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Loading and recovery behavior of the human lumbar spine under static flexion

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LOADING AND RECOVERY BEHAVIOR OF THE HUMAN LUMBAR SPINE UNDER STATIC FLEXION

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To my husband Yilmaz, my son Koral, and my parents M. Ozkan and Gulseren

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ABSTRACT

Musculoskeletal disorders account for 32 percent of work related injuries and illnesses. Extensive studies have been conducted to understand and explore the factors that affect the development of musculoskeletal disorders to provide effective control strategies.

One of the most important factors that allow effective control strategies is the biomechanical factors. Among the biomechanical risk factors, exposures to repetitive static and vibratory activities are known to result in musculoskeletal disorders. A ligamento–muscular reflex activity exists in humans and animals to stabilize the knee, shoulder, elbow, ankle joints, and spine. This reflex activity decreases over time when static loading is applied, which leads to joint instability and decreased safety of the spine.

However, the recovery process of this reflexive muscle activity to its original state is a complicated phenomenon and the research efforts in this area are limited to a few animal based experiments. Although the recovery process is modeled and known for animals, the process for human muscle recovery is still unknown.

This study concentrates on the behavior and recovery of the human lumbar muscle when subjected to static loading. Ten males participated in this study where the electrical activity of their erector-spinae muscle for two different lumbar levels was recorded for twenty minutes of loading and three hours of recovery period to observe and model the muscle behavior.

The behavior of recovery was modeled using a bi-exponential structure previously developed for animal experiments. However, the analysis of the EMG showed that the sitting task during rest periods created additional static loading on the lower back. The

effects of this task were introduced to the model by adding a third exponential component referred to as the “daily activity” factor.

Overall, the results support the previous findings about the behavior of the lower back muscles that were developed by animal experiments. However, the mathematical description of the phenomenon is modified by taking the daily activity factor into account. Understanding the time periods and phases for the recovery is essential, since a better understanding of the phenomenon can lead to optimal design of work/rest periods in occupational as well as sports activities.

CHAPTER 1 - INTRODUCTION

Workplace musculoskeletal injuries are classified under two categories: idiopathic and traumatic (Kumar, 2001). Idiopathic injuries are mediated through mechanical degradation and cannot be attributed to a specific act or incident (such as physical degeneration). Traumatic injuries are associated with an incident or an action including overexertion, sudden imbalance, pulling apart, crushing, impact, slip and fall.

A musculoskeletal disorder (MSD) is defined as an injury or disorder of the muscles, nerves, tendons, joints, cartilage, or spinal discs (disorders caused by slips, trips, falls, motor vehicle accidents or similar accidents are not included) (BLS, 2004).The risk factors for MSDs can be divided into four major categories (Kumar, 2001). These are genetic, morphological, psychosocial, and biomechanical. Although genetic and morphological factors have an important role in prevention of MSDs, only biomechanical and psychosocial factors allow effective control strategies. Among the various biomechanical risk factors, exposures to repetitive static and vibratory activities are known to result in musculoskeletal disorders including soft tissue injuries and cumulative trauma disorders (CTDs). CTDs occur when loading of the tissues and muscles accumulates by repeated exposures, or exposures of sufficient durations (National Academy of Sciences, 2001). Factors that affect developing CTDs are posture, external forces, strength, and anthropometry. CTDs are usually associated with work because throughout a workday, individuals often repeat actions or spend long periods of time performing work activities in many occupations.

Work-related MSDs account for a major component of the cost of work-related illnesses. The costs associated with work-related MSDs range from \$13 to \$54 billion

annually (BLS, 1997). In 2004, according to the Bureau of Labor Statistics, MSDs accounted for 402,700, or 32 percent of the injuries and illnesses with days away from work. The occupations most susceptible to MSDs are laborers and material movers (33,590), nursing aides, orderlies, and attendants (29,980), and heavy and tractor-trailer truck drivers (17,770), which experienced the greatest number of total injuries and illnesses with days away from work in 2004. The MSDs that affect the back constitute 48.8 percent of the total 402,700 cases of the injuries and illnesses with days away from work. The National Academy of Sciences (2001) estimated that there were 5.6 million work-related back pain cases and 600,000 back injury cases in 1999 that cost the industry 10 to 14 billion dollars in workers' compensation costs annually (Occupational Safety and Health Administration, 1992; National Academy of Sciences, 2001). Previous studies indicate that low back pain or low back injury may occur when spinal load exceeds the tissue's tolerance limit (Herrin et al., 1986; McGill, 1999; Norman et al., 1998). Therefore, cumulative load exposure predisposes the spine to pain and/or injury and is a risk factor (Kumar, 1990).

The understanding of the muscle behavior is one of the important areas to properly understand and analyze musculoskeletal injuries and disorders. Since the 1950s, the behavior of different muscles under different loading conditions (cyclic, cumulative, static, etc.) has been studied extensively. However, the understanding of the recovery process of the muscles after exposure to loading is still very limited. This information is critical especially when designing the control measures for MSDs and proper work/rest cycles.

This study concentrates on the response of the lumbar spine when subjected to forward static flexion. As a part of this study, electromyography (EMG) signals were collected from the lumbar erector spinae muscle by using surface electrodes. Ten participants were observed

in an experiment setup consisted of three periods to analyze the muscle behavior including rest, loading and recovery periods. The collected EMG data were analyzed to model the behavior of the muscle for loading and recovery periods. A discussion of the model behavior and its components are also included in the study to increase the understanding of the overall mechanism of the muscle behavior under static loading.

The following chapters include a review of the literature and background information including a simple description of low back anatomy and physiology, electromyography, a summary of lower back related studies, and a review of mathematical models explaining muscle behavior. The experiment procedure and methodology is included with detailed data collection and processing techniques. The development of the mathematical models for muscle loading and recovery are also included with a detailed discussion of the recovery behavior followed by summary of the conclusions and future research directions.

CHAPTER 2 - BACKGROUND AND LITERATURE REVIEW

This chapter consists of the background information and review of the related literature including lower back anatomy and physiology, electromyography, lower back related studies and the related mathematical models of muscle behavior.

2.1. LOWER BACK ANATOMY AND PHYSIOLOGY

To analyze the effects of static loading on muscle tissue behavior and activity, it is important to understand the basic anatomical and physiological properties of the human lower back structure and its elements.

2.1.1. Ligaments and Tendons

Tendons attach muscles to the bones. These structures have high tensile strength and high modulus of elasticity (Kumar, 1999; Kumar, 2001). Their major function is transferring forces between muscles and bones. Ligaments connect bones at the articulations and provide joint support and stability. These structures help to distribute load to other structures in the joint. Therefore, they are important structures, which bear tensile loads in the joints (Garrett et al., 2000). Both of these connective tissues are made of collagen, which is visco-elastic in nature. Viscoelasticity is the tissue's ability to adjust or adapt to repetitive loading by changes in its length or its load over time. Ligaments and tendons differ in their microstructure and collagen fiber organization. In ligaments, the collagen fibers are arranged in the form of a flat sheet. In tendons, the fibers are in the form of a rope, running in one direction.

During loading, the collagenous fibers rearrange their position parallel to the axis of stress. As their loading is further increased to the failure point, the fibers start gliding upon

one another. This represents the breaking of the forces, which hold the fibers together. When these forces are broken, the fibers tear completely (Tkaczuk, 1968).

2.1.2. Vertebral Column

The vertebral column is a strong, flexible structure that bends anteriorly, posteriorly, laterally, and rotates. Its function is to protect the spinal cord, support the head, and serve as a point of attachment for the ribs and muscles of the back. The vertebral column consists of a series of bones called vertebrae. The adult vertebral column typically contains 26 vertebrae. There are seven cervical vertebrae in the neck region, 12 thoracic vertebrae posterior to the thoracic cavity, 5 lumbar vertebrae supporting the lower back, 5 sacral vertebrae fused into one bone called sacrum, and 4 coccygeal vertebrae fused into one or two bones called the coccyx. Between each vertebra from the first vertebra (atlas) to the sacrum, are intervertebral discs. The lumbar vertebrae (L1-L5) are the largest and strongest in the column since they support higher amounts of body weight than cervical and thoracic vertebrae (Tortora, 1995).

2.1.3. Intervertebrae Joints

Intervertebrae joints provide mobility to the spine. In addition to ligaments and the muscles, several studies have shown that they also have a major role in stabilizing the lumbar spinal segments (Bogduk and Twomey, 1987; Farfan et al., 1970; Gregersen and Lucas, 1967; Markolf, 1972, White and Panjabi, 1978). An intervertebral joint includes two vertebrae and an intervertebral disc between the vertebrae.

The intervertebral disc consists of four tissues: the nucleus pulposus, the annulus fibrosus, the cartilaginous endplates, and the vertebral body (NRCIM, 2001). The nucleus pulposus is a viscous, protein gel, made up of collagen and 80 percent water. The annulus fibrosus is a ligamentous tissue composed of collagen fibers. The cartilaginous endplates are

composed of mostly hyaline cartilage. They serve as the upper and lower surfaces of the intervertebral discs. The vertebral body consists of a porous bone core that is surrounded by a thin shell of dense bone. The major load-bearing role of the intervertebral disc is in axial compression and flexion (Adams et al., 1988; Tencer et al. 1982).

The previous studies have shown that the intervertebral disc changes its height and viscoelastic properties as a result of changes in the fluid content, which can be due to compression elicited by prolonged sitting and standing (Dolan et al., 1993; Hedman and Fernie, 1995; Leivseth and Drerup, 1997; McGill and Brown, 1992, Towney and Taylor, 1982), or as a result of cyclic, vibratory, or prolonged loading (Adams and Hutton, 1988; Botsford et al., 1994; Ekstrom et al., 1996; Goel et al., 1988; Keller et al., 1989; Wilder et al., 1985) by means of fluid loss of the disc. The reduced fluid content in the disc increases the laxity of the intervertebrae joints, allowing an increase in the intervertebral motion that causes instability (Solomonow et al., 1999). A joint is instable when it can not carry the required functional loads without pain and produce joint contact forces with abnormal intensities on articular cartilage surfaces when moving through a normal range of motion. This instability leads to decreased safety of the spine, leaving the vertebral column susceptible to injury and illness.

2.1.4. Spinal Cord

The spinal cord is located within the spinal canal of the vertebral column. Spinal cord conducts sensory impulses from the periphery to the brain and motor impulses from the brain to the periphery. The gray matter around the central canal of the spinal cord receives and integrates incoming and outgoing information (Tortora, 1995).

2.1.5. Erector Spinae Muscle

The erector spine muscle is chosen for this study because it is the largest and strongest muscle group, located on each side of the vertebral column. It is divided into three longitudinal columns: iliocostalis muscle, longissimus muscle, and spinalis muscle. It extends, flexes, and rotates the head and the vertebral column. Figure 1 shows the location of the erector spinae muscle group.

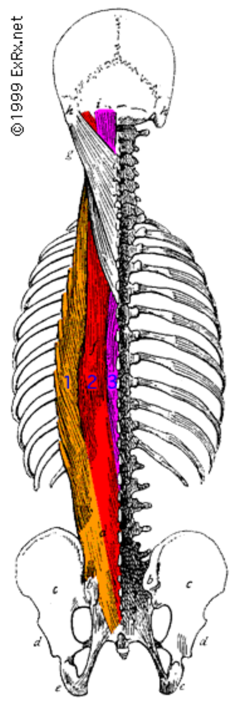

Figure 1. Erector Spinae Muscle Group (ExRx, 1999)

2.1.6. Mechanoreceptors

There are five different types of receptors in human body (Tortora, 1995). Mechanoreceptors detect mechanical pressure or stretching (Bronzino, 1995; Dorland, 1994). Thermoreceptors detect temperature changes; nociceptors detect pain; photoreceptors detect

light; and chemoreceptors detect chemicals in the mouth, nose, and body fluids. Mechanoreceptors are responsible for triggering the reflexive muscle activity to stabilize the joints when subjected to constant loading or displacement.

2.2. ELECTROMYOGRAPHY

Electromyography (EMG) is the recording of the electrical activity associated with contracting muscle, by means of using surface electrodes, needle electrodes, or wire electrodes. Surface electrodes are adhered to the skin. They consist of a silver disc shaped detection surface. Wire electrodes consist of a small diameter, flexible, non-oxidizing insulated wire that is inserted into the muscle. Like needle electrodes, the tip of the wire is not insulated (Baratta and Solomonow, 1995). In addition to the electrodes, any EMG collection system consists of amplifiers, filters, and a data acquisition device.

The EMG signal is the summation of all of the muscle fiber action potentials (MUAPs). In biomechanical studies, EMG is commonly used to evaluate if a muscle is "on" or "off" during a particular activity, to evaluate the relationship between EMG and the production of force, and as a fatigue index.

EMG signals are affected by anatomical and physiological muscle properties and the instrumentation used to collect the signal. Physiological properties include the amount of fat tissue overlaying the muscle, firing rate of the motor units, temperature and length of the muscle. Electrode type, size, location, and spacing can also affect the collected EMG signal (Baratta and Solomonow, 1995).

There are several processing techniques to evaluate the collected EMG signal. Most commonly used analyses include time domain analysis, frequency domain analysis, and normalization.

2.2.1. Time Domain Analysis

This processing method involves the wave form of the signal directly over a period of time. The most commonly used technique is rectification of the EMG signal. It can be full wave, where all the negative deflections are inverted (absolute value of the signal is used), or half wave, where only the positive deflections are rendered.

Other most commonly used parameters are the Mean Absolute Value (MAV), Root Mean Square (RMS) of the signal, and integrated EMG (IEMG). MAV is the averaged rectified signal used to establish a relationship between EMG and muscle force. RMS is similar to MAV but it does not require the rectification of the signal. It is the square root of the integration of the squared EMG signal divided by the time period of the integration. IEMG is the area under the rectified curve. It begins at a preset time and continues throughout the total time of muscle activity (Baratta and Solomonow, 1995).

2.2.2. Frequency Domain Analysis

The analysis involves determining the frequency spectrum of the signal, usually by using a Fast Fourier Transform (FFT). The power density spectrum (PSD) is then obtained by squaring the FFT and determining the magnitude, which can be used to calculate the median, the mean, and the mode frequencies. These parameters are commonly used to evaluate fatigue (Baratta and Solomonow, 1995). The mean frequency is the frequency which divides the amplitude spectrum into two equal halves, the median frequency of the spectrum is the frequency which divides the power spectrum into two equal halves, and the mode frequency is the frequency at which the spectrum reaches its maximum.

2.2.3. Normalization

MAV, RMS, and IEMG values can be normalized with respect to a maximum measurable value such as time, to compare EMG signals from different individuals or muscles. The most commonly used maximal peak value is the peak EMG value obtained during a maximal voluntary contraction (MVC) (Baratta and Solomonow, 1995).

2.2.4. Fatigue Evaluation with Electromyography

Surface EMG can be used to evaluate muscle fatigue and is shown to have advantages for fatigue assessment as well as providing evidence of physiological phenomena related to the back musculature (Farina et al., 2003). Fatigue can be defined as failure to maintain the required or expected force, and is accompanied by changes in electrical activity in the muscles (Dimitrova and Dimitrov, 2003). A variety of parameters have been used to evaluate muscle fatigue using EMG signals. These include full wave rectified integral, root mean square, number of turning points (number of points at which the EMG signal changes its slope from negative to positive and vice versa), number or zero crossings (number of points at which the EMG signal crosses the zero voltage level), and average amplitude (averaged amplitudes of the rectified signal).The most commonly used EMG parameter to assess fatigue from the back muscles is the median power spectral frequency (MDF) (Farina et al., 2003). MDF has very similar properties to the mean frequency (MNF) (Farina and Merletti, 2000). The slope of the MDF (the rate of change over time) is associated to the fatigability of the motor units of the muscle fibers (De Luca, 1993). In addition, a decline in median frequency and an increase in the amplitude of the EMG signal is an accepted fatigue indicator (Koumantakis et al., 2001).

2.3. REVIEW OF MUSCLE BEHAVIOR RELATED STUDIES

Previous studies have shown that the lumbar intervertebral discs and ligaments are easily injured when the spine is subjected to a forward bending moment and that the risk of bending related injury to the lumbar discs, ligaments, and muscles depends on the loads applied to the spine, the loading rate, and the loading history (Adams and Dolan, 1996). Wang et al. (2000) studied the effects of loading rate on the load sharing and safety margin of spinal structures by applying compression (2000 N) and shear loads (200 N) to the L2–L3 lumbar motion segment. The higher loading rate increased the peak intradiscal pressure by 12.4 percent and the bending moment by 20.7 percent, as well as the total ligament forces by 11.4 percent. The higher loading rate reduced the safety margin of passive spinal elements.

The literature indicates that repetitive prolonged static displacement or load induces creep (time dependent elongation) in passive tissues of the spine including ligaments, disc, and joint capsule (Jackson et al., 2001; Solomonow et al., 1999; Williams et al., 2000).This resulting creep in the ligaments or the laxity developed in the viscoelastic structure causes instability of the spine and pain. Numerous studies indicate that lumbar ligaments play a limited role in stabilizing the spine (Abumi, 1990; Andersson et al., 1985; Berkson et al., 1979; Crisco et al. 1992; Lucas and Bresler, 1961; McGill and Norman, 1986; Posner et al., 1982; White and Panjabi, 1978); where as the musculature around the spine is the major stabilizing structure (Bogduk and MacIntosh, 1984; Bojadsen et al., 2000; Gardner-Morse and Stokes, 1998; Granata and Marras, 1995; Indahl et al., 1995; Kaigle et al., 1995; Kumar et al., 1996; McGill and Norman, 1986; Panjabi, 1992; Pope et al., 1986; Zetterberg et al., 1987). Recent studies have shown that a ligamento–muscular reflex activity exists in humans and animals, to stabilize the knee, shoulder, elbow, ankle joints and spine. This reflex activity is

triggered by mechanoreceptors in the ligaments, discs, and facet capsules to the multifidus and longissimus muscles, allowing the musculature and the visco–elastic tissues of the spine to act synergistically (Guanche et al., 1995; Hirokawa et al., 1991; Knatt et al., 1995; Lewis et al., 1996; Phillips et al, 1997; Solomonow et al., 1996; Solomonow et al., 1987; Solomonow et al., 1998; Stubbs et al., 1998; Williams et al., 2000). In other words, the interaction of the musculature and the visco-elastic tissues of the spine create a combined effect that is greater than the sum of their individual effects.

In 1999, Solomonow et al. studied the effects of cyclic loading of the lumbar spine in an in vivo feline model by applying a passive cyclic loading of 0.25 Hz. to L4–L5 lumbar level for 50 minutes. The loading was followed by a 10-minute rest period and the same loading cycle was repeated for a second and a third time. They found out that cyclic loading exposes the spine to instability and injury because of three physiologic mechanisms. The laxity developed in the ligamentous tissue desensitizes the mechanoreceptors within the ligament, making them unable to initiate the muscular forces to prevent instability. The creep deformation of the viscoelastic structures such as ligament and disc also causes laxity increase in the intervertebrae joint, which has a desensitizing effect on the receptors within. The third physiologic mechanism is the loss of the muscular forces when subjected to fatigue.

In addition, innervations and mechanical stimulation of the ligament produces reflex spasms, which could cause low back pain and discomfort (Eversull et al., 2001; Pedersen et al., 1956; Williams et al., 2000; Yahia and Newman, 1993). The reflexive activity of the musculature decreases significantly when the viscoelastic structures are subjected to creep, leaving the muscular forces unable to prevent instability (Gedalia et al., 1999; Solomonow et al., 1999; Jackson et al., 2001).

Although the soft tissue and muscle behavior under certain types of loading are known, to this day scientists are still trying to find answers to the required rest period that will allow the creep, the laxity in the viscoelastic structures, and the muscles to recover and restore the reflexive muscular activity. Several studies investigated the time for recovery under different types of load exposures. McGill and Brown (1992) investigated the creep induced in the human lumbar spine during flexion in seated position for 20 minutes. Both male and female subjects participated in the experiment. In males, after 20 minutes of rest, the recovery of the creep was 50 percent. The recovery was faster for females. They concluded that the viscoelastic recovery of the creep was exponential and required longer time for full recovery. In their study of an in vivo porcine spine, after applying vibratory stimulus for one hour, Ekstrom et al. (1996) concluded that after one hour of rest, full recovery was not possible. Crisco et al. (1997) investigated creep and recovery in the human wrist ligaments. They concluded that 2 hours of rest produced only partial recovery after repetitive loading exercise of the wrist. However, when they examined their subjects after 24 hours of rest, the creep was fully recovered. These studies concentrated on mostly the recovery of the ligaments and therefore the reflexive muscular activity.

In an in vivo feline study, Gedalia et al. (1999) observed the recovery of reflexive muscular stability with rest. L4–L5 lumbar level was subjected to a 0.25 Hz. cyclic loading for 50 minutes and the electromyographic activity from the multifidus muscles of L1–L2 to L6–L7 lumbar levels were recorded. After a rest period of 2 hours (twice as long as the loading duration), the laxity developed in the lumbar viscoelastic structures demonstrated only a partial recovery. The prolonged rest periods did not allow sufficient recovery of reflexive muscular activity.

Solomonow et al. (2000) also studied the reflexive muscular activity after prolonged cyclic loading and the necessary rest period for recovery in the lumbar spine using an in vivo feline model. After 50 minutes of cyclic loading at 0.25 Hz, a mean time of 7 hours was required for a full recovery of the multifidus muscles. Another important finding was that once the reflexive activity of the multifidus muscles recovered fully, some muscles became hyperexcitable, generating more electromyographic response to viscoelastic deformation than at the beginning of loading, indicating that recovery occurs at different speeds and levels for different tissues.

Jackson et al. (2001) used a displacement-controlled method to determine the recovery of the tension-relaxation as well as the recovery of reflexive EMG activity in the multifidus after prolonged static flexion. They observed that prolonged static flexion results in fast exponential decrease of reflexive muscular activity, and that the recovery of the muscle activity with rest is characterized by an initial period of hyperexcitability, followed by a biexponential increase.

In 2002, Hatipkarasulu looked at the effects of short term static flexion of the feline spine. Four different magnitudes of load (20N, 30N, 50N, and 70N) were applied for 20 minutes to the lower back of the subjects. And the recovery of the reflexive multifidus muscle activity as well as the creep developed was measured for 7 hours during the rest period. Full recovery of the reflexive muscular activity was not observed for any of the load magnitudes that were applied. It was estimated that full recovery may take from 24 hours to a week.

Recently, Claude et al. (2003) investigated the biomechanical effects of cyclic loading on the lower back by using a feline model. Their subjects underwent 20 minutes of cyclic

loading of different magnitudes of load and 7 hours of rest period. Full recovery of reflexive muscular activity was never observed for any loading period.

In a similar study, Solomonow et al. (2003) looked at the flexion-relaxation response to static flexion in males and females where the subjects performed three bouts of lumbar flexion-extension before and after a 10 min period of static lumbar flexion. They measured the surface EMG response from the erector spinae muscles and the flexion angle. They concluded that creep developed during a short static lumbar flexion, which elicited significant changes in the muscle behavior. Although they used humans as subjects, they did not investigate the recovery after the flexion period.

In 2003, Solomonow et al. used a feline model to assess the mechanical and neurological processes that are present in the development of a cumulative trauma disorder associated with repetitive exposure to lumbar flexion. The spines of the felines were exposed to three 10 min flexion sessions with each session followed by a 10 min rest, and EMG response from the multifidus muscle was measured. The loading period was followed by a 7-hour recovery period. They found out that the creep developed during each 10 min loading period did not recovery fully in the following 10 min rest period and at the end of the 7 hr recovery period.

Lu et al. (2004) studied the effects of repetitive loading by using a feline model, where the spines of felines were subjected to 20 min of cyclic loading at frequencies of 0.1 Hz and 0.5 Hz while monitoring the EMG activity of the multifidus muscles. The results showed that cyclic lumbar flexion causes a transient neuromuscular disorder consisting of EMG spasms. They concluded that repetitive loading at frequencies induces larger creep in the viscoelastic tissues and is indeed a risk factor.

Most recently, Olson et al. (2004) looked at the flexion-relaxation response to cyclic lumbar flexion in males. The subjects performed cyclic lumbar flexion at 0.1 Hz for 9 min while recording paraspinal electromyogram and kinematic data. It was found that cyclic lumbar flexion and extension develops laxity in the lumbar viscoelastic structures and increased frequency of spasms towards the end of the exercise. They did not observe the recovery behavior for this experiment set up. This study used a similar posture and experiment set up where both loading (flexion) period and recovery period were observed.

2.4. MATHEMATICAL MODELS OF MUSCLE BEHAVIOR AND RECOVERY

Gedalia et al. (1999) conducted a study on recovery after 50 minutes of cyclic loading in a feline setup and developed models for both muscle activities during loading and recovery period. A bi-exponential model was developed for recovery which consisted of two exponential components (including two separate time constants), and a residual. The two exponential components were a fast component and a slow component, which are time dependent. These models are presented in Equations (1) and (2). The model for the muscle activity and load during cyclic loading is defined as:

$$Y_C(t) = Ae^{-t/T_1} + Be^{-t/T2} + r \qquad (1)$$

where,

$Y_C(t) =$ The integrated electromyographic (IEMG) response or load as a function of time during laxity development (% of initial value)

$A =$ The amplitude of the first viscoelastic component (% of initial value)

$T_1 =$ The time constant of the first viscoelastic component (min)

$B =$ The amplitude of the second viscoelastic component (% of initial value)

T_2 = The time constant of the second viscoelastic component (min)

r = The final residual value (% initial value)

The model for the recovery period is defined as:

$$Y_r(t) = C(1 - e^{-t/T3}) + D(1 - e^{-t/T4}) + r \quad (2)$$

where,

$Y_r(t)$ = The IEMG response or load as a function of time during recovery (% of initial value)

C = The amplitude of the first viscoelastic component (% of initial value)

T_3 = The time constant of the first viscoelastic component (min)

D = The amplitude of the second viscoelastic component (% of initial value)

T_4 = The time constant of the second viscoelastic component (min)

r = The value at start of recovery (% initial value)

In 2000, Solomonow et al. developed a biexponential recovery model of lumbar viscoelastic laxity and reflexive muscular activity after they applied prolonged cyclic loading to the spine of feline preparations for 50 minutes. The recovery data was modeled as follows:

$$Y_r(t) = C(1 - e^{-t/T_1}) + D(1 - e^{-(t-\tau)/T2}) + r \quad (3)$$

where,

$Y_r(t)$ = The IEMG or load as a function of time during recovery (% of initial value)

C = The amplitude of the first viscoelastic component (% of initial value)

T_1 = The recovery time constant of the first viscoelastic component (min)

D = The amplitude of the second viscoelastic component (% of initial value)

T_2 = The recovery time constant of the second viscoelastic component (min)

τ = The time delay to second recovery component

r = Residual value at start of recovery (% initial value)

In their study on tension-relaxation (displacement controlled method), Jackson et al. (2001) added a third component to this structure to describe the initial transient hyperexcitability phenomena, where the normalized IEMG (NIEMG) values exhibited large initial increase at the beginning of the rest period, which was followed by minor decrease. This model was defined as.

$$Y(t) = A(1 - e^{-t/T_1}) + B(1 - e^{-(t-\tau)/T2}) + Cte^{-t/T_3} + R \tag{4}$$

where,

A = The amplitude of the fast recovery component (% of initial value)

T_1 = The time constant of the fast recovery component (min)

B = The amplitude of the slow recovery component (% of initial value)

T_2 = The time constant of slow recovery component (min)

τ = The time delay of the slow recovery component

R = The residual component left at the end of the 20-minute flexion period (% initial value)

C = The amplitude of the transient hyperexcitability component (% of initial value)

T_3 = The time constant of the transient hyperexcitability component (min)

In Equation (4), the term "$A(1-e^{-t/T_1})$" describes a fast increasing exponential and believed to account for the behavior of the several ligaments and capsules of the lumbar spine. The term "$B(1-e^{-(t-\tau)/T2})$" describes the response of the intervertabral discs in which the recovery is slower and is associated with a time delay (τ). Finally, the third term "Cte^{-t/T_3}" represents the initial transient hyperexcitability phenomena.

In 2002, Hatipkarasulu modeled the muscle behavior and its recovery during 20 minutes application of four different loads (static flexion) and during a 7-hour recovery period. An exponential model was developed to represent the classical response of viscoelastic materials to loads and/or elongation. The pooled NIEMG data of the six lumbar levels from the multifidus muscle were modeled in the form of exponential functions. The model structure for NIEMG in the loading period is similar to the ones developed by Solomonow et al. (2000) and Jackson et al. (2001), which takes the form of:

$$NIEMG_{(t)} = Ae^{-t/T_1} + NIEMG_o \qquad (5)$$

where,

$NIEMG_{(t)}$ = Normalized integrated EMG as a function of time

A = Exponential component initial amplitude

T_1 = Exponential decay rate constant

$NIEMG_o$ = Steady state NIEMG amplitude

t = Time

A graphical representation and the fit of the model are shown in Figure 2. This graph shows the average NIEMG values for 70 N loading for 6 different lumbar levels of the felines and the graphical representation of the model.

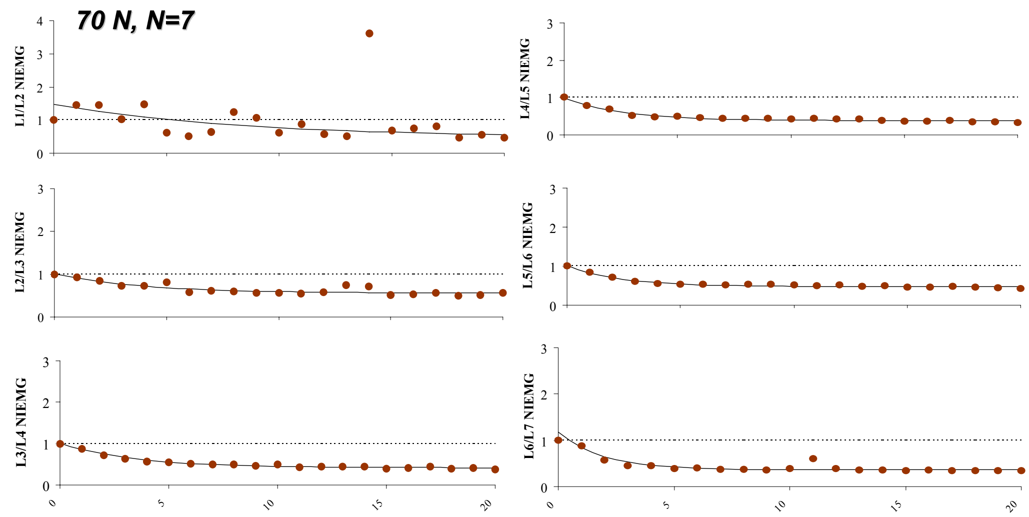

Figure 2. NIEMG Model for Six Lumbar Levels during 20-minute Loading Period

In Figure 2, the data points represent the NIEMG values and the smoothed line represents the model. The y-axis in these graphs are the average NIEMG values for lumbar levels L1/L2, L2/L3, L3/L4, L4/L5, L5/L6, and L6/L7 (no unit since they are normalized values); and the x-axis are time in minutes. As it can be seen from the graphs, the average NIEMG values decrease exponentially over time for all of the lumbar levels.

Similarly, exponential models were chosen to describe the NIEMG during the 7 hours recovery period. Although the muscles' behavior during the loading period was similar to the previous studies, a different phenomena, which was a delayed transient hyperexcitability

period, was observed through out the collected data and was added as a component to the model. This model is presented in Equation (6).

$$NIEMG_{(t)} = E\left(1-e^{-t/T_4}\right) + tBe^{-t/T_5} + C(t-T_d)e^{-(t-T_d)/T_6} + NIEMG_o \tag{6}$$

where,

$E\left(1-e^{-t/T_4}\right)$ = The steady state recovery component

tBe^{-t/T_5} = Transient hyperexcitability component

$C(t-T_d)e^{-(t-T_d)/T_6}$ = Delayed transient hyperexcitability component

$NIEMG_o$ = Residual response at the end of the 20-minute loading period

Figure 3 shows the NIEMG values for the recovery period, the model, and the components of the recovery period.

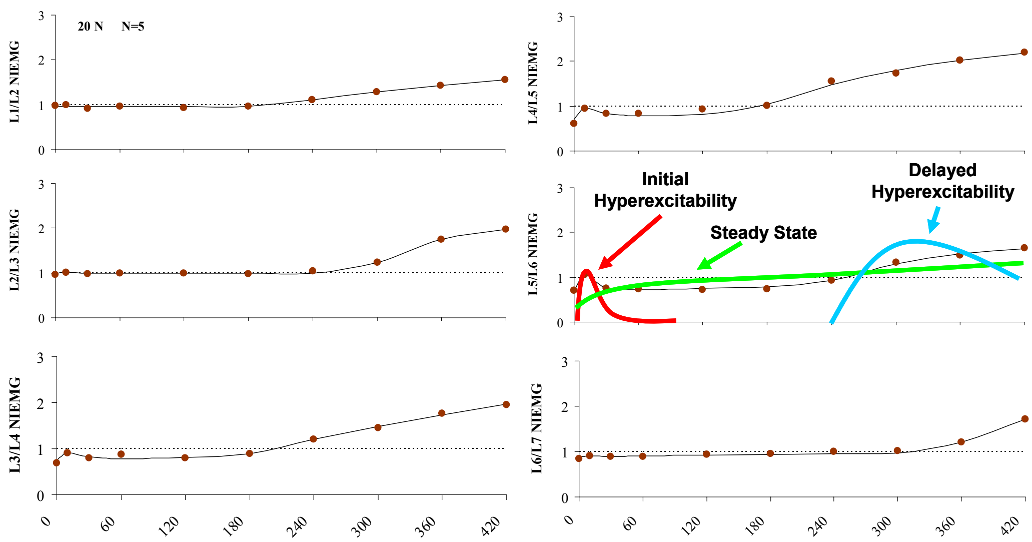

Figure 3. NIEMG Model for Six Lumbar Levels during Recovery Period.

In this model, the constraint of $E + NIEMG_o = 1$ was used to insure that full recovery results in a normal response (the maximal value of total recovery is one). $NIEMG_o$ (value at the start of recovery) was directly extracted as the final point of the laxity at the end of the 20-minutes of static flexion. The parameters for the first and second hyperexcitability periods were estimated by the amplitude and duration of the periods. Time constants were estimated through iteration. The maximal value of the total recovery (minus the parameters $NIEMG_o$) was used as E.

Most recently, Claude et al. (2003) investigated the biomechanical effects of cyclic loading on the lower back by using a feline model. Their subjects underwent 20 minutes of cyclic loading of different magnitudes of load and 7 hours of rest period, and as a result of their study, they developed the following models for loading and recovery periods.

For the loading period:

$$NIEMG_{(t)} = Ae^{-t/T_1} + NIEMG_{ss} \tag{7}$$

where,

$NIEMG_{(t)}$ = Normalized integrated EMG as a function of time

A = Exponential component initial amplitude

T_1 = Exponential decay time constant (minutes)

$NIEMG_{ss}$ = Steady state NIEMG amplitude

t = Time

For the recovery period:

$$NIEMG_{(t)} = E\left(1 - e^{-t/T_4}\right) + tBe^{-t/T_5} + C(t - T_d)e^{-(t-T_d)/T_6} + NIEMG_{ss} \qquad (8)$$

where,

$E\left(1 - e^{-t/T_4}\right)$ = The steady state recovery component

tBe^{-t/T_5} = Transient hyperexcitability component

$C(t - T_d)e^{-(t-T_d)/T_6}$ = Delayed transient hyperexcitability component

$NIEMG_o$ = Residual response at the end of the 20-minute loading period.

This model structure that includes steady state, transient hyperexcitability, and delayed hyperexcitability components have been applied in Lu et al.'s (2004) cyclic lumbar flexion study on felines.

In their study on cumulative lumbar disorders, Solomonow et al. (2003) used the same model structure to describe the muscle behavior during loading and rest periods when the spine of felines were subjected to a series of three 10 min sessions of static lumbar flexion with each session followed by a 10 min rest period. A 7 hr rest period was observed after the series of three flexion-rest sessions. In order to convert the models (for both the loading and the rest periods) to describe a series of work periods spaced by rest periods, they defined two new time components. The models are shown in Equation (9) and Equation (10) for work and rest periods, respectively:

$$NIEMG(t) = A_n \exp\left(\frac{-\left[t - n(T_w + T_R)\right]}{T_{n1}}\right)\Bigg|_{n(T_w + T_R)}^{(n+1)T_w + nT_R} + NIEMG_{0n} \tag{9}$$

$$\begin{aligned} NIEMG(t) = {} & \left(t - \left[(n+1) + T_w + nT_R\right]\right)B_n \\ & \times \exp\left(\frac{t - \left[(n+1)T_w + nT_R\right]}{T_{n2}}\right)\Bigg|_{(n+1)T_w + nT_R}^{(n+1)(T_w + T_R)} + NIEMG_{0n} \end{aligned} \tag{10}$$

where,

$NIEMG_{(t)}$ = Normalized integrated EMG as a function of time

A_n = Exponential component initial amplitude

T_w = The time period over which work was performed by/to the spine (minutes)

T_R = The period of rest between any two work periods (minutes)

$NIEMG_{0n}$ = Steady state NIEMG amplitude

CHAPTER 3 - GOALS AND OBJECTIVES

3.1. RATIONALE

The models developed previously make it possible to mathematically explain the behavior of muscles under certain loading and displacement conditions. However, these models were developed based on the observations of animal experiments (in vivo feline models) in highly controlled environments. Although the muscle behavior of felines is similar to humans, felines are quadrupeds, have 7 lumbar vertebrae, and have a gravity vector perpendicular to the spine, where as humans have 5 lumbar vertebrae, biped, and have gravity vector parallel to the spine.

Furthermore, the displacement-controlled methods, flexion, and cyclic loading to determine the recovery of the tension-relaxation and creep, as well as the recovery of reflexive EMG activity in the multifidus and other paraspinal muscles, have been studied and analyzed using different animal models, but not human models. This information is essential for designing control measures for optimal design of work/rest periods and musculoskeletal disorders in occupational activities as well as sports.

3.2. GOALS AND OBJECTIVES

The goal of this study is to observe and model the lower back erector spinae muscles' behavior under static flexion using humans as test subjects. To achieve this goal, the following objectives were pursued:

1. Observation of the lumbar erector spinae muscle behavior under static flexion using EMG.
2. Observation of the lumbar erector spinae muscle during a 3-hour recovery period.

3. Analysis of the EMG data and mathematical modeling of the lumbar erector spinae muscle behavior under static flexion and recovery.
4. Evaluation of erector spinae muscle fatigue.

In addition to the mathematical modeling of the human muscle behavior, two phenomena were also tested as defined in the following hypotheses:

Phenomenon I:

H_0: The reflexive muscle activity to stabilize the spine does not decrease over time throughout the loading period

H_1: The reflexive muscle activity to stabilize the spine decreases over time throughout the loading period

Phenomenon II:

H_0: The muscle fully recovers at the end of the three hour recovery period

H_1: The muscle does not recover fully at the end of the three hour recovery period

The decision criteria for the first and second phenomena are the calculated NIEMG values during the loading and recovery periods respectively. The detailed calculation methods for NIEMG values are described in Chapter 4.

CHAPTER 4 - METHODS AND PROCEDURE

This chapter includes a description of the experiment procedure, data collection and processing methodology.

4.1. EXPERIMENT PROCEDURE AND DATA COLLECTION

This section consists of detailed description of the experiment equipment and procedure including the descriptive information for each participant.

4.1.1. Equipment

The Bagnoli 2 EMG System (Delsys Inc, Boston, MA) was used to collect the EMG data with two single differential surface and a disposable reference electrodes (Delsys (a), 2003). The electrode housing is constructed with waterproof polyurethane plastic, and the electrode contacts are made from 99.9% pure silver bars measuring 10 mm in length, 1 mm in diameter and spaced 10 mm apart for optimal signal detection. The system consists of two single differential electrodes with a 5 ft electrode cable, a reference electrode, a belt-mounted amplifier unit, and an output cable. The belt-mounted amplifier unit supplies power to the EMG electrodes, receives and conditions the EMG signals, and provides 2 analog channel outputs. Figure 4 shows the different components of the Bagnoli 2 EMG System used in this study.

Two pairs of single differential surface EMG electrodes were adhered to the skin on the lumbar erector spinae muscles at the L2 and L4 lumbar levels on the right side of the spine, 2.0 cm lateral from the midline. L2 and L4 levels were chosen to observe high and low load bearing lumbar levels. Although L1 level is exposed to the highest magnitude of loads, the placement of electrode on this level creates discomfort for the participants. Therefore, L2

lumbar level, which is the second highest load bearing lumbar level, was chosen for this study. The second electrode was placed at L4 level for two reasons. First is to reduce the channel cross talk between the two lumbar levels by creating a distance between electrodes. The second reason is to observe the difference in the EMG magnitudes and muscle behavior in a less load bearing level.

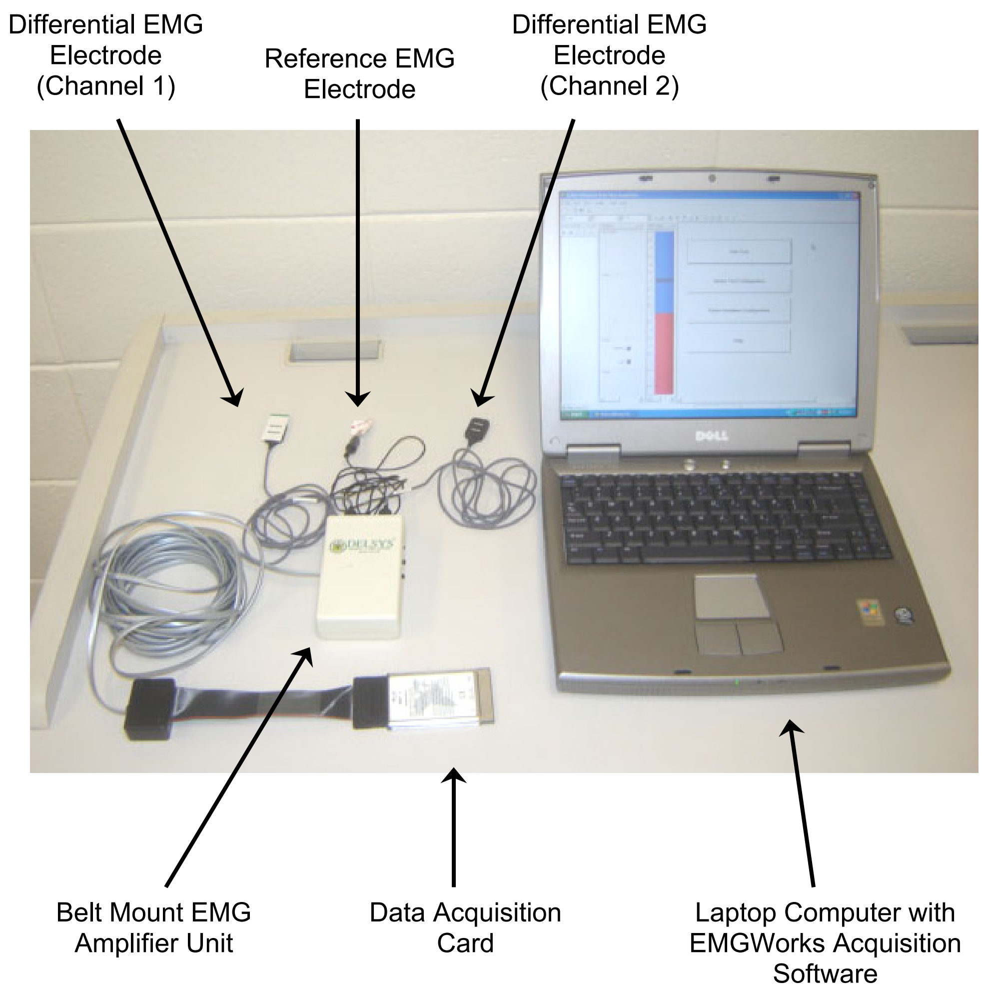

Figure 4. The Bagnoli 2 EMG System, Delsys Inc, Boston, MA

The reference electrode was placed on the elbow of the participants. Electrodes were placed so that the silver bar pairs on each electrode were perpendicular to the muscle fibers. A band pass filter of 20-450 Hz and a sampling rate of 2000 Hz were used with a preamplification factor of 10. The EMG response from each channel was monitored on the computer screen and stored in the computer by using Delsys System equipment and EMGWorks Acquisition software (Delsys (a), 2003).

4.1.2. Experiment Setup

Ten males volunteered to participate in this experiment signing a consent form which was approved by the Institutional Review Board of Louisiana State University. None of the participants reported having any history of back pain or back related disorders. A copy of the consent form used in the study can be found in Appendix A.

Before the EMG measurements, age, height, weight, and back strength of each participant was recorded for control purposes (age 28.80 ± 4.57 yr old; height 177.00 ± 6.50 cm; weight 169.20 ± 26.49 lb). The participant information and back strength measurement for each participant is shown in Table 1.

Back strengths of the participants were measured by using Isometric Strength Testing Device (Prototype Design and Fabrication, Ann Harbor, MA) and DT_VEE software. Figure 5 shows the back strength measurement set up.

The participants were asked to sit on a chair in front of the static strength measurement platform. A strap was wrapped around the participants shoulders which was connected to the handle of the platform. The participants were asked to extend their back slowly and steadily until maximum exertion is reached and hold for three seconds. The DT_VEE program runs until the test is performed three times and the coefficient of variation is then calculated. If it’s

less than 10 percent, the program calculates the average of the three tests as the back strength value. If not, the test is repeated for the fourth time and best three out of four are selected.

Table 1. Participant Information

Subject	Date	Height (cm)	Weight (lb)	Age	Back Strength (lb)
1	08/15/2005	170	162	28	34.08
2	08/12/2005	170	142	29	37.98
3	08/07/2005	172	150	23	37.92
4	08/10/2005	178	181	24	40.76
5	10/15/2005	172	144	23	35.00
6	10/16/2005	184	222	31	40.72
7	08/09/2005	179	182	29	18.23
8	10/16/2005	173	139	36	33.32
9	08/07/2005	186	188	35	44.30
10	10/16/2005	186	182	30	54.14
	Average	**177.00**	**169.20**	**28.80**	**37.65**
	StDev	**6.50**	**26.49**	**4.57**	**9.14**

After the back strength test, the skin around the lumbar region at L2 and L4 levels was cleaned with alcohol and the surface electrodes were placed on the participants as described above and a signal check was performed to ensure the quality of the EMG signal. Figure 6 shows the placement of the differential surface EMG electrodes and the reference electrode as well as the belt mounted amplifier unit.

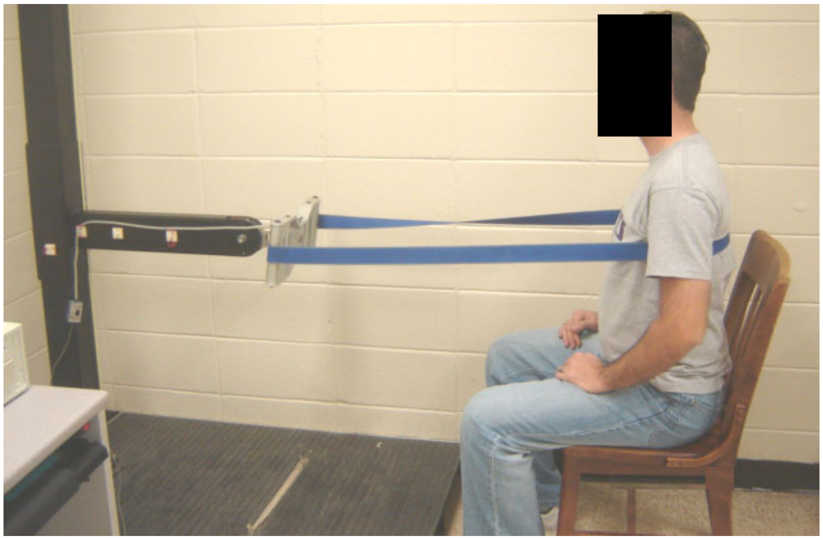

Figure 5. Back Strength Measurement Setup with Isometric Strength Testing Device

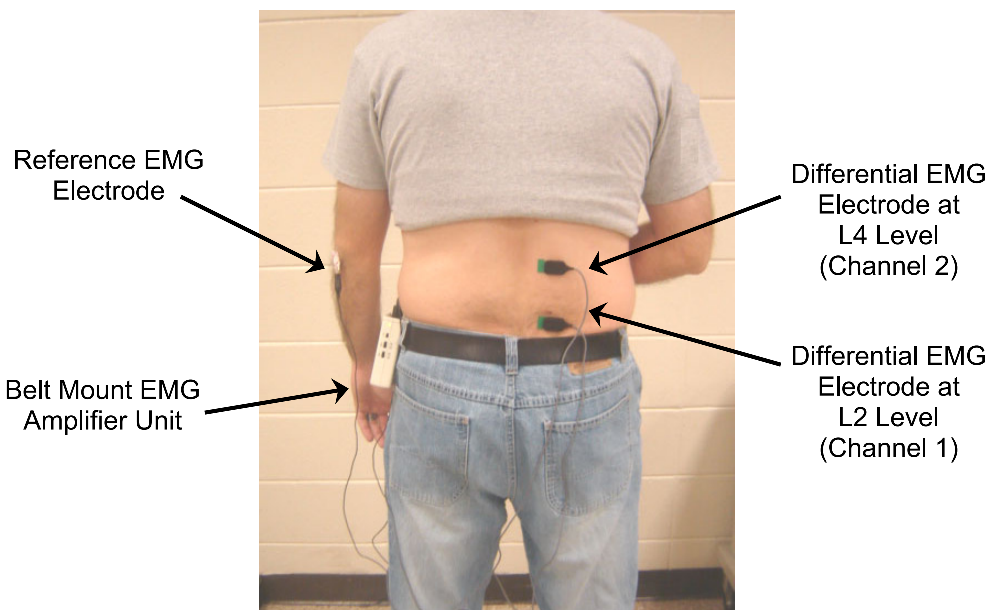

Figure 6. Electrode Placement

The participants were asked to sit on a chair comfortably for 20 minutes during which at rest measurements were recorded as control data. The measurements were taken continuously for 20 minutes to observe the behavior of the muscle when the back is at resting position and to assure any EMG activity recorded afterwards was indeed due to the static flexion of the spine. Figure 7 shows a participant while at rest measurements were taken.

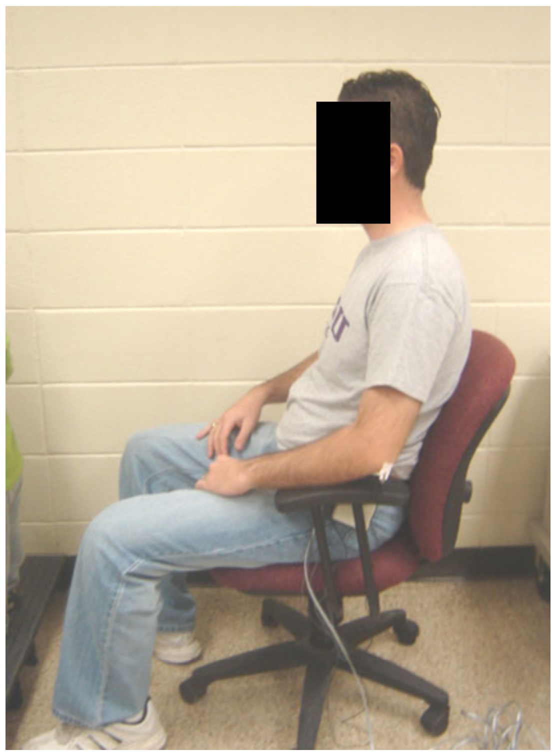

Figure 7. The Experiment Setup for the Rest Period

For the loading period, the participants were asked to sit on a hard mattress. A triangular foam wedge (7” H x 24” W x 24” L) was provided as a support for their knee joints to decrease additional muscle strain and discomfort of the participants. While on the mattress, the participants were asked to bend forward as much as they can, and to hold still for 20

minutes at fully bent position. EMG measurements were taken continuously for the 20 minute loading period. The static loading posture is presented in Figure 8.

Figure 8. The Experiment Setup for the Loading Period

After the loading period, the participants were asked to sit on the chair for the recovery period measurements. Recovery measurements were collected after 10, 30, 60, 120, and 180 minutes of rest. At the end of each rest period, the participants were asked to repeat the forward flexion for 2 minutes while continuous EMG measurements were taken during the 2-minute flexion to assess the recovery of the electrical activity in the muscle.

4.2. DATA PROCESSING

In this section, data processing and preparation methods are described step-by-step with sample illustrations. Matlab scientific software (Ver. 7) and its toolboxes were used for raw data processing and modeling of the recorded EMG files. (Matlab (a), 2005).

4.2.1. Raw Data Conversion

Loading and recovery data were recorded using seven different EMG files (at rest, loading, and five recovery files) for each participant at a sampling rate of 2000 Hz. At rest and loading measurements includes twenty minutes of continuous EMG measurements. The recovery files were recorded continuously for two minutes per period. All data files were saved in binary ".emg" format, which was converted to Matlab using a code script written by Delsys Inc. (Delsys, 2003(2)).

Figure 9 shows a sample raw EMG output where channel 1 and channel 2 labels represent L2 and L4 lumbar levels, respectively. Raw EMG recordings for all participants are included in Appendix A. In Figure 3, at rest and loading windows include 2,400,000 data points (2000 Hz x 1200 seconds) and each recovery window includes 120,000 data points (2000 Hz x 120 seconds). For each participant, raw EMG data consist of 10,800,000 data points at two lumbar levels.

A number of sharp spikes are observed in several locations in Figure 9 that can be interpreted as muscle spasms resulting in high amplitude electrical activity. Although the spikes show abnormal muscle activity, their value in describing muscle behavior is very limited. These high value data points are smoothed or eliminated as the raw EMG data is processed thorough rectification, integration and normalization processes. It is also observed that the EMG discharge amplitude for channel 2 is higher than channel 1; this observation is consistent for all participants of the study. The EMG signal is the summation of the motor unit action potentials in the muscle fibers and the discharge is related to the number of active action potentials. The difference between channels 1 and 2 simply shows that there are more active motor unit action potentials at L4 lumber level of the erector spinae muscle.

PARTICIPANT 5

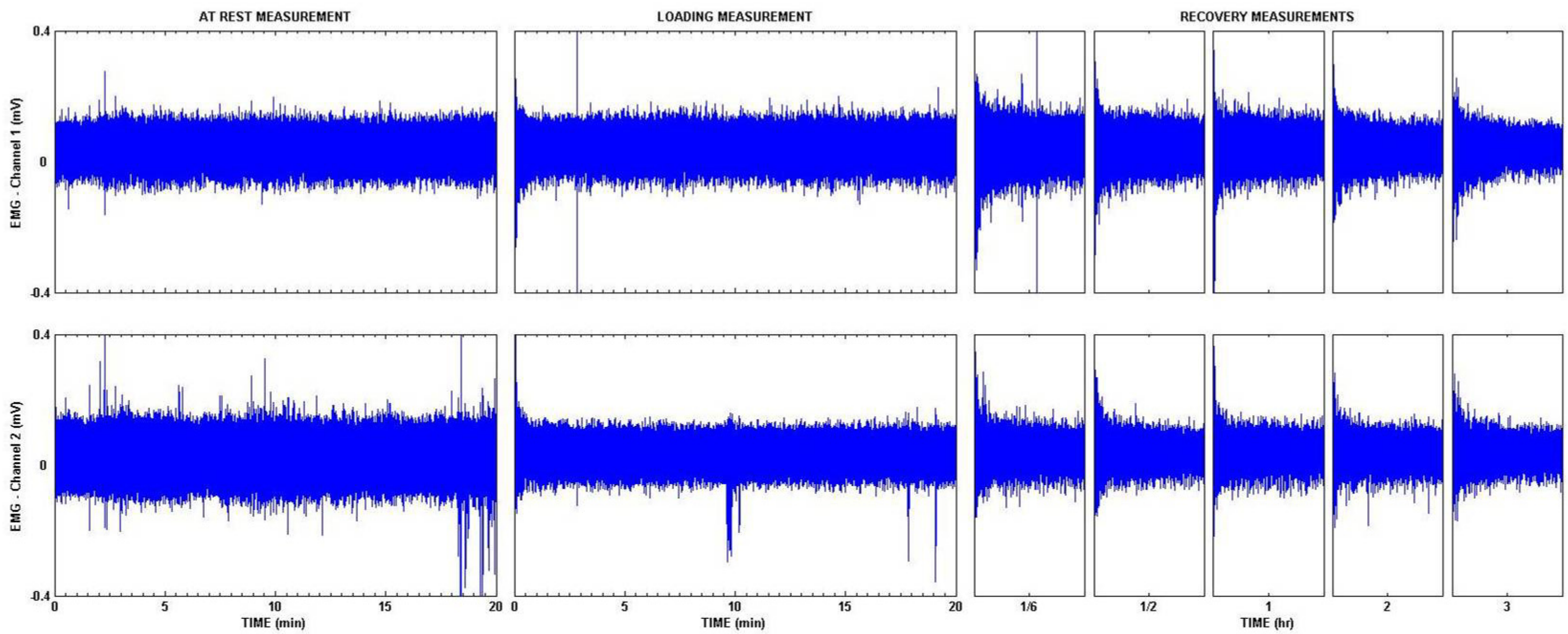

Figure 9. Sample Raw EMG Data for At Rest, Loading and Recovery Periods

4.2.2. Full Wave Rectification

The full wave rectification of the EMG data includes converting all negative data points to positive values. To accomplish the rectification, first, the raw EMG data is filtered through a band pass operation with a bandwidth of 6 to 500 Hz which eliminates any artifacts (noise) due to equipment conditions and environmental effects from the data set. The filtering operation uses "Butterworth" filter with an order level of 2 (Matlab (b), 2005). The filtered data is then full wave rectified using "abs" function in Matlab (Matlab (c), 2005).

Figure 10 shows a filtered full wave rectified EMG output for L2 and L4 lumbar levels. Full wave rectified EMG recordings for all participants are included in Appendix C. Compared to the raw EMG signal in Figure 3, Figure 4 shows a relatively noise free signal.

4.2.3. Normalized Integration

The integrated EMG (IEMG) is the area under the rectified curve ($m(t)$) that can be defined as (Basmajian and De Luca, 1986):

$$IEMG(|m(t)|) = \int_0^t |m(t)| dt \qquad (11)$$

In this study, each EMG sample was integrated over a 3-second window for at rest, loading, and recovery periods. This process converts 2,400,000 data points for a 20 minute time period down to 400 data points (2,400,000 / (2000 Hz x 3 seconds)). Matlab's "cumtrapz" function was used to calculate the area under the curve for the selected 3-second windows (Matlab (c), 2003).

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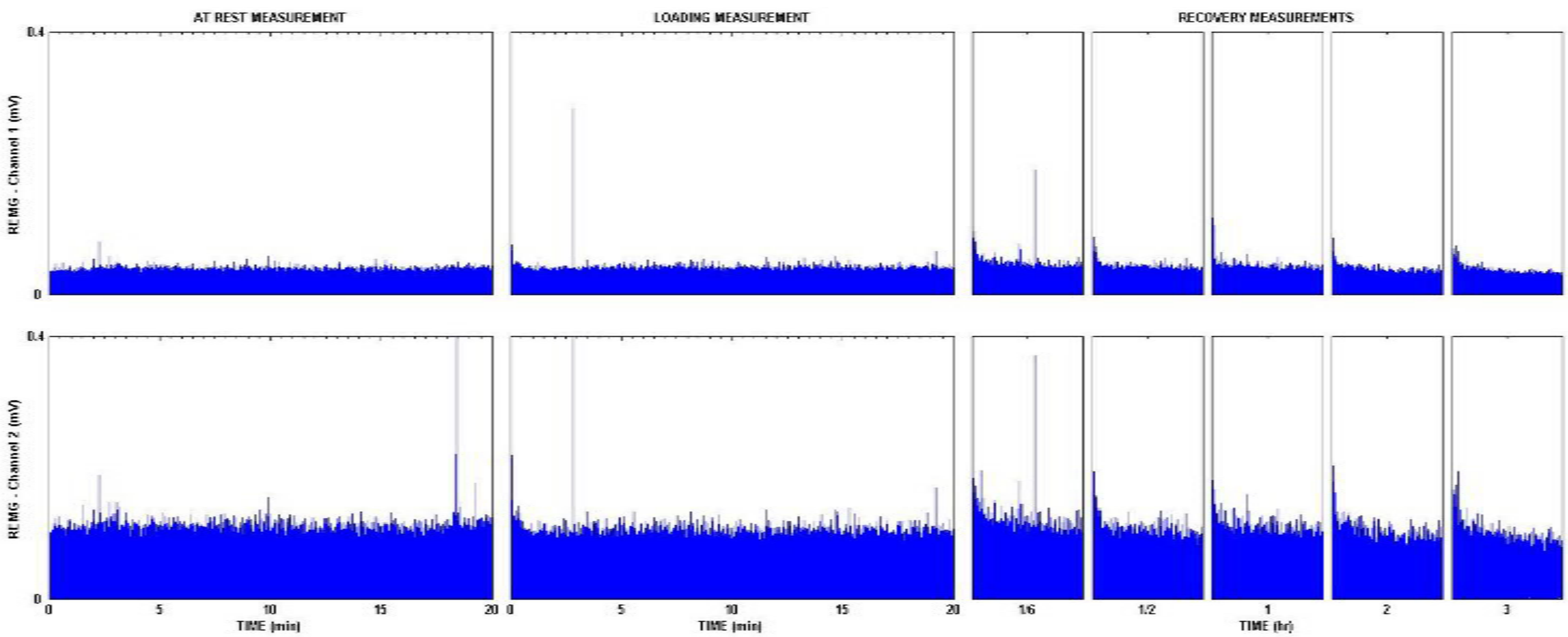

Figure 10. Sample Full Wave Rectified EMG Data for At Rest, Loading and Recovery Periods

To illustrate the integration process, a simplified integration calculation is provided in Figure 11 that shows three calculation steps of recording the raw EMG, filtering and rectifying the signal, and calculating the integrated EMG values. The y-axis represents the recorded EMG signals in mVs and the x-axis represents the time in seconds. The first graph shows a 1/8 second section of the raw EMG data from channel 2 of one of the participants during the loading period that consist of 250 data points. It can be clearly seen that the EMG signal changes its slope from negative to positive and crosses zero voltage level a number of times. After the data is filtered to eliminate the effects of noise (unwanted signal), the filtered EMG data is rectified by taking the absolute value of the filtered EMG signal. This step is presented in the middle graph in Figure11. After this step, the area under the rectified curve is calculated as the IEMG value.

To compare EMG signals from different lumbar levels and different participants, the IEMG values were normalized by dividing each 3-second integrated value by the value obtained from the first window over the time periods of at rest, loading, and recovery. The average normalized IEMG (NIEMG) values were then calculated for each minute throughout the rest and loading periods. For the recovery periods, two minute averages were calculated converting the IEMG values into a single NIEMG value.

Figure 12 shows a sample NIEMG data for at rest, loading, and recovery periods for each lumbar level. NIEMG values for all participants are provided in Appendix D.

To prepare the data set for the modeling process, the averaged NIEMG values for all the participants were pooled and averaged. The averaged NIEMG values at two lumber levels were used as the model input. Tables 2 and 3 show the NIEMG values and their averages for each participant at lumber levels L2 (channel 1) and L4 (channel 2), respectively.

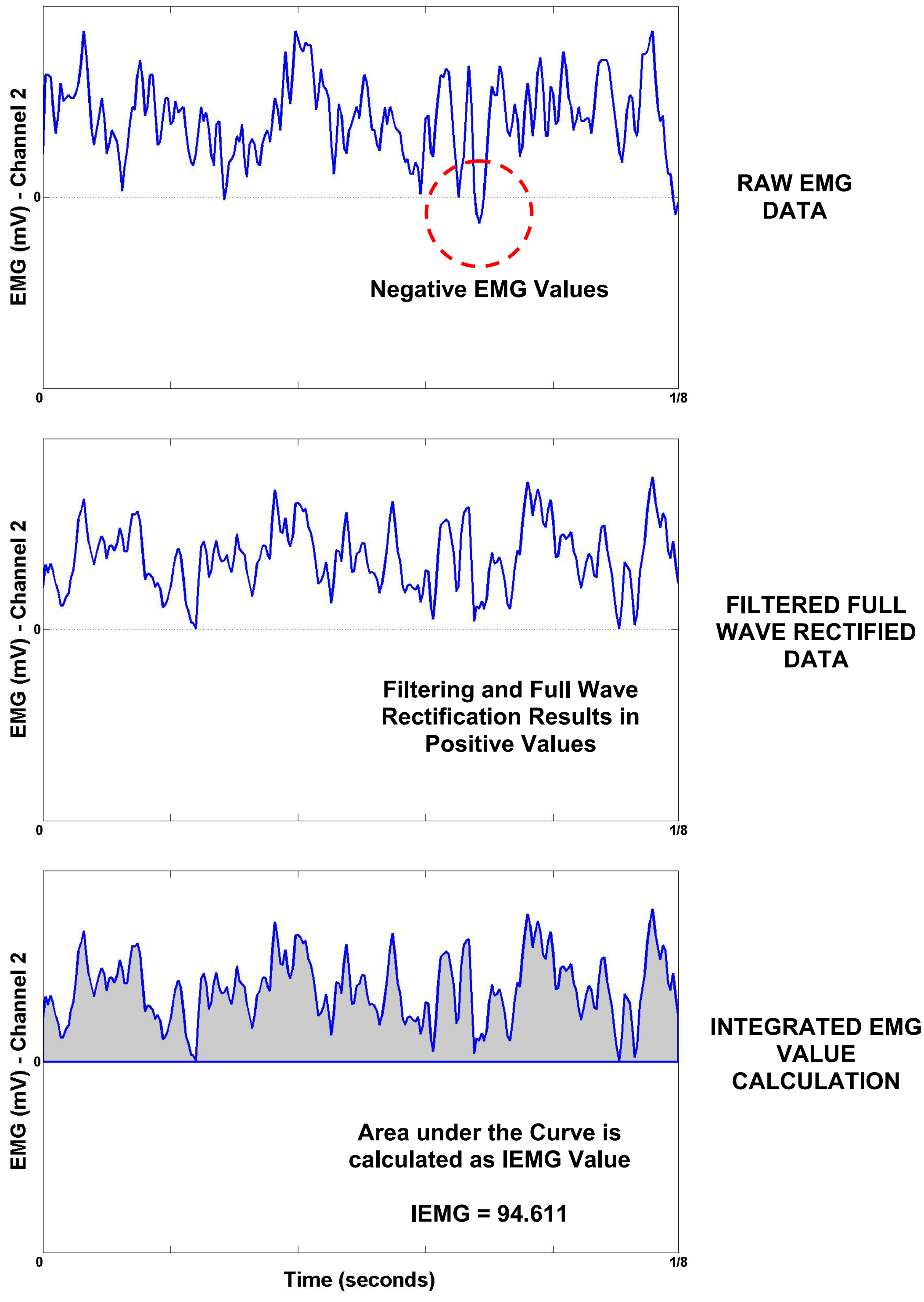

Figure 11. A Sample Full Wave Rectified Integration Procedure

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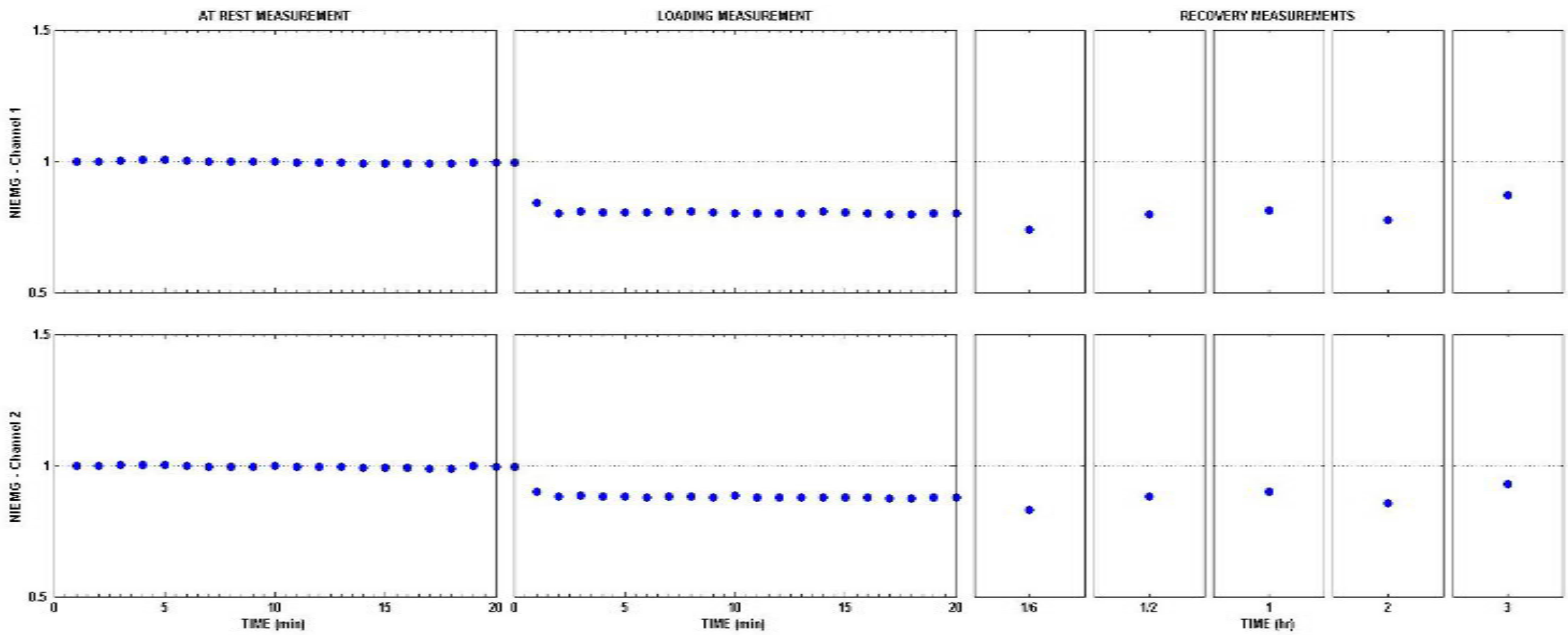

Figure 12. Sample Normalized Integrated EMG Data for At Rest, Loading and Recovery Periods

Table 2. NIEMG Values for All Participants - Channel 1

Period	Unit	Time	PARTICIPANT 1	2	3	4	5	6	7	8	9	10	Average	Std
At Rest	(min)	1	1.0005	1.0023	1.0111	1.0182	0.9982	0.6919	1.0020	0.9580	1.0023	1.0094	**0.9694**	**0.0988**
		2	0.9996	1.0072	0.9959	1.0074	0.9972	0.6056	1.0090	0.9417	0.9991	0.9992	**0.9562**	**0.1247**
		3	0.9981	1.0079	0.9943	1.0087	1.0034	0.6028	1.0102	0.9480	0.9989	0.9947	**0.9567**	**0.1256**
		4	0.9968	0.9848	1.0057	1.0162	1.0043	0.6671	1.0075	0.9544	0.9964	0.9945	**0.9628**	**0.1052**
		5	0.9968	0.9852	0.9888	1.0164	1.0048	0.6001	1.0080	0.9431	0.9958	0.9935	**0.9533**	**0.1257**
		6	0.9957	0.9863	0.9896	1.0383	1.0014	0.5995	1.0061	0.9498	0.9925	0.9960	**0.9555**	**0.1270**
		7	0.9949	0.9838	0.9881	1.0194	0.9975	0.6039	1.0010	0.9306	0.9930	0.9959	**0.9508**	**0.1240**
		8	0.9944	0.9833	0.9853	1.0003	0.9982	0.6055	1.0015	0.9274	0.9901	0.9916	**0.9478**	**0.1222**
		9	0.9939	0.9815	0.9850	0.9993	0.9966	0.6280	0.9989	0.9273	0.9889	0.9908	**0.9490**	**0.1148**
		10	0.9936	0.9826	0.9928	1.0000	0.9998	0.6065	0.9994	0.9349	0.9896	0.9906	**0.9490**	**0.1219**
		11	0.9924	0.9829	0.9910	1.0052	0.9962	0.6030	0.9987	0.9282	0.9889	0.9897	**0.9476**	**0.1229**
		12	0.9915	0.9818	1.0067	0.9994	0.9951	0.6009	0.9968	0.9317	0.9903	0.9893	**0.9483**	**0.1238**
		13	0.9912	0.9858	1.0170	0.9966	0.9947	0.6043	0.9969	0.9306	0.9869	0.9895	**0.9493**	**0.1232**
		14	0.9912	0.9899	1.0291	0.9973	0.9920	0.5972	0.9948	0.9300	0.9870	0.9883	**0.9497**	**0.1262**
		15	0.9904	0.9853	1.0542	0.9950	0.9922	0.5987	0.9954	0.9310	0.9914	0.9879	**0.9522**	**0.1276**
		16	0.9899	0.9889	1.0137	0.9934	0.9923	0.6007	0.9953	0.9280	0.9893	0.9857	**0.9477**	**0.1239**
		17	0.9894	0.9836	1.0003	0.9918	0.9892	0.5957	0.9958	0.9292	0.9871	0.9833	**0.9445**	**0.1242**
		18	0.9888	0.9874	0.9986	0.9922	0.9900	0.6171	0.9971	0.9368	0.9845	0.9828	**0.9475**	**0.1174**
		19	0.9882	0.9984	1.0172	0.9892	0.9929	0.6088	1.0076	0.9352	0.9834	0.9821	**0.9503**	**0.1219**
		20	0.9877	0.9978	0.9862	0.9925	0.9933	0.6001	1.0199	0.9350	0.9837	0.9835	**0.9480**	**0.1240**
Loading	(min)	1	0.8659	0.8001	0.7713	0.6899	0.8398	0.8482	0.7768	0.8832	0.7642	0.8488	**0.8088**	**0.0592**
		2	0.8632	0.7246	0.7469	0.6546	0.7995	0.8281	0.7479	0.7829	0.7436	0.8396	**0.7731**	**0.0623**
		3	0.8412	0.7073	0.7456	0.6370	0.8073	0.7684	0.7491	0.7635	0.7213	0.8397	**0.7580**	**0.0623**
		4	0.8386	0.6966	0.7594	0.6347	0.8052	0.7809	0.7599	0.7366	0.7122	0.8399	**0.7564**	**0.0645**
		5	0.8341	0.6708	0.7448	0.6355	0.8030	0.7644	0.7874	0.7283	0.7094	0.8396	**0.7517**	**0.0675**
		6	0.8352	0.6819	0.7487	0.6350	0.8036	0.7526	0.8116	0.7301	0.7106	0.8395	**0.7549**	**0.0679**
		7	0.8333	0.6994	0.7549	0.6350	0.8068	0.7560	0.8181	0.7320	0.7091	0.8391	**0.7584**	**0.0666**
		8	0.8330	0.6866	0.7642	0.6379	0.8066	0.7521	0.8387	0.7258	0.7102	0.8394	**0.7595**	**0.0699**
		9	0.8312	0.6839	0.7850	0.6381	0.8034	0.7513	0.8178	0.7253	0.7108	0.8391	**0.7586**	**0.0679**
		10	0.8319	0.6867	0.7449	0.6354	0.8021	0.7623	0.8239	0.7258	0.7074	0.8387	**0.7559**	**0.0684**
		11	0.8307	0.6645	0.7530	0.6367	0.7997	0.7602	0.8269	0.7272	0.7076	0.8386	**0.7545**	**0.0708**
		12	0.8304	0.6798	0.7690	0.6358	0.7996	0.7497	0.8195	0.7278	0.7115	0.8388	**0.7562**	**0.0680**
		13	0.8316	0.7305	0.7707	0.6444	0.8017	0.7481	0.8318	0.7254	0.7066	0.8383	**0.7629**	**0.0638**
		14	0.8304	0.7112	0.7762	0.6398	0.8069	0.7502	0.8418	0.7239	0.7074	0.8380	**0.7626**	**0.0676**
		15	0.8301	0.7278	0.7641	0.6556	0.8063	0.7453	0.8840	0.7251	0.7075	0.8390	**0.7685**	**0.0700**
		16	0.8294	0.7227	0.7585	0.6383	0.8020	0.7586	0.8675	0.7229	0.7062	0.8384	**0.7645**	**0.0704**
		17	0.8289	0.7302	0.7458	0.6402	0.7974	0.7623	0.8491	0.7248	0.7070	0.8376	**0.7623**	**0.0663**
		18	0.8287	0.6877	0.7478	0.6347	0.7980	0.7638	0.8236	0.7241	0.7076	0.8372	**0.7553**	**0.0676**
		19	0.8296	0.6651	0.7405	0.6445	0.8009	0.7686	0.8339	0.7240	0.7061	0.8374	**0.7550**	**0.0705**
		20	0.8279	0.6643	0.7596	0.6439	0.8014	0.7856	0.7497	0.7237	0.7053	0.8373	**0.7499**	**0.0658**
Recovery	(hr)	1/6	0.9164	0.9902	0.7958	0.6540	0.7385	0.8686	0.7903	0.8723	0.8316	0.9468	**0.8404**	**0.1006**
		1/2	0.8626	0.9972	0.7358	0.6571	0.7958	0.9619	0.8426	0.8110	0.9277	0.9307	**0.8523**	**0.1062**
		1	0.7652	0.9996	0.7576	0.7533	0.8105	0.9865	0.7463	0.8399	0.8945	0.9449	**0.8498**	**0.0997**
		2	0.9270	0.9959	0.8497	0.7305	0.7745	0.9668	0.6868	0.8833	0.9258	0.9270	**0.8667**	**0.1041**
		3	0.9634	0.9917	0.7079	0.6278	0.8690	0.8671	0.8957	0.8143	0.8938	0.9634	**0.8594**	**0.1157**

Mean and standard deviation of the NIEMG values were calculated and plotted on a signal versus time plot for each of the experiment periods. This averaged data set was used as the basis for the mathematical models. Figure 13 shows the averaged NIEMG values for all participants for at rest, loading, and recovery periods. In this figure, the y axis represents the

averaged NIEMG values and x-axis represents time. The vertical bars show the standard deviation of the data points.

Table 3. NIEMG Values for All Participants - Channel 2

			PARTICIPANT										Average	Std
			1	**2**	**3**	**4**	**5**	**6**	**7**	**8**	**9**	**10**		
At Rest	**(min)**	**1**	1.0010	1.0065	1.0160	1.0175	0.9995	0.7535	0.9991	0.9858	0.9983	1.0105	**0.9788**	**0.0797**
		2	1.0001	1.0061	0.9961	1.0082	0.9989	0.6541	1.0014	0.9767	0.9961	1.0043	**0.9642**	**0.1093**
		3	0.9988	1.0025	0.9962	1.0106	1.0009	0.6519	0.9992	0.9813	0.9969	0.9995	**0.9638**	**0.1098**
		4	0.9974	0.9801	1.0053	1.0157	1.0001	0.6949	0.9968	0.9902	0.9959	0.9994	**0.9676**	**0.0963**
		5	0.9967	0.9792	0.9945	1.0148	1.0000	0.6503	0.9964	0.9802	0.9933	0.9989	**0.9605**	**0.1094**
		6	0.9964	0.9787	0.9943	1.0263	0.9987	0.6746	0.9958	0.9862	0.9933	1.0032	**0.9648**	**0.1027**
		7	0.9946	0.9775	0.9935	1.0107	0.9945	0.6885	0.9938	0.9678	0.9943	1.0006	**0.9616**	**0.0967**
		8	0.9942	0.9752	0.9919	1.0039	0.9952	0.6761	0.9943	0.9591	0.9928	0.9967	**0.9579**	**0.0999**
		9	0.9936	0.9744	0.9911	1.0029	0.9945	0.6700	0.9925	0.9572	0.9903	0.9964	**0.9563**	**0.1014**
		10	0.9942	0.9731	0.9979	1.0035	0.9969	0.6541	0.9921	0.9599	0.9915	0.9961	**0.9559**	**0.1069**
		11	0.9921	0.9730	0.9923	1.0064	0.9948	0.6638	0.9924	0.9599	0.9895	0.9944	**0.9559**	**0.1034**
		12	1.0172	0.9714	1.0017	1.0025	0.9933	0.6516	0.9907	0.9644	0.9902	0.9934	**0.9576**	**0.1086**
		13	1.0128	0.9720	1.0029	1.0001	0.9930	0.6519	0.9902	0.9620	0.9880	0.9941	**0.9567**	**0.1081**
		14	0.9906	0.9750	1.0067	1.0002	0.9906	0.6477	0.9894	0.9599	0.9886	0.9924	**0.9541**	**0.1084**
		15	0.9899	0.9745	1.0156	0.9981	0.9900	0.6504	0.9887	0.9640	0.9954	0.9923	**0.9559**	**0.1082**
		16	0.9897	0.9771	1.0025	0.9952	0.9895	0.6472	0.9887	0.9707	0.9887	0.9900	**0.9539**	**0.1081**
		17	0.9889	0.9755	0.9947	0.9932	0.9869	0.6478	0.9885	0.9587	0.9896	0.9873	**0.9511**	**0.1071**
		18	0.9884	0.9750	0.9954	0.9930	0.9880	0.6605	0.9981	0.9728	0.9885	0.9868	**0.9547**	**0.1037**
		19	0.9876	0.9855	1.0121	0.9907	0.9986	0.6545	1.0046	0.9707	0.9875	0.9864	**0.9578**	**0.1072**
		20	0.9868	0.9788	0.9896	0.9938	0.9951	0.6539	1.0062	0.9701	0.9887	0.9869	**0.9550**	**0.1062**
Loading	**(min)**	**1**	0.8654	0.8616	0.8187	0.6647	0.8998	0.8771	0.8752	0.8423	0.8327	0.9111	**0.8449**	**0.0694**
		2	0.8582	0.8055	0.8042	0.6412	0.8809	0.8438	0.8594	0.7723	0.8160	0.9062	**0.8188**	**0.0741**
		3	0.8487	0.7926	0.8031	0.6312	0.8854	0.8114	0.8454	0.7637	0.8002	0.9061	**0.8088**	**0.0762**
		4	0.8473	0.7874	0.8082	0.6305	0.8818	0.8154	0.8393	0.7518	0.7968	0.9063	**0.8065**	**0.0767**
		5	0.8451	0.7781	0.8024	0.6311	0.8803	0.8067	0.8439	0.7479	0.7957	0.9063	**0.8037**	**0.0770**
		6	0.8459	0.7808	0.8037	0.6301	0.8795	0.7992	0.8448	0.7497	0.7969	0.9060	**0.8037**	**0.0770**
		7	0.8442	0.7884	0.8064	0.6293	0.8808	0.8009	0.8528	0.7525	0.7955	0.9055	**0.8056**	**0.0772**
		8	0.8438	0.7844	0.8111	0.6304	0.8808	0.8002	0.8618	0.7464	0.7962	0.9059	**0.8061**	**0.0783**
		9	0.8430	0.7816	0.8229	0.6326	0.8785	0.7990	0.8510	0.7462	0.7958	0.9055	**0.8056**	**0.0769**
		10	0.8428	0.7852	0.8023	0.6291	0.8846	0.8052	0.8593	0.7474	0.7946	0.9052	**0.8056**	**0.0786**
		11	0.8420	0.7735	0.8054	0.6296	0.8779	0.8037	0.8579	0.7487	0.7950	0.9052	**0.8039**	**0.0778**
		12	0.8411	0.7796	0.8137	0.6296	0.8783	0.7963	0.8504	0.7521	0.7962	0.9051	**0.8042**	**0.0769**
		13	0.8420	0.8063	0.8123	0.6349	0.8772	0.7957	0.8512	0.7457	0.7937	0.9049	**0.8064**	**0.0756**
		14	0.8412	0.7931	0.8175	0.6321	0.8778	0.7969	0.8560	0.7460	0.7944	0.9048	**0.8060**	**0.0767**
		15	0.8407	0.7997	0.8130	0.6413	0.8788	0.7938	0.8834	0.7461	0.7946	0.9053	**0.8097**	**0.0770**
		16	0.8402	0.8003	0.8112	0.6319	0.8778	0.8017	0.8687	0.7436	0.7938	0.9048	**0.8074**	**0.0778**
		17	0.8396	0.7969	0.8030	0.6329	0.8753	0.7989	0.8576	0.7457	0.7943	0.9043	**0.8049**	**0.0761**
		18	0.8391	0.7812	0.8036	0.6291	0.8755	0.8015	0.8426	0.7443	0.7942	0.9037	**0.8015**	**0.0764**
		19	0.8389	0.7725	0.7987	0.6393	0.8761	0.8014	0.8441	0.7447	0.7934	0.9039	**0.8013**	**0.0743**
		20	0.8379	0.7716	0.8108	0.6381	0.8773	0.8128	0.8127	0.7445	0.7932	0.9037	**0.8003**	**0.0736**
Recovery	**(hr)**	**1/6**	0.9411	0.9887	0.8731	0.6487	0.8321	0.9209	0.8584	0.8894	0.8874	0.9761	**0.8816**	**0.0960**
		1/2	0.8873	0.9884	0.8023	0.6673	0.8797	0.9812	0.8956	0.8327	0.9655	0.9598	**0.8860**	**0.0995**
		1	0.8093	0.9796	0.8356	0.7443	0.9003	0.9920	0.8154	0.8547	0.8640	0.9745	**0.8770**	**0.0831**
		2	0.9321	0.9596	0.9006	0.7354	0.8554	1.0044	0.7830	0.9064	0.9363	0.9321	**0.8945**	**0.0819**
		3	0.9703	0.9814	0.7873	0.6223	0.9292	0.9063	0.9406	0.8463	0.8781	0.9703	**0.8832**	**0.1103**

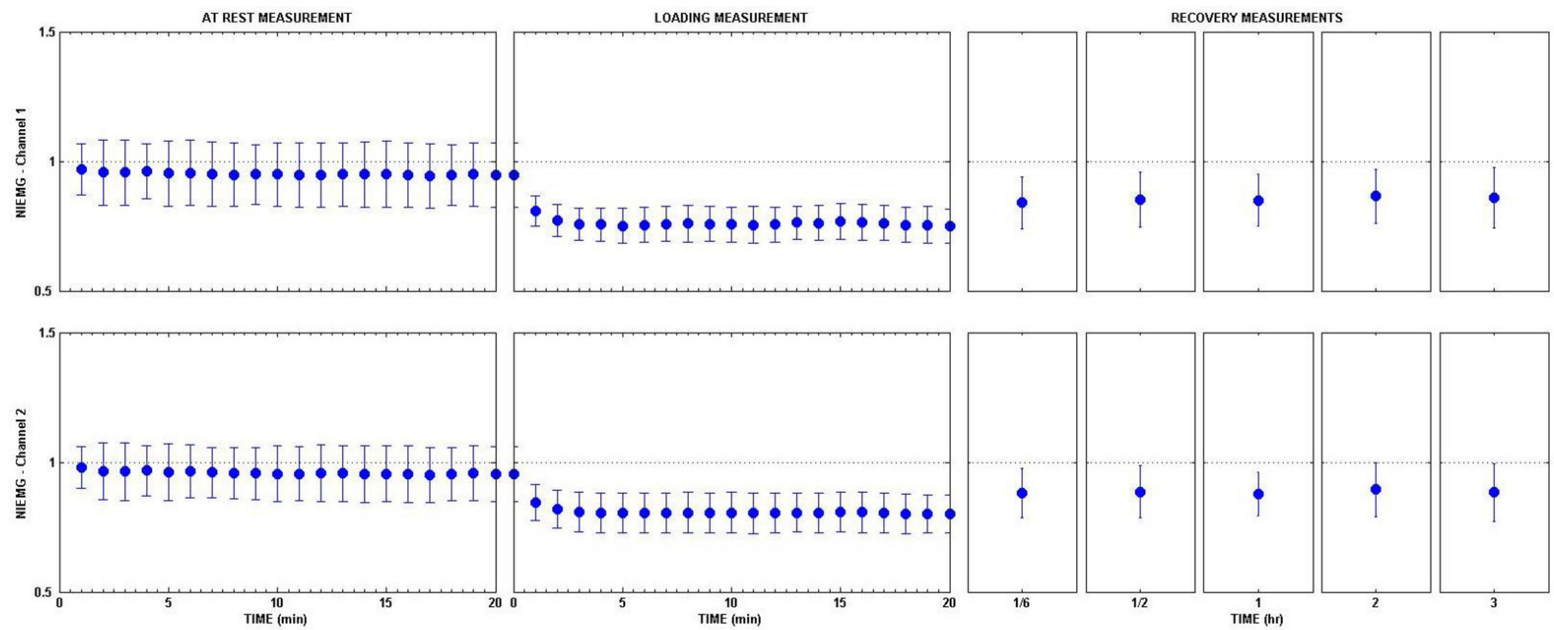

Figure 13. Averaged Normalized Integrated EMG Values for At Rest, Loading and Recovery Periods

CHAPTER 5 - MODELING AND ANALYSIS OF RESULTS

In this chapter, the observations from the collected EMG data are summarized followed by the hypothesis testing of the two phenomena based on the NIEMG values. In addition, the NIEMG values for the loading and recovery periods are analyzed and the muscle behavior is described using bi-exponential mathematical models. The significance of the experiment variables on the change of the behavior are tested to ensure the validity of the models. Analysis of muscle fatigue using median frequency values during the loading period is also included in this chapter.

5.1. OBSERVATIONS AND HYPOTHESIS TESTING

In general, the average NIEMG values of the erector spinae muscles of the participants demonstrate an exponential decrease for channel 1 (L2) and channel 2 (L4) lumbar levels during the 20-minute loading period, decreasing to 74.99 and 80.03 percent from their initial value of one, respectively (Table 2, Table 3). At the first minute of the loading period the average NIEMG values for L2 and L4 were 80.88 percent and 84.49 percent, respectively. There was a decrease of 5.89 percent and 4.46 percent for L2 and L4, respectively from the first minute of loading. Most of the exponential decrease occurred during the initial 3 minutes of the static loading, where the average NIEMG values decreased from 80.88 percent to 77.31 percent during the second minute of loading, and to 75.80 percent during the third minute of loading for L2. The total decrease in three minutes for L2 was 5.89 percent. The average NIEMG values for L4 decreased from 84.49 percent to 81.88 percent during the second

minute of loading, and to 80.88 percent during the third minute of loading. The total decrease in average NIEMG values for L4 was 3.61 percent.

A number of spasms were observed in the EMG data for each participant during the 20-minute static loading period with various amplitudes and frequencies. These spasms were random and unpredictable which are expected in any EMG data recording (Solomonow et al., 2000; Jackson et al., 2001). Figure 14 shows sample raw EMG outputs with high and low magnitude discharges with spasms at different amplitudes to illustrate the range of the collected data. Samples (a) and (b) were collected from two different participants during the 20-minute loading period.

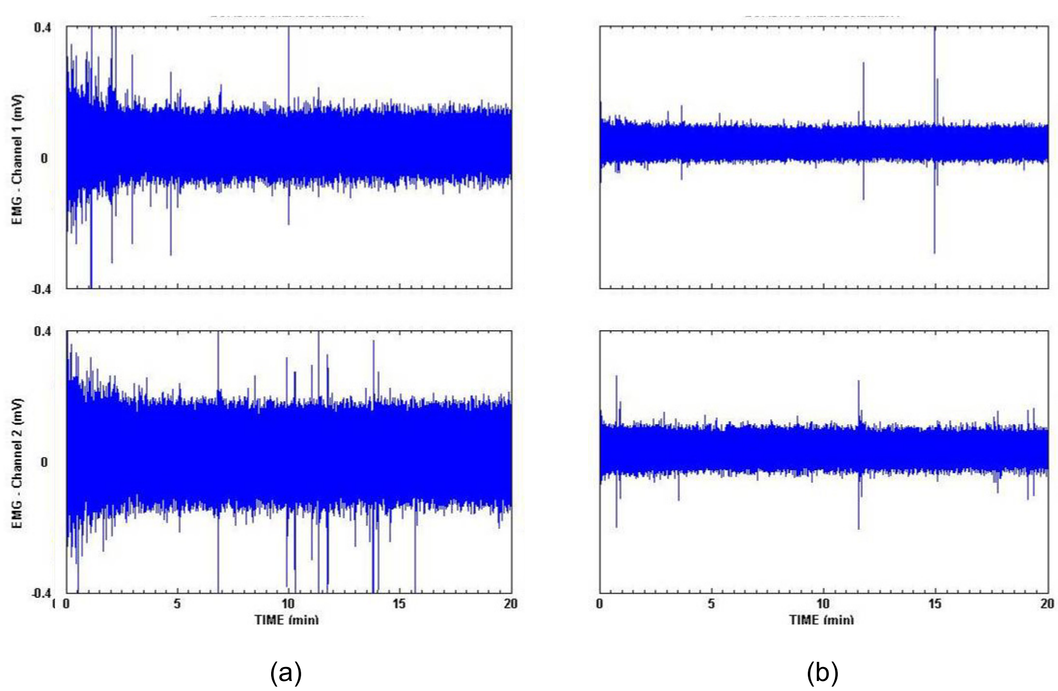

Figure 14. Sample Raw EMG Readings for High (a) and Low (b) Magnitude Discharges

During the recovery period, EMG values gradually increased over time which indicated that the recovery of the reflexive muscular activity has started. As shown in Table 2, the average NIEMG values for L2 increased to 84.04 during the first 10 minutes of recovery from 74.99 percent, an increase of 9.05 percent. After thirty minutes of rest, the average NIEMG value increased to 85.23 percent. At the end of one hour, two hour, and three hour recovery periods the average NIEMG values were 84.98 percent, 86.67 percent, and 85.94 percent, respectively. The total recovery for L2 was 10.95 percent.

At L4 level, the average NIEMG values increased to 88.16 percent during the first ten minutes of recovery from 80.03 percent, an increase of 8.13 percent (Table 3). After thirty minutes of rest, the average NIEMG value for L4 increased to 88.60 percent. At the end of one hour, two hour, and three hour recovery periods, the average NIEMG values were 87.70 percent, 89.45 percent, and 88.32 percent, respectively. The total recovery for L4 was 9.29 percent. A transitory hyperexcitability was observed for all participants during the first 30 minutes of the recovery.

Based on the NIEMG values, two phenomena were tested that were defined as:

Phenomenon I:

H_0: The reflexive muscle activity to stabilize the spine will does not decrease over time throughout the loading period

H_1: The reflexive muscle activity to stabilize the spine decreases over time throughout the loading period

Phenomenon II:

H_0: The muscle will fully recover at the end of the three hour recovery period

H_1: The muscle will not recover fully at the end of the three hour recovery period

The decision criteria for both phenomena are the comparison of the average NIEMG values during the at rest, loading and recovery periods that are presented in Table 2 and 3. It is assumed that a full recovery of the muscle occurs when the average NIEMG values return to the value of 1.

As illustrated in Figure 13 and Appendix D (for all participants), the average NIEMG values, which were used to evaluate the reflexive muscular activity to stabilize the spine, decreased exponentially over time throughout the loading period. For channel 1 (L2), the average NIEMG value at the end of the rest period was 94.80 percent, which decreased to 74.99 percent at the end of the loading period. For channel 2 (L4), the average NIEMG value at the end of the rest period was 95.50 percent, which decreased to 80.03 percent at the end of the loading period. Therefore, the null hypothesis for the first phenomenon is rejected. In other words, the reflexive muscle activity to stabilize the spine decreased over time throughout the loading period.

The hypothesis testing of the second phenomenon requires the comparison of the average NIEMG values before and after the recovery period. Although the average NIEMG values at the end of the recovery period were higher than the values at the end of the loading period, the values did not reach the full recovery value of 1. For channel 1 (L2), the average NIEMG value at the end of the loading period was 74.99 percent, which increased to 85.94 percent at the end of the 3-hour recovery period. For channel 2 (L4), the average NIEMG value at the end loading period was 80.03 percent, which increased to 88.32 percent at the end of the 3-hour recovery period. Therefore, the null hypothesis for the second phenomenon is also rejected. In other words, the muscles did not recover fully at the end of the 3-hour recovery period.

It should be noted that there is a magnitude difference between the average NIEMG values for L2 and L4 lumbar levels. L2 and L4 levels were chosen to observe high and low load bearing lumbar levels. L2 level is second highest load bearing and L4 is a lesser load bearing lumbar level. In Figure 13 and Tables 2 and 3, the decrease in the average NIEMG values during the loading period was higher for L2. The average NIEMG value at the end of the 20-minute rest period for L2 was 94.80 percent, which decreased to 80.88 percent during the first minute of loading. At the end of the loading period, the average NIEMG value for L2 was 74.99 percent. The average NIEMG values decreased a total of 19.81 percent. For L4, the average NIEMG value at the end of the 20-minute rest period was 95.50 percent, which decreased to 84.99 percent during the first minute of loading. At the end of the loading period, the average NIEMG value for L4 was 80.03 percent. The average NIEMG values decreased a total of 15.47 percent. These results support the original concept of observing different load bearing levels. Since L2 level has been exposed to a higher magnitude of load, the decrease in the average NIEMG value for that level is higher compared to L4.

Similarly, the total recovery of the erector spinae muscle activity was 10.95 percent and 8.29 percent for L2 and L4 lumbar levels, respectively (Table 2 and Table 3). This result corroborates with the observations of animal experiments (Hatipkarasulu, 2002; Solomonow et al., 2003) that the lumbar levels which are subjected to higher magnitudes of load show a higher initial recovery rate.

5.2. LOADING MODEL

An exponential structure was chosen for the loading model as it represents the classical response of viscoelastic tissues to loads and/or elongation. The average NIEMG data of the two lumbar levels from the erector spinae muscle were fitted to the model, in the form

of exponential functions. The model structure is similar to the ones developed by Solomonow et al. (2000) and Jackson et al. (2001), which takes the form shown in Equation (12).

$$NIEMG_{(t)} = Ae^{-t/T_1} + NIEMG_o \tag{12}$$

where,

$NIEMG_{(t)}$ = Normalized integrated EMG as a function of time

A = Exponential component initial amplitude

T_1 = Exponential decay rate constant (minute)

$NIEMG_o$ = Steady state NIEMG amplitude

t = Time (minute)

The model defined in Equation (12) was applied to the average NIEMG data set for both of the lumbar levels. The parameters for this model were obtained by using a Marquardt-Levenberg non-linear regression algorithm (Marquardt, 1963; Levenberg, 1944). The graphical interface in "Curve Fitting Toolbox" of Matlab software is used for the modeling process (Matlab (d), 2005).

The final point of the loading at the end of the 20 minute period was used directly as the $NIEMG_o$ parameter and the rest of the parameters were estimated through iteration. R^2 values were calculated using the difference between the collected data set and the values generated by the model. Figures 15 and 16 show the graphical representation and the fit of the model for channel 1 and channel 2. The data points represent the averaged NIEMG values and the function curve represents the model fit.

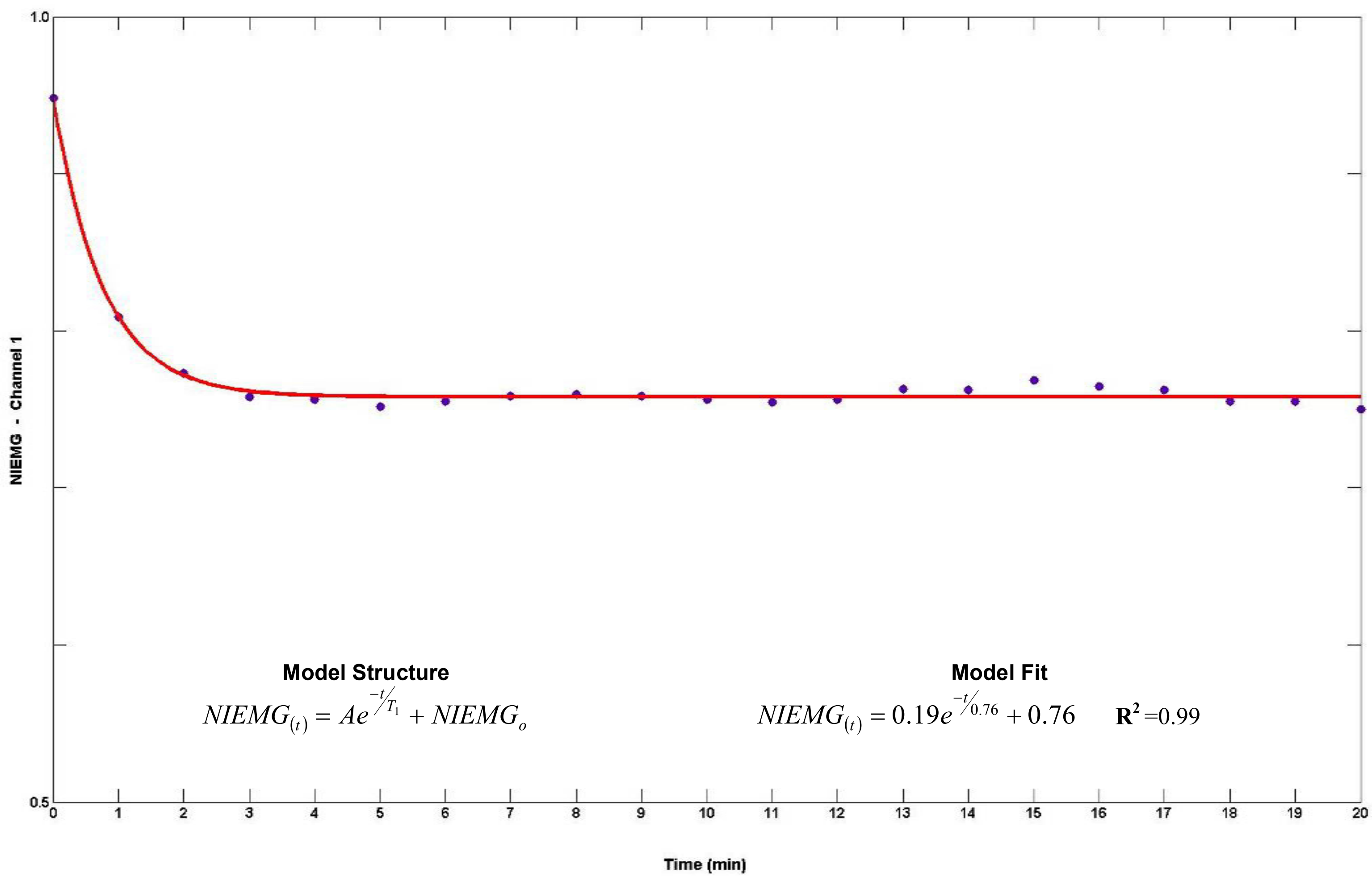

Figure 15. Loading Model using Averaged NIEMG Values - Channel 1

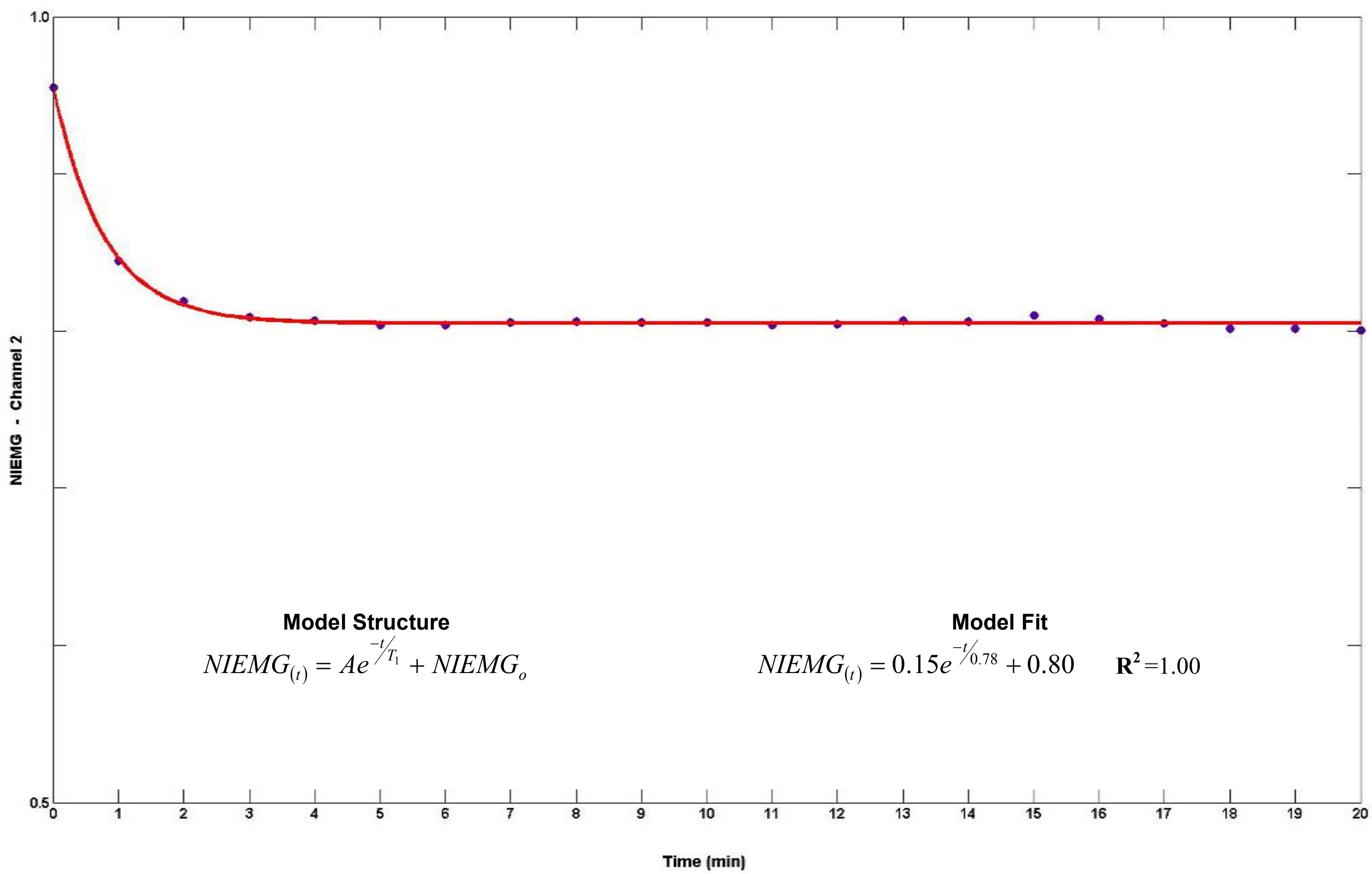

Figure 16. Loading Model using Averaged NIEMG Values - Channel 2

Considering the high range of the data values due to occurrence of spasms, 80 percent or better R^2 values are considered a good fit. The calculated parameters for channel 1 and 2 are presented in Table 4.

Table 4. Loading Model Parameters using Averaged NIEMG Values

	Channel 1	Channel 2
$NIEMG_0$ **(95% Conf. Interval)**	0.7509 (0.7555 – 0.7602)	0.8049 (0.8037 – 0.8060)
A **(95% Conf. Interval)**	0.1902 (0.1801 – 0.2002)	0.1499 (0.1450 – 0.1549)
T_1 **(95% Conf. Interval)**	0.7567 (0.6526 – 0.8607)	0.7785 (0.7127 – 0.8442)
R^2	0.9891	0.9957

The R^2 values for the fit of channel 1 and channel 2 are 0.9891 and 0.9957, respectively, indicating an almost perfect fit of the model. These values confirm that the exponential model structure represents the behavior of the muscles during loading accurately.

5.3. RECOVERY MODEL

An exponential model structure was also chosen to describe the behavior during the 3 hour recovery period. Through animal experiments, bi-exponential models were previously developed by Gedalia et al. (1999), Solomonow et al. (2000), and Hatipkarasulu (2002) which included time dependent components. These studies demonstrated that the recovery of the reflexive muscular activity is composed of two phases with two different time constants: a fast recovery phase and a slow recovery phase. The fast recovery component represents the

recovery behavior of the ligamentous structures while the slow recovery component describes the recovery of the discs. Two different components were chosen because although the ligaments and discs are both viscoelastic in nature, their behavior over time differs. The ligaments are composed of collagen fibers and the discs are composed of fluid, which are enclosed by collagen. When subject to loading, the fluid content of the disc decreases slowly and the restoration of this fluid loss with rest is also slow when compared to the ligamentous structures (Solomonow et al., 2000). Based on this explanation and to test for the similarities of the muscle behavior between the animal models and humans, a two component model was fitted to the collected data defined as:

$$NIEMG_{(t)} = B(1 - e^{-t/T2}) + tCe^{-t/T3} + NIEMG_o \qquad (13)$$

where,

$B(1 - e^{-t/T2})$ = The steady state recovery component (slow recovery component)

$tCe^{-t/T3}$ = Transient hyperexcitability component (fast recovery component)

$NIEMG_o$ = Residual response at the end of the 20-minute loading period

t = Time

The graphical representation of this model is presented in Figure 17. The overall model is the summation of the fast and slow recovery components, and the residual response at the end of the 20 minute loading period. As mentioned before, the fast recovery component represents the recovery of the visco-elastic structures and is characterized by $tCe^{-t/T3}$. It is also referred to as the hyperexcitability component. The slow recovery component represents the recovery of the discs (restoration of the disc's fluid) and is characterized by $B(1 - e^{-t/T2})$.

It is also referred as the steady state component since the increase in average NIEMG values happen over longer periods of time with a slower recovery rate.

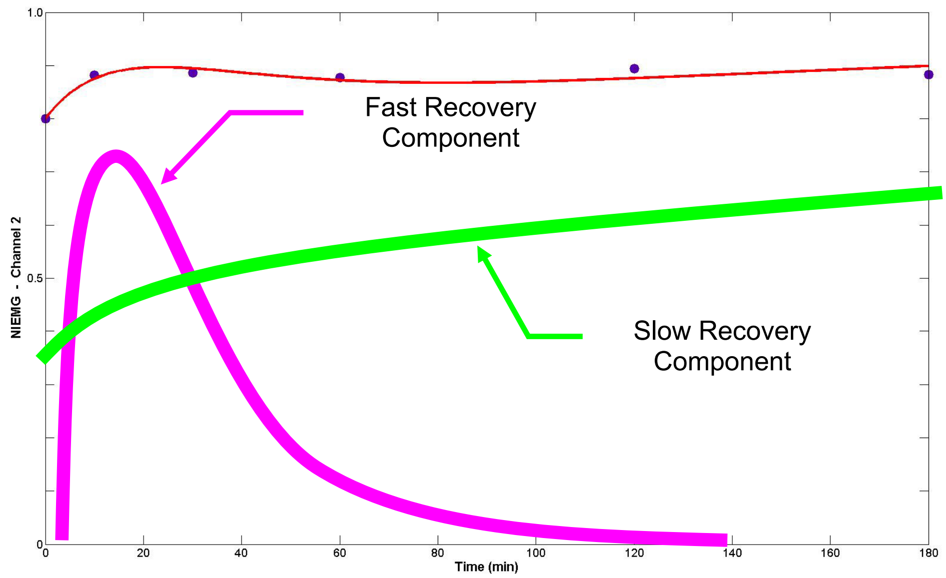

Figure 17. Recovery Model Components

In Equation 13, the difference between the total recovery value (1.0) and the residual response at the end of the loading period ($NIEMG_o$) was used as the steady state coefficient (B). In other words, the constraint of ($B + NIEMG_o = 1$) was used to insure that full recovery results in a normal response. These initial parameters were applied in a Marquardt-Levenberg non-linear regression algorithm for final iterations using Matlab's "Curve Fitting Toolbox" (Matlab (d), 2005). The model components were highly dependent upon each other; therefore, once the model format was set, the fits were relatively insensitive to changes in parameters. It should also be noted that the parameters shown here are approximate in nature. They reflect the overall behavior of the recovery response and are not precise determinations of this highly

variable phenomenon. Some of the model parameters are out of norm with others, likely due to larger deviation from the norm caused by frequent and severe occurrence of spasms.

The model parameters defined in Equation (13) are presented in Table 5. Figures 18 and 19 show the graphical representation and the fit of the model for channel 1 and channel 2 respectively. The data points represent the average NIEMG values and the function curve represents the model fit.

Table 5. Recovery Model Parameters using Averaged NIEMG Values

	Channel 1	**Channel 2**
$NIEMG_o$	0.7499	0.8002
B	0.2501	0.1998
T_2 (95% Conf. Interval)	261.2000 (126.3 – 396.1)	264.2000 (111.2 – 417.2)
C (95% Conf. Interval)	0.01121 (0.00147 – 0.02095)	0.01123 (0.00087 – 0.02159)
T_3 (95% Conf. Interval)	22.1100 (7.4620 – 36.7700)	19.6100 (6.3030 – 32.9200)
R^2	0.9105	0.8800

In Figures18 and 19 and Table 5, the R^2 values calculated for the goodness of the fit for channel 1 (L2) and channel 2 (L4) are 0.9105 and 0.8800, respectively. These are over 80 percent value and the model can be considered to be a good fit for both of the channels. Based on this result, it can be concluded that the models developed based on animal experiments captures the behavior of human muscle behavior properly.

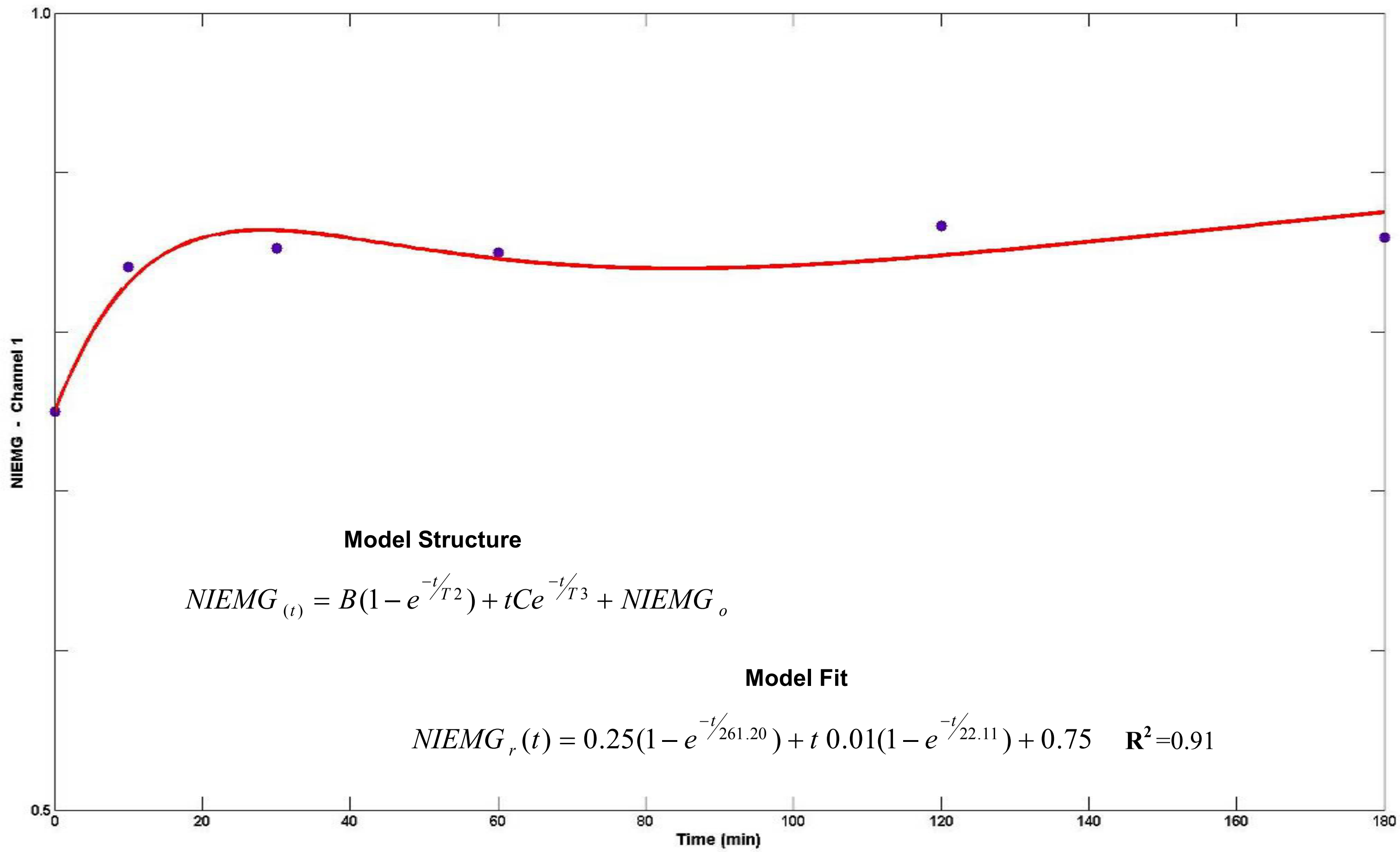

Figure 18. Recovery Model using Averaged NIEMG Values - Channel 1

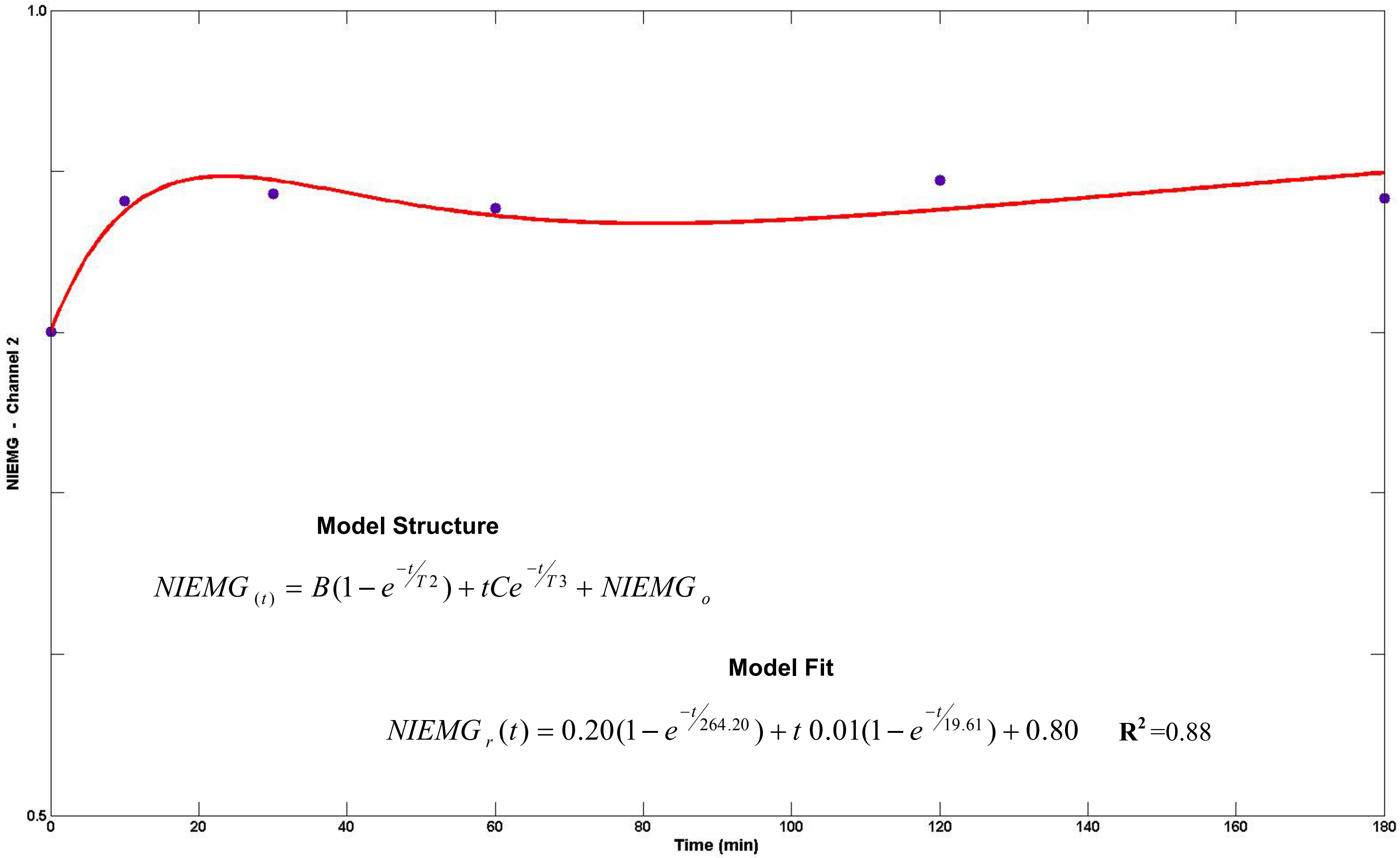

Figure 19. Recovery Model using Averaged NIEMG Values - Channel 2

The model structure in Equation (13) was developed based on the data collected from in vivo feline experiments in highly controlled environments. During the recovery periods, the back of the felines were not exposed to any additional load. However, full control of the exposure against loading is very difficult for human experiments.

The human back musculature is exposed to load continuously throughout the day, regardless of the activity and posture. This continuous load exposure is expected to have a negative impact on the progression of recovery and the recovery rate of the muscles. The two component animal models do not take this factor into account. To model the human muscle behavior recorded in this study, a third component was needed to account for the continuous load exposure.

5.4. ADJUSTED RECOVERY MODEL

A third component was added to the model structure defined in Equation (13) to represent the continuous loading effect on the lower back. The adjusted recovery model with the continuous loading component, which is referred as the "daily activity factor", defined as:

$$NIEMG_{(t)} = B(1 - e^{-t/T2}) + tCe^{-t/T3} - tDe^{t/T4} + NIEMG_o \tag{14}$$

where,

$B(1 - e^{-t/T2})$ = The steady state recovery component (slow recovery component)

$tCe^{-t/T3}$ = Transient hyperexcitability component (fast recovery component)

$tDe^{t/T4}$ = Continuous loading component (daily activity factor)

$NIEMG_o$ = Residual response at the end of the 20-minute loading period

t = Time

The graphical representation of this model is presented in Figure 20. The overall model is the summation of the first, second, and third components.

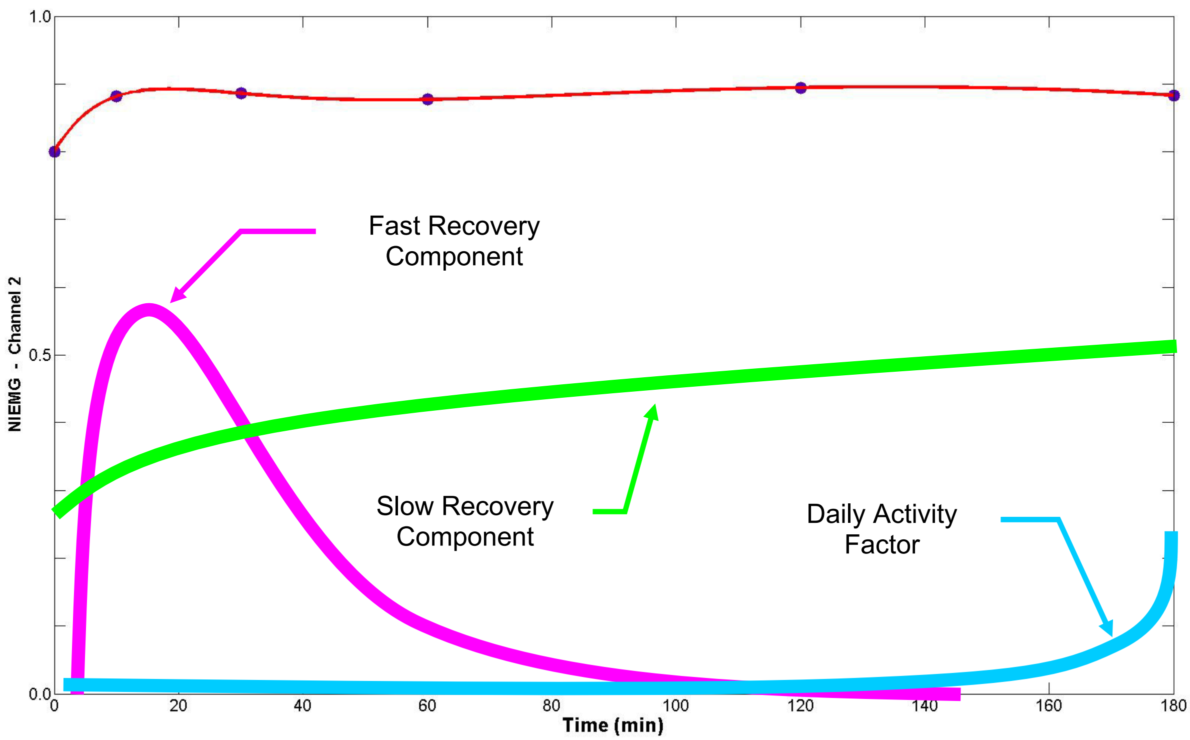

Figure 20. Adjusted Recovery Model Components

In Equation 14, the difference between the total recovery value (1.0) and the residual response at the end of the loading period ($NIEMG_o$) was used as the steady state coefficient (B) to insure that full recovery results in a normal response. These initial parameters were applied in a Marquardt-Levenberg non-linear regression algorithm for final iterations using Matlab's "Curve Fitting Toolbox" (Matlab (d), 2005). Figures 21 and 22 show the graphical representation and the fit of the model for channel 1 and channel 2, respectively.

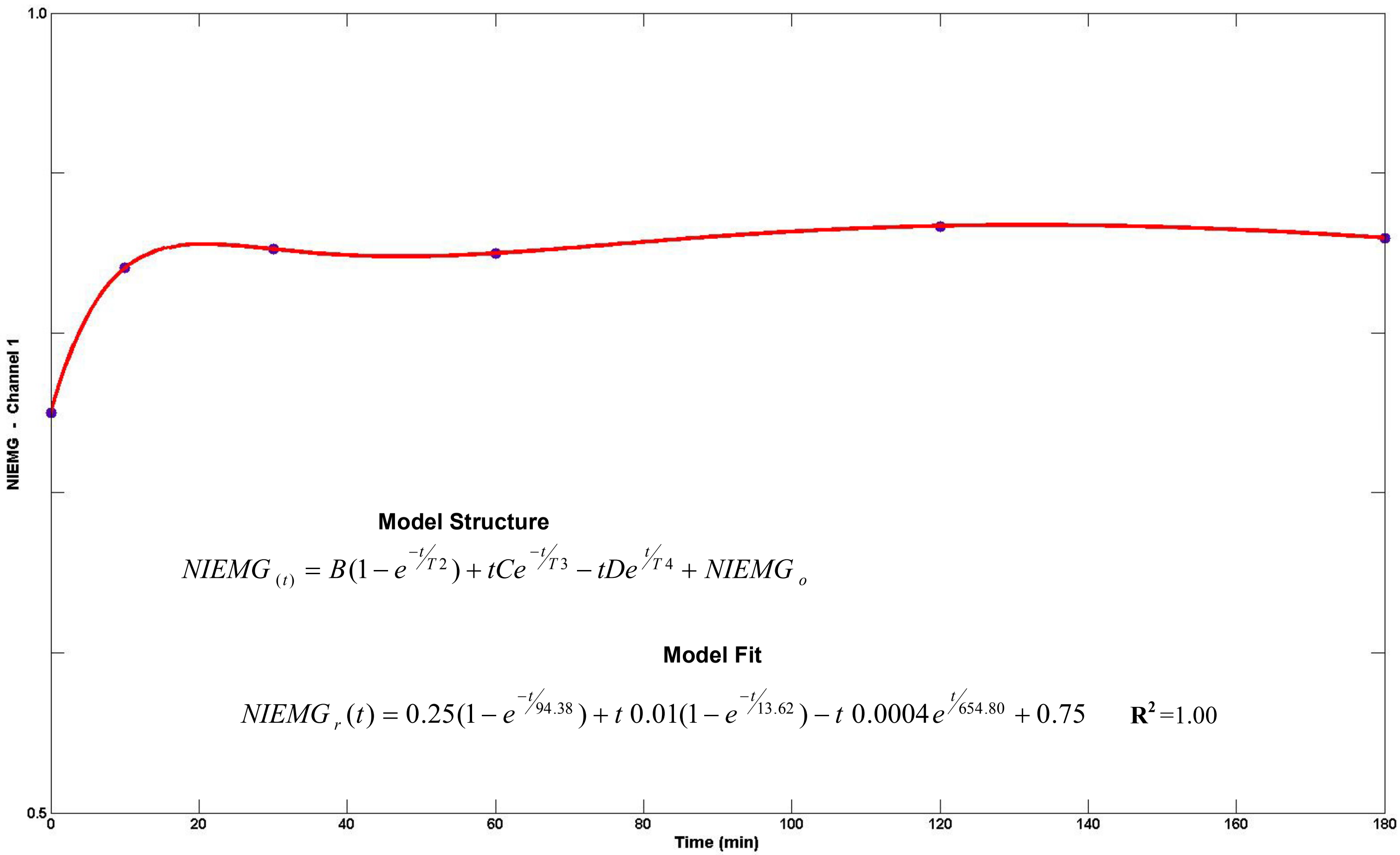

Figure 21. Adjusted Recovery Model using Averaged NIEMG Values - Channel 1

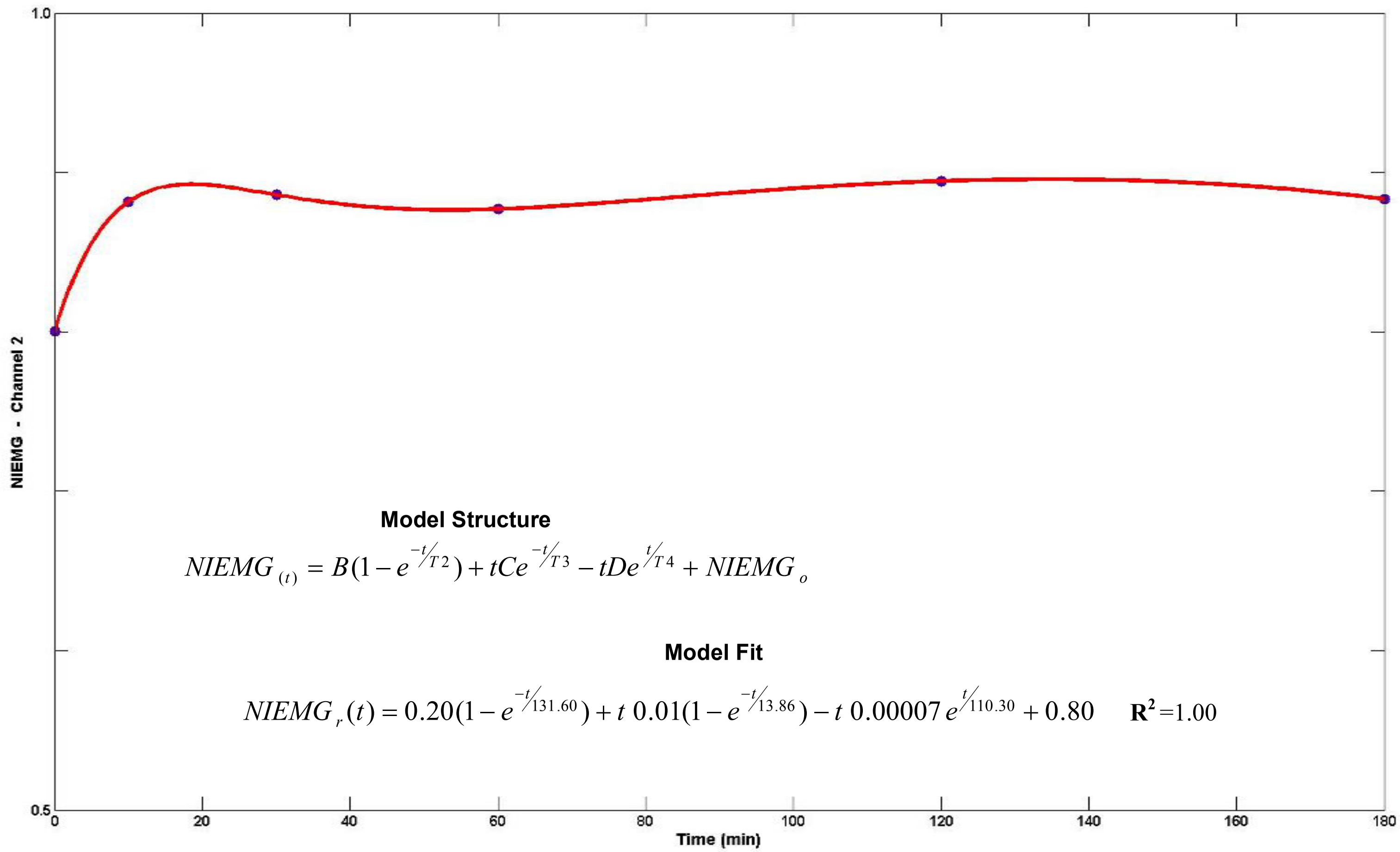

Figure 22. Adjusted Recovery Model using Averaged NIEMG Values - Channel 2

The adjusted NIEMG model parameters are presented in Table 6. With the adjusted model, the R^2 values increased to 100% and 100% from 91% and 88% for channel 1 and channel 2, respectively.

Table 6. Adjusted Recovery Model Parameters using Averaged NIEMG Values

	Channel 1	**Channel 2**
$NIEMG_o$	0.7499	0.8002
B	0.2501	0.1998
T_2 (95% Conf. Interval)	94.3800 (15.97 – 172.8)	131.6000 (131 – 132.2)
C (95% Conf. Interval)	0.0145 (0.0119 – 0.01719)	0.0139 (0.01397 – 0.01390)
T_3 (95% Conf. Interval)	13.6200 (10.1600 – 17.0900)	13.8600 (13.8400– 13.8800)
D (95% Conf. Interval)	0.00043 (-0.00070 – 0.00157)	0.000007 (0.0000069 – 0.0000074)
T_4 (95% Conf. Interval)	654.8000 (-4226.0000 – 5535.0000)	110.0300 (108.5000 – 112.1000)
R^2	1.0000	1.0000

Figures 21 and 22 clearly show that there is a slight decrease in the NIEMG values at the end of the 3- hour rest period. The average NIEMG value for L2 at the end of the 2 hour recovery period was 86.67 percent. At the end of the 3 hour recovery period it showed a

decrease of 0.73 percent, reaching to 85.94 percent. For L4, the average NIEMG value decreased from 89.45 percent to 88.32 percent, a decrease of 1.13 percent. This can be explained by taking into account that although the participants' backs were resting at the back rest of the chair, their lower backs were exposed to some amount of force since they were still carrying some of the weight from the upper extremities. Evidently, after sitting continuously for three hours, the sitting task itself becomes the static load on the lower back and creates a negative effect on the speed and behavior of the recovery. This is a significant difference from the animal models and an important finding of this study. It should also be noted that although the average NIEMG values show a decreasing trend at the end of the recovery period, how they will change over extended hours of rest beyond three hours is still unclear and unpredictable due to the highly dynamic nature of the data.

5.5. STATISTICAL ANALYSIS OF VARIABLES

To determine whether age, height, weight, and back strength of the participants, and time are significant factors on the changes in the electrical activity in the two lumbar levels, a repeated analysis of variance test was performed for both the loading and recovery periods. The SAS Statistical Analysis System Release 9.1 was used to perform this analysis. A PROC MIXED procedure, with a significance level of 0.05 (α=0.05) was applied for this purpose since it provides a variety of covariance structures such as the random-effect parameters, which are additional unknown random variables assumed to influence the variability of the data. In this case, each participant is considered as the random-effect parameter. Age, height, weight, back strength, and time factors are considered as fixed effects. Summary of the results obtained for the 20-minute loading and 3-hour recovery periods are presented in Tables 7 and 8, respectively.

Table 7. Results of the Repeated Analysis of Variance for Loading Period

Independent Variables	Channel 1		Channel 2	
	Pr > F	F Value	Pr > F	F Value
Age	0.7279	0.12	0.9422	0.01
Height	0.8210	0.05	0.7305	0.12
Weight	0.8875	0.02	0.7207	0.13
Back Strength	0.8557	0.03	0.9312	0.01
Time	<.0001	4.32	<.0001	6.88

Table 8. Results of the Repeated Analysis of Variance for Recovery Period

Independent Variables	Channel 1		Channel 2	
	Pr > F	F Value	Pr > F	F Value
Age	0.1159	2.57	0.3315	0.96
Height	0.4347	0.62	0.7225	0.13
Weight	0.5659	0.33	0.8218	0.05
Back Strength	0.2391	1.43	0.5457	0.37
Time	0.0047	3.71	0.0007	5.25

The results presented in Tables 7 and 8 show that only time had a significant effect on the average NIEMG values for both of the lumbar levels during the loading and recovery periods with a Pr>F value of <.0001 for channel 1 and a Pr>F value of <.0001 for channel 2 during loading; and a Pr>F value of 0.0047 for channel 1 and a Pr>F value of 0.0007 for channel 2 for the recovery period. Age, height, weight, and back strength factors did not have a significant effect with a significance level greater than α=0.05.

It should be noted that the participants in this experiment have limited population characteristics and the statistical significance analysis results represent this limited population which also had a small sample size. It is possible to observe effects of some of these factors for a larger range of values. For example, the ages of the participants in this experiment were between 23 and 36 years and the age did not have a significant effect on the average NIEMG values. It is possible that this result may change if the population were to include other participants changing the range to 15 to 65 years of age.

To measure whether the change in the NIEMG values at the beginning and at the end of the loading period was statistically significant, a paired t-test was performed on the beginning of the loading and the end of the loading NIEMG values of each of the participant as before and after values. The results of the paired t test analysis are summarized in Table 9. The Pr values for both channels are smaller than the confidence level of 0.05; therefore, the change in NIEMG values between the beginning of the loading period and the end of the loading period is statistically significant for both lumbar levels.

Table 9. Results of Paired T-Test Analysis of Loading NIEMG Values

	Degree of Freedom	**Pr > | t |**	**t Value**
Channel 1	9	0.0032	3.98
Channel 2	9	0.0049	3.71

To measure whether the change in the NIEMG values at the end of the loading period and the end of the rest period was statistically significant, a paired t-test was performed on the end of the loading and the end of the recovery NIEMG values of each of the participant as before and after values. The results of the paired t test analysis are summarized in Table 10.

The Pr values for both channels are smaller than the confidence level 0.05; therefore, the change in NIEMG values between the end of the loading period and the end of the rest period is statistically significant for both lumbar levels.

Table 10. Results of Paired T-Test Analysis of the Loading and Recovery NIEMG Values

	Degree of Freedom	Pr > | t |	t Value
Channel 1	9	0.0096	-3.28
Channel 2	9	0.0043	-3.78

5.6. ANALYSIS OF FATIGUE

EMG data can be used to evaluate muscle fatigue and a decline in median frequency (MDF) is an accepted fatigue indicator among others (Koumantakis et al., 2001). In this study, the median frequency calculations were performed using Delsys Bagnoli 2 EMG Software. The MDF values were calculated from the filtered data sets for the 20-minute loading period using a 3-second moving window, which made it possible to apply the calculation to the any desired length of data and repeat the calculation for the entire data series.

The MDF calculations consisted of three major steps. First, a Fast Fourier Transform (FFT) was calculated for the data within 3-second windows. The Power Spectrum Density (PSD) of the window was then determined by squaring the FFT and determining the magnitude. Finally, the MDF values were calculated by determining the frequency that divided the PSD in two regions that have the same amount of power.

Similar to the EMG integration process, the MDF values were calculated for every 3 seconds throughout the loading period. Average MDF values were than calculated for each minute for each participant. The average MDF values for each participant are provided in Table 11.

Table 11. Averaged Median Frequency Values for All Participants

		PARTICIPANT									
		1	2	3	4	5	6	7	8	9	10
Loading Period MDF Values - Channel 1 (min)	1	103.1905	76.0635	78.4762	49.0159	75.8254	53.4603	66.8254	48.0635	97.8730	112.2698
	2	96.8333	73.9000	89.4167	53.0667	57.7500	48.6833	64.5667	45.6167	103.4000	126.0667
	3	105.9167	80.4167	89.2500	59.1333	55.2333	65.1667	67.0333	48.7833	98.8500	124.9667
	4	102.7500	75.8000	72.4000	59.9167	51.3667	63.2333	61.7333	50.5500	92.9000	126.6500
	5	106.3167	78.3333	91.6000	59.0000	52.2833	82.2333	57.7000	51.6167	92.9333	128.7000
	6	103.2667	80.6833	85.1833	58.1167	53.4667	78.9167	56.3167	53.7833	93.7000	130.0667
	7	101.4333	72.3000	78.3000	58.7500	53.2000	81.3000	55.9167	57.5500	91.7500	127.2000
	8	102.7500	69.3333	69.5000	55.1667	52.0000	84.0833	56.9500	49.8000	91.7833	129.9667
	9	104.0667	70.8167	55.6833	55.5667	53.7167	84.0167	59.0333	53.8167	88.5500	129.9167
	10	99.5000	67.4833	93.3667	54.7000	57.5667	70.4667	64.0167	52.4000	89.4833	125.9333
	11	100.3333	70.9667	75.4000	53.7500	56.8167	71.4667	62.1333	51.0333	89.2500	128.5333
	12	100.8333	69.8167	65.8667	53.0000	53.5667	77.6000	59.9167	57.8167	85.6000	126.3833
	13	98.2500	57.3833	64.5833	57.9167	52.7333	83.7500	56.1833	52.1500	87.0667	138.4333
	14	97.6833	55.8333	65.8333	59.3833	52.8333	79.8333	57.1000	55.6167	87.3667	137.3500
	15	95.5000	56.8833	67.2167	70.1000	56.2667	77.8667	60.6000	50.9667	85.6167	131.3500
	16	94.5500	62.9167	89.6500	57.4833	57.6000	61.8333	57.1333	52.1333	87.4667	134.1500
	17	92.8833	64.9667	93.7667	60.8167	60.2833	57.2667	55.0833	51.4333	85.9833	140.0333
	18	90.9167	68.9333	88.8500	58.1500	57.8833	58.7000	59.4167	49.5833	86.5667	140.3667
	19	92.6167	69.4500	93.3500	75.3000	56.4500	56.0000	56.4167	49.7667	89.6167	144.5667
	20	90.5088	72.1930	82.7544	76.8772	56.0000	51.4912	75.1579	53.5614	88.9474	140.9649
Loading Period MDF Values - Channel 2 (min)	1	55.7778	54.2381	80.7619	43.9048	56.6825	40.4762	48.6349	42.5238	116.5079	88.4286
	2	54.7833	55.2000	84.1333	40.5667	52.2000	41.4500	46.8167	39.9167	116.4167	107.8167
	3	57.9833	62.0833	84.5833	41.9167	54.9667	46.8500	49.0000	37.0000	113.2667	99.5500
	4	53.8833	63.6000	85.1333	39.7500	54.8667	61.9167	52.4667	36.1167	108.1333	109.6833
	5	50.4167	63.5833	85.0833	40.3333	51.8333	69.3167	55.6167	32.5000	106.7500	102.5667
	6	52.4000	63.2667	86.0833	40.4833	53.0333	81.3000	60.5333	33.0833	113.6667	99.4833
	7	51.7000	59.6500	85.7833	43.3500	53.9667	79.3000	61.5000	34.7167	108.7500	96.0500
	8	52.1667	56.4667	86.8167	43.4833	53.2833	82.4833	62.6000	31.6167	99.0667	97.8333
	9	47.1833	56.2000	86.0333	41.3333	54.6833	77.9833	61.6167	32.6000	105.9500	92.7833
	10	49.1833	62.3667	86.4833	43.3333	79.4667	63.3333	64.2667	32.9833	105.3000	82.8667
	11	50.2333	63.8500	84.3333	43.0667	59.4000	59.9167	65.0500	32.7500	101.2167	82.7833
	12	50.4333	61.1000	84.6333	42.9667	50.6167	66.1167	65.0833	42.5667	99.2667	79.1167
	13	50.6000	53.6167	86.4500	47.2167	52.4500	75.6833	64.8000	34.0667	103.4667	93.8333
	14	57.3333	53.1500	85.1833	45.6667	52.9333	68.8000	62.7333	34.2167	101.6667	89.3167
	15	51.8500	53.5833	80.9167	47.6000	61.0667	69.5833	62.8000	32.6000	105.1833	95.2000
	16	52.2667	59.9000	83.1167	41.6833	57.1833	52.5833	63.1833	32.9333	104.4333	93.7333
	17	49.5000	52.8667	85.2500	42.4833	55.3667	52.5333	67.7333	31.5833	100.5667	96.4500
	18	52.1833	61.6833	86.2333	43.9167	61.1167	48.4333	70.6833	31.4167	97.4333	87.1667
	19	59.4167	69.2000	87.3167	78.6167	56.0833	50.6833	70.1500	31.3000	103.7167	91.5000
	20	53.7544	65.9123	84.5088	57.3158	55.5263	52.2982	70.5088	32.6491	99.6316	89.7895

A sample MDF graph for the loading period is illustrated in Figure 23. Graphical representation of the MDF values for all participants can be found in Appendix E. In Figure

23, the y-axis represents the MDF in Hz, and the x-axis represents time in minutes for both channel 1 and channel 2.

As illustrated in Figure 23 and Appendix E, the results of the MDF analysis did not show a consistent increasing or decreasing trend for any of the participants. It must be noted that none of the participants experienced fatigue throughout the loading period. However, since the muscles were used continuously for 20 minutes, fatigue must have started to develop within with no significant indicator to the participant.

Therefore, to measure whether the change in the MDF values at the beginning and at the end of the loading period was statistically significant, a paired t-test was performed on the first and last minute MDF values of each of the participant as before and after values. The results of the paired t test analysis are summarized in Table 12.

Table 12. Results of Paired T-Test Analysis of Median Frequency Values

	Degree of Freedom	Pr > | t |	t Value
Channel 1	9	0.6007	-0.54
Channel 2	9	0.3809	-0.92

In Table 12, the data set used for the paired t-test analysis includes the first and last minute MDF values of all participants that consist of a total of 20 data points. The Pr values for both channels were greater than the confidence level 0.05 therefore there were no statistically significant difference between the MDF values of the first and the last minutes of the loading period for both of the channels. The results of the t-test corroborate with the participants expressing that there was no fatigue.

PARTICIPANT 5

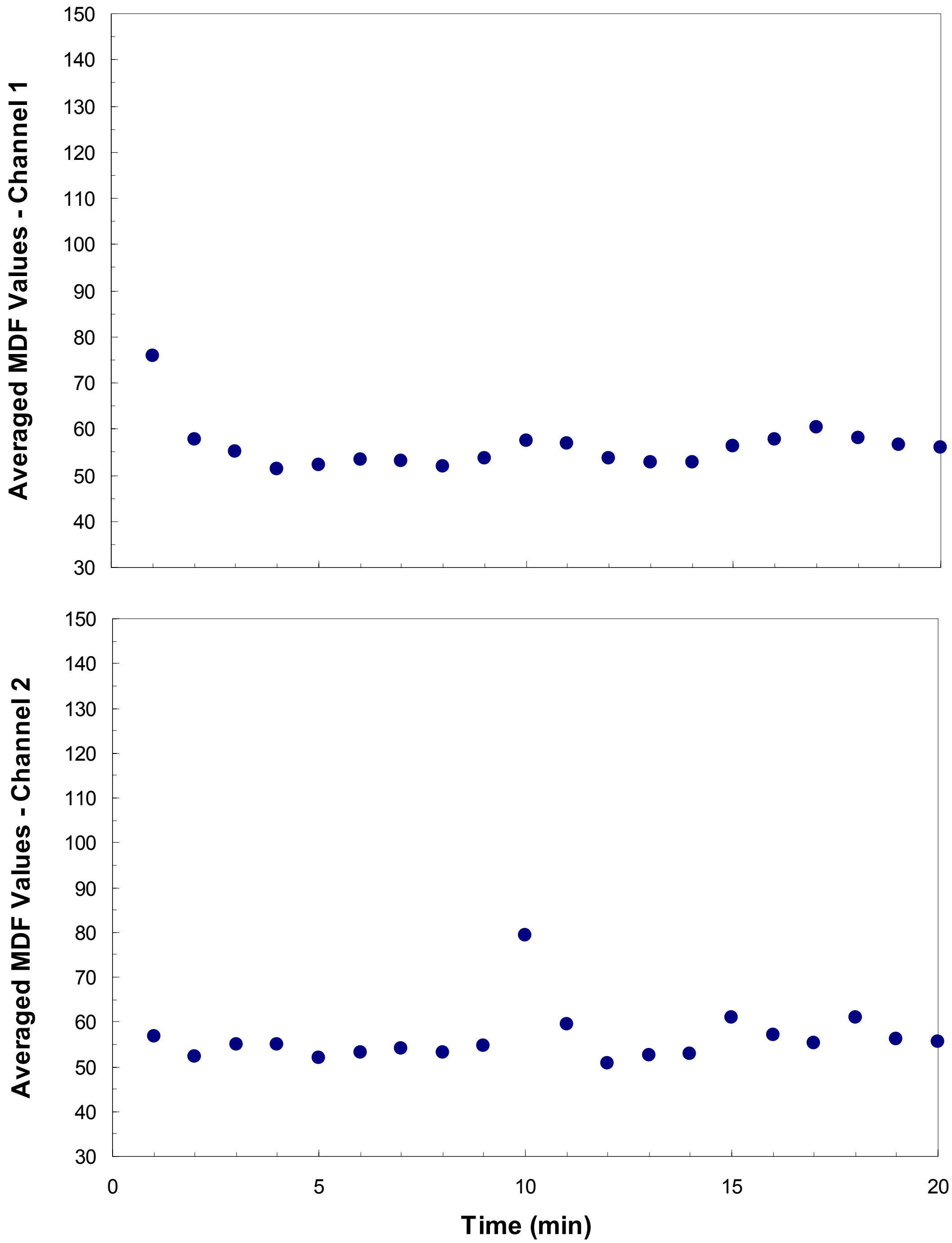

Figure 23. Sample Median Frequency Values for Loading Period

CHAPTER 6 - CONCLUSIONS

Reflexive muscle behavior phenomena under different loading conditions such as static and cyclic loading have been examined since the late 1950s. It has been established that this reflexive muscle behavior to stabilize the spine decreases exponentially throughout the loading period and the restoration of the reflexive muscle activity to its original state takes at least twice as much time as the loading period. Although the muscle mechanisms for the loading period have been investigated for both human and animal models, the only long term recovery process studies have been done on animal models. This study analyzed the constant loading effect on lower back and the recovery of the reflexive muscle activity using a human model and resulted in several important findings. As a part of the study, mathematical models were also developed to represent the lower back muscle behavior for both the loading and recovery periods.

6.1. CONCLUSIONS

Recent studies have shown that a ligamento-muscular reflex arc exists to stabilize the spine. Short term loading of the spine decreases this reflexive muscle activity, causing loss of stability of the spine or injury to the spine, and low back pain (Adams et al., 1990; Adams et al., 1987; Adams and Dolan, 1996; Gedalia et al., 1999). It is also reported that this reflexive muscle activity recovers partially with a short-term rest and the initial rest period is the most effective period in recovery (Gedalia et al., 1999). However, these findings were based on animal experiments. The results of this study confirm the previous findings from animal experiments for both loading and the recovery periods for humans.

The reflexive muscle activity to stabilize the spine decreased exponentially throughout the loading period for both of the channels. For L2, the average NIEMG value at the end of the 20 minute at rest period was 95 percent (±0.12). This value decreased to 81 percent (±0.06) in the first minute of loading (a decrease of 14 percent) and continued to decrease to 75 percent (±0.07), an additional decrease of 6 percent, at the end of the loading period. For L4, the average NIEMG value at the end of the 20 minute at rest period was 96 percent (±0.11). This value decreased to 85 percent (±0.07) in the first minute of loading (a decrease of 11 percent) and continued to decrease to 80 percent (±0.07) with an additional decrease of 5 percent at the end of the loading period. Although the percentage change does not seem to be too much, it has to be kept in mind that the load applied to the spine due to static flexion was minimal.

In 2002, Hatipkarasulu showed that the magnitude of load applied to the spine has a significant effect on the average NIEMG values in felines. Four different magnitudes of load (20N, 30N, 50N, and 70N) were applied to lumbar spine of the felines for 20 minutes. It was observed that higher magnitudes of loads resulted in higher decrease in the average NIEMG values during the loading period. For L1/L2 lumbar level of the feline, the average NIEMG values were observed as 98 percent, 94 percent, 61 percent, and 48 percent for 20N, 30N, 50N, and 70N of static loading, respectively. This reflects to the average NIEMG values throughout the recovery period as well. The recovery happens at a faster rate during the initial recovery period when the spine is subjected to higher magnitudes of loading. For L1/L2 lumbar level of the feline, the average NIEMG values during the initial recovery period increased 2 percent, 8 percent, 22 percent, and 26 percent for 20N, 30N, 50N, and 70N of static loading, respectively.

In this study using human participants, since the spine was not subjected to high magnitudes of loads, the percentage change in the average NIEMG values were minimal. L2 and L4 levels were chosen to observe high and low load bearing lumbar levels. The average NIEMG values decreased a total of 19.81 percent for L2 and 15.47 percent for L4. These results support the original concept of observing different load bearing levels. Since L2 level has been exposed to a higher magnitude of load, the decrease in the average NIEMG value for that level is higher compared to L4. The total recovery of the erector spinae muscle activity was 10.95 percent and 8.29 percent for L2 and L4 lumbar levels, respectively (Table 2 and Table 3). This result corroborates with the observations of animal experiments (Hatipkarasulu, 2003; Solomonow et al., 2003) that the lumbar levels which are subjected to higher magnitudes of load show a faster initial recovery.

In the recovery period, a transitory hyperexcitability in the initial rest period was observed followed by an exponential increase. It can be assumed that in the hyperexcitability period, the musculature is recruiting higher than normal muscle fibers/forces to protect the injured tissues from further damage and to increase the stiffness of the spine. In addition, a delayed decrease of muscle activity was observed with a time delay of 3 hours. The total recovery for L2 and L4 were 10.95 percent and 9.29 percent, respectively. Even after a short term (20 minutes) static loading, 3–hours of rest while sitting was not enough to restore the reflexive muscle activity and the NIEMG values at the end of the 3 hr showed a decrease.

Most of the recovery occurred during the first thirty minutes of the recovery period. The recovery rate after the first thirty minutes was not as high. The average NIEMG value at the end of the 20 minute loading period for channel 1 increased to 84 percent (±0.10) from 75 percent (±0.07) at the end of the first 10 minutes of rest. At the end of the 30 minute rest

period, the average NIEMG value increased to 85 percent (±0.11). The average NIEMG value at the end of the 3 hour rest period only showed a 1% increase and reached up to 86 percent (±0.12) of its original value. The average NIEMG value at the end of the 20 minute loading period for channel 2 increased to 88 percent (±0.10) from 80 percent (±0.07) at the end of the first 10 minutes of rest. At the end of the 30 minute rest period, the average NIEMG value increased to 89 percent (±0.10). The average NIEMG value at the end of the 3 hour rest period showed a 1 percent decrease and reached up to 88 percent (±0.11) of its original value. This decrease may be attributed to the continuous loading effect on the muscle, where 3 hours of continuous sitting itself constitutes static loading and has a negative impact on the recovery process.

Considering the effects of sitting posture on the lower back, an important suggestion can be constructed from the finding of this study. When lower back is exposed to static loading due to activities such as manual material handling, the process of muscle recovery can be substantially improved by laying down instead of sitting during the rest periods. This simple posture control measure eliminates the additional loading on the lower back

Full recovery of the erector spinae muscles was not observed for any of the lumbar levels. It can be concluded based on the results of this study that a short term static loading of the spine caused significant changes in the reflexive muscle activity, and that 3 hours of sitting rest was not enough for the reflexive muscle activity to be restored, and that for 20 minute static loading, at least thirty minute rest period was required since most of the recovery happened during the first thirty minutes of the recovery period.

When the 2 component recovery model, presented in Equation (13), was extrapolated, the time of full recovery was estimated to be between 22 and 25 hours. This corroborates with

the findings of Solomonow et al. (2003) that the recovery of the reflexive muscle activity in felines will take more than 24 hours, maybe even days. This phenomenon was refered to as the "morning after effect." However, during the recovery of their subjects for 7 hours, a secondary (delayed) hyperexcitability period was observed after the subjects were exposed to high magnitudes of loading (20N, 30N, 50N, and 70N). Other than the initial loading, no additional load was applied to the back of the felines during recovery. In this study, recovery was measured for only three hours after the loading period, therefore, no delayed hyperexcitability was observed and the load magnitude was minimal.

However, in this study, there was additional loading of the spine due to sitting for prolonged hours. If recovery measurements were taken over longer hours with higher load exposure, a delayed hyperexcitability was expected for the participants of this study. Based on the model extrapolation, short term static loading effect might be carried to the next day as the morning after effect.

Although it is concluded that animal models capture the behavior of the human muscles for both loading and recovery, the effects of individual subject/participant characteristics, such as the effect of age, height, weight, and back strength, were impossible to address in animal models. Only the effect of time and loading type could have been tested. In this study, a repeated analysis of variance was performed to test whether these factors affect the recorded EMG values significantly. The results showed only time had a significant effect on the recorded EMG values for both the loading and recovery periods (Pr>F=<.0001 for channel 1 and Pr>F=<.0001 for channel 2 for loading; Pr>F=0.0047 for channel 1 and Pr>F=0.0007 for channel 2 for the recovery period). This result corroborates with the results of the previous studies that the EMG values are highly time dependent. Although factors such

as age, height, weight, and back strength may have some effect, it is not significant on the behavior of the muscle response under static loading conditions.

Another important factor that should be considered is the effect of fatigue. In this study, none of the participants expressed muscle fatigue throughout the experiment. MDF values were calculated for the loading period and a paired t test was conducted to measure whether the MDF values at the beginning of the loading period were significantly different than the values at the end. The results showed no significant change between the beginning and the end of the experiment for both of the lumbar levels. Since the experiment includes continuous loading, it is expected that fatigue might have started to set in, however, the duration of the experiments in this study was not long enough to see a significant shift in MDF. With longer loading periods, significant shifting of the MDF values are expected.

Previous studies have shown that muscle spasms occur while under different loading conditions in in vivo animal models (Solomonow et al., 2000; Williams et al., 2000; Jackson et al., 2001; Hatipkarasulu, 2002). In this study, muscle spasms were recorded from both lumbar levels with different amplitudes, frequencies and timing. It can also be concluded that the animal models represent the presence of spasms accurately.

Although the observed NIEMG data behavior was similar to the cyclic (Gedalia et al., 1999), constant displacement (Jackson et al., 2001), and constant static loading (Hatipkarasulu, 2002) behavior during the loading period, a relatively different model was necessary to model the recovery period in this study. An additional exponential term was modeled to represent the continuous loading effect (daily activity factor) with a different time constant to accurately represent the impact of resting for long hours while in a sitting position. The two component model that was developed by Gedalia et al. (1999) and Solomow et al.

(2000) was applied to the data sets collected for this study and R^2 values over 80% were obtained. However, when the daily activity factor adjustment was added to the model, the R^2 value for the goodness of the model fit increased to 100 percent for both of the lumbar levels, indicating that this adjustment factor has to be taken into account when human models are considered.

Overall, this study supports the previous findings about the initial and secondary recovery of the electrical activity of the lower back muscles (based on animal models) after a short-term period of loading. However, the mathematical description of the phenomenon is modified by taking the daily activity factor into account. Understanding the time periods and phases for the recovery is essential since a better understanding of the phenomenon can lead to optimal design of work/rest periods in occupational as well as sports activities.

6.2. RECOMMENDATIONS FOR FUTURE RESEARCH

Several research directions can be suggested as further investigation of the muscle recovery phenomenon. Although it might require the similar experiment set up, first direction is the observation of behavior under cumulative loading conditions. In all of the previous studies, the efforts have been concentrated on the behavior after a single stimulus (loading, displacement, etc.). However, the repetition of the stimulus may result in a cumulative effect, similar to the occurrence of the cumulative musculoskeletal disorders. The stimulus can be introduced repetitively with constant loading and rest periods several times, simulating a behavior that can be observed in a manual material handling task.

The second research direction is the modeling and the validation of the required time period for the full recovery of the NIEMG. Although the model developed as a part of this study can be used to extrapolate the required time periods for the full recovery, the

calculations should be validated with a series of experimentation, where the subjects are observed for longer durations of time, with a slight difference in the rest posture. It is proven in this study that sitting for long hours after performing a static task significantly affects the recovery process. Therefore an alternative resting posture has to be considered.

A third possible research direction is to explore the effects of gender. Although this study looked at the effects of different individual characteristic factors, the population was limited to males.

A fourth possible future research is to investigate the effect of fatigue on the reflexive muscle activity and its recovery. In this study, fatigue was not observed due to the duration of the experiments and the minimal loading of the spine. However, by using a different type of loading, such as dynamic and repetitive loading with longer loading periods, the effect of fatigue could also be investigated.

A fifth possible research direction is to investigate the effect of different types of loading (static, dynamic, and cumulative) on back strength. A series of back strength measurements could be taken throughout the experiment, such as at the beginning and at the end of the loading period, and at the end of the rest periods, to evaluate the effect of different types of loading on back strength.

REFERENCES

Abumi, K., Panjabi, M., Duranceau, J., Oxland, T., Crisco, J., 1990. "Biomechanical evaluation of lumbar spine stability after graded facectomies." Spine, 15: 1142-7

Adams, M. A., Dolan, P., 1996, "Time dependent changes in the lumbar spine's resistance to bending." Clinical Biomechanics, Vol. 11, No. 4, pp. 194-200

Adams, M. A., Dolan, P., Hutton, W. C., 1988. "The lumbar spine in backwards bending." Spine, Vol. 13, pp. 1019-26

Adams, M. A., Hutton, W. C., 1988. "Mechanics of the intervertebral disc." In Gosh, P. edition. The biology of the intervertebral disc. Boca Raton, FL: CRC Press, pp. 38-71

Andersson, C., Chaffin, D., Herrin, C., Mathews, L., 1985. "A biomechanical model of the lumbosacral joint during lifting activities." Journal of Biomechanics, 18: 578-84

Baratta, R., Solomonow, M., 1995. "A practical guide to electromyography for ergonomists and biomechanists." LSUHSC, Department of Orthopaedics, Bioengineering Laboratory, Louisiana State University.

Basmajian, J. V., Deluca C. J.,1986. "Muscles alive: Their functions revealed by electromyography.", 5th edition, Williams and Wilkins, Baltimore.

Berkson, M., Nachemson, A., Schultz, A., 1979. "Mechanical properties of the human lumbar spine motion segment: II. Response in compression and shear." Journal of Biomechanical Engineering, 101: 53-7

BLS. 1997. Workplace injuries and illnesses in 1997. Washington, DC. U.S. Department of Labor, Bureau of Labor Statistics.

Bogduk, N., Twomey, L. T., 1987. "Clinical anatomy of the lumbar spine." New York Churchill Livingstone.

Bogduk, N., MacIntosh, J., 1984. "The applied anatomy of the throcalumbar fascia." Spine, 9: 164-70

Bogduk, N., 1983. "The myotomes of the human multifidus." Journal of Anatomy, Vol. 136, pp. 148-9

Bogduk, N., Wilson, A. S., Tynan, W., 1982. "The human lumbar dorsal rami." Journal of Anatomy, Vol. 134, pp. 383-97

Bojadsen, T. W. A., Silva, E. S., Rodrigues, A. J., Amadio, A. C., 2000. "Comperative study of Mm. Multifidi in lumbar and thoracic spine." Journal of Electromyography and Kinesiology, Vol. 10, 143-149

Botsford, D. J., Esses, S., Ogilvie-Harris, D. J., 1994. "In vivo diurnal variation in intervertebral disc volume and morphology." Spine, Vol. 19, No. 8, pp. 935-940

Bronzino, J. D., 1995. "The biomedical engineering handbook." CRC Press, IEEE Press, MA, ISBN: 0-8493-8346-3

Claude, L., Solomonow, M., Zhou, B., Baratta, R.V., Zhu, M., 2003. "Neuromuscular Disorder Elicited By Cyclic Lumbar Flexion." Muscle & Nerve, 27:348-358

Crisco, J., Panjabi, M., Yamamoto, I., Oxland, T., 1992. "Stability of the human ligamentous spine: II. Experiment." Clinical Biomechanics, 7: 27-32

Delsys (a), 2003. "The Bagnoli 2 EMG system Users Manual", March 2003 Edition, Copyright 2003, by Delsys, Inc., available at www.delsys.com

Delsys (b), 2003. "Matlab Script: LOADEMG3.m – Version 1.5." Copyright 2003, by Delsys, Inc., available at http://www.delsys.com/download/files/loademg3.m

De Luca, C.J., 1993. "Use of the surface EMG signal for performance evaluation of back muscles." Muscle & Nerve, 16: 210-216

Dimitrova, N.A, Dimitrov, G.V., 2003. "Interpretation of EMG changes with fatigue: facts, pitfalls, and fallacies." Journal of Electromyography and Kinesiology, 13: 13-36

Dolan, P., Benjamin, E., Adams, M., 1993. "Diurnal changes in bending and compressive stresses acting on the lumbar spine." Journal of Bone and Joint Surgery, 75, (Suppl). 22

Dorland's Medical Dictionary, 1994. W.B. Saunders Company, 28th edition, Philadelphia, ISBN: 0-7216-2859-1

Ekstrom, L., Kaigle, A., Holt, E., Holm, S., Rostgat, M., Hansson, T., 1996. "Intervertebral disc response to cyclic loading: An animal model." Proc Inst Mech Eng, Vol. 210, pp. 249-58

Eversull, E., Solomonow, M., Zhou, B., Baratta, R., Zhu, M., 2001. "Neuromuscular neutral zones sensitivity to lumbar displacement rate." Clinical Biomechanics, 16: 102–113

ExRx, 1999. “Exercise Prescription on the Net”, ExRx.net LLC, Available at http://www.exrx.net/Muscles/ErectorSpinae.html

Farfan, H. F., Cossett, J. W., Robertson, G. H., 1970. "The effects of torsion on the lumbar intervertebral joints: the role of torsion in the production of disc degeneration." Journal of Bone and Joint surgery, Vol. 52A, pp. 468-497

Farina, D., Gazzoni, M., Merletti, R., 2003. “Assessment of low back muscle fatigue by surface EMG signal analysis: methodological aspects.” Journal of Electromyography and Kinesiology, 13: 319-332

Farina, D., Merletti, R., 2000. “Comparison of algorithms for estimation of EMG variables during voluntary isometric contractions.” Journal of Electromyography and Kinesiology, 10: 337-350

Gardner-Morse, M., Stokes, I., 1998. "The effect of abdominal muscle co activation on lumbar spine stability." Spine, 23: 86-92

Garrett, Jr. W. E., Speer, K. P., Kirkendall, D. T., 2000. "Principles and practice of orthopaedic sports medicine." Lippincott, Williams&Wilkens, Philadelphia, ISBN: 0-7817-2578

Gedalia, U., Solomonow, M., Zhou, B., Baratta, R., Harris, M., 1999. "Biomechanics of increased exposure to lumbar injury caused by cyclic loading: Part 2. Recovery of reflexive muscular stability with rest." Spine, 23: 2461–2467

Goel, V., Voo, L., Weinstein, J., King, L., Okuma, T., Njus, G., 1988. "Response of ligamentous lumbar spine to cyclic bending." Spine, Vol. 13, pp. 294-300

Granata, K., Marras, W., 1995. "The influence of trunk muscle co activity on dynamic spinal loading." Spine, 20: 913-19

Gregersen, G. G., Lucas, D. B., 1967. "An in vivo study of the axial rotation of the human throcalumbar spine." Journal of Bone and Joint Surgery, Vol. 49A, pp. 247-262

Guanche, C., Knatt, T., Solomonow, M., Lu, Y., Baratta, R., 1995. "The synergistic action of the capsule and the shoulder muscles." Am J Sports Med, 23: 301-6

Hatipkarasulu, G. S., 2002. “Visco-Elastic Creep in the Lumbar Spine Associated with Static Flexion: A Feline Model.” A masters’ thesis, Louisiana State University, Industrial and Manufacturing Systems Engineering Department

Hedman, T., Fernie, G., 1995. "In vivo measurements of lumbar spinal creep in two seated postures using magnetic resonance imaging." Spine, Vol. 20, pp. 178-83

Herrin, G. A., Jaraiedi, M., Anderson, C. K., 1986. "Prediction of overexertion injuries using biomechanical and psychophysical models." Journal of American Industrial Hygiene Association, Vol. 47, pp. 322-30

Hirokawa, S., Solomonow, M., Luo, Z., Lu, Y., D’Ambrosia, R., 1991. "Muscular co contraction and control of knee stability." Journal of Electromyography and Kinesiology, 1: 199-208

Indahl, A., Kaigle, A., Reikeras, O., Holm, S., 1995. "Electromyographic response of the porcine multifidus musculature after nerve stimulation." Spine, 20, 24: 2652–2658

Jackson, M., Solomonow, M., Zhou, B., Baratta, R., Harris, M., 2001. "Multifidus EMG and Tension–Relaxation recovery after prolonged static lumbar flexion." Spine, 26, 7: 715–723

Kaigle, A., Holm, S., Hansson, T., 1995. "Experimental instability in the lumbar spine." Spine, 20, 4: 421–430

Keller, T., Hansson, T., Holm, S., Pope, M., Spengler, D., 1989. "In vivo creep behavior of the normal and degenerated porcine disc." Journal of Spinal Disorder, Vol. 1, pp. 267-78

Knatt, T., Guanche, C., Solomonow, M., Lu, Y., Baratta, R., Zhou, B. H., 1995. "The glenohumeral biceps reflex in the feline." Clinical Orthopeadics, Vol. 314, pp. 247-252

Koumantakis, G.A., Arnall, F., Cooper, R.G., Oldham, J.A., 2001. “Paraspinal muscle EMG fatigue testing with two methods in healthy volunteers. Reliability in the context of clinical applications.” Clinical Biomechanics, 16: 263-266

Kumar, S., 2001. "Theories of musculoskeletal injury causation." Ergonomics, 44,1: 17–47

Kumar, S., 1999. "Biomechanics in ergonomics." Taylor & Francis Ltd, ISBN 0 7484 0704 9

Kumar, S., 1996. "Narayan Y, Zedka. An electromyographic study of unresisted trunk rotation with normal velocity among healthy subjects." Spine, 21: 1500-12

Kumar, S., 1990. "Cumulative load as a risk factor for back pain." Spine, Vol. 15, No. 12, pp. 1311-1316

Leivseth, G., Drerup, B., 1997. "Spinal shrinkage during work in sitting posture compared to work in standing posture." Clinical Biomechanics, Vol. 12, pp. 409-18

Levenberg, K., 1944. "A Method for the Solution of Certain Problems in Least Squares," Quart. Appl. Math, Vol. 2, pp. 164-168.

Lewis, J., Donatto, K., Solomonow, M., Zhou, B. H., Lu, Y., Baratta, R., Zhu, M. P., 1996. "Ligamento-muscular protective reflex in the feline ankle." Orthopaedics International, Vol. 4, pp. 359-365

Lu, D., Solomonow, M., Zhou, B., Baratta, R.V., Li, L., 2004. “Frequency-dependent changes in neuromuscular responses to cyclic lumbar flexion.” Journal of Biomechanics, Vol. 37, 845-855

Lucas, D., Bresler, B., 1961. "Stability of the ligamentous spine." Biomechanics La, UCSF/B. Report No. 40, 1-41

Macintosh, J. E., Valencia, F., Bogduk, N., Munro, R. R., 1986. "The morphology of the lumbar multifidus muscles." Clinical Biomechanics, Vol. 1, pp. 196-204

Markolf, K. L., 1972. "Deformation of the throcalumbar intervertebral joints in response to external loads." Journal of Bone and Joint Surgery, Vol. 54A, pp. 511-533

Marquardt, D., 1963. "An Algorithm for Least Squares Estimation of Nonlinear Parameters," SIAM J. Appl. Math, Vol. 11, pp. 431-441.

Matlab (a), 2005. “Matlab Scientific Programming Language – Version 7”, Mathworks Inc., New York

Matlab (b), 2005. “Signal Processing Toolbox – Version 1.8”, Mathworks Inc., New York

Matlab (c), 2005, “Matlab Version 7 – User Manual”, Mathworks Inc., New York

Matlab (d), 2005. “Curve Fitting Toolbox – Version 1.1.4”, Mathworks Inc., New York

McGill, S.M., 1999. "The biomechanics of low back injury: Implications on current practices in industry and the clinic." Journal of Biomechanics, Vol. 30, pp. 465-75

McGill, S. M., Brown, S., 1992. "Creep response of the lumbar spine to prolonged full flexion." Clinical Biomechanics,Vol. 7, pp. 43-6

McGill, S. M., Norman, R., 1986. "Partitioning of the L4–L5 dynamic moment into disc, ligamentous, and muscular components during lifting." Spine, 11, 7: 666–668

National Academy of Sciences, 2001. "Musculoskeletal disorders and the workplace: low back and upper extremities." National Academy Press, Washington, D.C.

National Institute for Occupational Safety and Health, 1991. "DHHS, Work Practices Guide for Manual Lifting." National Institute of Occupational Safety and Health, Washington, DC.

NIOSH, 1999. "Survey of Occupational Injuries and Illnesses." National Institute of Occupational Safety and Health, Washington, DC.

NRCIM, 2001. "Musculoskeletal disorders and the workplace. Low back and upper extremities." National Research Council and Institute of Medicine, National Academy Press.

Norman, R. W., Wells, R., Newmann, P., 1998. "A comparison of peak vs. cumulative physical work exposure risk factors for the reporting of low back pain in the automotive industry." Clinical Biomechanics, Vol. 13, pp. 561-73

Occupational Safety and Health Administration, 1992. "OSHA Technical Manual, TED 1-0.15A." National Institute of Occupational Safety and Health, Washington, DC.

Olson, M.W., Li, L., Solomonow, M., 2004. "Flexion-relaxation response to cyclic lumbar flexion." Clinical Biomechanics, 19, pp. 769-776.

Panjabi, M. M., 1992. "The stabilizing system of the spine: I. Function, dysfunction, adaptation, and enhancement." Journal of Spinal Disorder, 5:383-9

Pedersen, H., Blunck, C., Gardner, E., 1956. "The anatomy of lumbosacral posterior rami and meningeal branches of spinal nerves." Journal of Bone and Joint Surgery, Vol.38-A, No.1, pp. 377-391

Phillips, D., Petrie, S., Solomonow, M., Zhou, B. H., Guanche, C., D'Ambrosia, R., 1997. "Ligamentomuscular protective reflex in the elbow." The Journal of Hand Surgery, Vol. 22A, No. 3, pp. 473-478

Pope, M., Andersson, G., Broman, H., Sevensson, M., Zetterberg, C., 1986. "Electromyographic studies of the lumbar trunk musculature during development of axial torque." Journal of Orthopaedic Research, 4: 288-97

Posner, I., White, A., Edwards, T., Hayes, W., 1982. "A biomechanical analysis of the clinical stability of the lumbar and lumbosacral spine." Spine, 7: 374-89

SIC, 2005. “Sports Injuries Index by Body Area.” Available at www.sportsinjuryclinic.net. Last Accessed April 1, 2005.

Solomonow, M., Zhou, B.H., Baratta, R.V., Burger, E., 2003. “Biomechanics and electromyography of a cumulative lumbar disorder: response to static flexion.”, Clinical Biomechanics, Vol. 18, pp. 890-898

Solomonow, M., Baratta, R.V., Banks, A., Freudenberger, C., Zhou, B. H., 2003. “Flexion-relaxation response to static lumbar flexion in males and females.” Clinical Biomechanics, Vol. 18, pp. 273-279

Solomonow, M., Hatipkarasulu, S., Zhou, B.H., Baratta, R.V., Aghazadeh, F.,2003. “Biomechanics and electromyography of a common idiopathic low back disorder.” Spine, 28 (12):1235-48

Solomonow, M., Zhou, B. H., Baratta, R., Lu, Y., Zhu, M., Harris, M., 2000. "Biexponential recovery model of lumbar viscoelastic laxity and reflexive muscular activity after prolonged cyclic loading." Clinical Biomechanics, Vol. 15, pp. 167-175

Solomonow, M., Zhou, B. H., Baratta, R., Lu, Y., Harris, M., 1999. "Biomechanics of increased exposure to lumbar injury caused by cyclic loading: Part I. Loss of reflexive muscular stabilization." Spine, Vol. 24, No. 23, pp. 2426-2434

Solomonow, M., Zhou, B., Harris, M., Lu, Y., Baratta, R., 1998. "The ligamento-muscular stabilizing system of the spine." Spine, 23: 2552-62

Solomonow, M., Guanche, C., Wink, C., Knatt, T., Baratta, R., Lu, Y., 1996. "Mechanoreceptors and reflex arc in the feline shoulder." J Elbow Shoulder Surgery, 5: 139-46

Solomonow, M., Baratta, R. V., Zhou, B., Shoji, H., Bose, W., Beck, C., D’Ambrosia, R., 1987. "The synergistic action of the ACL and thigh muscles in maintaining joint stability." Am J Sports Med, 15: 207-13

Stubbs, M., Harris, M., Solomonow, M., Zhou, B., Lu, Y., Baratta, R., 1998. "Ligamento-muscular protective reflex in the lumbar spine of the feline." Journal of Electromyography and Kinesiology, Vol. 8, pp. 197-240

Tencer, A. F., Ahmed, A. M., Burke, D. L., 1982. "Some static mechanical properties of the lumbar intervertebral joint, intact and injured." Journal of Biomechanical Engineering, Vol. 104, pp. 193-201

Tkaczuk, H., 1968. "Tensile properties of human lumbar longitudinal ligaments." Acta Orthopaedica Scandinavica Supplementum, 115: 8–69

Tortora, G., 1995. "Principles of human anatomy." Seventh edition, HarperCollins College Publishers, ISBN: 0-673-99074-5

Towney, L., Taylor, J., 1982. "Flexion creep deformation and hysterisis in the lumbar vertebral column." Spine, Vol. 7, pp. 116-22

Wang, J., Parnianpour, M., Shirazi-Adl, A., Engin, A., 2000. "Viscoelastic finite-element analysis of a lumbar motion segment in combined compression and sagittal flexion." Spine, Vol. 25, no. 3, pp. 310-318

White, A. A. III., Panjabi, M. M., 1978. "Clinical biomechanics of the spine." Philadelphia, JB Lippincott

Wilder, D., Frymoyer, J., Pope, M., 1985. "The effect of vibration on the spine of the seated individual." Automedica, Vol. 6, pp. 3-35

Wiliams, M., Solomonow, M., Zhou, B., Baratta, R., Harris, M., 2000. "Multifidus spasms elicited by prolonged lumbar flexion." Spine, Vol. 25, No. 22, pp. 2916-2924

Yahia, H., Newman, N., 1993. "A scanning electron microscopic and immunohistochemical study of spinal ligaments innervation." Ann Anat, 175: 111–114

Zetterberg, C., Andersson, G., Schultz, A., 1987. "The activity of individual trunk muscles during heavy physical loading." Spine, 12: 1035-40

APPENDIX A - STUDY CONSENT FORM

CONSENT FORM

Study Title: Loading and Recovery Behavior of the Human Lumber Spine under Static Loading.

Performance Sites: Louisiana State University, IE Department, Human Factors Laboratory

Investigators: The following investigators are available for questions about this study, M- F, 8:00 a.m. – 4:30 p.m.

Dr. Fred Aghazadeh 225 578 5367

G. Selen Hatipkarasulu 225 819 8830

Purpose of the Study: The goal of the proposed study is to observe and model the lower back multifidus muscles' behavior under static flexion by using humans as test subjects. To achieve this goal, the following objectives are going to be completed: Observation of the multifidus muscle behavior under static flexion by using EMG, Observation of the multifidus muscle at the recovery period, Filtering and reduction of recorded EMG data, Analysis of the filtered EMG data, Mathematical modeling of the multifidus muscle behavior under static flexion and recovery, Comparison of animal models to human models, Fatigue evaluation.

Number of Subjects: 10

Study Procedures: The skin around the lumbar region will be cleaned by using alcohol. The surface electrodes, filled with Redux gel to increase the conductivity between the skin and the electrodes, will be attached to the skin. First, the subjects will be asked to sit on a chair comfortably for 20 minutes during which at rest measurements will be taken to be used as control data. This will be done to assure that in a neutral position, there will be little or no electrical activity and that any EMG activity recorded after the experiment begins will indeed be due to the static flexion of the spine. The subjects then will be asked to sit down on a hard mattress and place a triangular shaped foam pillow underneath their knee joints so as they feel as comfortable as possible. The subjects will be asked to bend forward as much as they can

while sitting and hold still for 20 minutes (this will be the loading period). EMG measurements will be taken every 2 minutes for the 20 minute loading period. After the loading period is complete, the subjects will be asked to sit down on a chair for the recovery period measurements. At rest measurements will be collected after 10 minutes of rest, 30 minutes of rest, 60 minutes of rest and every hour thereafter for 3 hours. During the 3 hours of rest, 20sec tests will be applied to assess the recovery of the electrical activity in the muscle.

Benefits: There are no potential benefits to be gained by the participants. The expected benefit of this study is observation of the behavior of human lower back (multifidus) muscles' recovery and modeling of the behavior. Such information is essential for optimal design of work/rest periods in occupational activities.

Risks/Discomforts: The risks associated with this project are muscle fatigue, muscle soreness, physical discomfort due to static flexion, and possible muscle spasms. A foam wedge will be placed under the participant's knee joint for support and to reduce the discomfort. If needed, after the experiment a warm compress will be provided to be placed around the lower back region. If this research project causes any physical injury to participants, treatment is not available at Louisiana State University, nor is there any insurance carried by the university or its personnel applicable to cover any such injury. Treatment and financial compensation for such injury must be provided through the participant's own insurance program. In case of an emergency, the local emergency service (911) will be contacted.

Right to Refuse: Subjects may choose not to participate or to withdraw from the study at any time without penalty or loss of any benefit to which they might otherwise be entitled.

Privacy: Results of this study may be published, but no names or identifying information will be included in the publication. Subject identity will remain confidential unless the law requires disclosure.

Signatures:

The study has been discussed with me and all my questions have been answered. I may direct additional questions regarding study specifics to the investigators. If I have questions about subjects' rights or other concerns, I can contact Dr. Robert Mathews, Institutional Review Board, (225) 578 1492. I agree to participate in the study described above and acknowledge the investigator's obligation to provide me with a signed consent form.

Signature of the Participant:-------------------------------- **Date**:----------------------------

APPENDIX B - RAW EMG DATA

PARTICIPANT 1

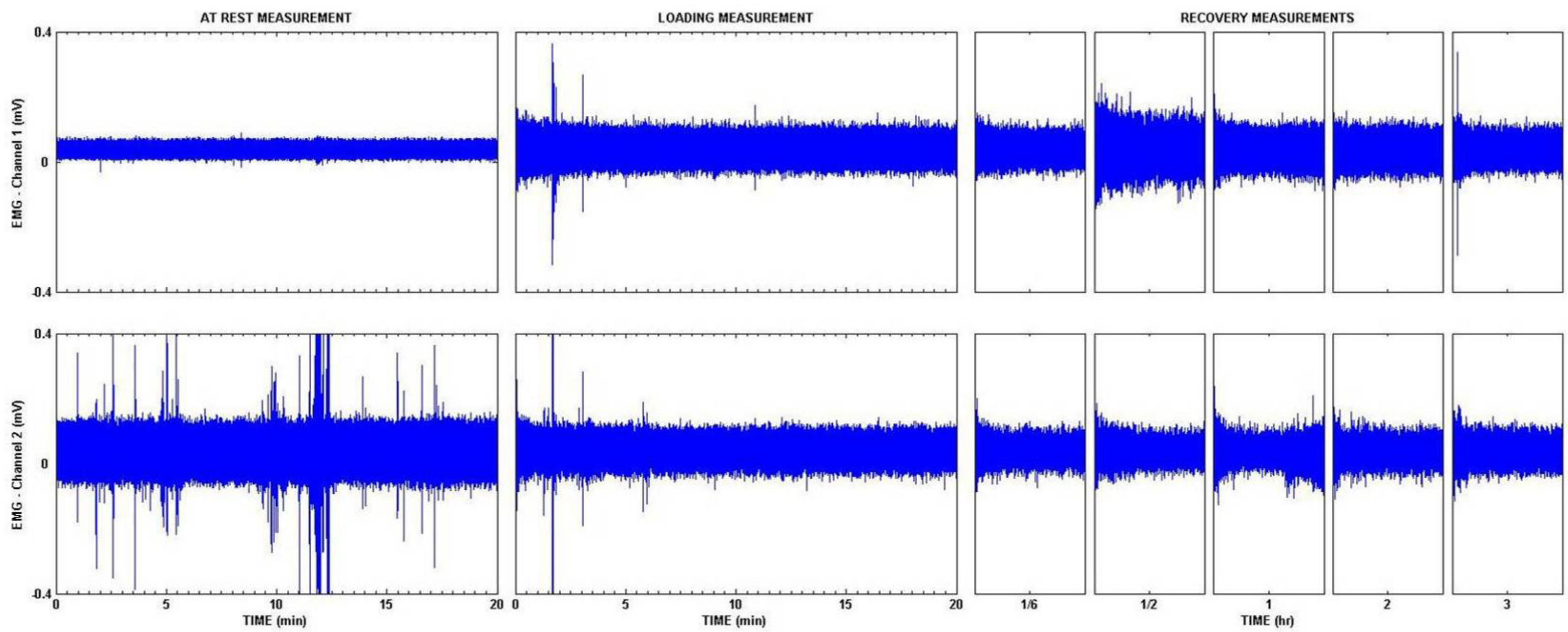

Figure 24. Raw EMG Data - Participant 1

PARTICIPANT 2

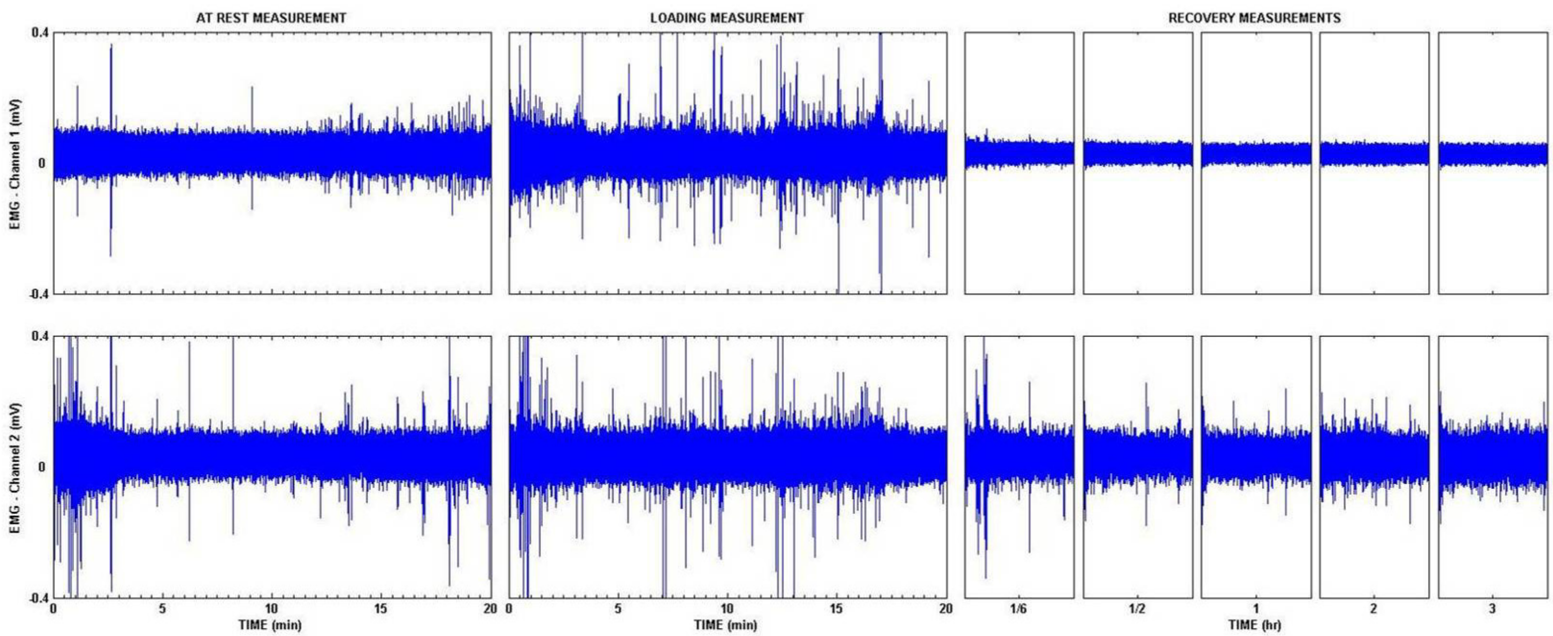

Figure 25. Raw EMG Data - Participant 2

PARTICIPANT 3

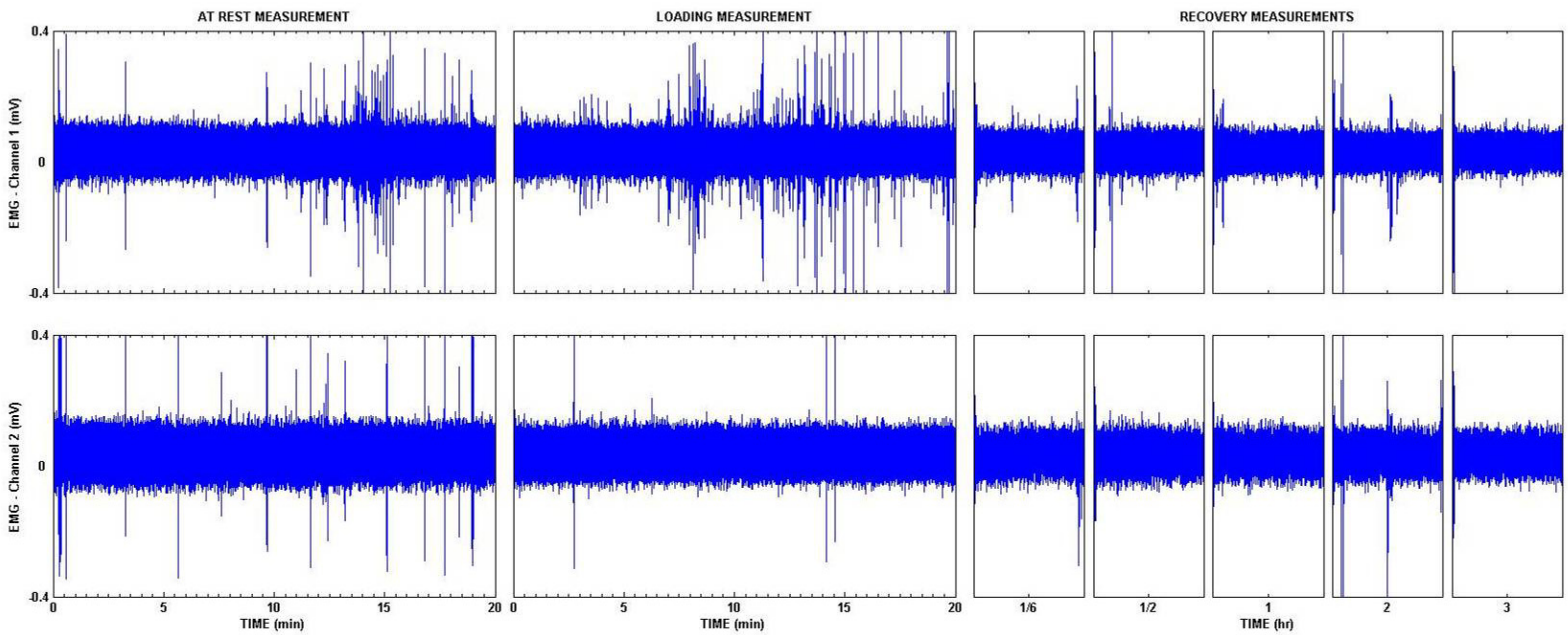

Figure 26. Raw EMG Data - Participant 3

PARTICIPANT 4

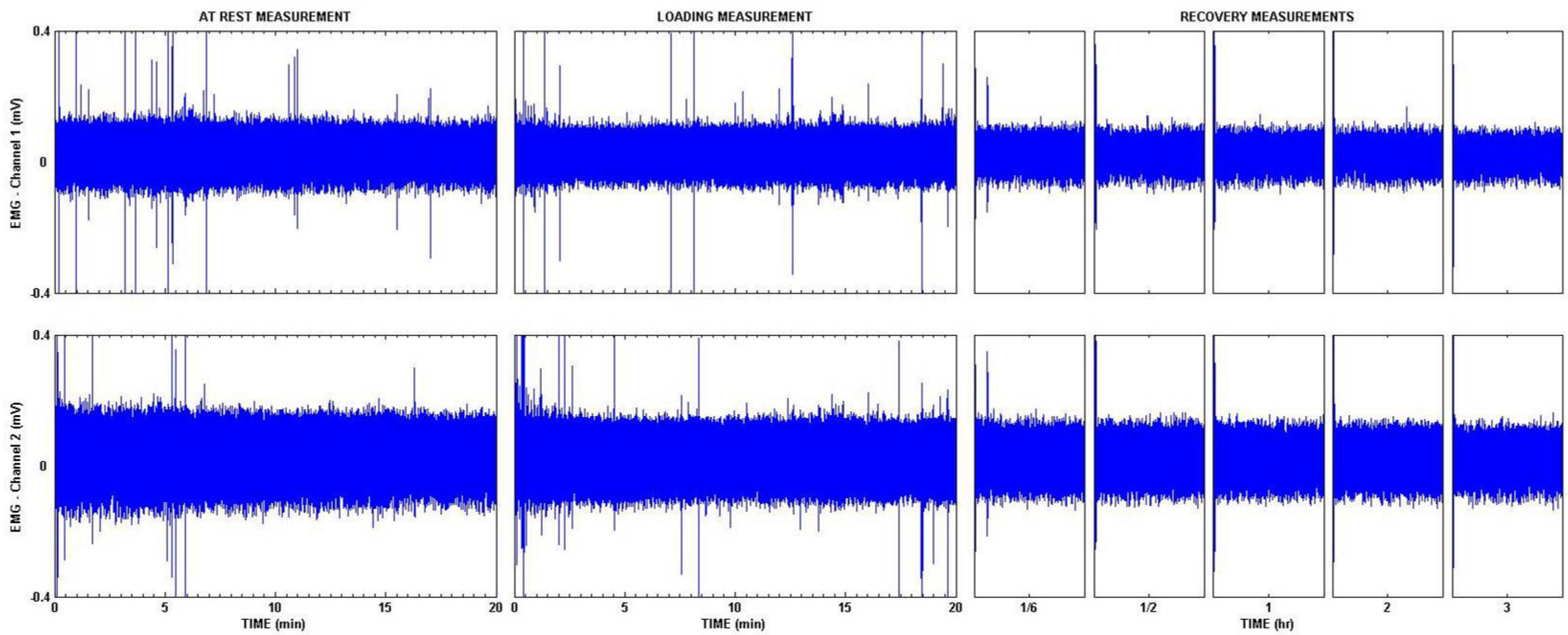

Figure 27. Raw EMG Data - Participant 4

PARTICIPANT 5

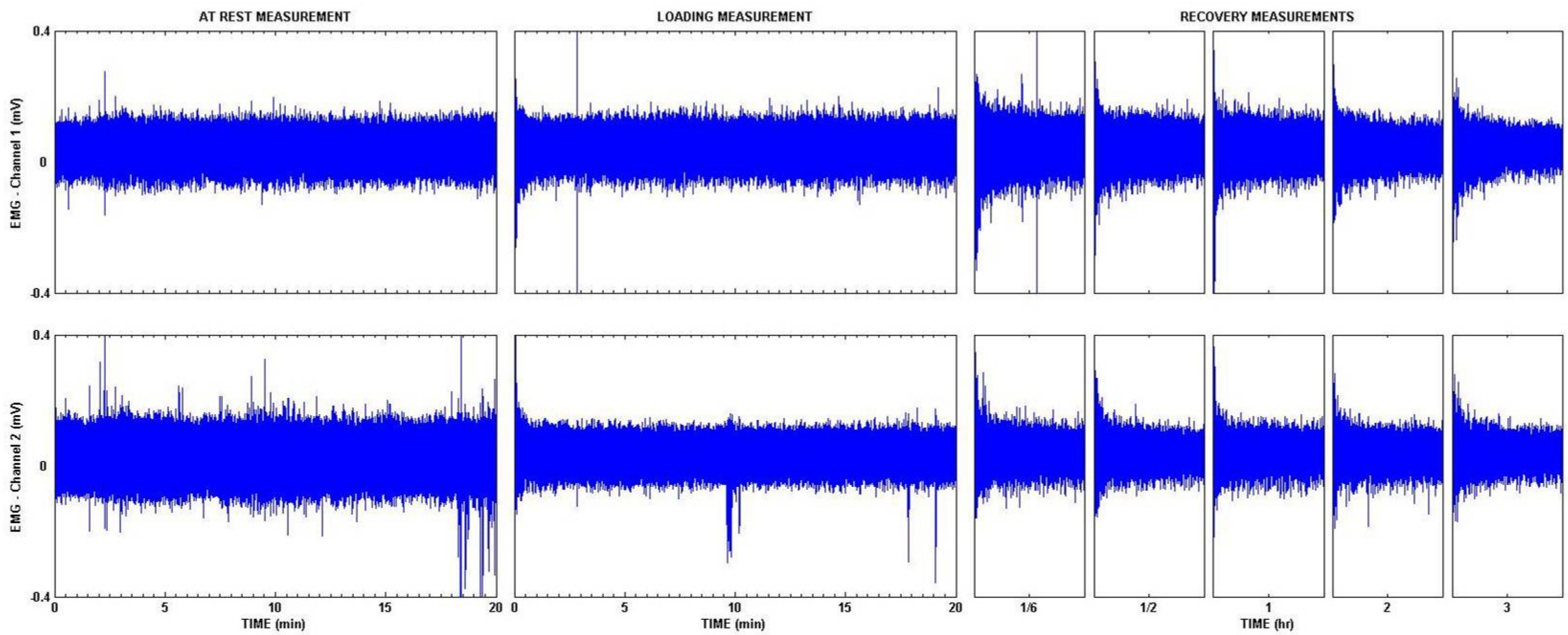

Figure 28. Raw EMG Data - Participant 5

PARTICIPANT 6

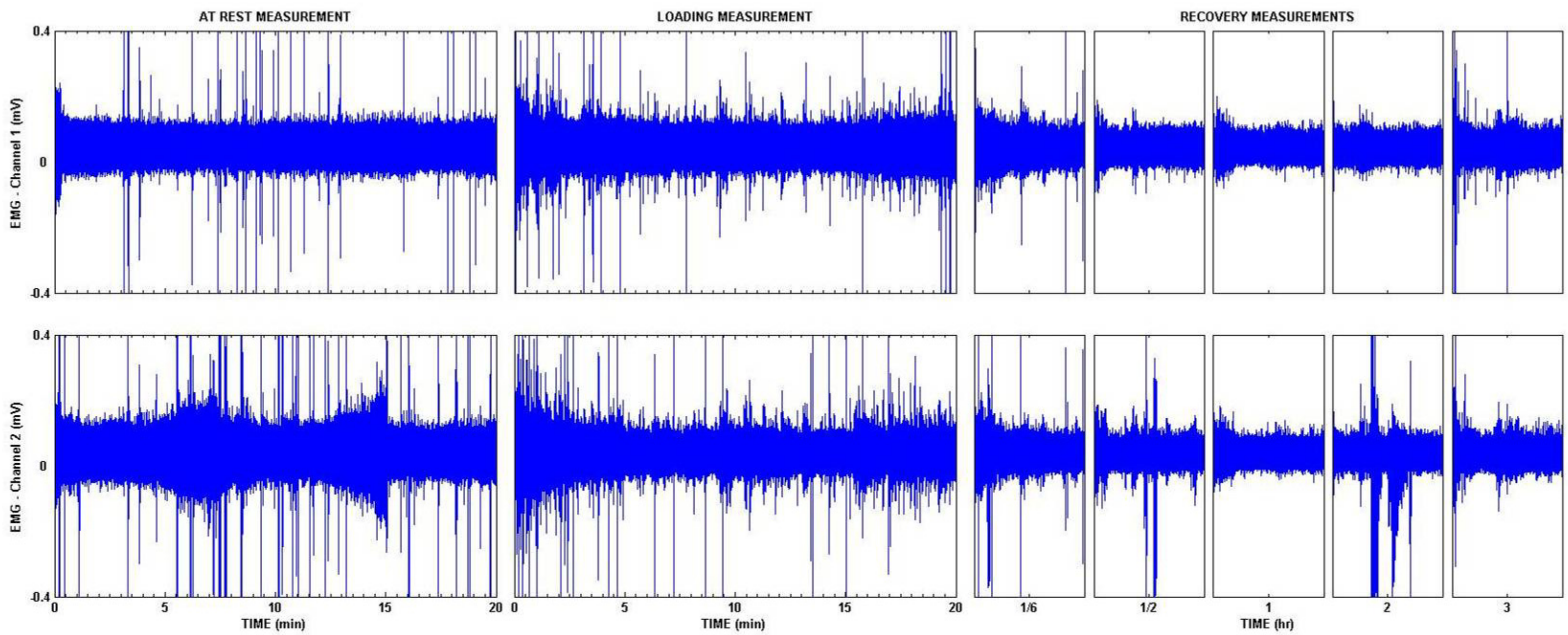

Figure 29. Raw EMG Data - Participant 6

PARTICIPANT 7

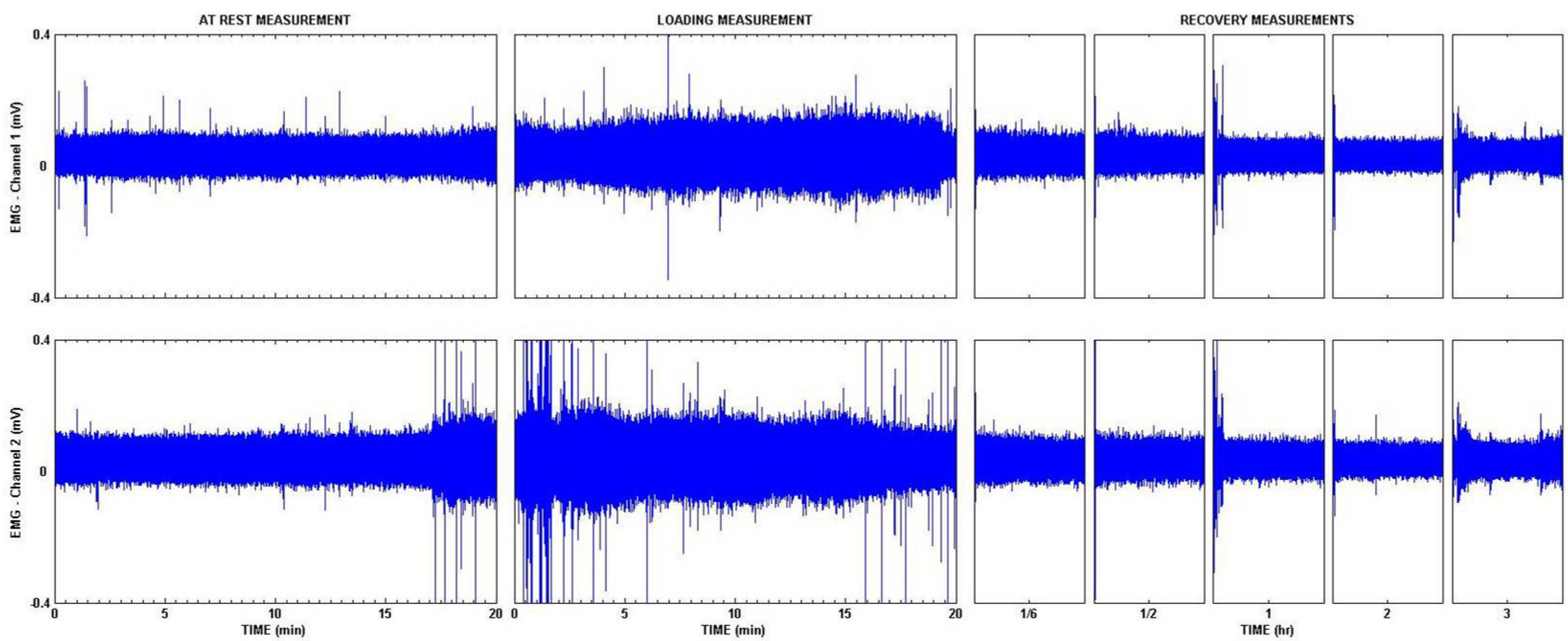

Figure 30. Raw EMG Data - Participant 7

PARTICIPANT 8

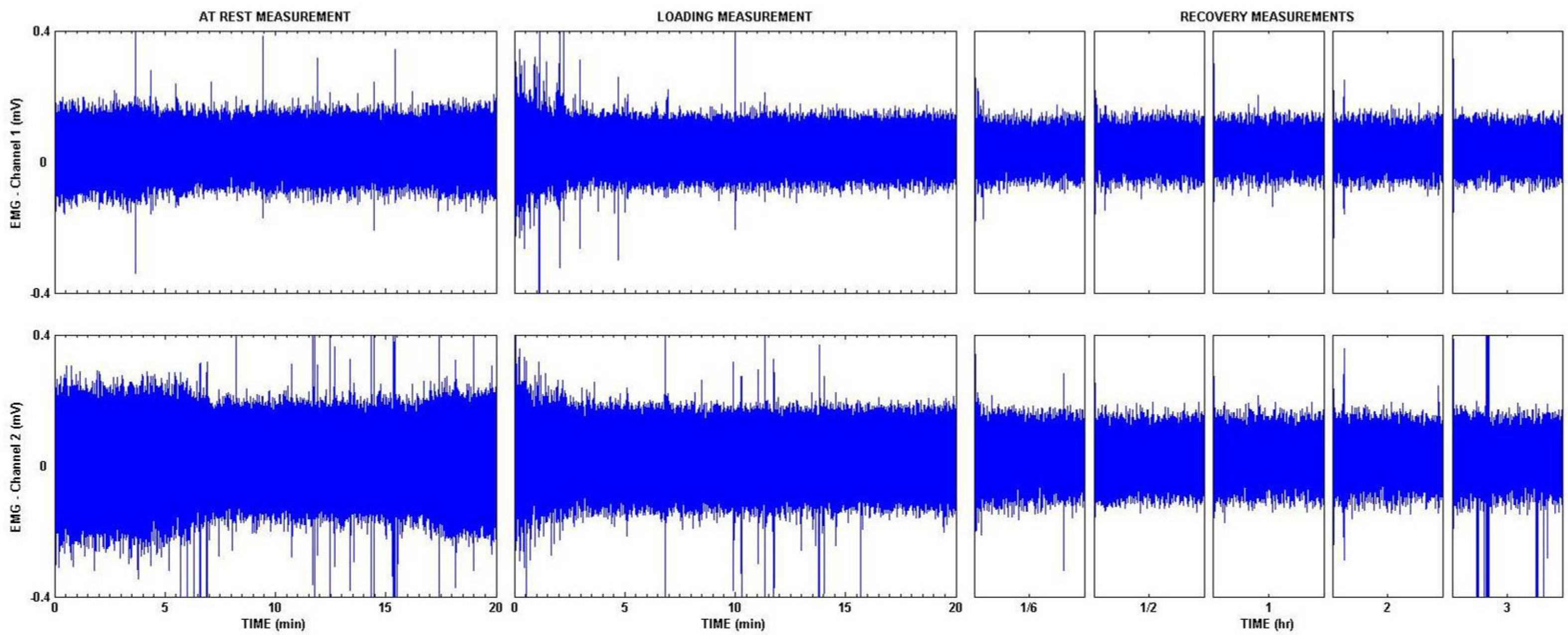

Figure 31. Raw EMG Data - Participant 8

PARTICIPANT 9

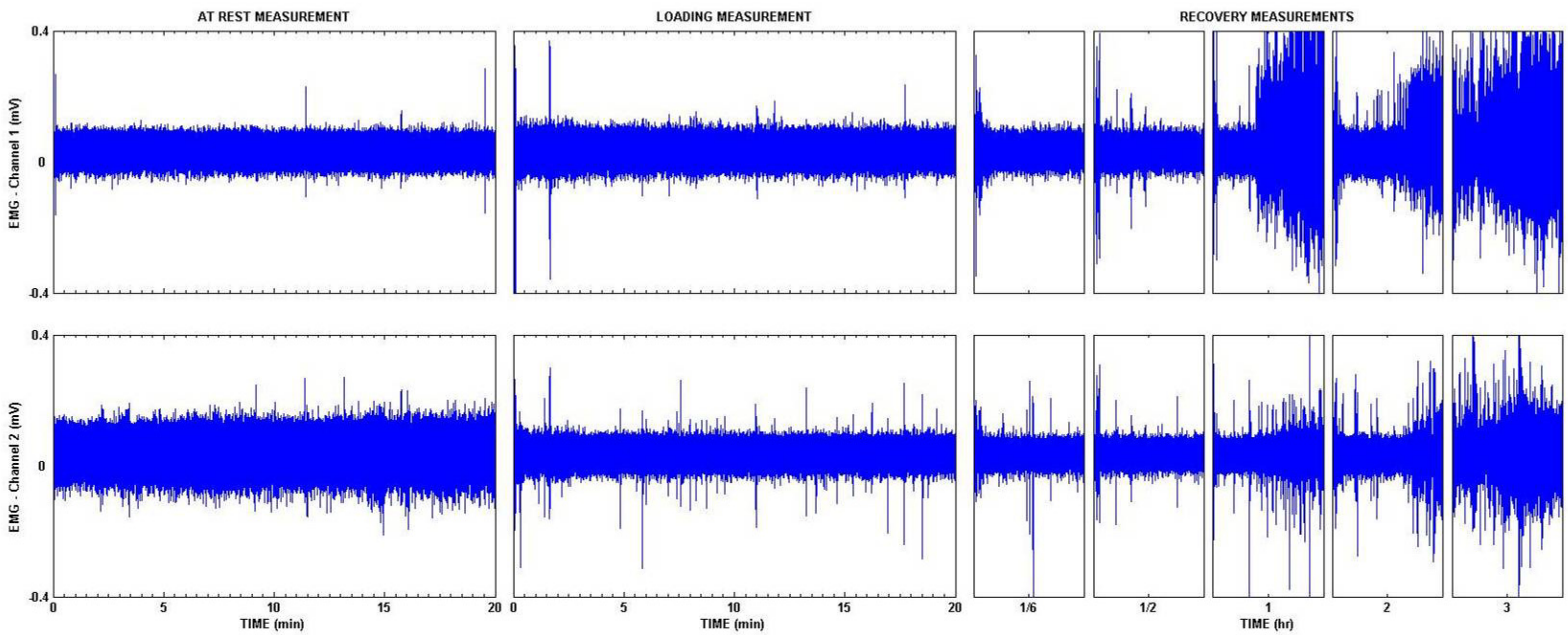

Figure 32. Raw EMG Data - Participant 9

PARTICIPANT 10

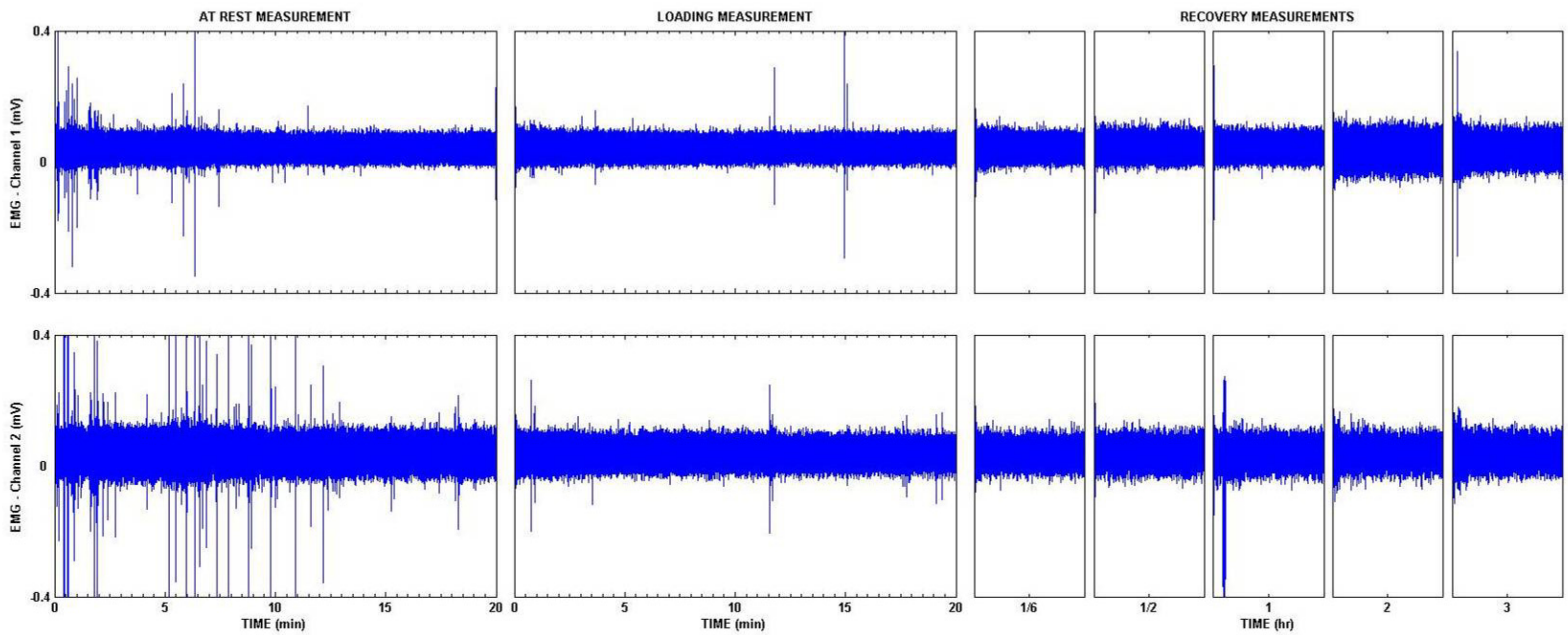

Figure 33. Raw EMG Data - Participant 10

APPENDIX C - FULL WAVE RECTIFIED EMG DATA

PARTICIPANT 1

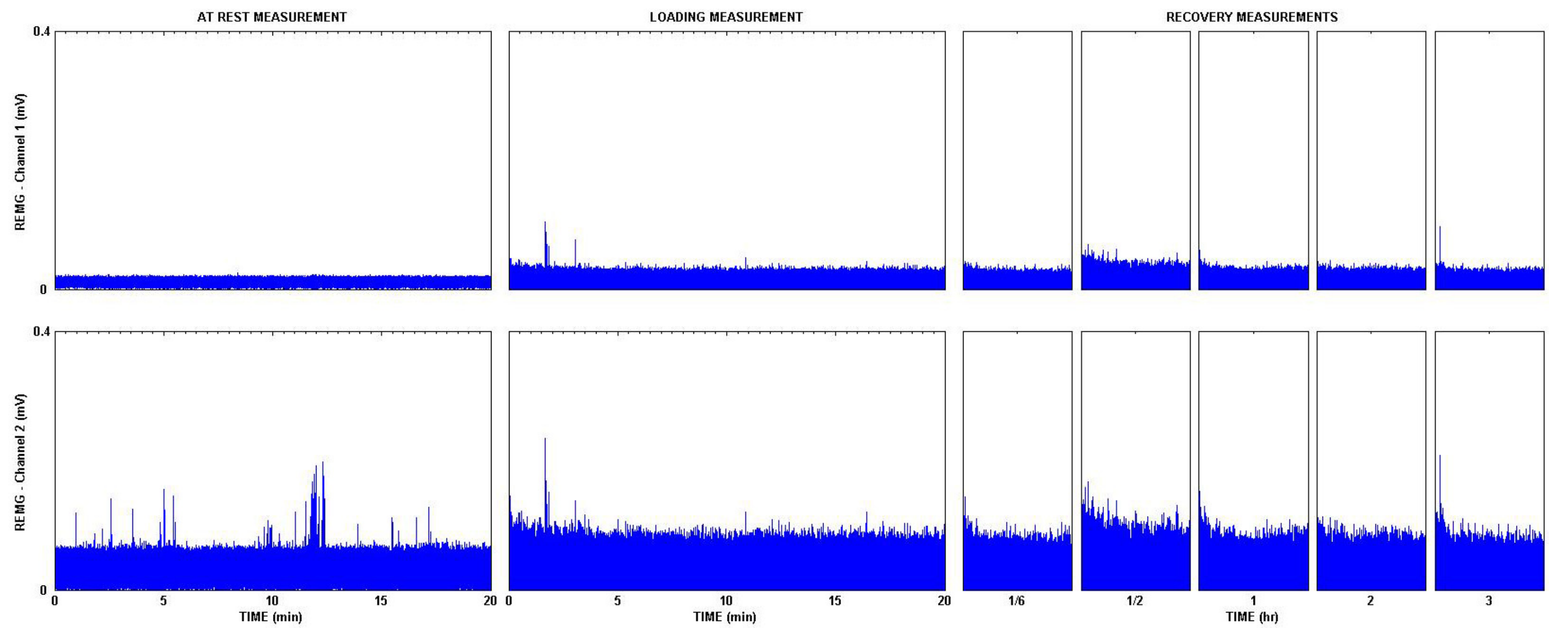

Figure 34. Full Wave Rectified EMG Data - Participant 1

PARTICIPANT 2

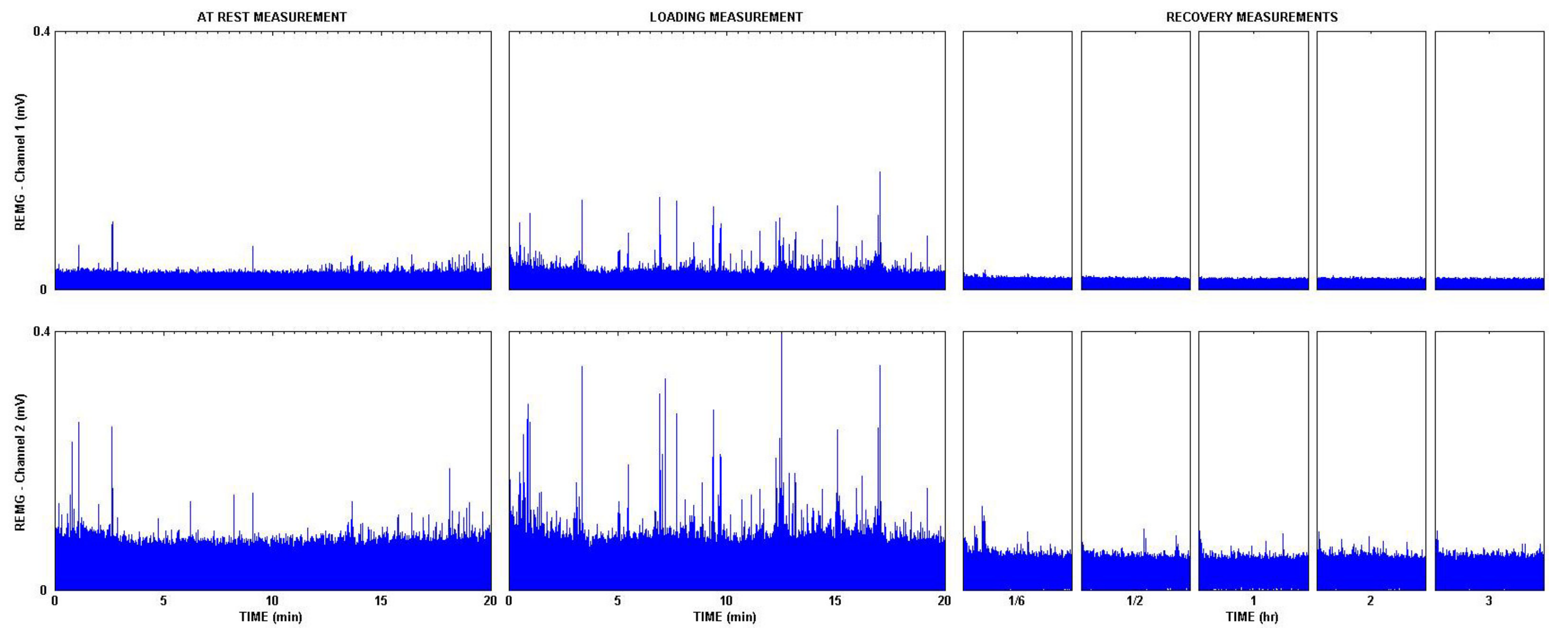

Figure 35. Full Wave Rectified EMG Data - Participant 2

PARTICIPANT 3

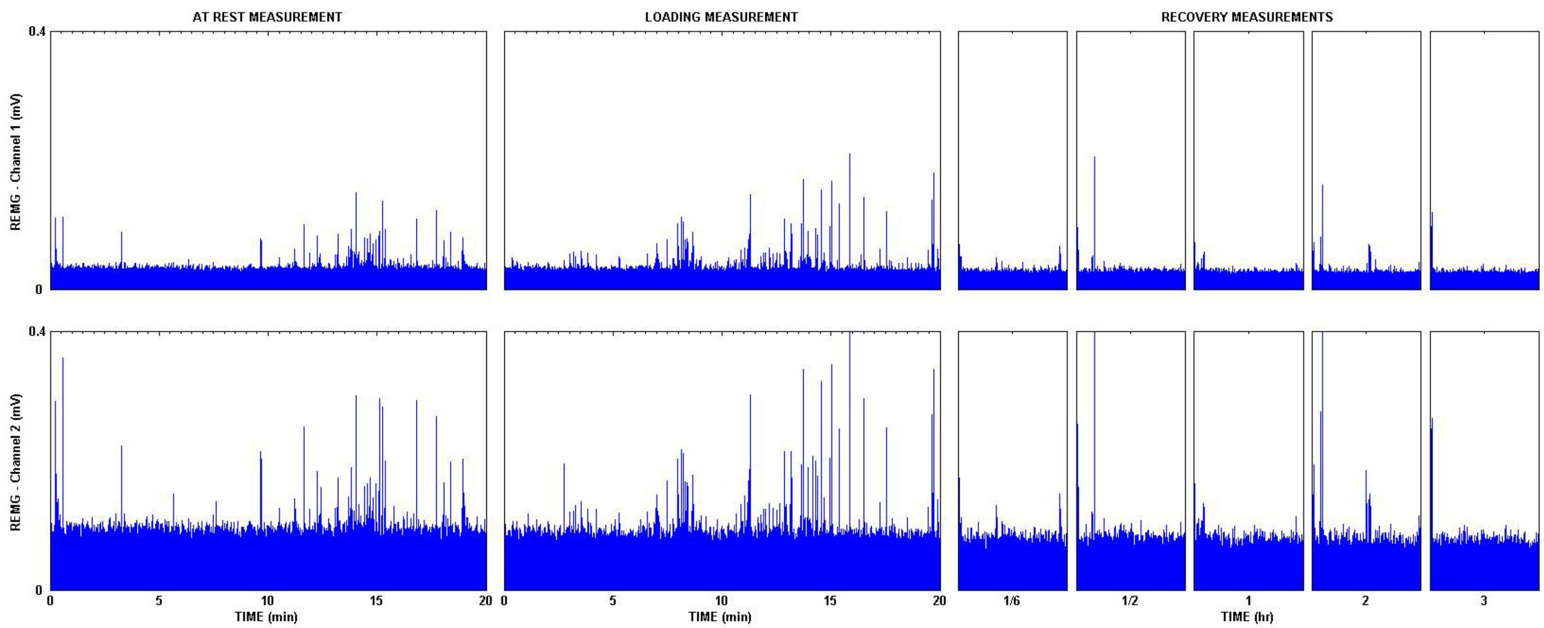

Figure 36. Full Wave Rectified EMG Data - Participant 3

PARTICIPANT 4

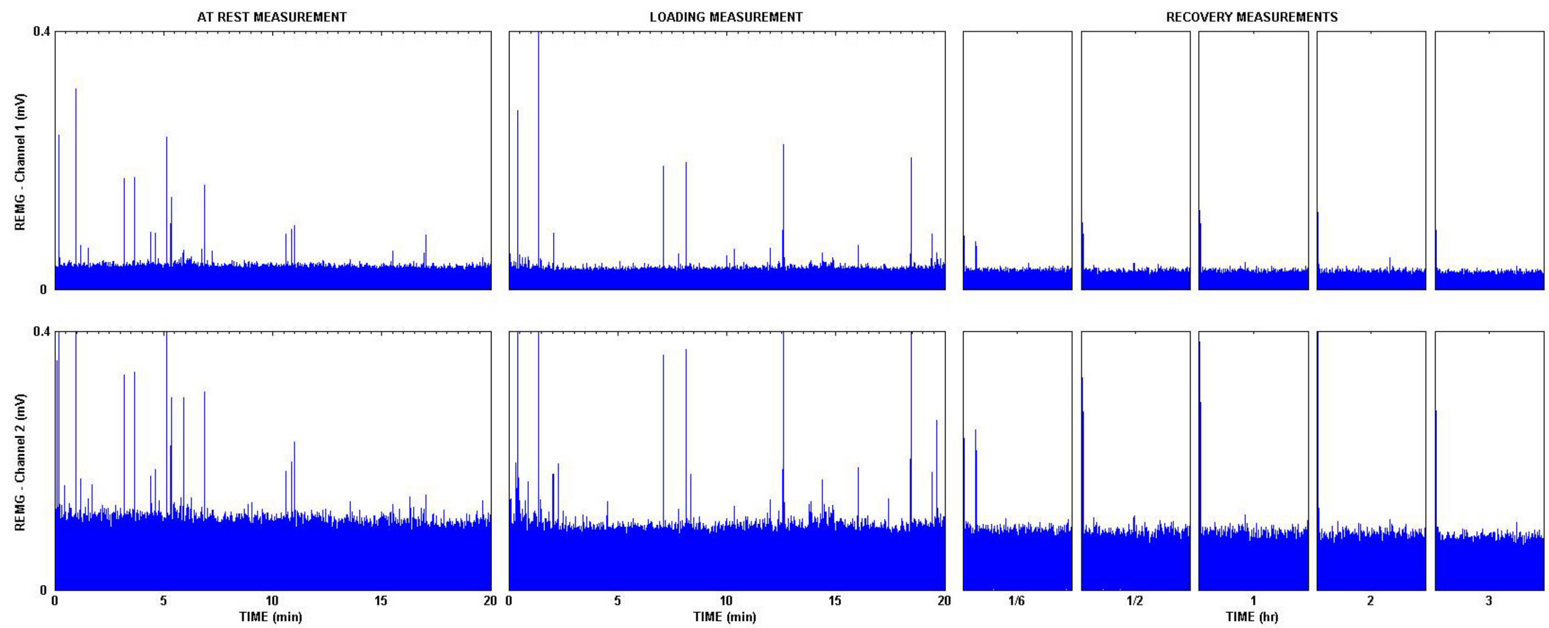

Figure 37. Full Wave Rectified EMG Data - Participant 4

PARTICIPANT 5

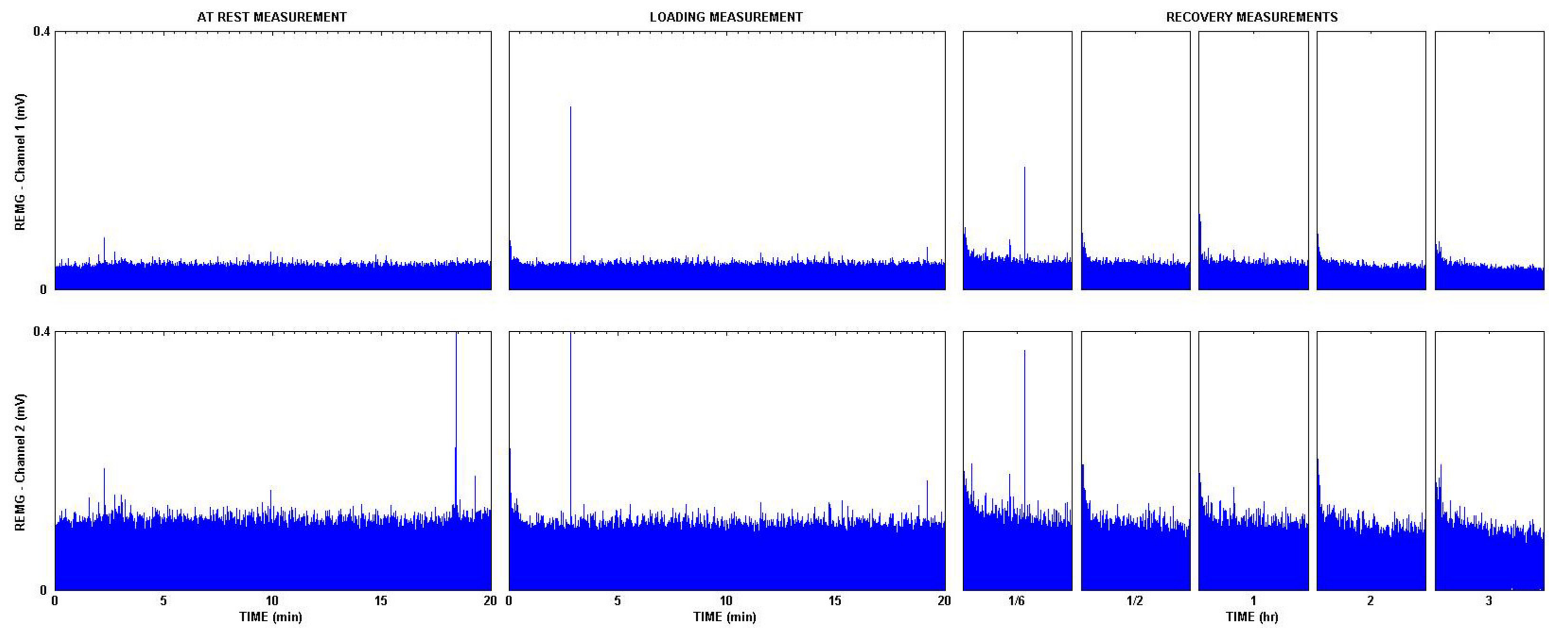

Figure 38. Full Wave Rectified EMG Data - Participant 5

PARTICIPANT 6

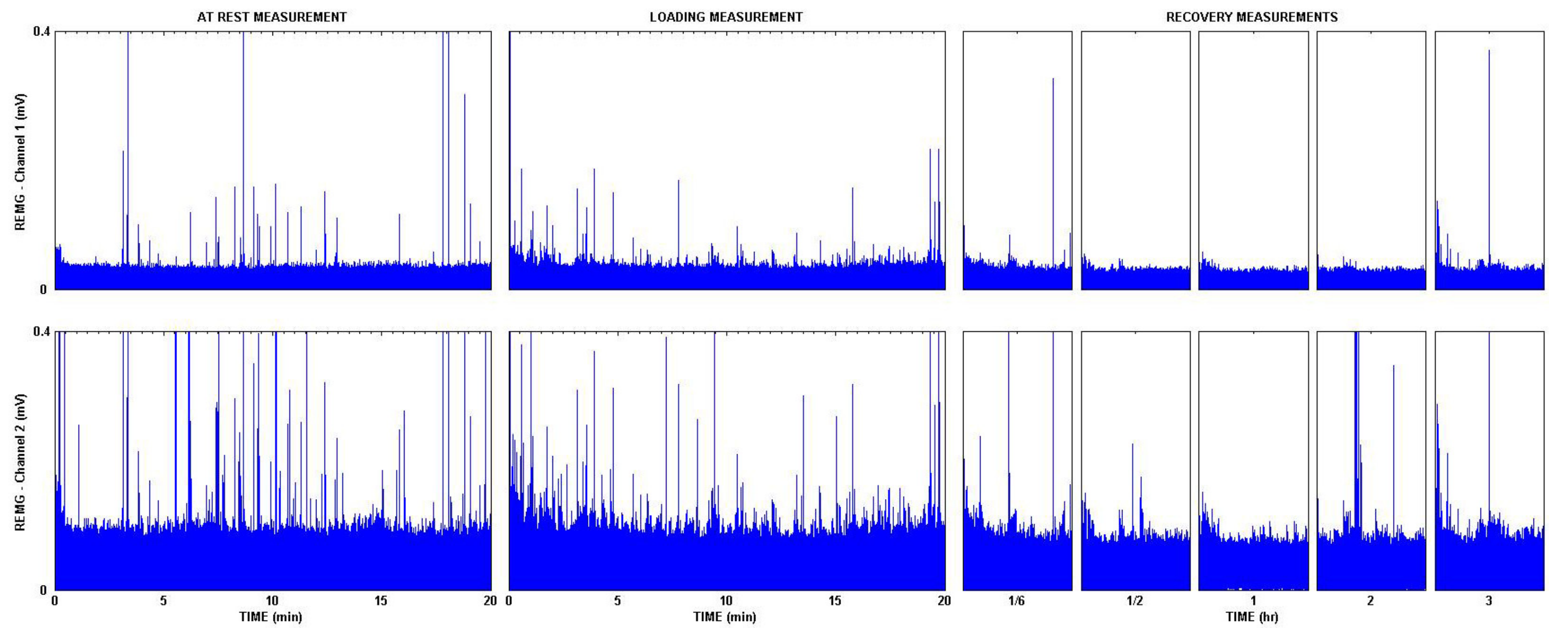

Figure 39. Full Wave Rectified EMG Data - Participant 6

PARTICIPANT 7

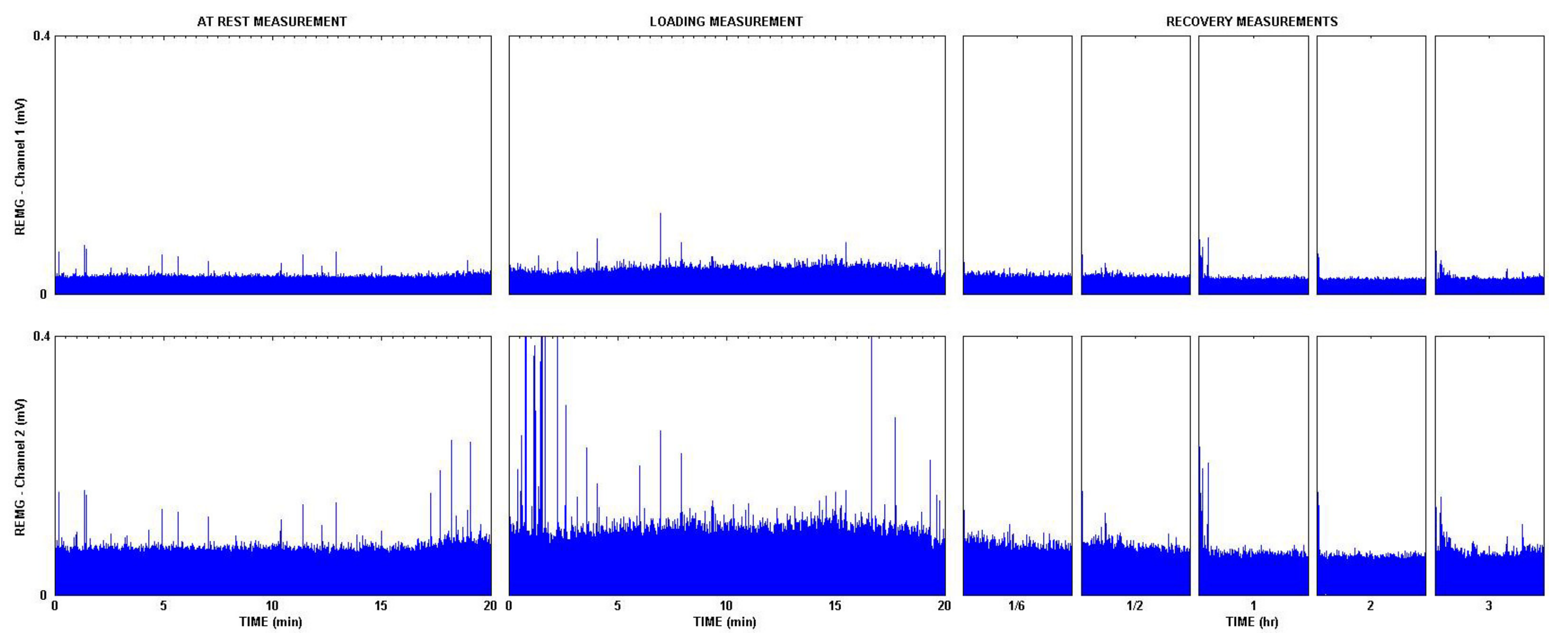

Figure 40. Full Wave Rectified EMG Data - Participant 7

PARTICIPANT 8

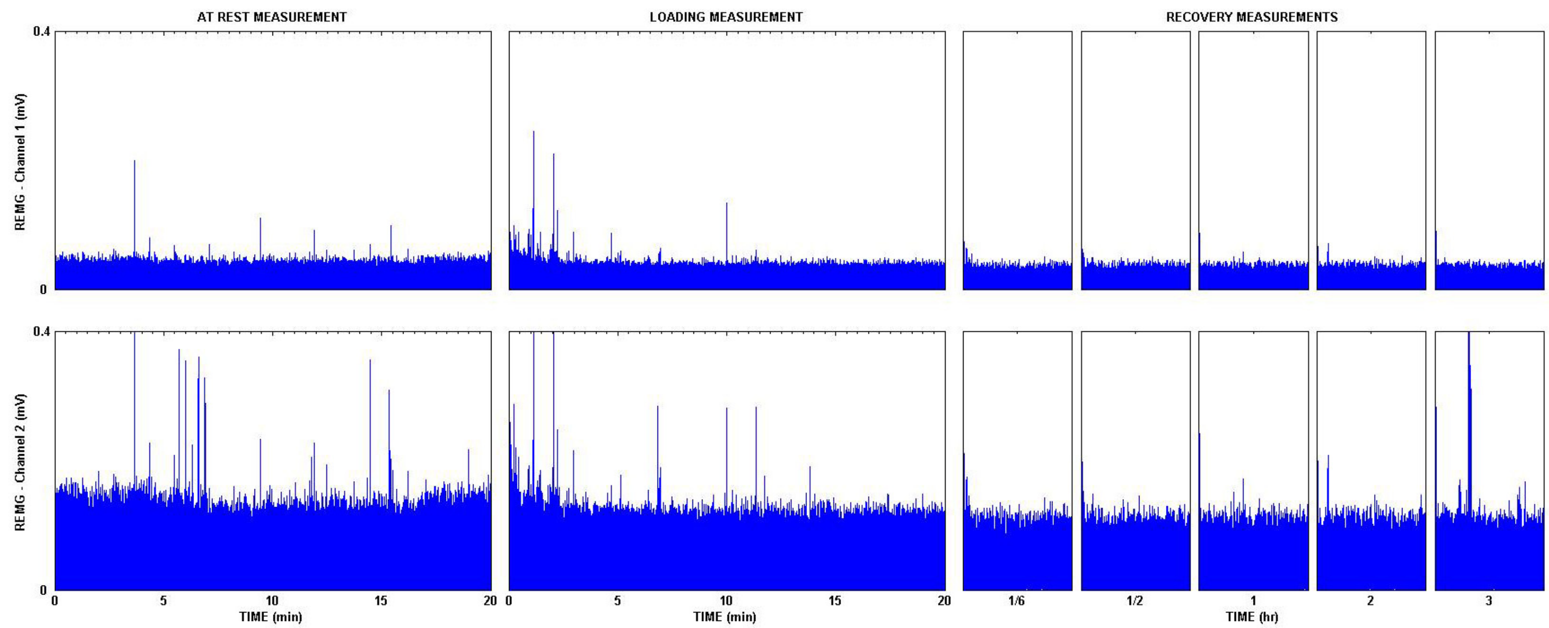

Figure 41. Full Wave Rectified EMG Data - Participant 8

PARTICIPANT 9

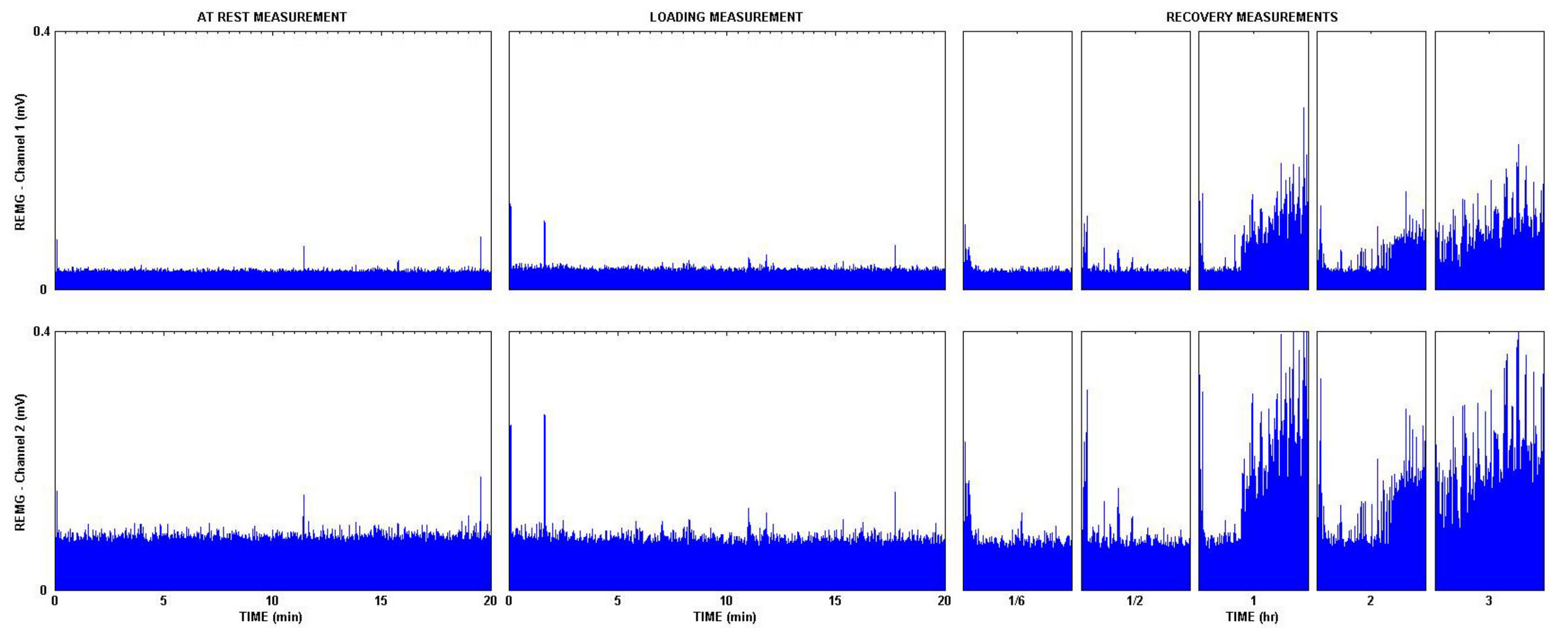

Figure 42. Full Wave Rectified EMG Data - Participant 9

PARTICIPANT 10

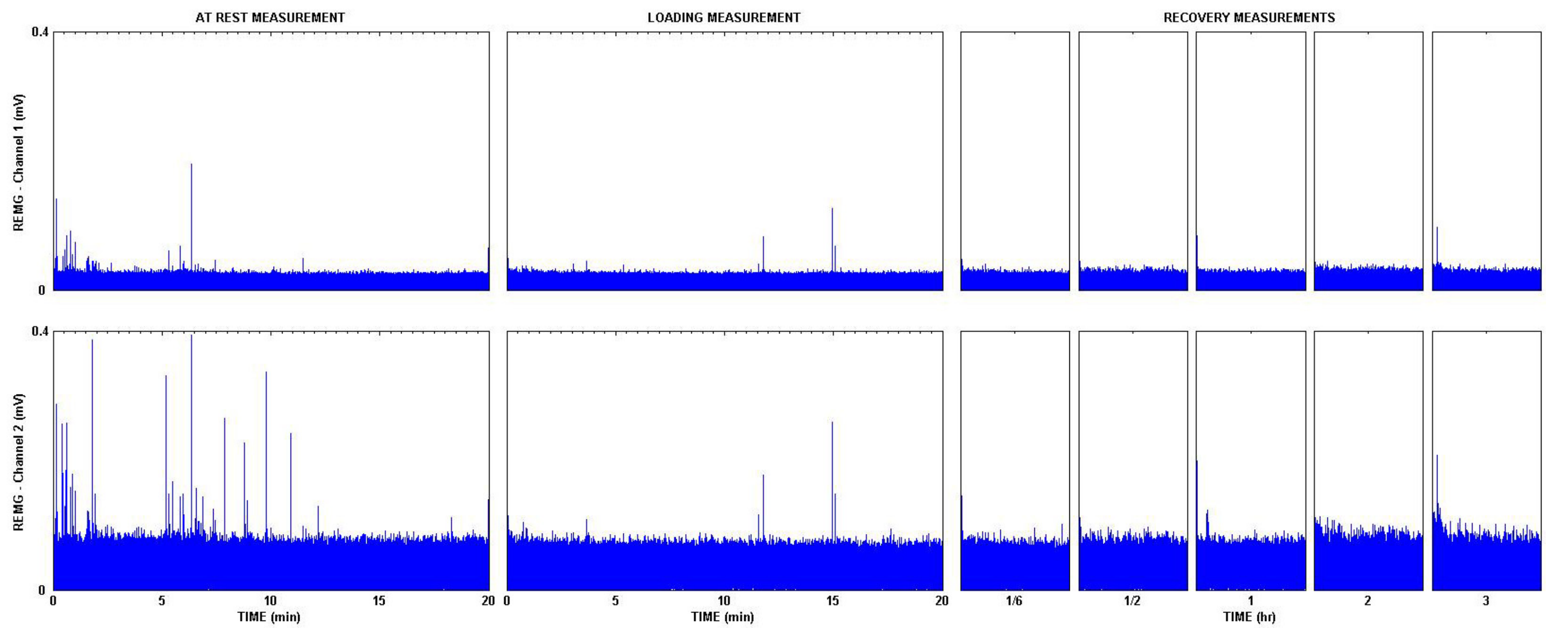

Figure 43. Full Wave Rectified EMG Data - Participant 10

APPENDIX D - NORMALIZED INTEGRATED EMG DATA

PARTICIPANT 1

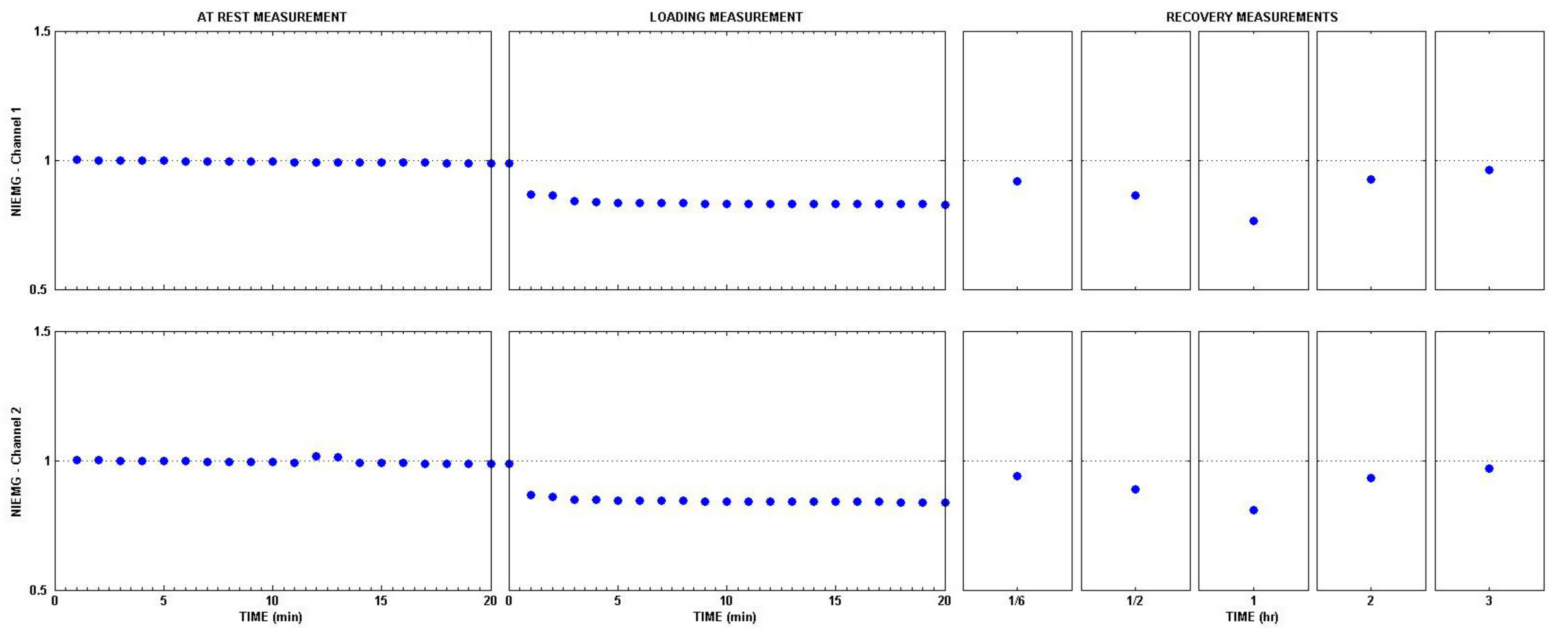

Figure 44. Normalized Integrated EMG Data - Participant 1

PARTICIPANT 2

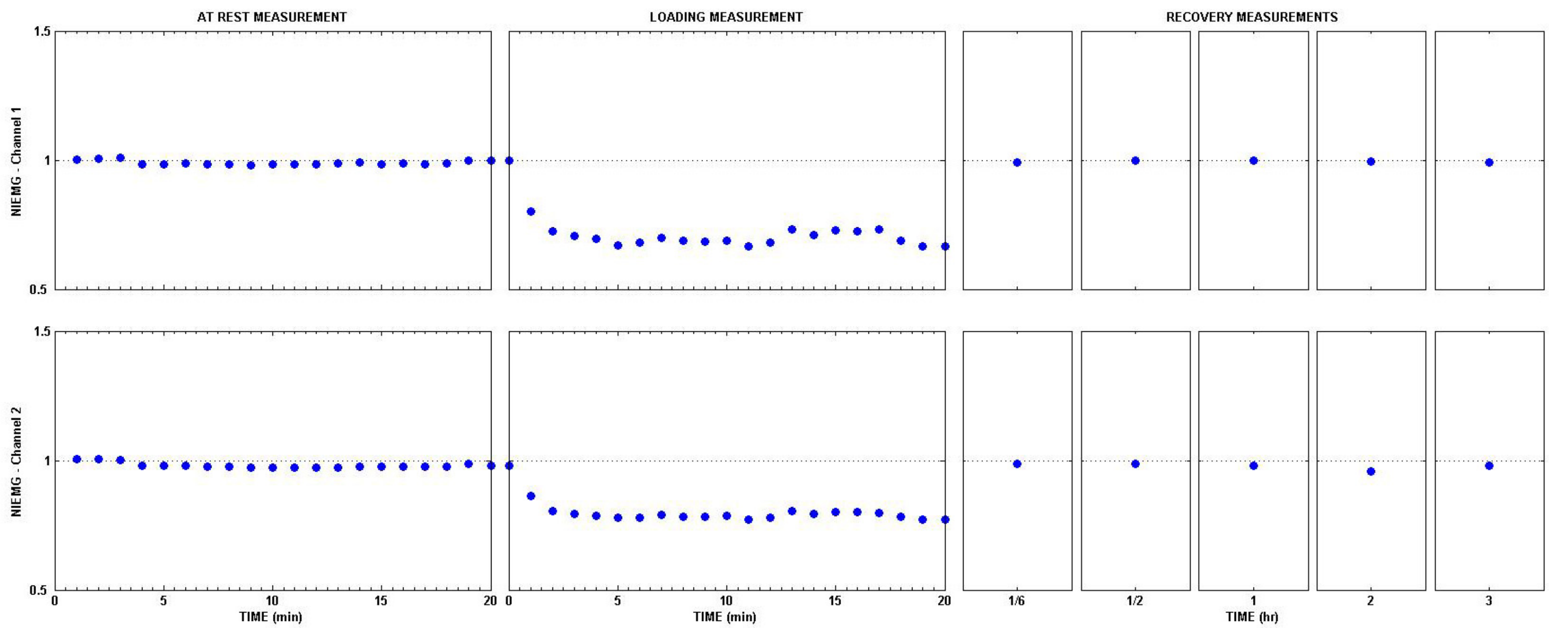

Figure 45. Normalized Integrated EMG Data - Participant 2

PARTICIPANT 3

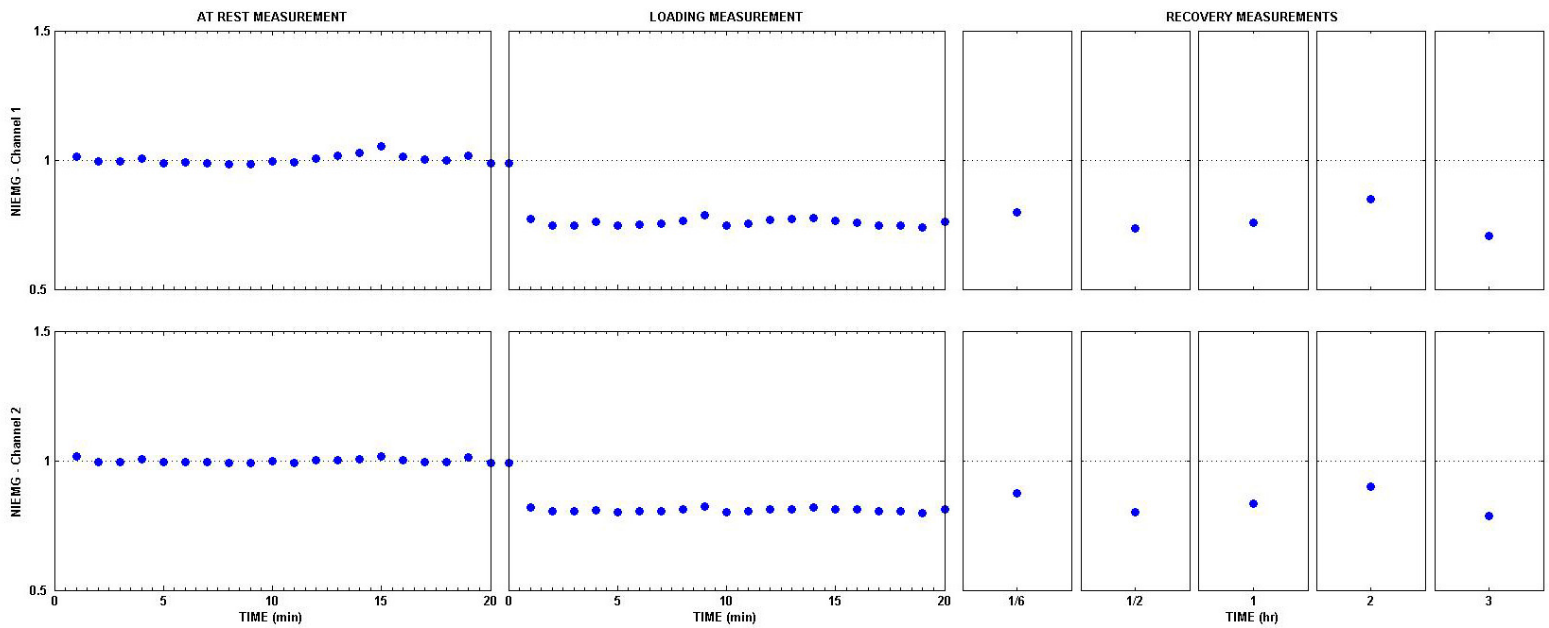

Figure 46. Normalized Integrated EMG Data - Participant 3

PARTICIPANT 4

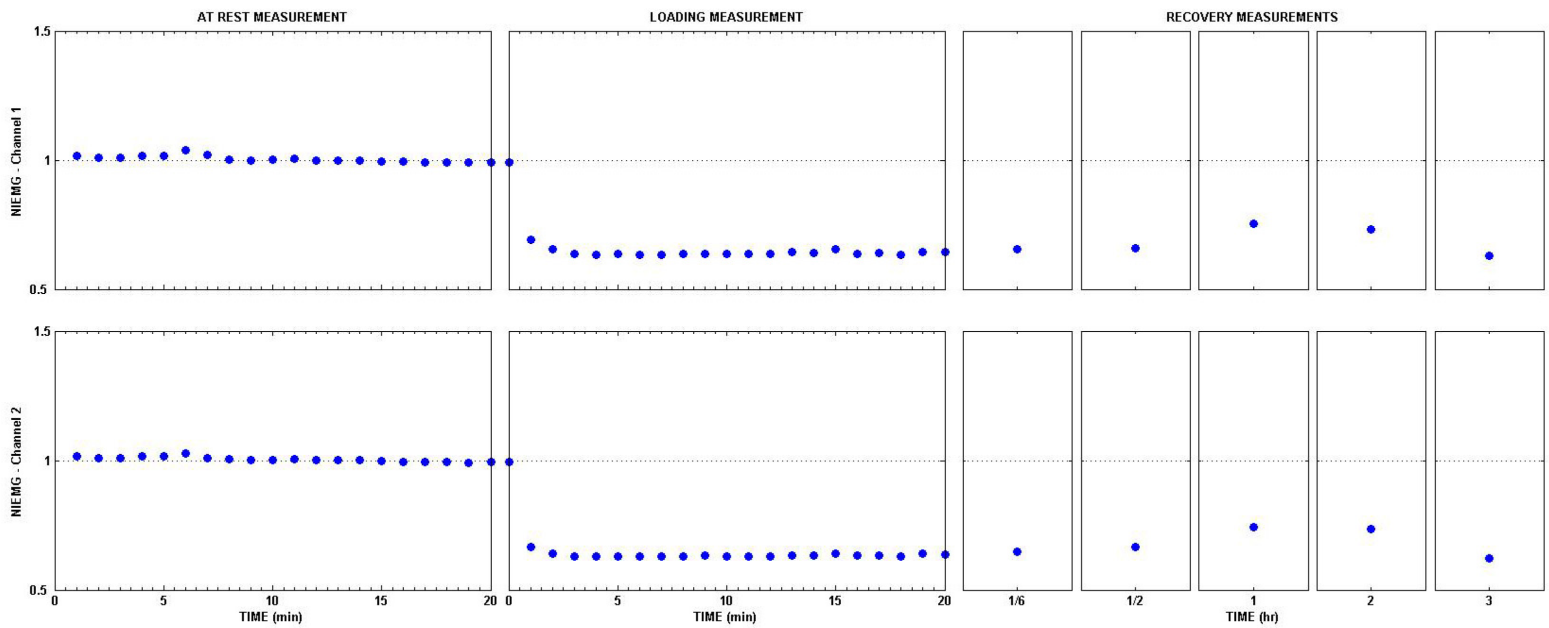

Figure 47. Normalized Integrated EMG Data - Participant 4

PARTICIPANT 5

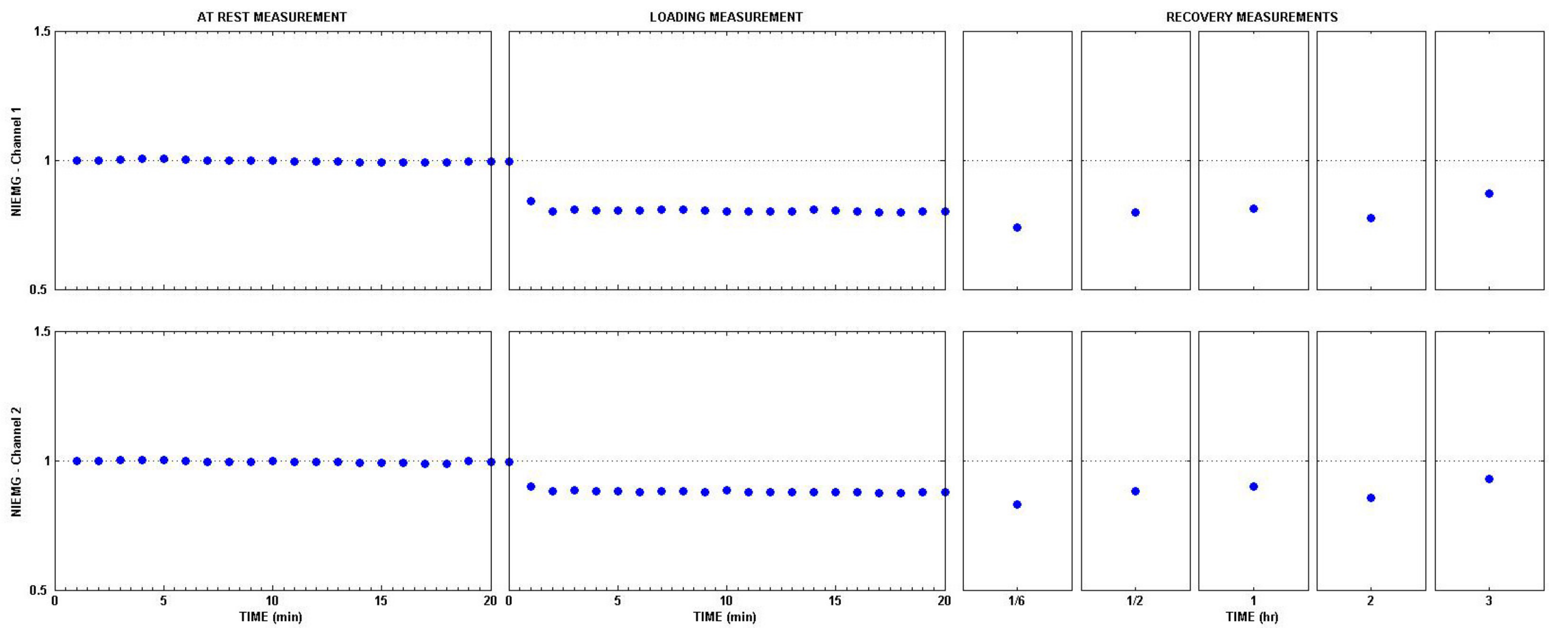

Figure 48. Normalized Integrated EMG Data - Participant 5

PARTICIPANT 6

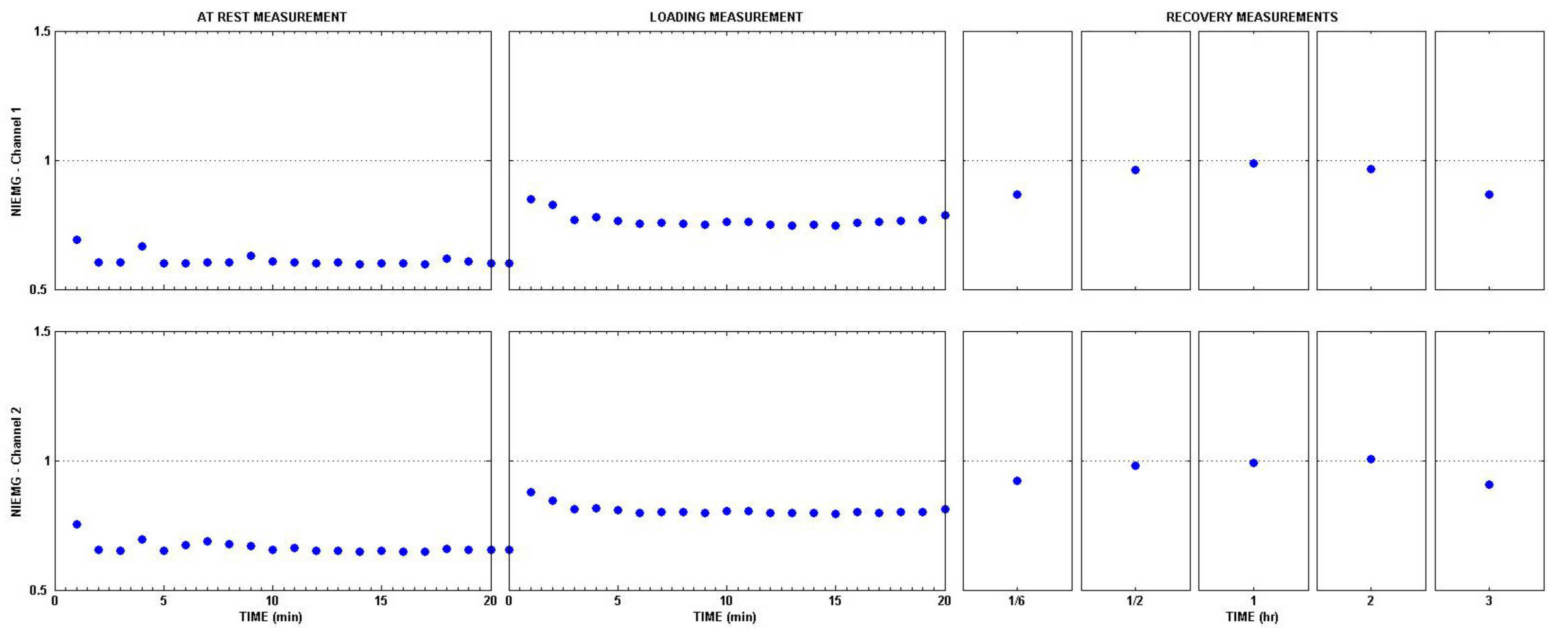

Figure 49. Normalized Integrated EMG Data - Participant 6

PARTICIPANT 7

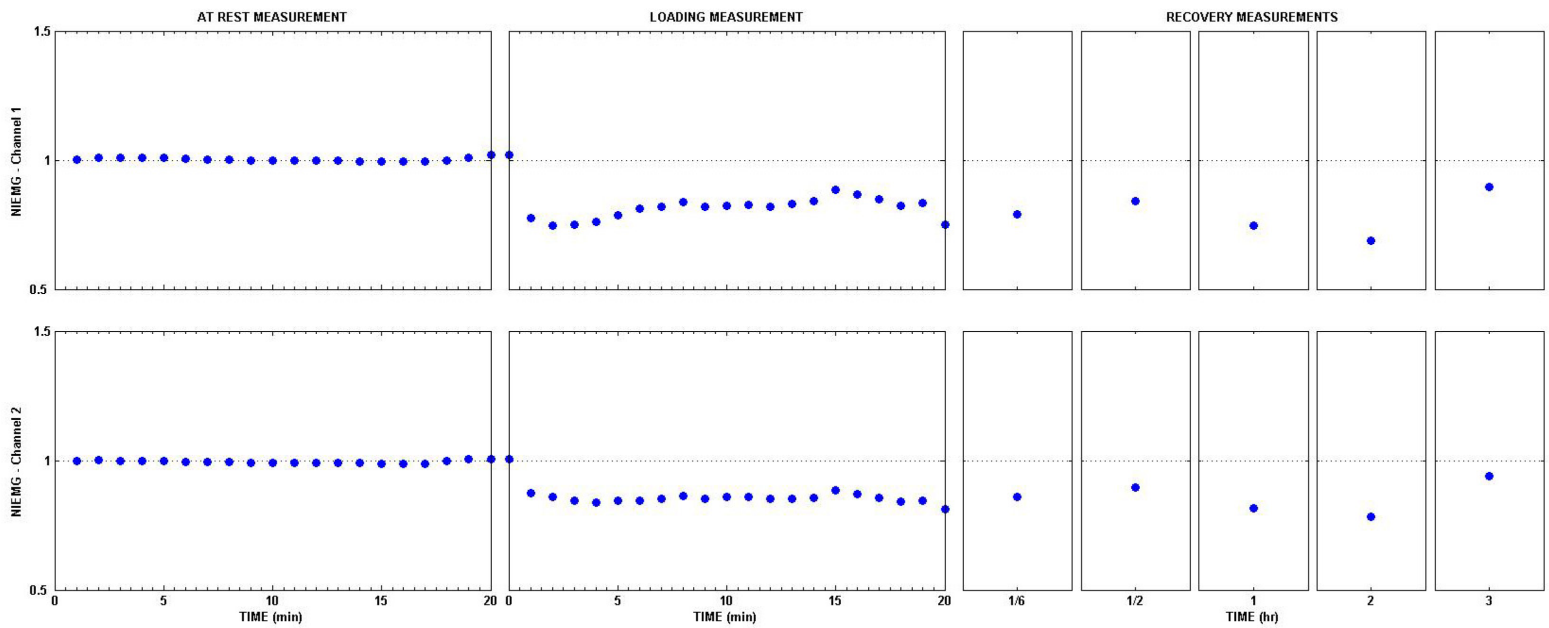

Figure 50. Normalized Integrated EMG Data - Participant 7

PARTICIPANT 8

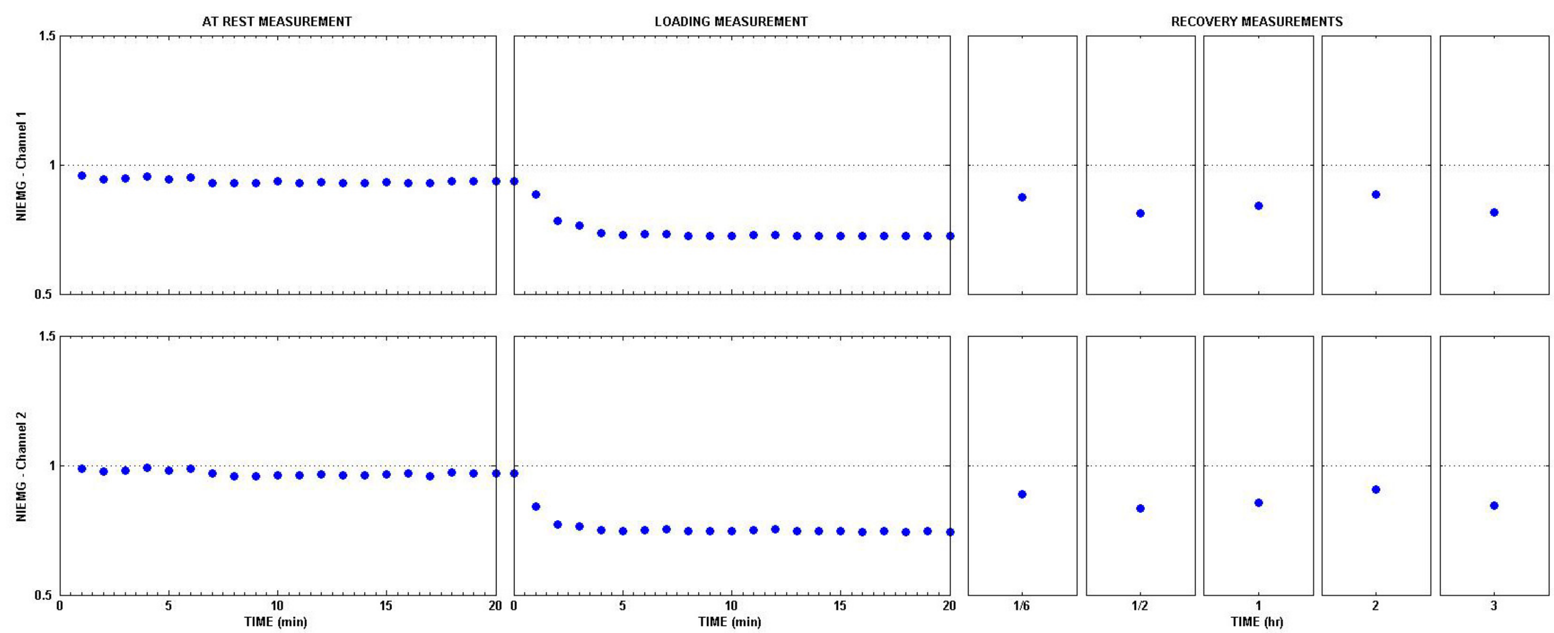

Figure 51. Normalized Integrated EMG Data - Participant 8

PARTICIPANT 9

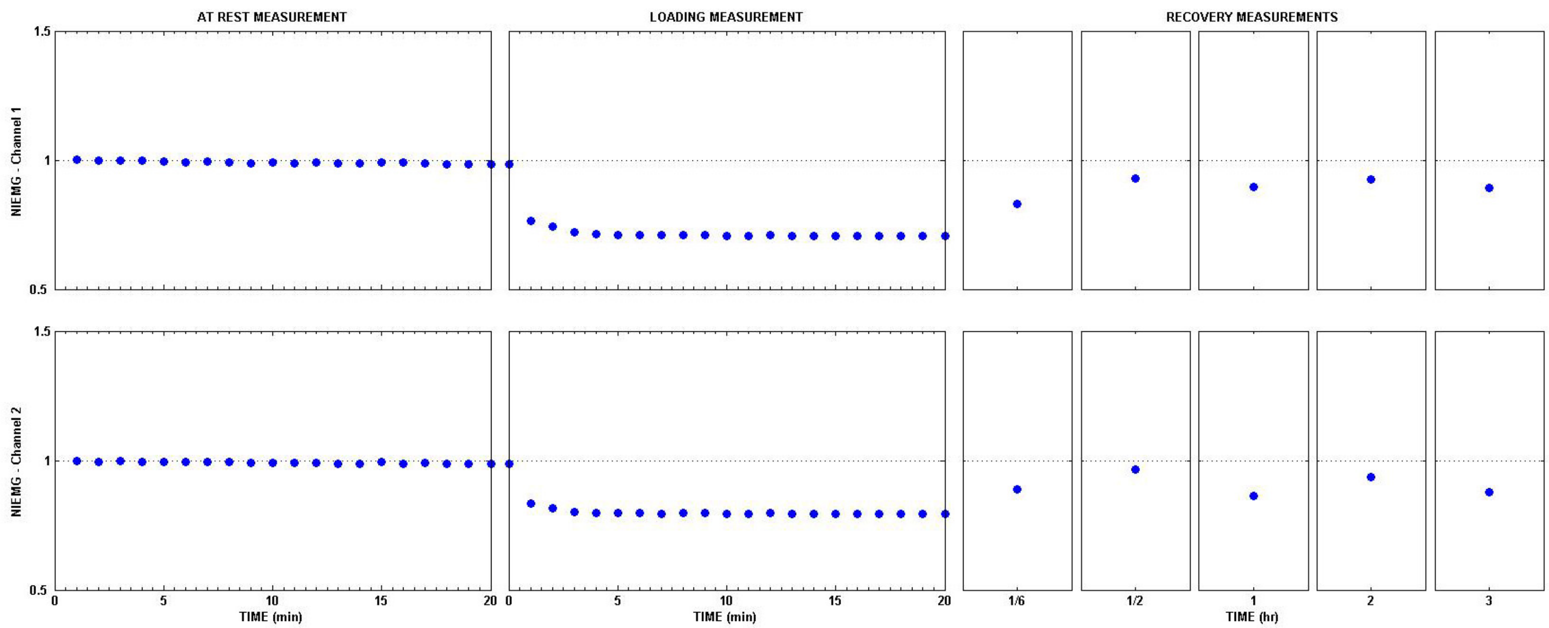

Figure 52. Normalized Integrated EMG Data - Participant 9

PARTICIPANT 10

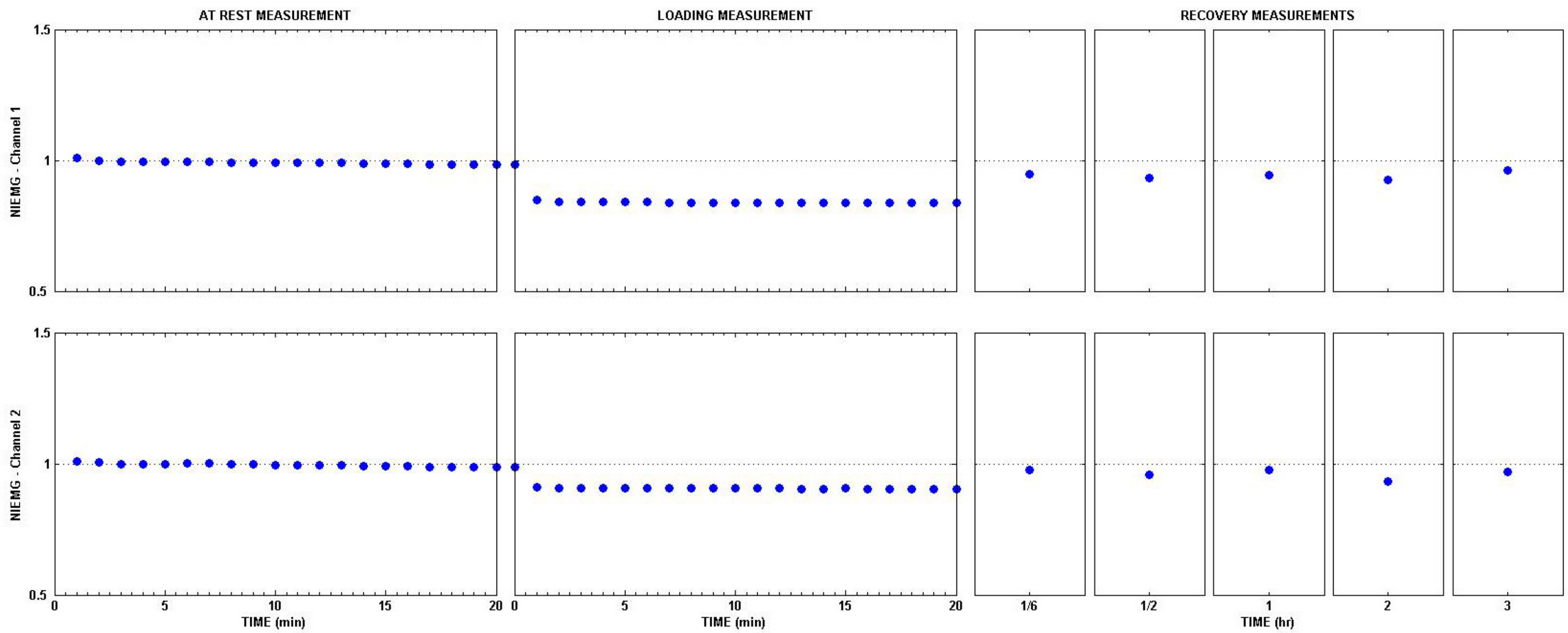

Figure 53. Normalized Integrated EMG Data - Participant 10

APPENDIX E - MEDIAN FREQUENCY DATA FOR FATIGUE ANALYSIS

PARTICIPANT 1

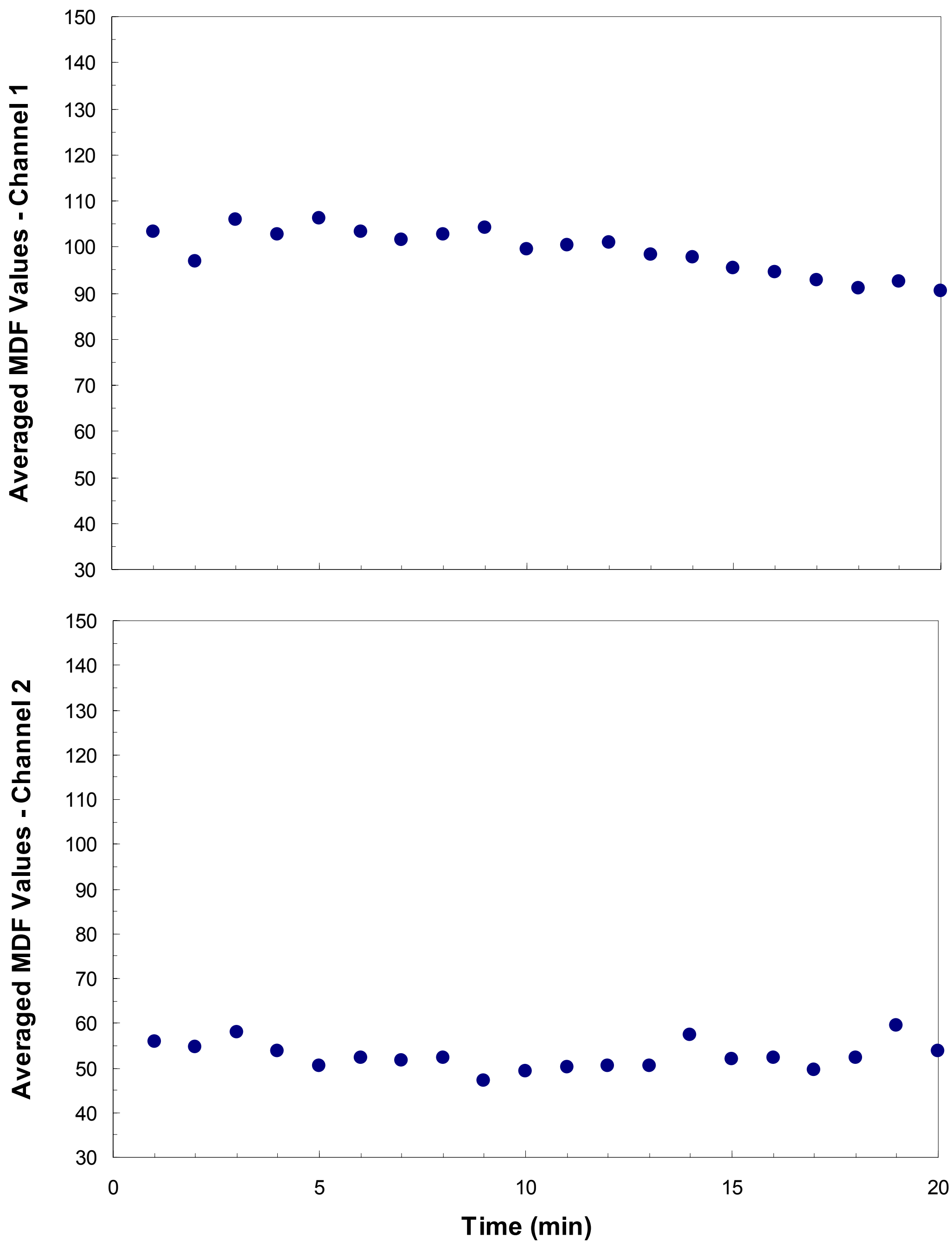

Figure 54. Median Frequency Values for Loading Period - Participant 1

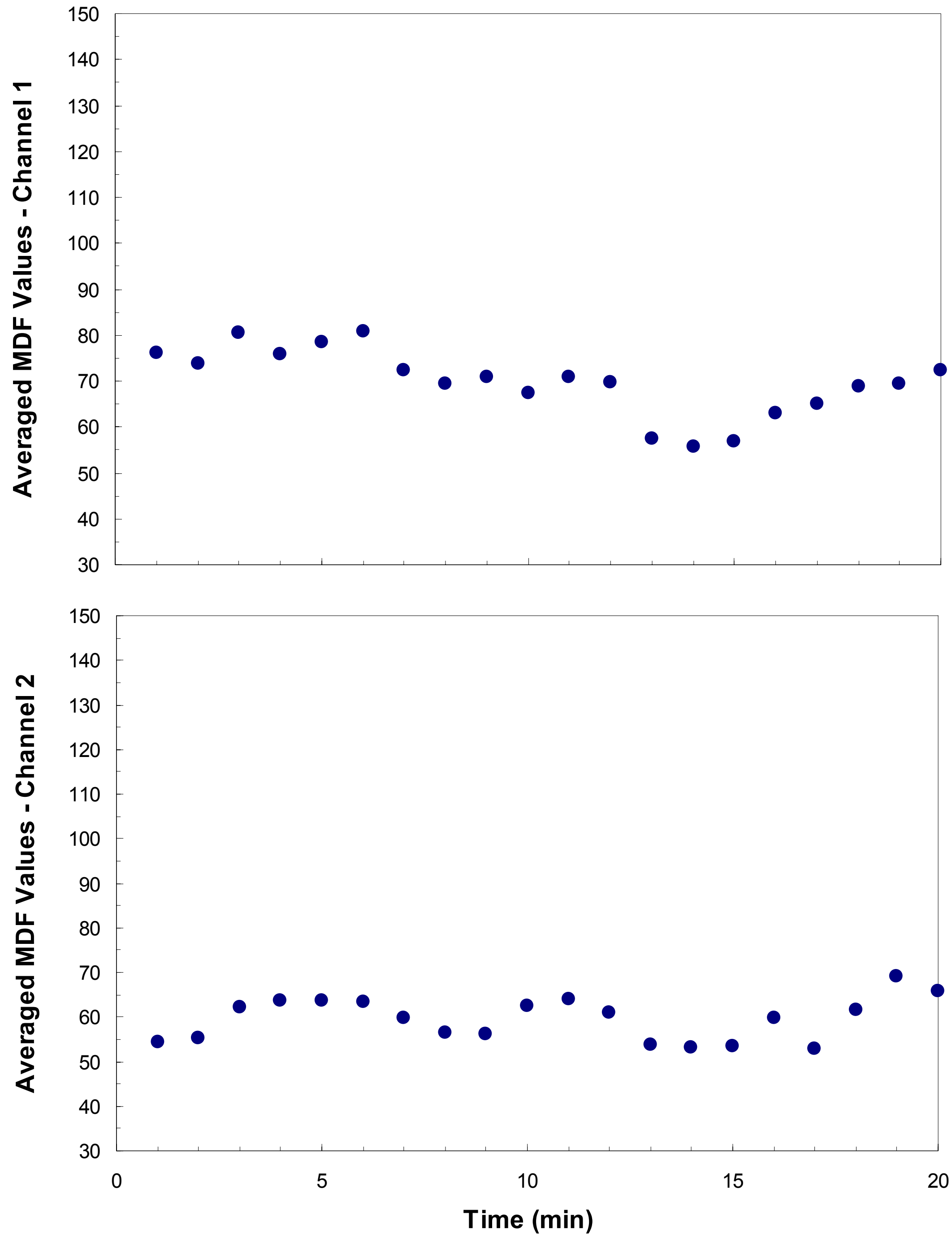

Figure 55. Median Frequency Values for Loading Period - Participant 2

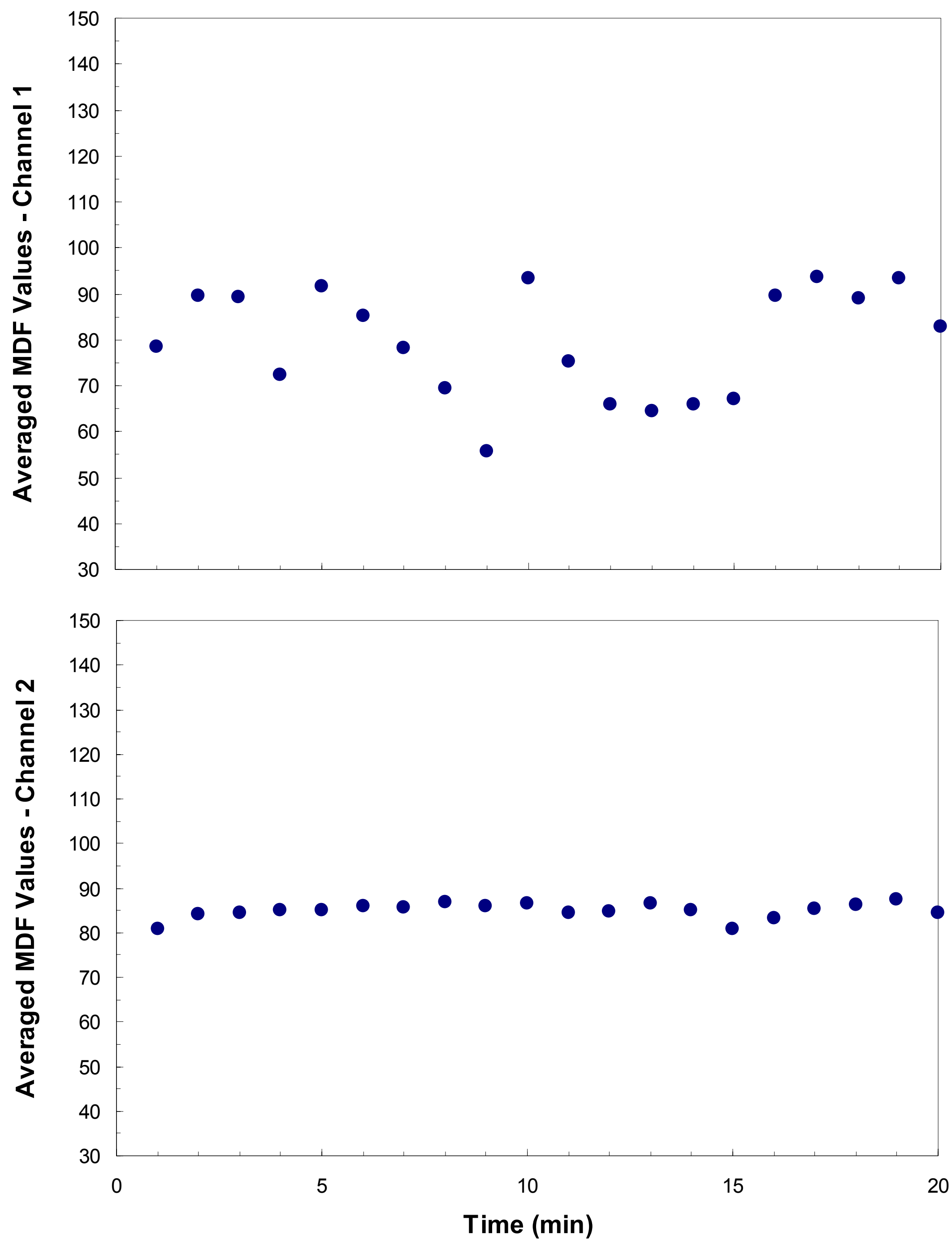

Figure 56. Median Frequency Values for Loading Period - Participant 3

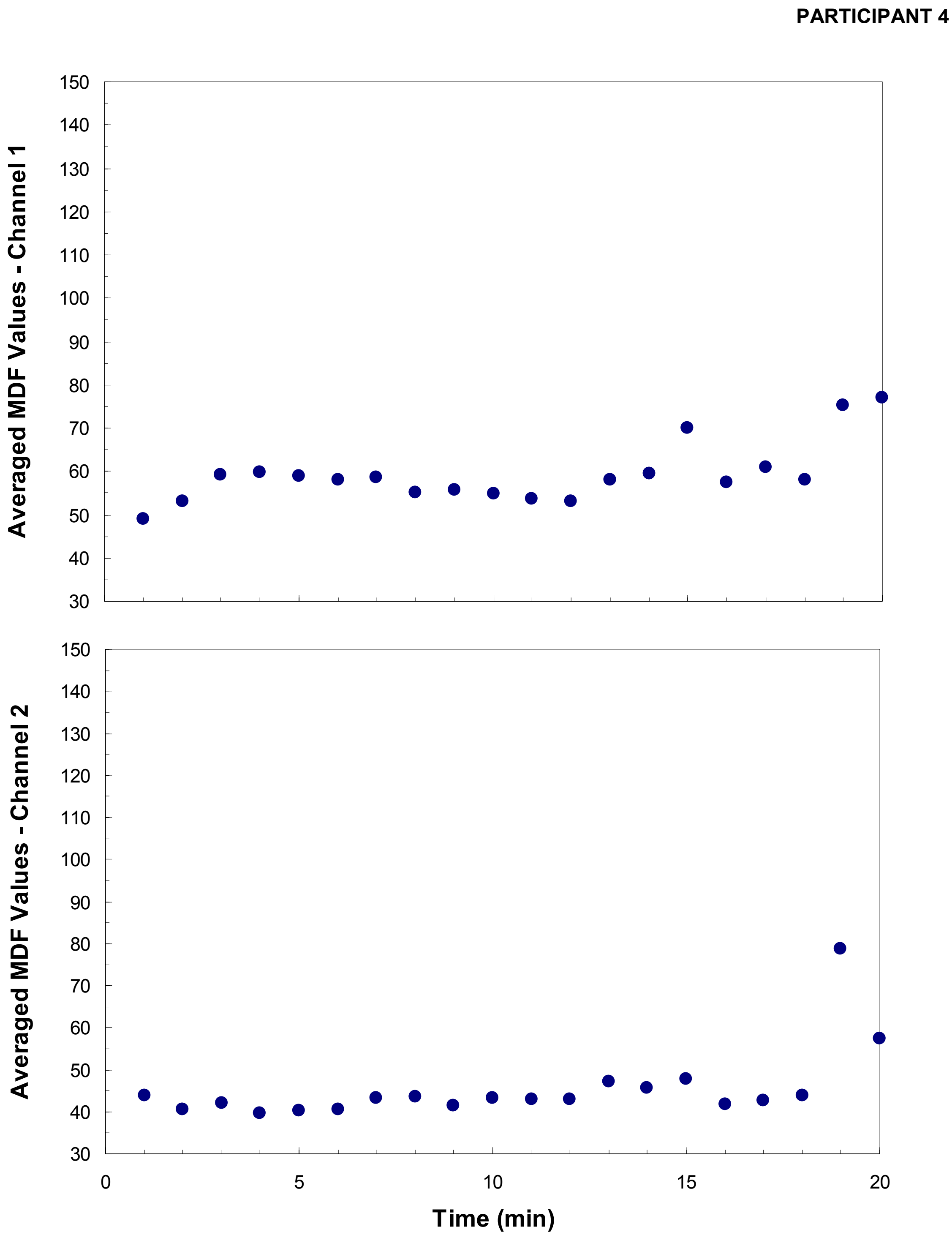

Figure 57. Median Frequency Values for Loading Period - Participant 4

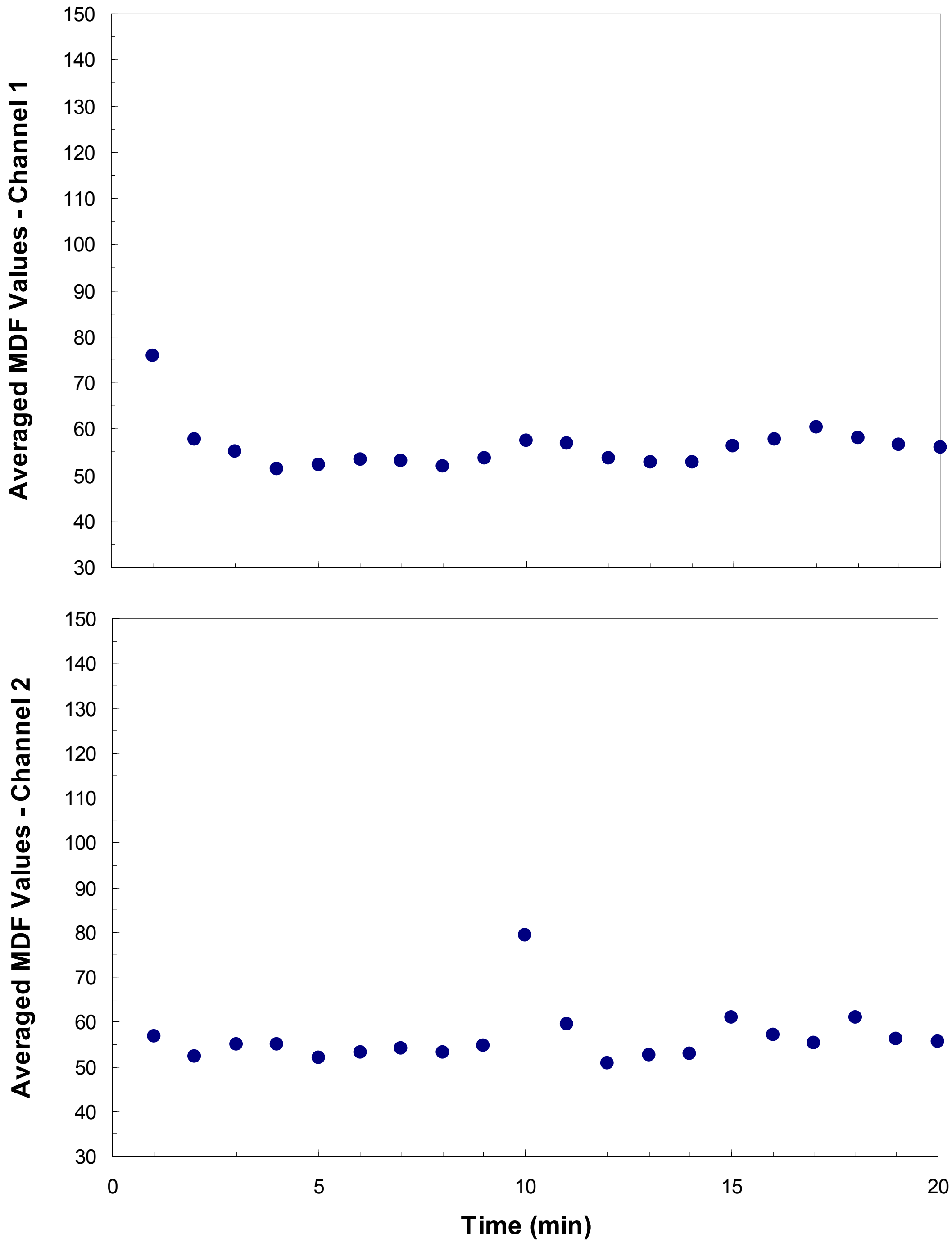

Figure 58. Median Frequency Values for Loading Period - Participant 5

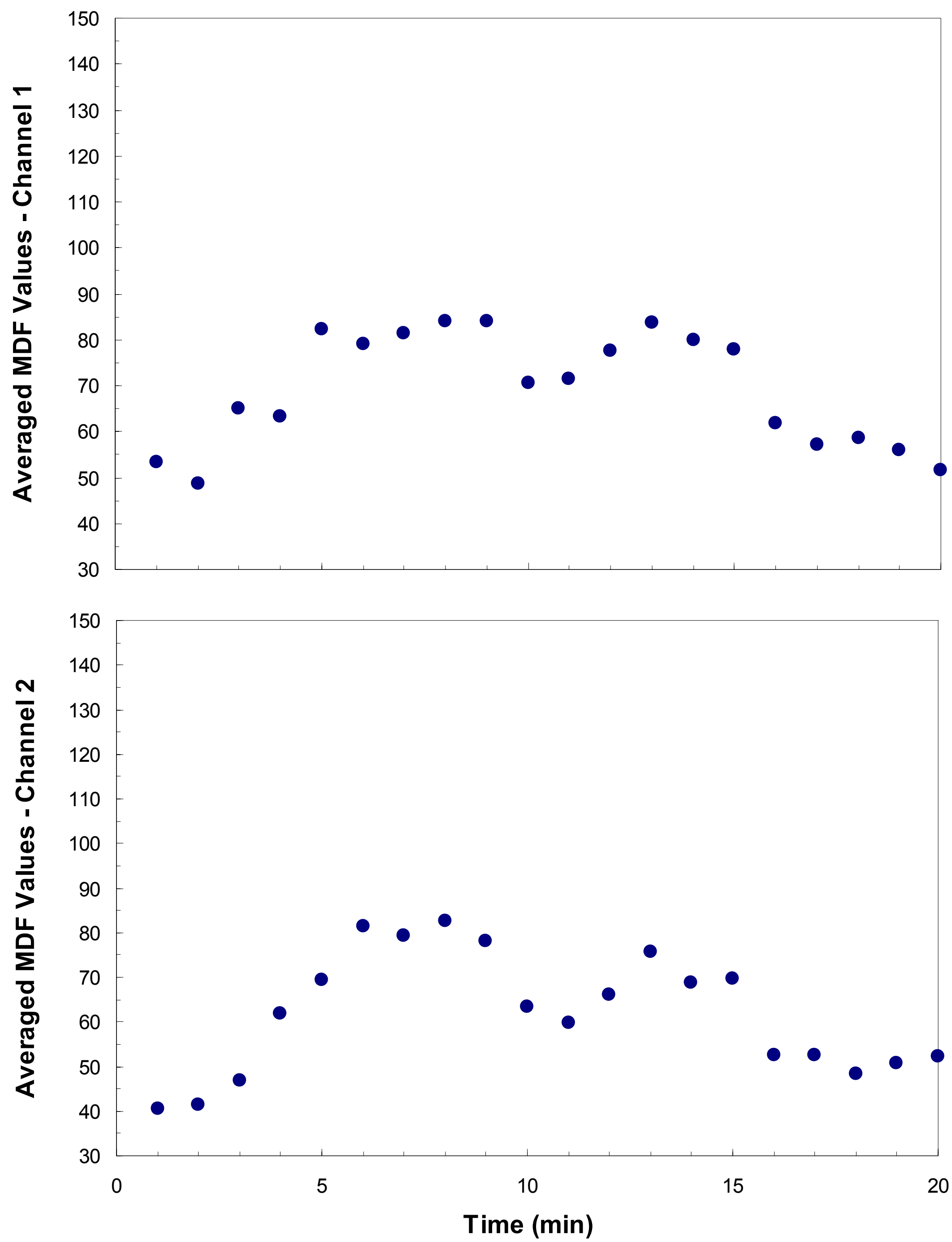

Figure 59. Median Frequency Values for Loading Period - Participant 6

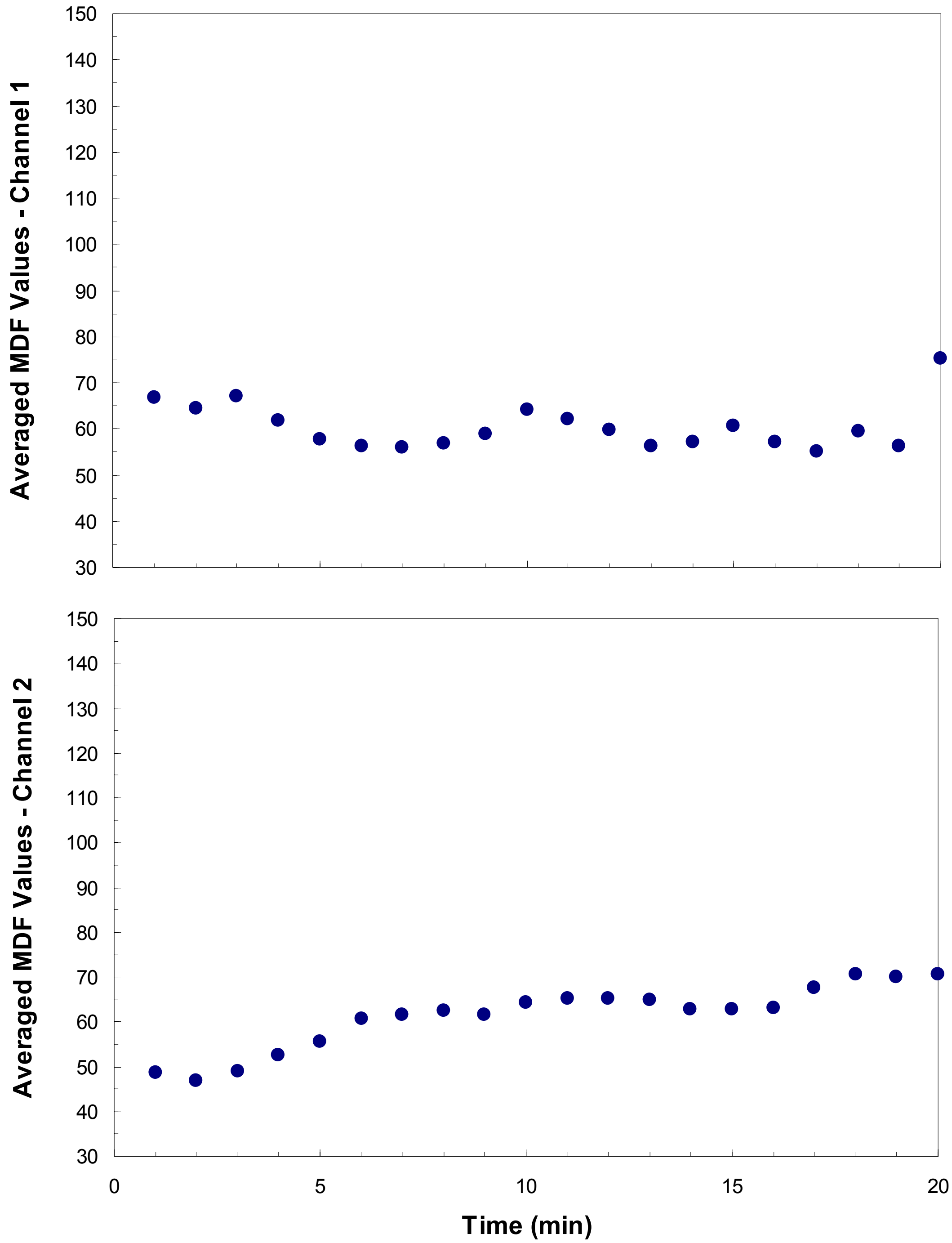

Figure 60. Median Frequency Values for Loading Period - Participant 7

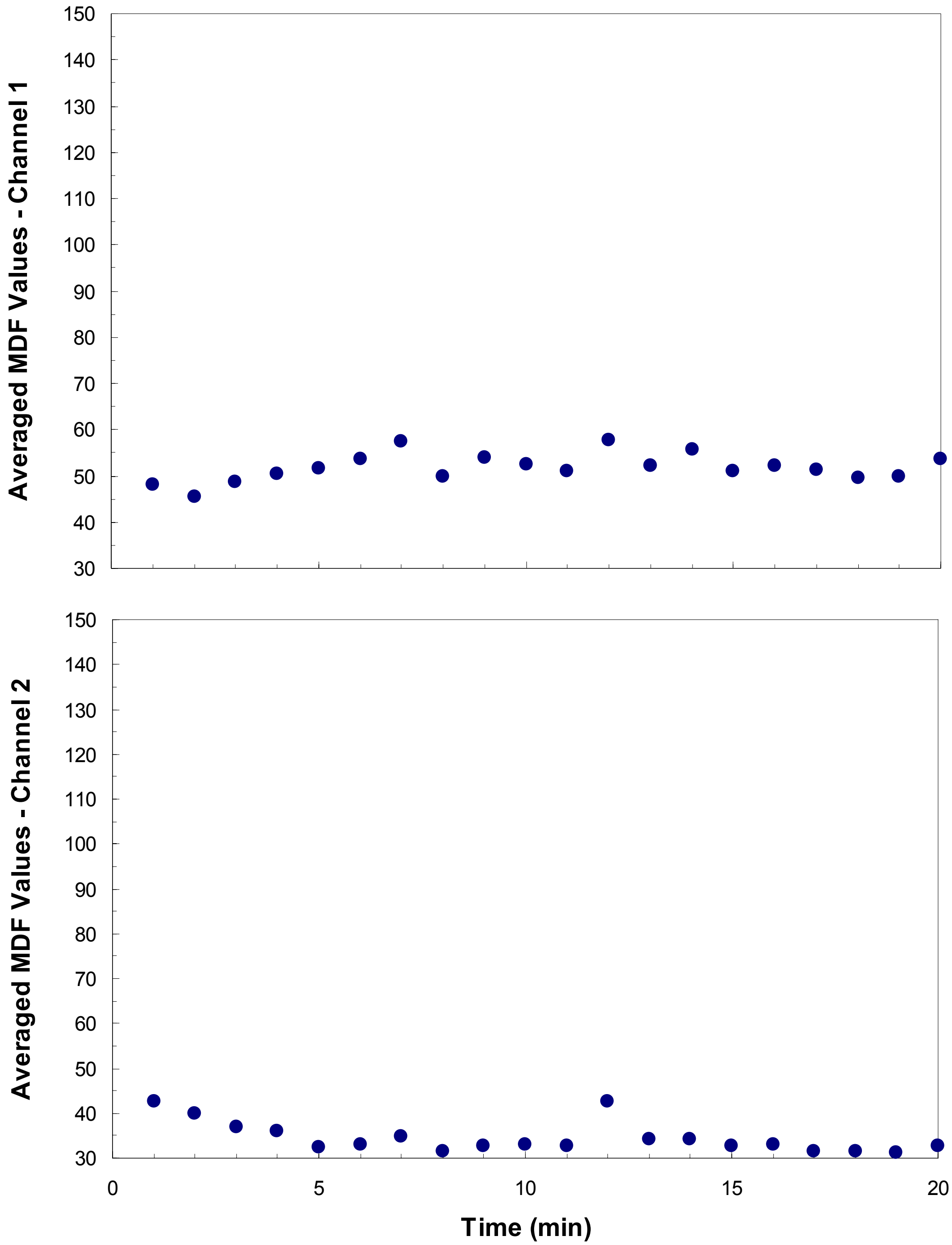

Figure 61. Median Frequency Values for Loading Period - Participant 8

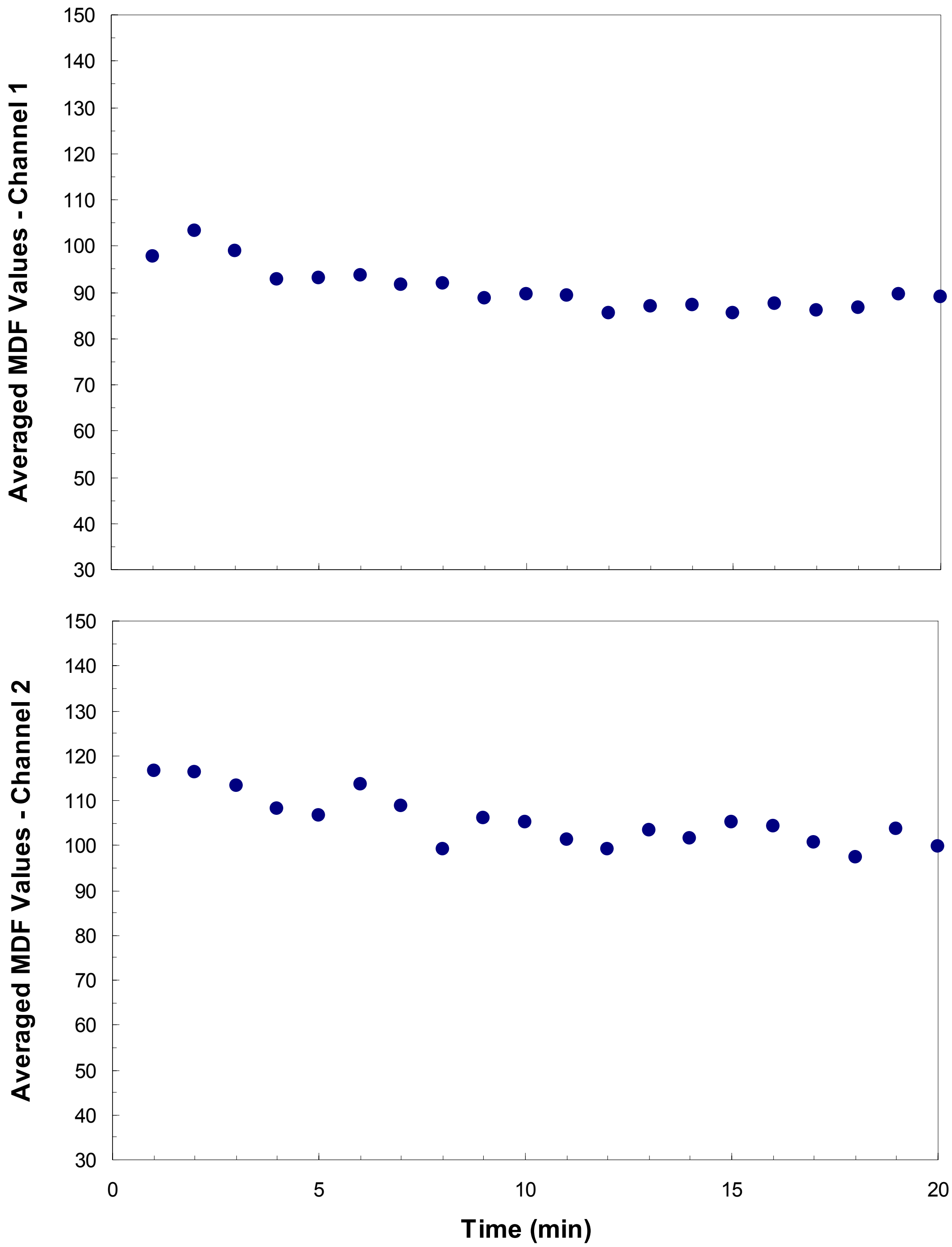

Figure 62. Median Frequency Values for Loading Period - Participant 9

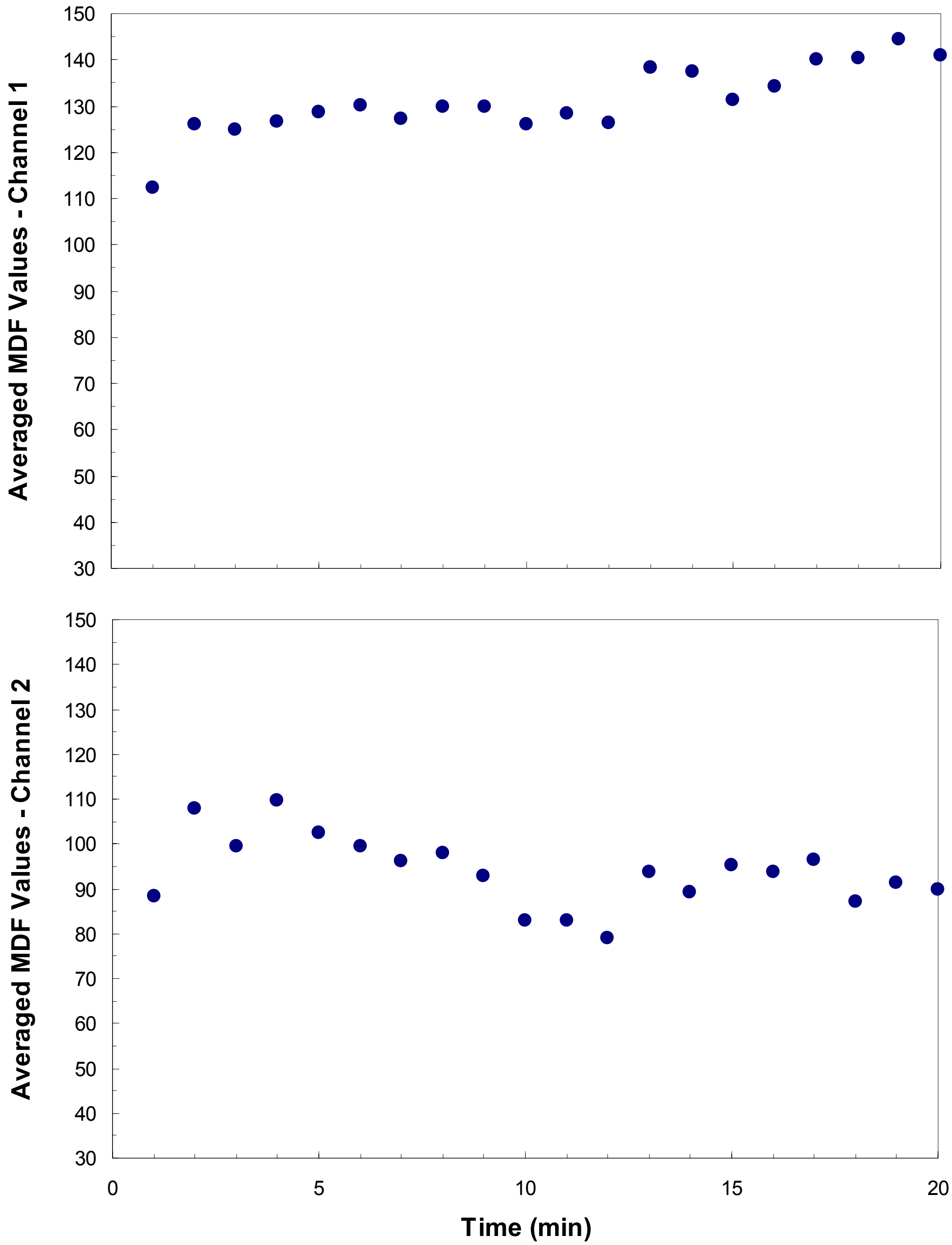

Figure 63. Median Frequency Values for Loading Period - Participant 10

VITA

Guntulu Selen Hatipkarasulu was born in Adana, Turkey, on November 18, 1974. She holds a Bachelor of Science degree in Civil Engineering from Çukurova University in Adana, Turkey (February 1997), and the degree of Master of Science in Industrial Engineering from Louisiana State University in Baton Rouge, Louisiana (May 2002). She is currently a candidate for the Doctor of Philosophy in engineering science degree at Louisiana State University, Baton Rouge, Louisiana. She is expected to graduate in May 2006.

Before she began her studies at Louisiana State University, she worked as a design and technical engineer for a highway construction project in Turkey for three years. Her professional interest lies in the areas of ergonomic, work place design, and occupational biomechanics. She is a member of the Union of Chambers of Engineers and Architects of Turkey. She is also a member of the National Society of Collegiate Scholars.