values of k and more closer to the Seth's model.

In the development of the fatigue model, 5 main parameters were used unlike R.Ma's model which uses 4 parameters and L.Ma's 3 parameters. The parameters are fatigue rate for agonist and antagonist muscle groups, dynamic variability factor, co-contraction factor and Γ_{cem} calculated theoretical. The new parameters have been created to indicate the fatigue resistance and recovery rate of an individual. Although only 10 subjects have been tested experimentally for the fatigue resistance in arm muscle groups, it can be validated for more subjects if needed and it is believed that all the parameters are valid to quantify fatigue and recovery properties in dynamic conditions. If the distribution was achievable for some other postures and experimental conditions, these parameters could be very important for evaluating the fatigue and recovery rate, of a certain population.

6.2 Future Perspectives

In our work we take into consideration only the physical human factors. The external environmental factors and other internal factors can also be included in the study for better understanding.

In our fatigue and recovery model, only the human arm has been tested in the experiments in our current research. In the future research, more experiments should be carefully designed to test the availability of our model in other muscle groups, such as back/hip, neck etc. The recovery model is experimentally and theoretically compared with other existing models in the literature. More effort should be made to validate recovery model for other work postures. The combination of the fatigue and recovery model can be applied in the measurement for work-rest allowance. For the fatigue rate and recovery rate normalization for large population more experiments are needed. Till now our model is validated for one planar motion and a simple task. There is a need of validation of our model for different planar motions and complex three dimensional real life tasks. For this some changes in the model may be required, like for more complex motions more muscles will come into effect and more co-contraction factors have to be introduced for different type of motions, similarly to characterize the muscle fatigue and recovery model for the whole body there would be a need of various other fatigue and physical factors.

In this thesis, we have a case study with dynamic conditions. In dynamic operations, with the change in posture of arms or other parts of the body, the fatigue and recovery process might be different from the study we did in this thesis. It could be interesting to find out the applicability of our model in different postures.

In our study, we used 5 parameters to study muscle fatigue. $k_{agonist}$ and $k_{antagonist}$ have different behavior in flexion and extension phase. In extension phases, we observed almost no activity from other muscle groups. So some work needed to observe the viability of the reducing the fatigue rate factor to one in place to two for different group of muscles. Efforts have to be made to experimentally establish generalized the fatigue rate ' k ' for different group of people according to their activity and classes like sports, industrial worker, students, old people, working woman etc.

We can also study the influence of anthropometric data in studies of muscle fatigue, for example, for the group of male subjects of the same height and weight group can give different values of fatigue rate and strength as compared to mixed groups.

With the increase in fatigue during experiment we observed the change in co-contraction between muscles. There is a possibility of some work in analyzing the relation between accuracy of any dynamic task and co-contraction. It is believed that change in co-contraction can affect the accuracy of the task performed.

Some work has to be done to integrate our model into ergonomic tools. There could be a need for the development of a muscle fatigue and strength database to define various working postures and fatigue characteristics. This database could be applied to

an ergonomic tool for the fatigue prediction according to the posture. The human work behavior and environmental conditions should also be taken into account in order to assess an operation completely. For prediction of these behaviors and to integrate it with fatigue model some separate study may be required and it could be a complete research topic for study.

Furthermore, there is a lot of scope of application of muscle fatigue model to the health sectors and different area of bio-mechanics research. One of our research direction would be constructing detailed musculoskeletal systems and study the muscle behavior for prosthesis and injured muscle part. This could include the study of the force reaction of muscles, tendons, and bones. It is expected that detailed study of several parameters of muscle fatigue can also help us in avoiding and analyzing the source of MSDs.

In summary, there are still lots of fields available to study the muscle fatigue model in certain aspect and conditions. The final aim of this model is that it should be able to predict the human's postures as real as possible and as quick as possible. This could be done with the devices like motion capture system, light tracker, etc. This could be helpful in reducing the various risks in motion and posture prediction and hence help in better study of the muscle fatigue and recovery characteristics.

Bibliography

- [1] Jouni Freund and Esa-Pekka Takala. A dynamic model of the forearm including fatigue. *Journal of Biomechanics*, 34:597–605, 2001.
- [2] Jing Z. Liu, Robert W. Brown, and Guang H. Yue. A dynamical model of muscle activation, fatigue and recovery. *Biophysical Journal*, 82:2344–2359, 2002.
- [3] Ting Xia and Laura A. Frey Lawa. A theoretical approach for modeling peripheral muscle fatigue and recovery. *Journal of Biomechanics*, 41:3046–3052, 2008.
- [4] Jorge M. Martins Miguel T. Silva, Andre F. Pereira. An efficient muscle fatigue model for forward and inverse dynamic analysis of human movements. *Procedia IUTAM*, 2:262–274, 2011.
- [5] Liang Ma, Damien Chablat, Fouad Bennis, and W Zhang. A new muscle fatigue and recovery model and its ergonomics application in human simulation. *Virtual and Physical Prototyping*, 5:123–137, 2008.
- [6] Ruina Ma, Damien Chablat, and Fouad Bennis. Human muscle fatigue model in dynamic motions. *Latest Advances in Robot Kinematics*, 0:349–356, 2012.
- [7] Tim Taylor. Muscular system. <https://www.innerbody.com/image/musfov.html>, October 2012.
- [8] A Patricia and Pierce. *Fatigue: Neural and Muscular Mechanisms*, volume 384. Springer Science & Business Media, 2013.
- [9] OpenStax College. Anterior and posterior views of muscles. https://commons.wikimedia.org/wiki/File:1105_Anterior_and_Posterior_Views_of_Muscles.jpg, May 2013.
- [10] Tim Taylor. Muscles of the arm and hand. HowToMedia, Inc., October 2012.
- [11] R H Edwards, DK Hill, D A Jones, and P A Merton. Fatigue of long duration in human skeletal muscle after exercise. *The Journal of Physiology*, 272(3):769–778, 1977.
- [12] B Bigland-Ritchie, f Furbush, and J J Woods. Fatigue of intermittent voluntary

- contraction: Central and peripheral factors. *Journal of Applied Physiology*, 61:421–429, 1986b.
- [13] B Bigland-Ritchie, E Cafarelli, and N K Vollestad. Fatigue of submaximal static contractions. *Acta Physiologica Scandanavica*, 128:137–148, 1986a.
- [14] J A Kent-Braun, K K McCully, and B Chance. Metabolic effects of training in human a p-rms study. *Journal of applied Physiology*, 69:1165–1170, 1990.
- [15] R G Miller, Carson P J, Moussavi R S, A Green, A Baker, M D Boska, and M W Weiner. Factors which influence alteration of phosphate and ph in exercising human skeletal muscle: measurement error, reproducibility and effects of fasting, carbohydrate loading and metabolic acidosis. *Muscle and Nerve*, 18(60-67), 1995.
- [16] Vollestad and Nina K. Measurement of human muscle fatigue. *Journal of Neuroscience Methods*, 74:219–227, 1997.
- [17] S Thorn, K Sogaard, L A C Kallenberg, L Sandsjo, G Sjogaard, H J Hermens, R Kadefors, and M Forsman. Trapezius muscle rest time during standardised computer work – a comparison of female computer users with and without self-reported neck/shoulder complaints. *Journal of Electromyography and Kinesiology*, 17(4):420–427, August 2007.
- [18] Simon N. Young. How to increase serotonin in the human brain without drugs. *Journal of Psychiatry and Neuroscience*, 32(6):394–399, 2007.
- [19] George Brooks, Thomas Fahey, and Kenneth Baldwin. *Exercise Physiology: Human Bioenergetics and Its Applications*. McGraw-Hill, 4 edition, 2004.
- [20] Asmussen E and B. Mazin. Recuperation after muscular fatigue by ”diverting activities”. *European Journal of Applied Physiology*, 38:1–7, 1978.
- [21] Jane A. Kent-Braun. Central and peripheral contributions to muscle fatigue in humans during sustained maximal effort. *European Journal of Applied Physiology and Occupational Physiology*, 20:57–63, may 1999.
- [22] Matt Middlesworth. The definition and causes of musculoskeletal disorders (msds). <http://ergo-plus.com/musculoskeletal-disorders-msd/>, May 2015.
- [23] Arthritis. Health in aging foundation, newyork. <http://www.healthinaging.org/aging-and-health-a-to-z/topic:arthritis/>, March 2012.
- [24] American Academy of Orthopaedic Surgeons. Information about common musculoskeletal diseases. <http://www.aaos.org/research/stats/patientstats.asp>, 2009.
- [25] National Institute of Arthritis, Musculoskeletal, and Skin Diseases. Handout on health: Back pain. http://www.niams.nih.gov/Health_Info/Back_Pain/

- [default.asp](#), April 2012.
- [26] Bureau of Labor Statistics. Injuries, illnesses, and fatalities-2011-2014. <http://www.bls.gov/iif/oshsum.htm>, October 29 2015.
- [27] Centers for Disease Control and Prevention. Work-related musculoskeletal disorders (wmsds) prevention. <http://www.cdc.gov/workplacehealthpromotion/evaluation/topics/disorders.html>, october 23 2013.
- [28] Unites Stated Department of Labour. Estimated costs of occupational injuries and illnesses and estimated impact on a company's profitability worksheet. https://www.osha.gov/dcsp/smallbusiness/safetypays/estimator_text.html, November 2015.
- [29] Canadian Centre for Occupational Health and Safety. Work-related musculoskeletal disorders (wmsds). <https://www.ccohs.ca/oshanswers/diseases/rmirsi.html>, January 08 2014.
- [30] Punnett L and Wegman DH. Work-related musculoskeletal disorders: the epidemiologic evidence and the debate. *Journal of Electromyography and Kinesiology*, 14(1):13–23, 2004.
- [31] World Health Organization. Preventing musculoskeletal disorders in the workplace. Protecting workers 'Health series; no. 5', Geneva 27, Switzeland, 2003. ISBN 92 4 159053 X.
- [32] Nurhayati Mohd Nur, Siti Zawiah Md Dawal, and Mahidzal Dahari. The prevalence of work related musculoskeletal disorders among workers performing industrial repetitive tasks in the automotive manufacturing companies. In *Proceedings of the 2014 International conference on inductrial engineering and operations management, Bali, Indonesia*, 2014.
- [33] Laura Punnett and David H. Wegman. Work related musculoskeletal disorders: the epidemiologic evidence and the debate. *Journal of Electromyography and Kinesiology*, 14:13–23, 2004.
- [34] Angelika Huppe, Kristin Muller, and Heiner Raspe. Is the occurrence of back pain in germany decreasing? two regional postal surveys a decade apart. *European Journal of Public Health*, 17:318–322, 2006.
- [35] RC Lawrence, CG Helmick, FC Arnett, RA Deyo, DT Felson, EH Giannini, SP Heyse, R Hirsch, MC Hochberg, GC Hunder, MH Liang, SR Pillemer, VD Steen, and F Wolfe. Estimates of the prevalence of arthritis and selected musculoskeletal disorders in the united states. *Arthritis & Rheumatism*, 41(5):778–799, 1998.
- [36] Don B Chaffin, Gunnar B J Andersson, and Bernard J Martin. *Occupational Biomechanics*. Wiley - Interscience, third edition, 1999.

- [37] Andre F. Pereira, Miguel T. Silva, Jorge M. Martins, and Mamede de Carvalho. Development of a hill-type muscle model with fatigue for the calculation of the redundant muscle forces using multibody dynamics. *The 1st Joint International Conference on Multibody System Dynamics, Lappeenranta, Finland*, 0:–, 2010.
- [38] A.V. Hill. The heat of shortening and dynamic constant of muscle. *Proceedings of the Royal Society of Biological Sciences*, 126:135–195, 1938.
- [39] Ruina Ma. *Modélisation de la fatigue musculaire dynamique et son application pour la analyse ergonomique*. PhD thesis, STIM-IRCCyN, 2012.
- [40] Ruina Ma. *Modelisation de la fatigue musculaire dynamique et son application pour l'analyse ergonomique*. PhD thesis, IRCCyN, Ecole Centrale de Nantes, 2012.
- [41] J Ding, AS Wexler, and SA Binder-Macleod. Mathematical models for fatigue minimization during functional electrical simulation. *journal of electromyography and kinesiology*, 13:575–588, 2003.
- [42] Elena V. Galiamova Vladimir V. Syuzev, Alexander M. Gousskov. Human skeletal muscle - mechanical and mathematical models. *ICABB-2010, Venice, Italy*, 0, 2010.
- [43] Ruina Ma, Damien Chablat, Fouad Bennis, and Liang Ma. A framework of motion capture system based human behaviours simulation for ergonomic analysis. In *HCI International 2011, 9-14 July, Hilton Orlando Bonnet Creek, Orlando, Florida, USA*, 2011.
- [44] Olivier Missenard, Denis Mottet, and Stephane Perrey. Muscular fatigue increases signal-dependent noise during isometric force production. *Neuroscience Letters*, 437:154–157, 2008.
- [45] Jean-Luc Gennisson, Thomas Deffieux, Emilie Macea, Gabriel Montaldo, Mathias Fink, and Mickaeal Tanter. Viscoelastic and anisotropic mechanical properties of in vivo muscle tissue assessed by supersonic shear imaging. *Elsevier-Ultrasound in Med.and Biol.*, 36:789–801, 2010.
- [46] J.S.Leedham and J.J.Dowling. Force-length, torque-angle and emg-joint angle relationships of the human in vivo biceps brachii. *Eur J Appl. Physiol*, 70:421–426, 1995.
- [47] Adel al Jumaily Yee Mon Aung. Estimation of upper limb joint angle using surface emg signal. *International journal of advanced robotics systems*, 10:369–376, 2013.
- [48] Liang Ma. *Contributions pour l'analyse ergonomique de mannequins virtuels*. PhD thesis, IRCCyN, Ecole Centrale de Nantes, France, 2009.
- [49] Thomas Deffieux, Jean-Luc Gennisson, Mickael Tanter, and Mathias Fink. Ultrafast ultrasonic imaging of in vivo muscle contraction using ultrasound. *IEEE Ultrasonics Symposium*, pages 1001 – 1004, 2006.

-
- [50] John C Cogshall and G A Bekey. A stochastic model of skeletal muscle based on motor unit properties. *Elsevier Publication Mathematical Biosciences*, 7:405–419, 1970.
- [51] K. M. Pell and J. W. Stanfield Jr. Mechanical model of skeletal muscle. *American Journal of Physical Medicine & Rehabilitation*, 50:23–28, 1972.
- [52] C.N. Christakos and S. Lal. Lumped and population stochastic models of skeletal muscle: implications and predictions. *Biological Cybernetics - Springer*, 36:73–85, 1980.
- [53] R.D. Woittiez, P.A. Huijing, H.B.K. Boom, and R.H. Rozendal. A three-dimensional muscle model: a quantified relation between form and function of skeletal muscles. *Journal of Morphology - Wiley Online Library*, 182:95–113, 1984.
- [54] B. Hannaford. A nonlinear model of the phasic dynamics of muscle activation. *IEEE transaction on Biomedical Engineering*, 37:1067–1075, 1990.
- [55] A.B. Schultz, J.A. Faulkner, and V.A. Kadhiresan. A simple hill element nonlinear spring model of muscle contraction biomechanics. *Journal of Applied Physiology*, 70:803–812, 1991.
- [56] A.S. Wexler, J. Ding, and S.A. Binder-Macleod. A mathematical model that predicts skeletal muscle force. *IEEE Transactions on Bio-Medical Electronics*, 44:337–348, 1997.
- [57] J. Bobet and R.B. Stein. A simple model of force generation by skeletal muscle during dynamic isometric contractions. *IEEE Transactions on Bio-Medical Engineering*, 45:1010–1016, 1998.
- [58] L.M. Studer, D.G. Rugg, and J.P. Gabriel. A model for steady isometric muscle activation. *Biological Cybernetics - Springer*, 80:339–355, 1999.
- [59] A.J. Fuglevand, D.A. Winter, and A.E. Patla. Models of recruitment and rate coding organization in motor-unit pools. *Journal of Neurophysiology*, 70:2470–2488, 1993.
- [60] R.D. Herbert and S.C. Gandevia. Twitch interpolation in human muscles: mechanisms and implications for measurement of voluntary activation. *Journal of Neurophysiology*, 82:2271–2283, 1999.
- [61] P.A. Merton. Voluntary strength and fatigue. *Journal of Physiology*, 123:553–564, 1954.
- [62] B. Bigland-Ritchie. Emg/force relations and fatigue of human voluntary contractions. *ACSM Exercise and Sport Sciences Reviews*, 9:75–117, 1981.
- [63] R.M. Enoka and D.G. Stuart. Neurobiology of muscle fatigue. *Journal of applied*

- Physiology*, 72:1631–1648, 1992.
- [64] S.C. Gandevia A.J. McComas, R.G. Miller. Fatigue brought on by malfunction of the central and peripheral nervous systems. *Fatigue Neural and Muscular Mechanisms*, Plenum Press, New York, pages 495–512, 1995.
- [65] David A. Hawkins and M.L. Hull. An activation-recruitment scheme for use in muscle modeling. *Journal of Biomechanics*, 25:1467–1476, 1992.
- [66] David A. Hawkins and M.L. Hull. Muscle force as affected by fatigue: Mathematical model and experimental verification. *Journal of Biomechanics*, 26:1117–1128, 1993.
- [67] I W Charlton and G R Johnson. Application of spherical and cylindrical wrapping algorithms in a musculoskeletal model of the upper limb. *Journal of Biomechanics*, 34:1209–1216, 2001.
- [68] M Gatton, M Pearcy, and G Pettet. Modelling the line of action for the oblique abdominal muscles using an elliptical torso model. *Journal of Biomechanics*, 34:1203–1207, 2001.
- [69] E Desailly, P Sardain, N Khouriand D Yepremian, and P Lacouture. The convex wrapping algorithm: A method for identifying muscle paths using the underlying bone mesh. *Journal of Biomechanics*, 43:2601–2607, 2010.
- [70] F C T Van Der Helm, H E J Veeger, G M Pronk, L H V Van Der Woude, and R H Rozendal. Geometry parameters for musculoskeletal modelling of the shoulder system. *Journal of Biomechanics*, 25:129–144, 1992.
- [71] F C T Van Der Helm. A finite element musculoskeletal model of the shoulder mechanism. *Journal of Biomechanics*, 27:551–569, 1994.
- [72] S C Gandevia. Spinal and supraspinal factors in human muscle fatigue. *Physiological Reviews*, 81:1729–1771, 2001.
- [73] Y Giat, J Mizrahi, and M Levy. musculotendon model of the fatigue profiles of paralyzed quadriceps muscle under fes. *IEEE Transactions on Biomechanical Engineering*, 40:664–674, 1993.
- [74] W Rohmert. Determination of the recovery pause for static work of man. *Internationale Zeitschrift Fur Angewandte Physiologie, Einschliesslich Arbeitsphysiologie*, 18:123–164, 1960.
- [75] H Monod and J Scherrer. The work capacity of a synergic muscular group. *Ergonomics*, 8:329–338, 1965.
- [76] M Hagberg. Work load and fatigue in repetitive arm elevations. *Ergonomics*, 24:543–555, 1981.

-
- [77] J M M Huijgens. A model for quantifying static load, incorporating muscle fatigue. *Biomechanics symposium. Boulder, CO, American Society of Mechanical Engineers*, pages 97–99, 1981.
- [78] G. Sjogaard. Intramuscular changes during long-term contraction. *The Ergonomics of Working Postures Models, Methods and Cases, Taylor & Francis, London*, 14:136–143, 1986.
- [79] L. Rose, M. Ericsson, and R. Ortengren. Endurance time, pain and resumption in passive loading of the elbow joint. *Ergonomics*, 43:405–420, 2000.
- [80] A Garg, K Hermann, B Schwoerer, and J Kapellusch. The effects of maximum voluntary contraction on endurance times for the shoulder girdle. *International Journal of Industrial Ergonomics*, 30:103–113, 2002.
- [81] K El Ahrache, D Imbeau, and B Farbos. Percentile values for determining maximum endurance times for static muscular work. *International Journal of Industrial Ergonomics*, 26:99–108, 2006.
- [82] Liang Ma., W. Zhang, Damien Chablat, Fouad Bennis, and F. Guillaume. Multi-objective optimization method for posture prediction and analysis with consideration of fatigue effect and its application case, computers & industrial engineering, 57(4), 1235-1246.. *Computers & Industrial Engineering*, 57(4):1235–1246, 2009.
- [83] L A Frey Law and K G Avin. Endurance time is joint specific: A modelling and meta-analysis investigation. *Ergonomics*, 53:109–129, 2010.
- [84] J Ding, A Wexler, and S A Binder-Macleod. A predictive model of fatigue in human skeletal muscles. *Journal of Applied Physiology*, 89:1322–1332, 2000.
- [85] R M Vignes. Modeling muscle fatigue in digital humans. *Masters thesis, Graduate College of The University of Iowa*, 2004.
- [86] J R Bendall. Effects of the "marsh factor" on the shortening of muscle fibre models in the presence of adenosine triphosphate. *Nature*, 170:1058–1060, 1952.
- [87] J Mizrahi, O Levin, A Aviram, E Isakov, and Z Susak. Muscle fatigue in interrupted stimulation: Effect of partial recovery on force and emg dynamics. *J. Elecfromyogr. Kinesiol.*, 7:51–65, 1997.
- [88] JP Gacesa, T Ivancevic, N Ivancevic, FP Paljic, and N Grujic. Non-linear dynamics in muscle fatigue and strength model during maximal self-perceived elbow extensors training. *Journal of Biomechanics*, 43:2440–2443, 2010.
- [89] A Beelen, A J Sargeant, D A Jones, and C J de Ruyter. Fatigue and recovery of voluntary and electrically elicited dynamic force in humans. *Journal of Physiology*, 484:227–235, 1995.

- [90] S. Alireza Fayazi. Optimal pacing in a cycling time trial considering cyclist's fatigue dynamics. *American control conference ACC, Washington, USA*, pages 6442–6447, 2013.
- [91] Vaughan G Macefield, Andrew J Fuglevand, John N Howell, and Brenda Bigland-Ritchie. Discharge behaviour of single motor units during maximal voluntary contractions of a human toe extensor. *Journal of Physiology*, 528:227–234, 2000.
- [92] Liang Ma, Damien Chablat, Fouad Bennis, and W Zhang. A new simple dynamic muscle fatigue model and its validation. *International Journal of Industrial Ergonomics*, 39(1):211–220, 2009.
- [93] Liang Ma, Damien Chablat, Fouad Bennis, and W Zhang. A new muscle fatigue and recovery model and its ergonomics application in human simulation. *Virtual and Physical Prototyping*, 5(3):123–137, 2010.
- [94] M. Hagberg. Muscular endurance and surface electromyogram in isometric and dynamic exercise. *Journal of Applied Physiology*, 51:1–7, 1981.
- [95] T Sato, Y Ohsumi, and Y Anraku. Substrate specificities of active transport systems for amino acids in vacuolar-membrane vesicles of *saccharomyces cerevisiae*. evidence of seven independent proton/amino acid antiport systems. *Journal of Biological Chemistry*, 259:11505–11508, 1984.
- [96] N. Corlett and J. Wilson and I. Manenica. A technique for postural load assessment. *The Ergonomics of Working Postures, Taylor & Francis, Londres*, 29:270–277, 1986.
- [97] W. Rohmert, M. Wangenheim, J. Mainzer, P. Zipp, and W. Lesser. A study stressing the need for a static postural force model for work analysis. *Ergonomics*, 29:1235–1249, 1986.
- [98] L. Rose, M. Ericsson, B. Glimskar, B. Nordgren, and R. Ortengren. Ergo-index. development of a model to determine pause needs after fatigue and pain reactions during work. *Computer Applications in Ergonomics, Occupational Safety and Health, Elsevier Science Publishers B.V, North-Holland*, pages 461–468, 1992.
- [99] W Rohmert. Problems in determining rest allowances part 1: use of modern methods to evaluate stress and strain in static muscular work. *Applied Ergonomics*, 4:91–95, 1973.
- [100] S Nanthavanij. Quantitative analysis of heart rate recovery profile during recovery from physical work. *International journal of industrial ergonomics*, 9(4):329–342, 1998.
- [101] S Konz. Work/rest: Part 2 the scientific basis for the guide. *International Journal of Industrial ergonomics*, 22(1):73–99, 1998.
- [102] H Shin and J Kim. Measurement of trunk muscle fatigue during dynamic lifting and

-
- lowering as recovery time changes. *International Journal of Industrial Ergonomics*, 37(6):545–551, 2007.
- [103] D Wood, D Fisher, and R Andres. Minimizing fatigue during repetitive jobs: optimal work-rest schedules. *Human factors*, 39:83–101, 1997.
- [104] B Duong, M Low, A Moseley, R Lee, and R Herbert. Time course of stress relaxation and recovery in human ankles. *Clinical Biochemistry*, 16:601–607, 2001.
- [105] M Yassierli, H Iridiastadi, and L Wojcik. The influence of age on isometric endurance and fatigue is muscle dependent: a study of shoulder abduction and torso extension. *Ergonomics*, 50:26–45, 2007.
- [106] N Milner, E Corlett, and C OBrien. A model to predict recovery from maximal and submaximal isometric exercise. *The ergonomics of working postures: models, methods and cases: the proceedings of the First International Occupational Ergonomics Symposium, Zadar, Yugoslavia*, pages 126–136, 1985.
- [107] W Rohmert and J Rutenfranz. Recovery and pause. practical physiology of work. *Stuttgart: G Thieme*, 1983.
- [108] B Elfving, D Liljequist, A Dederling, and G Nemeth. Recovery of electromyograph median frequency after lumbar muscle fatigue analysed using an exponential time dependence model. *European Journal of Applied Physiology*, 88:85–93, 2002.
- [109] R Westgaard and J Winkel. Guidelines for occupational musculoskeletal load as a basis for intervention: a critical review. *Applied Ergonomics*, 27:79–88, 1996.
- [110] Liang Ma, Wei Zhang, Su Wu, and Zhanwu Zhang. A new simple local muscle recovery model and its theoretical and experimental validation. *International Journal of Occupational Safety and Ergonomics*, 21:86–93, 2016.

List of Publications

Publications

- **Paper:** New Dynamic Muscle Fatigue Model to Limit Musculo-Skeletal Disorder.
 - **Conference:** Laval Virtual International Conference and Exhibition - VRIC 2016, March 23 - 25, 2016, Laval, France.
 - **Author:** Deep Seth, Damien Chablat, Sophie Sakka, Fouad Bennis, Marc Jebeau, Antoine Nordez.
 - **Status:** Accepted and under publication. ISBN 978 – 1 – 4503 – 4180 – 6/16/03
 - **DOI:** [http : //dx.doi.org/10.1145/2927929.2927935](http://dx.doi.org/10.1145/2927929.2927935)
- **Paper:** Experimental Validation of a New Dynamic Muscle Fatigue Model
 - **Conference:** 18th International Conference on Human-Computer Interaction. 17 - 22 July 2016, Toronto, Canada.
 - **Author:** Deep Seth, Damien Chablat, Sophie Sakka, Fouad Bennis.
 - **Status:** Accepted and will be presented in July.
 - **DOI:** *Springer International Publishing Switzerland*, 2016, *V.G.Duffy (Ed.)* : *DHM2016, LNCS9745, pp.112, 2016. DOI* : 10.1007/978–3–319–40247–5_6

Oral Presentation

1. CRIEC 2014, Inter Ecole Centrale annual conference, 2014, Ecole centrale de Lyon, France.
2. Virtual International Conference and Exhibition, VRIC 2016, March 23 to 25, 2016, Laval, France.
3. JDOC, EDSTIM, Doctoral School annual conference, 2015, University of Angers, France.

Appendices

.1 Load the data into MATLAB and indexing

Loading the data into Matlab

For importing the raw data into Matlab we have used this command:

```
data = importdata('Users\deep\Dropbox\subject9\6.txt','t');
```

Indexing parameters to loaded data in Matlab

```
t=data(:,2);    %//data for Torque//  
p=data(:,4);    %//data for Position//  
v=data(:,3);    %//data for velocity//  
bi=data(:,5);   %//data for Biceps EMG//  
tri=data(:,6);  %//data for Triceps EMG//  
tra=data(:,7);  %//data for Trapezius EMG//
```

After loading the raw data into Matlab each column of the data is indexed to specific character.

.2 Filter, Detrend, Rectification

% For filtering (bandpass FIR filter with Hamming window)

Here we have defined the environment for filtering the raw data:

```
lowfreq = 6; % [Hz] % Set low frequency
highfreq = 400; % [Hz] % Set high frequency
nyquistfreq = 2000/2; % Set nyquist frequency
Wn = [lowfreq/nyquistfreq highfreq/nyquistfreq] % Filter window;
b = fir1(4, Wn);
filtsignal1 = filter(b, 1, bi);
% De-mean bicep data
mv = mean(filtsignal1);
filtdeameanssignalbi = filtsignal1 - mv;
bi1 = filtdeameanssignalbi;
filtsignal2 = filter(b, 1, tri);
% De-mean tricep data
mv = mean(filtsignal2);
filtdeameanssignaltri = filtsignal2 - mv;
tri1 = filtdeameanssignaltri;
filtsignal3 = filter(b, 1, tra);
% De-mean trapezius data
mv = mean(filtsignal3);
filtdeameanssignaltra = filtsignal3 - mv;
tra1 = filtdeameanssignaltra;

% Full wave rectification of Data
bi2 = detrend(bi1); % Detrend
recbi = abs(bi2); % Rectification
tri2 = detrend(tri1);
rectri = abs(tri2);
tra2 = detrend(tra1);
rectra = abs(tra2);
```

.3 Butterworth Filter

```
%Here we have applied Butter worth Low pass filter to EMG data
[b, a] = butter(4, 20/2000, 'low');
filterbi = filtfilt(b, a, recbi);
b1 = abs(filterbi);
[b, a] = butter(4, 20/2000, 'low');
filtertri = filtfilt(b, a, rectri);
t1 = abs(filtertri);
[b, a] = butter(4, 20/2000, 'low');
filtertra = filtfilt(b, a, rectra);
T1 = abs(filtertra);
[b, a] = butter(4, 20/2000, 'low');
v = filtfilt(b, a, v);
t = filtfilt(b, a, t);
p = filtfilt(b, a, p);
```

.4 Cycles separation on the basis of velocity

```
%cycle separation
m = 0;
k = 0;
l = 1;
n1 = size(N);
for i = 1 : n1;
    if N(i) == 0;
        m = 1;
        k = k + 1;
        Vm, k = NN(i);
        m = m + 1;
    end
end
```

.5 Time scale Normalization with interpolation

```
% Data points interpolation to percentage scale
k = 0;
zv = [];
for i = 1 : size(V, 2);
k = k + 1;
zv(:, k) = interpft(cell2mat(V(:, i)), 100); %Interpolation
end
```

.6 Cycles separation on the basis of EMG activity

EMG cut point before activation starts

```
for i = 1 : (n - 2);  
    if b1(i) == 0 & b1(i + 1) == 0 & b1(i + 2) > 0;  
        y(l) = i;  
        l = l + 1;  
    end  
end
```

```
    u = [];  
l = 1;
```

EMG cut point after activation ends

```
for i = 1 : (n - 2);  
    if b1(i) > 0 & b1(i + 1) == 0 & b1(i + 2) == 0;  
        u(l) = i + 60;  
        l = l + 1;  
    end  
end
```

.7 Experimentally Calculated Values of Γ_{cem} or MVC

S. No.	$\Gamma_{cem}Extension$									
	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 9	Subject 10
1	44.092	60.098	46.488	67.199	64.815	69.890	54.954	78.526	47.286	71.142
2	39.109	50.585	40.677	40.826	52.055	52.780	48.422	73.414	40.754	62.622
3	35.636	53.454	37.399	30.992	48.865	50.315	48.848	55.806	36.210	59.072
4	34.126	51.038	36.952	33.823	45.095	46.110	49.274	52.114	44.162	62.054
5	37.448	51.944	36.505	21.9	40.455	46.255	50.694	69.012	30.672	—
6	37.146	49.830	31.290	18.178	42.920	36.975	50.694	48.706	38.908	—
7	37.599	48.924	33.227	—	42.485	38.715	50.978	42.032	33.796	—
8	32.767	47.565	—	—	37.120	40.745	52.398	43.594	22.436	—
9	37.448	48.018	—	—	34.510	39.730	52.114	47.144	—	—
10	36.995	48.018	—	—	—	39.150	54.670	—	—	—
11	36.995	46.659	—	—	—	38.715	54.386	—	—	—
12	35.183	43.639	—	—	—	42.485	53.108	—	—	—
13	35.636	—	—	—	—	35.380	52.114	—	—	—
14	38.052	—	—	—	—	38.860	53.108	—	—	—
15	36.089	—	—	—	—	36.975	52.824	—	—	—
16	32.163	—	—	—	—	41.325	50.836	—	—	—
17	36.542	—	—	—	—	35.090	52.966	—	—	—
18	31.106	—	—	—	—	31.610	52.114	—	—	—
19	—	—	—	—	—	31.610	52.114	—	—	—
20	—	—	—	—	—	35.815	50.978	—	—	—
21	—	—	—	—	—	35.235	53.108	—	—	—
22	—	—	—	—	—	31.030	52.966	—	—	—
23	—	—	—	—	—	—	52.540	—	—	—
24	—	—	—	—	—	—	52.824	—	—	—
25	—	—	—	—	—	—	50.978	—	—	—
26	—	—	—	—	—	—	51.120	—	—	—
27	—	—	—	—	—	—	49.416	—	—	—
28	—	—	—	—	—	—	53.392	—	—	—
29	—	—	—	—	—	—	51.688	—	—	—
30	—	—	—	—	—	—	53.108	—	—	—
31	—	—	—	—	—	—	50.694	—	—	—
32	—	—	—	—	—	—	53.250	—	—	—
33	—	—	—	—	—	—	53.108	—	—	—

Table 1: Experimentally measured values of $\Gamma_{cem}Extension$

S. No.	$\Gamma_{cemFlexion}$									
	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 9	Subject 10
1	44.020	58.890	47.560	59.015	51.185	60.320	51.830	66.740	47.996	76.822
2	43.878	47.112	36.830	42.485	39.440	52.2	49.416	52.256	34.080	66.314
3	42.742	48.471	34.220	21.750	38.425	50.460	47.996	60.492	31.666	65.746
4	43.594	48.169	32.480	24.940	35.380	48.285	47.286	59.924	28.542	63.616
5	47.570	46.055	25.665	17.110	30.015	43.935	47.286	54.812	32.234	—
6	47.854	44.092	26.535	17.110	38.280	44.805	47.286	57.510	36.068	—
7	41.464	43.337	28.710	—	35.525	42.485	47.854	45.440	24.850	—
8	40.044	37.297	—	—	32.335	41.615	47.570	38.056	26.838	—
9	45.440	35.183	—	—	29	49.010	48.848	41.606	—	—
10	44.730	40.166	—	—	—	41.760	49.7	—	—	—
11	45.014	40.468	—	—	—	43.210	50.410	—	—	—
12	36.778	38.354	—	—	—	46.690	48.706	—	—	—
13	47.712	—	—	—	—	45.675	48.138	—	—	—
14	39.618	—	—	—	—	51.910	48.848	—	—	—
15	41.606	—	—	—	—	43.645	48.706	—	—	—
16	44.446	—	—	—	—	43.645	46.292	—	—	—
17	42.032	—	—	—	—	41.470	43.736	—	—	—
18	38.198	—	—	—	—	42.920	43.310	—	—	—
19	—	—	—	—	—	34.365	43.452	—	—	—
20	—	—	—	—	—	40.6	43.594	—	—	—
21	—	—	—	—	—	37.410	42.174	—	—	—
22	—	—	—	—	—	38.280	43.452	—	—	—
23	—	—	—	—	—	—	42.174	—	—	—
24	—	—	—	—	—	—	42.742	—	—	—
25	—	—	—	—	—	—	44.304	—	—	—
26	—	—	—	—	—	—	43.310	—	—	—
27	—	—	—	—	—	—	41.748	—	—	—
28	—	—	—	—	—	—	41.038	—	—	—
29	—	—	—	—	—	—	44.162	—	—	—
30	—	—	—	—	—	—	44.446	—	—	—
31	—	—	—	—	—	—	45.724	—	—	—
32	—	—	—	—	—	—	43.026	—	—	—
33	—	—	—	—	—	—	41.464	—	—	—

Table 2: Experimentally measured values of $\Gamma_{cemFlexion}$

Thèse de Doctorat

Deep SETH

Contribution to the Evaluation of Muscle Fatigue Model and Recovery Model

Contribution à l'évaluation du Modèle de Fatigue Musculaire et de Récupération

Résumé

L'automatisation a changé les conditions de travail dans les industries en changeant les processus de fabrication. Cependant, il existe encore beaucoup d'industries demandant des opérations manuelles ou de manutention pour diverses tâches. Ces opérations manuelles conduisent à l'apparition de troubles musculo-squelettiques (TMS) qui sont l'un des problèmes majeurs pour les travailleurs. La fatigue musculaire est l'une des causes des troubles musculo-squelettiques. Les entreprises doivent faire attention à ce problème en raison des nouvelles lois sur la pénibilité ou les tâches répétitives. Dans cette thèse, nous nous concentrons sur le développement d'un modèle dynamique de fatigue et de récupération musculaire pour des postures de travail dynamique, ses applications et sa validation par le biais d'expériences. Tout d'abord, nous avons introduit un nouveau modèle de fatigue musculaire dynamique incluant un facteur de co-contraction pour prédire la variation de la force physique dans des conditions dynamiques. Un modèle de récupération est utilisé pour prédire la récupération après la fatigue. Les deux modèles sont d'abord validés théoriquement et comparés avec les modèles précédents, puis validés expérimentalement. En théorie, les modèles d'analyse sont comparés en utilisant la méthode de régression. Deuxièmement, des expériences ont été menées sur 10 sujets pour le mouvement push-pull du bras pour étudier la force musculaire et le taux de récupération. Des signaux d'électromyographies (EMG) sont utilisés pour analyser l'activité musculaire et valider les modèles de fatigue musculaire et de récupération. Il a été constaté que lorsque la fatigue musculaire augmente, l'indice de co-contraction diminue. Aussi, la courbe de fatigue de la plupart des sujets a suivi la fonction exponentielle décroissante prédite par le modèle de fatigue. Avec l'utilisation du facteur de co-contraction, dynamique, le temps d'endurance maximale (DMET) diminue de 25,9 % par rapport au modèle de R. Ma. Enfin, les valeurs normalisées du taux de fatigue et le facteur co-contraction sont utilisés pour prédire la DMET pour diverses charges de travail. Le modèle de fatigue et de récupération proposé dans cette thèse peut être utile dans l'analyse des paramètres de fatigue musculaire et de récupération, d'analyser la posture humaine et dans l'amélioration de l'ergonomie des postes de travail.

Mots clés

Fatigue musculaire, troubles musculo-squelettiques (TMS), contraction maximale spontanée, modèle de la fatigue musculaire dynamique, modèle de récupération, co-contraction, électromyographie (EMG), temps d'endurance maximale dynamique

Abstract

Automation has changed the working conditions in industries and manufacturing process, but still many industries need manual operations and handling for various tasks. These manual operations lead to work-related Musculoskeletal Disorders (MSD) which is one of the major problems for industrial workers. Muscle fatigue is one of the reasons that leads to Musculoskeletal Disorder. The companies have to pay attention on this issue due to the new laws on penibility or repetitive tasks. In this thesis, we are focusing on the development of an adequate and realistic dynamic muscle fatigue and recovery model for dynamic work posture, its applications and validation through experiments. Firstly, we have introduced a new dynamic muscle fatigue model with a newly introduced co-contraction factor to predict the physical strength in dynamic conditions. A recovery model is used to predict the recovery after fatigue. Both the models are first validated theoretically and compared with previous models and then validated experimentally. In theoretical analysis models are compared using regression methods. Secondly, experiments were conducted on 10 subjects for push pull motion of the arm to study the muscle strength and rate of recovery. Electromyography (EMG) technique is used to analyze the muscle activity. At last, the experimental data are used to validate muscle fatigue and recovery model. It has been found that with the increase in muscle fatigue co-contraction index decreases and most of the subjects followed the exponential function predicted by fatigue model. With the use of co-contraction factor dynamic maximum endurance time (DMET) decreases by 25.9 % as compared to R.Ma's Model. At last, the normalized values of fatigue rate and co-contraction factor are used to predict the DMET for various work loads. The fatigue and recovery model proposed in this thesis can be useful in analyzing the muscle fatigue and recovery parameters, to analyze human posture and in improving working ergonomics conditions.

Key Words

Muscle fatigue, musculoskeletal disorder (MSD), maximum voluntary contraction (MVC), dynamic muscle fatigue model, strength recovery, recovery model, co-contraction, electromyography (EMG), dynamic maximum endurance time (DMET)