PRINCIPAL COMPONENT ANALYSIS DEMONSTRATES TRUNK MUSCLE PATTERN VARIATION WITH FALL DIRECTION

by

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Abstract

This study examined the role of the direction of a fall on neuromuscular response. Electromyography sensors were positioned on the erector spinae of twenty subjects. Falls, simulating slips, occurred in the anterior, posterior, and medial-lateral directions. The average activation curves for the four different fall directions displayed different characteristics. Observations were supported by principal component analysis (PCA). PCA coefficients related all fall directions to a single reflex-like response. Analysis of variance on the coefficients demonstrated that anterior falls had a significantly ($p<0.05$) stronger reflex-like response than posterior falls. There was also a significant ($p<0.05$) difference for the interaction between side of the erector spinae and fall direction for the medial-lateral fall directions. Here the contra-lateral muscle had a stronger reflex-like response than the ipsi-lateral muscle. Possible reasons for the observations could be to increase time available for active responses or decrease the energy transferred if the head impacted the ground.

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Chapter 1 Introduction

1.1 Significance

Falls are a significant health problem, accounting for more than 14,000 deaths and 22 million medical visits in 1996 [1]. Injuries commonly seen as a result of falls include hip and vertebral fractures. Although two-thirds of vertebral fractures remain silent, or lack symptoms, one-third are symptomatic [2]. The consequences of a vertebral fracture can include back pain, disability, loss of independence, and increased morbidity and mortality. In 1995, the direct cost of vertebral fractures was 746 million dollars [3], where the majority of the cost was associated with inpatient medical care or nursing homes. There was also an even greater burden due to the decrease in quality of life. About 10% of postmenopausal women have vertebral deformities that cause chronic pain [4].

To understand the etiology of spinal fractures, research has been performed to identify the amount of force required to fracture the spine. Studies have shown that a force as low as 500N can lead to a vertebral fracture and the average compressive failure force across several studies range from 8000N to 2000N for cadaver spines at 25 and 75 years of age [5]. Model predictions of peak impact forces for falls from standing height averaged 5600 N, well above the average failure force, even with dissipation through the soft tissues [6-8]. Although the model predictions estimate forces sufficient to lead to vertebral fracture, only 5% of falls result in fractures [9].

1.2 Background

Significance of Injuries. Falls are the largest single cause of accidental death in the elderly and account for over half of all these deaths [10]. Of all the injuries resulting from falls, 50% result from slips [11]. Some of the highest cost injuries, from a study on slips and trips, were musculoskeletal injuries sustained from a non-fall slips and trips [12]. Risk of injuries resulting from slipping and tripping is increased with age [13]. Understanding the contributing factors leading to an injury can help identify directions of research to work towards prevention of injury.

1.3 Current Study

While several possible contributors leading to injury from falls have been examined, the trunk muscle response during falls is currently unknown. It has been reported that muscle forces generated to stabilize the spine following a sudden perturbation are often several times larger than the external load and body weight combined [14], therefore the muscles themselves should be examined during a fall perturbation. This study examined the muscle activation during a fall. Since the fall direction has been shown to be a factor in determining risk of injury from falls [7, 15, 16], the specific aim was to determine if different fall directions illicit a different trunk muscle response.

Falls were created using a sliding platform to mimic slip-type falls. Data were collected using electromagnetic position sensors and eletromyographic sensors. The electromagnetic position sensor data were used in a separate study. The electromyography sensors collected muscle activation during falls in four different directions: anterior, posterior, and medial-lateral fall directions.

The activation of the erector spinae muscle was examined using two different methods. The first method examined the time series pattern created with the average activation for each fall direction. The second method utilized principle component analysis to focus on the reflex response period.

The PCA determined time series patterns, or eigenvectors, that explain the greatest variance within the data. Coefficients relating the eigenvector of greatest explanation of the data were determine for each of the four fall directions. An analysis of variance (ANOVA) with two within subjects' factors was used to determine if there was a significant difference in muscle activation between the different fall directions.

The results of the ANOVA can demonstrate if any of the four directions have a greater degree of erector spinae muscle activation and the pattern of that activation. This information could be used in future fall studies that are interested in examining the muscle response during falls.

Thesis Content. This work contains six chapters. Chapter 1 consists of an introduction of the area of study. Chapter 2 consists of an extensive review of relevant published literature. Chapter 3 consists of an in depth reporting of the methods involved for this study. Chapter 4 consists of the results of this study. Chapter 5 consists of a discussion of the results. Chapter 6 consists of a summary of the study. An appendix will be included which will contain information on the experimental design and programs used to analyze the data.

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Chapter 2 Background

2.1 Significance of Falls

Falls are a significant health problem, accounting for more than 14,000 deaths and 22 million medical visits in 1996 [1]. Acute medical care costs have been estimated at $17,483 with a standard deviation of $22,426 for a single fall event [2]. In addition to the direct cost associated with the fall, individuals may experience a decrease in mobility as a result of the fall. Decreases in mobility can be the result of physical injury as well as a self imposed decrease in physical activity due to an individual's fear that another fall event may occur [3-6]. This can be detrimental to the quality of life of an individual as personal mobility is necessary for functioning in everyday life to complete tasks such as preparing meals, bathing, and dressing [5, 7]. If the loss of mobility is severe enough independence may be completely lost.

Falls are a serious concern for the elderly. Studies have shown that about one third of older adults fall annually [8, 9]. Of those that fall, 50% do so repeatedly [4]. In addition to falls leading to an injury, a fall can affect an older adult's independence as falls are mentioned as a contributing factor for 40% of admissions to nursing homes [3, 9]. This is understandable when considering that nearly half of all injuries sustained by the elderly result from falls [1].

While the majority of falls result in minor or no physical injuries, a small percentage of falls cause severe injury. Studies have shown that only 5% of falls result in fractures or serious soft tissue damage [4-6, 10]. Fractures that are commonly seen as a result of falls include hip and vertebral fractures [11]. Of these types of fractures, hip fractures have been examined extensively due to the serious mobility issues that accompany a hip fracture. Less focus has been given to vertebral fractures resulting from falls.

Vertebral fractures have possibly received less attention due to the difficulty in diagnosing the occurrence of the fracture. However, in the clearly identified vertebral fractures, the mortality rates one year after fracture are similar to that of hip fractures. The survival rate one year following a vertebral fracture was reported as 72% and the survival rate for hip fracture was 78% [12]. The incidence of diagnosed vertebral fractures is less than the incidence of hip fractures however, only a third of vertebral fractures are symptomatic [11, 13]. As many of the vertebral fractures are non-symptomatic, many vertebral fractures may be unreported. The symptoms that are associated with vertebral fractures include back pain, physical impairment, increased risk for fractures, and an increase in mortality [14]. Multiple vertebral fractures can lead to changes in spine shape such as increased kyphosis and spine torsion ("dowager's hump").

Although a vertebral fracture may not be symptomatic, the risk of subsequent fractures of the hip, forearm, and vertebrae was found to be significantly greater than the general population [15]. The increased risk of a further vertebral fracture was four times that of the general population and was two times that of the general population for a hip fracture following a vertebral fracture [16].

Although hip fractures are typically considered the greatest cause of morbidity following a fall, up to 15% of vertebral fractures are associated with falls and account for significant morbidity [11]. The percentage of vertebral fractures associated with falls may be underreported as the subtle vertebral deformities and reductions in height make it difficult to consistently recognize vertebral fractures [17]. Cooper et al. (1992) found that 16% of vertebral fractures were identified during examination of radiographs for other problems [18]. However, almost 50% of acute, symptomatic vertebral fractures in people 60 years and older occurred with a fall [19].

2.2 Previous Approaches to Address Falls

Falls and the resulting injuries from falls are complex events with a multitude of possible factors that contribute to them. As a result, there have been several approaches to examine falls in order to decrease the occurrence and degree of the resulting injuries. One approach has examined risk factors to identify factors that contribute to falls. Other approaches have examined strategies to prevent falls from occurring. Another approach examined the injuries that are received from falls in order to create safety equipment to intervene during a fall to prevent injury.

Identification of Risk Factors

In order to reduce the number of injuries sustained from falls, the risk factors contributing to falls have been studied in an attempt to decrease the total number of falls. Factors that have been identified as being associated with risk of falls include: age, disease, degradation of the body sensory system, muscle strength, flexibility, balance, coordination, reaction time, and gait [20, 21].

Several studies have shown that age is linked to risk of falls. Tinetti et al. (1988) found that fall incident for people over the age of 65 years is 30%, but for people over the age of 80 years it increases to 40% [4, 20, 22].

Several diseases can affect the motor control of the body; a well known example would be Parkinson's disease [9]. Many diseases and disabilities that affect the bones, muscles, and joints, the components to maintain stability, have been shown to contribute to the risk of falling [5]. Decrease of input from the sensory system or the degradation of the body sensory system have also been shown as risk factors and may contribute to the loss of motor control of the body [3]. The decrease in input from the sensory system could include poor vision, poor vestibular sense, or loss of peripheral sensation that ultimately lead to increased difficulty in proprioception. Vision is an important sensory source that provides information to maintain postural balance [23]. Vision also provides information about the position and movements of the body [24]. A decrease in this information increases the difficulty of appropriate control, resulting in an increased risk of falls. Similarly, the vestibular and sensorimotor systems provide information about the body's orientation with respect to the environment around it. When the information from these systems decrease or become less reliable, the control of the body may become ineffective.

If the sensory systems are functioning properly, there could still be difficulties responding appropriately to the information if muscle strength or coordination is lacking [24-26]. The response of the muscles must be able to produce the force necessary and produce that force within the allotted time to prevent the fall from occurring [25, 27]. For these simple reasons, muscle strength and reaction time are also factors that can increase the risk of falls. Although the identification of risk factors can help to identify individuals with a higher risk of falls, the fall is still not prevented.

Prevention of Falls

Fall prevention studies have examined possible ways to prevent falls from occurring. Some of the studies have utilized the identified risk factors in determining different strategies to prevent falls. Knowing that muscle strength, flexibility, and balance are risk factors leading to falls, these factors were examined to identify the extremes at which recovery is no longer possible. Lean and release techniques have examined the degree from horizontal at which recovery is no longer possible. These studies have shown that there are significant differences between young (20-30 years of age) and older age (65+ years of age) groups. Specifically a decreased peak lower extremity joint torques and ranges of motion [28], as well as decreased lean angle at which balance could be recovered prior to a fall [29]. In addition to the lean and release technique, some studies have created a sliding mechanism to simulate slipping falls [30]. Many of these studies examined the reaction forces underneath the step and the reaction times of the step that was required to recover from the disturbance.

In addition to understanding what factors lead to a fall, some preventative exercise programs have been created to transform knowledge of these factors into prevention of falls [5, 31]. Many different exercise programs have been created; however, not all have been found effective in preventing falls [32]. Regimes that have been shown to be effective at preventing falls include tai chi [33], supervised strength and endurance training [34], and home exercise prescribed by a physiotherapist or specially trained nurse [35].

Prevention of Injury from a Fall

As some risk factors are unavoidable, other studies have examined possible solutions to minimize or prevent an injury from occurring when a fall does occur. Studies have examined the usefulness of safety equipment in the prevention of injuries from falls [36-39]. A popular injury prevention device is the hip protector pad designed to prevent hip fracture during impact of a fall. Among these are Robinovitch et al.'s (1995) examination of hip protection pads to reduce impact forces [37]. Unfortunately, the padded protectors have been unable to consistently protect from injuries in a community setting although the results for nursing homes is more positive [38]. Possible reasons reported for the difference include compliance to using the equipment. Individuals residing in nursing homes are encouraged by the nursing staff to comply with the hip protectors whereas individuals in the community rely on self administration of the hip protectors. The hip protectors need to be comfortable and unobtrusive enough to ensure compliance in addition to just decreasing the forces applied to the hip at impact.

Mechanics During a Fall

Due to the complex nature of falls, the mechanics during a fall is possibly the least understood area about falls. There have been previous studies that have focused on the impact phase of falls. They mainly focused on the force that is transferred at impact and where the forces were being applied. It has been shown, for example, that relaxing a body during a fall to the side reduces impact velocity [40].

Protective responses that have been examined include strategies to avoid falls, strategies to prevent impact to the hip, and strategies to absorb energy of the fall to prevent an injury from occurring. Hsiao et al. (1998) found postural stability was dependent on the direction of the perturbation with lateral directional perturbations being less likely to result in a fall [41]. When a lateral fall does occur, it has been shown that individuals actively respond by rotating the trunk, preventing the hip from impacting the ground [41].

A different strategy reported to occur during falls in order to protect from injury is bracing for impact with an arm. Slap falls, or falls in which energy is absorbed by the arms slapping the ground, have shown to have the lowest peak impact force on the hip and shoulder [1]. However, the inclusion of the bracing with the arm can lead to fractures within the wrist. Studies have examined the effect of direction and have determined that anterior falls had a significantly lower impact velocity on the distal radius than posterior falls [42].

Although several studies have examined the mechanics during a fall, the focus has usually been on the impact. The impact is believed to be the time at which the injury is sustained, but this does not explain how a significant number of musculoskeletal injuries resulted from non-fall slips and trips [43]. During the near slips and trips there was no impact phase, yet the injury was still sustained. It is possible that the neuromuscular system overreacts to the sudden perturbation, causing overload to the musculoskeletal system. The subsequent forces generated from the muscles could be much greater than the static condition where the force would be several times that of the external load and body weight combined [44]. Without information contradicting this theory, it is important to examine neuromuscular response during the fall phase.

2.3 Current study

The current study was designed to build upon the previous approaches of examining the role of the mechanics during the fall to examine neuromuscular response. As previous studies have shown fall direction can have a significant effect on the outcome, the role of direction of the fall on neuromuscular response was examined. Specifically, the initial reflex response of the erector spinae muscle was examined to determine if different fall directions lead to different reflex responses. This study will help identify if a particular fall direction would elicit a greater activation of the erector spinae muscle. Identification of this direction would be beneficial for further examination of trunk muscle reactions during falls. Identification of over-excitation within the erector spinae muscle could help explain how injuries can be received from non-fall slips and trips, one of the more costly injuries in Lipscomb's study of construction injuries from slips and trips [44].

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Chapter 3 Methods

Controlled falls were created to examine the trunk muscle activation responses in different fall directions. The falls were created by simulating a slipping motion. Although falls are commonly associated with gait, a sliding platform was created to generate a fall from a stationary subject. This approach removed influences of gait on the fall direction, allowing the fall direction to be safely controlled. The sliding platform was created so the platform could be moved out from underneath the stationary subject. The platform moved the subject's feet horizontally while inertia maintained the location of the subject's center of mass. The fall was initiated once the center of mass was outside the stable range of support. Subjects were instructed to let themselves fall onto a soft padded mat. Different fall directions were created by changing the direction the subject faced on the sliding platform, prior to platform release.

3.1 Experimental Design

A sliding platform was designed to allow falls to occur in multiple directions. The falls were created by horizontally moving a platform that the subject was standing on. The horizontal acceleration of the platform moved the subject's base of support while inertia delayed the motion of the subject's center of mass. The platform accelerated quickly causing the subject's base of support to move out from underneath the subject's center of mass. The fall was initiated once the center of mass was outside the base of support (Figure 1).

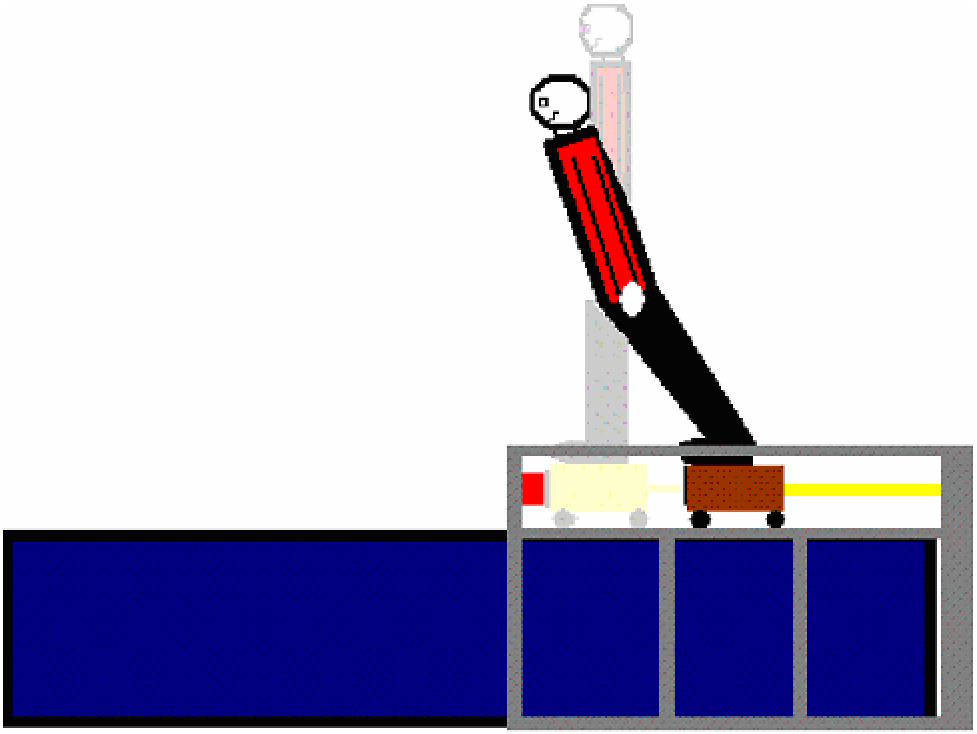

Figure 1. The sliding platform is raised above the soft padded mat used as a landing surface. The power mechanism is provided by four pre-loaded bungee cords (~30lbf) stretched from a base support to the sliding platform as it was locked in the beginning position by electromagnets. Once the sliding platform was released, the subject's feet were slid out from underneath them.

The sliding platform was constructed from wood to minimize possible distortion of the electromagnetic sensors used to measure motion. The sliding platform was constructed of a 2”x6” lumber frame. Particleboard provided the top cover for the frame, allowing a 16”x72” surface for the subject to stand on. To prevent injury in the event the subject landed on the platform, a 3/8” dense foam pad covered the top and front of the platform. Hard acrylic wheels were placed on the lower portion of the sliding platform to allow horizontal motion along a single axis. The 16” width of the platform aligned with the direction of platform motion. This width was selected based on two considerations. The first consideration was to provide an area large enough for the subject to feel comfortable while getting in position. The second consideration was to minimize the weight of the sliding platform and therefore the force required to accelerate the platform during the experiment. The 72” length was determined by the length required to span across the 5’x10’ large padded mat (Tiffin Athletic Mats, Elkton, MD – portable Gymnastics pit) that provided a safe landing area for the subject.

In order for the sliding platform to slide horizontally over the 2’ tall padded mat, a support base was constructed to elevate the sliding platform. The support base was also constructed of 2”x4” wooden framing to minimize effects on the electromagnetic sensors. The support base was created from three frame walls circling the mat. A cross beam under the padded mat substituted for structural support of the fourth wall that could not be placed where the subject would be falling. The 6’ long wall and cross beam support provided stability for the two support rails created to raise the sliding platform above the padded mat.

The support rails were constructed to minimize the elevation of the sliding platform over the mat. The support rails raised the surface of the platform 5.5” above the mat. Although additional elevation over the mat could change the fall characteristics reducing the similarities to a slip type fall, this height was selected to maintain platform rigidity, while minimizing the clearance above the mat.

In addition to providing elevation for the sliding platform, the rails provided a channel to control the platform’s range of motion. The support rails ran alongside the padded mat in the direction of the axis of motion for the sliding platform. The support rails contained a channel for the wheels of the sliding platform to prevent the platform from falling off the rails. The bottom of this channel was lined with a hard smooth acrylic surface to decrease the resistance to the motion of the platform. The support rails included a beam running above the platform in the direction of the platform motion. This beam was to prevent the platform from rotating and lifting wheels off of the support rails.

The support rails also contained a locking mechanism preventing movement of the sliding platform. The locking mechanism consisted of two 12V round electromagnets each rated for 180lbs maximum pull. Each of the two support rails contained one electromagnet, allowing both sides of the six foot long platform to be released simultaneously. The electromagnets were powered by a 12 volt power supply, creating a magnetic attraction to the small steel plates attached to the face of the sliding platform. Removing the power to the electromagnets released the locking mechanism allowing the platform motion to occur.

The motion of the platform was created using four preloaded bungee cords providing approximately 30 pounds of force and accelerating the platform at $6.6\frac{m}{s^2}$ for a 170lb subject. The bungee cords were stretched from the base support to the sliding platform locked in the initial position. The release of the locking mechanism initiated the platform motion, sliding the platform horizontally towards the back wall of the base support.

3.2 Protocol

20 subjects (10 male and 10 female) between 19 and 34 yrs old, with an average height and weight of 169.9 (SD 9.1) centimeters and 69.8 (SD 12.6) kilograms, were tested with the approval of the Human Subjects Committee from the University of Kansas, Lawrence KS (Appendix A). Motion data were collected at 100Hz from four electromagnetic motion sensors (Ascension Technology, VT) using the Motion Monitor software (Innsport, IL). The electromagnetic sensors were positioned around the pelvis at the sacrum level. The electromagnetic sensors were collected for a separate study and were not analyzed for this study. In conjunction with this system is an A/D board collecting analog data at 1500Hz. This allowed the synchronization of eight single differential surface electromyography (EMG) electrodes (Delsys, Boston, MA). The EMG sensors were attached to the skin over both sides of the trunk muscle groups: erector spinae (ES), rectus abdominus (RA), internal oblique (IO), and external oblique (EO).

The initial step was to determine the maximum voluntary EMG activation for the purpose of normalizing the EMG (Appendix B). This was accomplished by having the subject's hips and legs strapped to a bench to prevent motion of the lower body. The subject was instructed to complete a series of three maximal flexion, extension, and rotation contractions while their shoulders were held stationary. Different motions were completed to focus on specific muscle groups while performing the isometric activation. To determine the maximum EMG activation for the ES muscle group, the subject laid face down on a bench. They were then instructed to raise their chest off of the bench to complete a back extension while an investigator held their shoulders in place. Similarly, to determine the maximum EMG activation for the RA muscle group, the subject laid face up on the bench. When instructed, the subject raised their back off of the bench to perform a sit-up while an investigator held their shoulders in place. In order to determine the maximum EMG activation for the IO and EO, a slightly different motion was utilized. While an investigator held the subject's shoulders in place, the subject was instructed to raise one shoulder only in order to create a twisting motion. By raising the right shoulder off of the bench when laying face up, the maximum EMG activation was determined for the left side of the IO and the right side of the EO. Similarly, by raising the left shoulder off of the bench while laying face up, the maximum EMG activation was determined for the right side of the IO and the left side of the EO.

Following the maximum EMG activation trials, the subject completed four trials consisting of falls in each of eight different directions. The orders of these falls were randomized. The directions were in every 45˚ orientation including anterior, left lateral, posterior, and right lateral directions. Due to the sliding platform moving in a single direction, the fall direction was controlled by the initial positioning of the subject. Multiple fall directions were created by changing the direction the subject faced, relative to the platform sliding motion (Figure 2). During the data collection all directional falls were completed before the next trial began. The order for the fall directions within each individual subject was held constant during all four trial blocks.

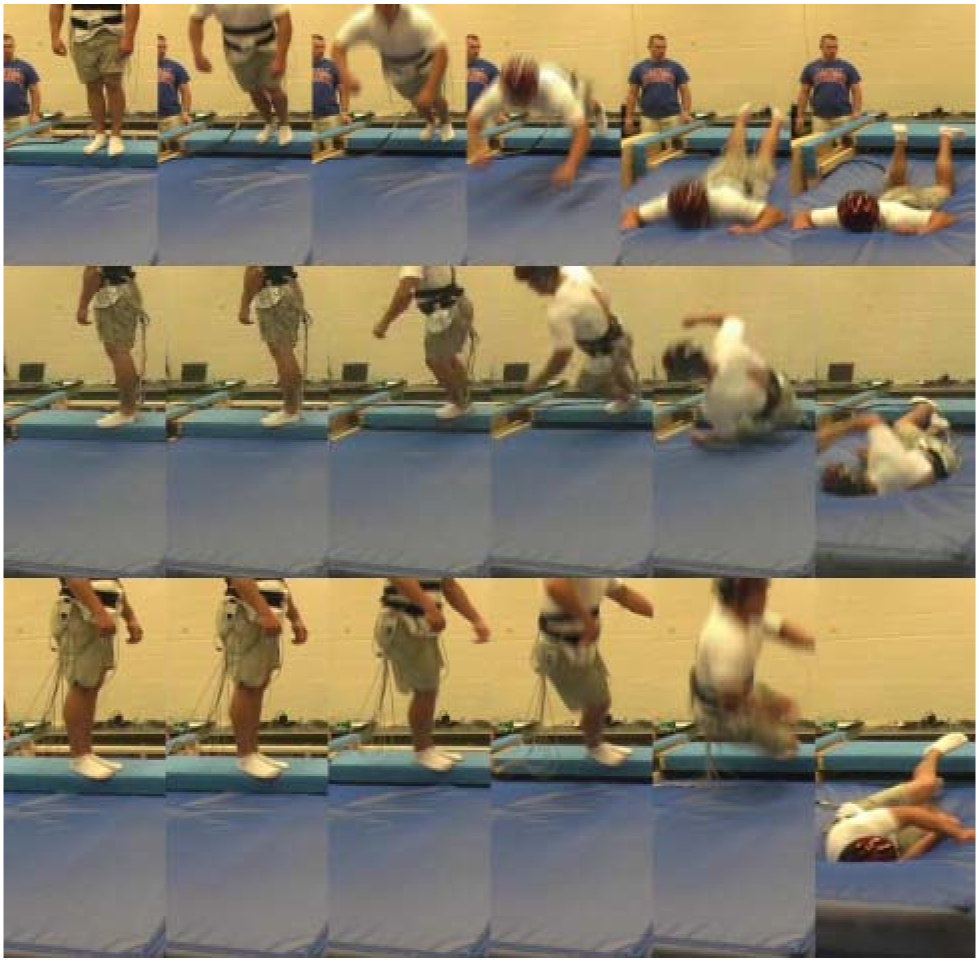

Figure 2. Rotating the initial direction of the subject prior to platform motion controlled fall direction during the experiment. Rotating every 45˚ starting with the forward directional fall resulted in 8 different fall directions. Three examples are shown above with forward (top), left-lateral (middle), and right-lateral (bottom) falls.

Due to the design of the sliding platform, the subject knew the direction of the fall prior to the fall occurring. Although the subjects were instructed to remain relaxed and allow the fall to occur, the subjects experienced different levels of anxiety in the various fall directions. General conversation during the experiment commonly resulted in the subject mentioning their fear of the falls. This fear decreased for most of the subjects after they experienced a few falls. Additionally, many subjects mentioned that their level of fear was dependent on the fall direction. Although these comments were not recorded, anterior and posterior directional falls were frequently mentioned as the scariest fall direction. Some believed that not being able to see where they were falling was the worst whereas others preferred to not see where they were falling.

3.3 Analysis

Raw EMG data were filtered to remove noise. The signals were band-pass filtered between 20 Hz and 250 Hz, a range commonly used with EMG [1-3]. In addition, several notch filters were necessary to remove distinct noise spikes occurring at 60 Hz, 100 Hz, 120 Hz, 180 Hz, 200 Hz, and 240 Hz. The EMG data were then rectified and integrated using a 100-point Hanning window. Finally the EMG data from the data trials were normalized using the integrated, maximum EMG activation data collected at the beginning of the experiment.

The data were examined to identify the general patterns using two different methods. The first method examined the average activation to assess the general patterns of activation for each fall direction. The second method utilized principal component analysis (PCA) for comparison to the general patterns for each fall direction observed by the average activation method. In addition to the general pattern comparison with the average activation method, the PCA calculated coefficients relating the degree of agreement for each trial to the PCA mode curves. These coefficients were then used to perform an ANOVA statistical analysis with two within subjects factors to determine the strength of any statistical significance between trunk muscle reactions in different fall directions.

3.3.1 Data Exclusion

Due to the motion involved in the data collection process, there were trials in which the sensors were loosened or even disconnected from the subject during the trial. These sensors continued collecting data and if included in the analysis would only add noise. Data were initially examined to assess the quality of the EMG signal. In order to reduce the noise added to the analysis, exclusion criteria were determined. Three exclusion criteria were determined: 1) the power spectrum (following the filtering) was dominated by spikes and did not demonstrate an EMG power profile suggesting loss of the muscular activity signal, 2) the maximum normalized, integrated EMG signal exceeded the average of the maximum activations plus two standard deviations, and 3) the initial normalized, integrated EMG signal exceeded the average of the initial activations plus two standard deviations.

The first criterion for exclusion was based on the power spectrum. The power spectrum of each trial was examined to determine the distribution of frequencies present. Trials were removed if there was a lack of distribution of frequencies commonly associated with muscle activation, namely a broad frequency spectrum between 20 and 250HZ [1-3]. Trials with the power spectrum dominated by noise, as identified by distinct spikes at specific frequencies and a lack of the broad frequency spectrum, were removed (Figure 3). This criterion resulted in the removal of 173 of the 640 trails.

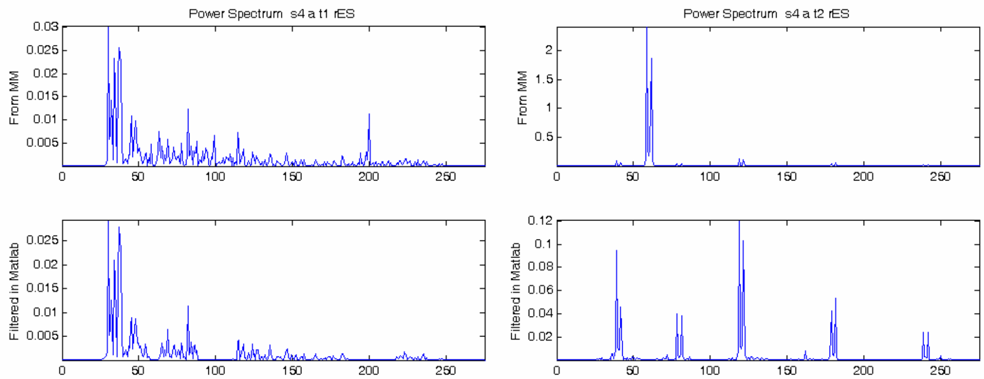

Figure 3. Power spectrum plots were examined to assure a distribution of frequencies commonly seen with muscle activation (left). Trials exhibiting strong distinct spikes were removed (right).

The second exclusion criterion was based on the maximum level of the normalized, integrated EMG activation during 600ms following the release of the platform. Although it was expected that the EMG could exceed 100% of the maximum voluntary contraction, there needed to be a cap to limit the values included in the analysis. The averages and standard deviations were calculated for the maximum EMG activation for each fall direction. Trials included in this calculation were all trials remaining following the first exclusion criteria. All trials exceeding the average of the maximum activation plus two times the standard deviation, in the respective fall direction, were removed. This criterion resulted in the removal of 29 trials. This was done because these relatively high maximum activations reached during the trial likely represent trials in which the sensor was partially dislodged resulting in increased electrical noise.

The third exclusion criterion was based on the level of EMG activation at the time of platform release. The averages and standard deviations were calculated for the normalized, integrated EMG activation at the time the platform was released for each fall direction. Again the average of the initial activation plus two times the standard deviation was the limit for trials to be included in the subsequent analysis. This criterion resulted in the removal of 26 trials. This was done because these high initial activations at the beginning of the trials likely represent a subject anticipating the platform release. This was tested by examining the normalized EMG activations for 200ms prior to the platforms release for the trials removed by this criterion; an example is given in the appendix (Appendix C). Inclusion of these trials would add difficulty in the understanding of the muscle response due to excessive premature activation.

3.3.2 Average Erector Spinae Activation

Once the data for analysis was determined, the general patterns within the data were identified. This was first accomplished by examining the average vectors of the normalized, integrated EMG activations. The individual trials were aligned with the release of the sliding platform as the initial time. This was marked by a rising edge in the circuit powering the electromagnets. The lengths of the trials were reduced to 600ms following the platform release. This length of time was chosen as it was shorter than the length of the shortest fall trial.

Once the EMG data were aligned, all trials for each subject for a specific fall direction were averaged into a single time vector. In this manner an average activation was determined for each muscle group (rES, lES, etc.), for each subject, in each fall direction. The subject averages for a particular fall direction were then combined into a group average activation. The subject averages were determined first to provide equal representation between all subjects. The average activations in the various fall directions were then compared to identify any difference in timing and magnitude of the initial peak activation.

3.3.3 Principal Component Analysis

In order to further describe the general patterns observed in the average activation method, principal component analysis was performed. Principle component analysis (PCA) is a method to identify complex patterns within data by expressing the data such that differences and similarities are highlighted [4]. The technique utilizes linear mathematical manipulations of the data to transform the data set to a “new” data set of fully orthogonal space of the original dimensionality [5]. Once the new data set is created, it is possible to rank the orthogonal space dimensions in relative order of variance explained. With the dimensions ranked, it is now possible to reduce dimensions of the data set to a more manageable data set while maintaining a high level of explanation of within data variance. This reduction allows a focus of the analysis to patterns having the greatest effect on the data, effectively narrowing the focus to the areas of greatest relevance. Once the data is reduced to the desired dimension, the data set is translated back to the original data space for subsequent analysis.

3.3.3.1 Mathematical Process

There are several steps involved in the performance of PCA. The first step is to organize the $m \times n$ input matrix ($\overline{D}$) for the analysis. We are interested in the effects of direction on the timing and magnitude of the initial response activations. Therefore our input matrix will be created using the subject average activations as time vectors in each column (Figure 4). In this manner, the pattern analysis will provide a time series as the result. The columns will be identified by our independent variables: the fall direction, side of ES, and the subject number.

Time (ms)	Subject 1	Subject 2
6	13.8	15.8
12	13.3	16.0
18	12.8	15.7
24	12.4	15.6
30	12.1	15.9
36	12.1	16.7
42	12.6	17.8
48	13.4	18.5
54	14.3	19.4
60	15.3	20.2
66	16.2	20.8
72	17.2	21.0
78	18.1	21.0
84	19.0	20.9
90	19.9	20.5
96	21.0	20.4
102	22.4	20.7
108	23.8	22.3
114	24.8	25.7
120	25.3	30.5

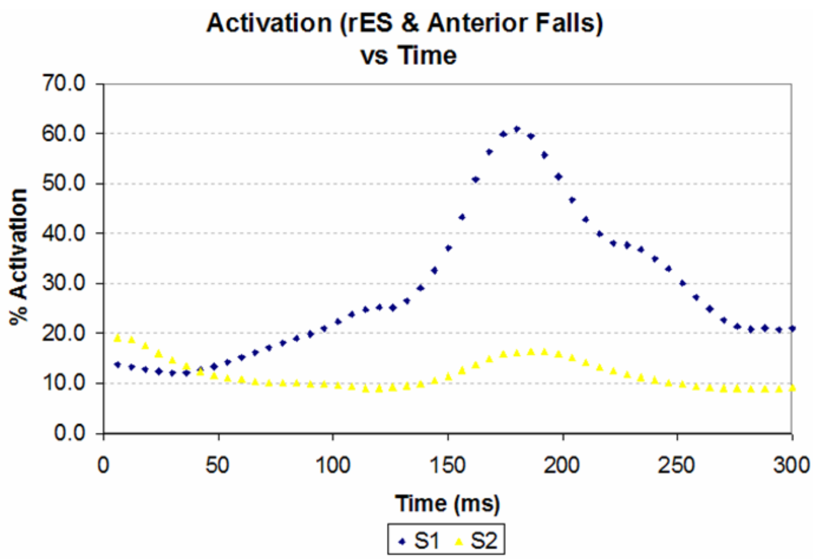

Figure 4. Example of the PCA input matrix made of the average activations of the first two subjects in the anterior fall direction for the right side erector spinae.

The second step is to subtract the column mean of the data such that the mean is zero. This aids in the impending calculations involved with PCA. Then the third step, calculating the covariance matrix C, can be completed. The covariance matrix provides a measure of the strength of the correlation between two random variates.

This is accomplished by $\operatorname{cov}(X,Y) = \dfrac{\sum_{i=1}^{n}(X_i - \overline{X})(Y_i - \overline{Y})}{(n-1)}$

where i is the row number, X is the first variate vector, and Y is the second variate vector. Recall that the data were previously demeaned , i.e. $\overline{X}$ and $\overline{Y}$ are equal to zero, reducing the above equation to

$$\operatorname{cov}(X,Y) = \frac{\sum_{i=1}^{n}(X_i - 0)(Y_i - 0)}{(n-1)} = \frac{\sum_{i=1}^{n}(X_i)(Y_i)}{(n-1)}$$

In this manner the covariate between each of the EMG signals included in the input matrix are calculated. Combining all the covariates produces the square covalence matrix C^{nxn} such that

$$C^{nxn} = \begin{pmatrix} \operatorname{cov}(x_1,x_1) & \operatorname{cov}(x_1,x_2) & \cdots & \operatorname{cov}(x_1,x_n) \\ \operatorname{cov}(x_2,x_1) & \operatorname{cov}(x_2,x_2) & \cdots & \operatorname{cov}(x_2,x_n) \\ \vdots & \vdots & \ddots & \vdots \\ \operatorname{cov}(x_n,x_1) & \operatorname{cov}(x_n,x_2) & \cdots & \operatorname{cov}(x_n,x_n) \end{pmatrix}$$

where x_i is the ith column relating to a specific data trial.

Now that the data matrix has been related to an $n \times n$ square matrix, the eigenvectors ($\vec{\zeta}$) and corresponding eigenvalues (λ) can be determined. The eigenvectors represent the data transformed into a new vector space. The eigenvectors, also called the characteristic vectors, are determined such that $C\vec{\zeta} = \lambda\vec{\zeta}$. In this manner n linearly independent eigenvectors, with unit length, are determined. These eigenvectors can then be ranked by ordering the corresponding eigenvalues (Figure 5). Eigenvalues are ordered from highest to lowest, ranking the corresponding eigenvectors from the greatest to least agreement with the data. This allows the analysis to identify the eigenvectors that best describe the data. The eigenvectors relative amount of agreement with the data, or percent contribution, can be determined by dividing the corresponding eigenvalues by the sum of all eigenvalues.

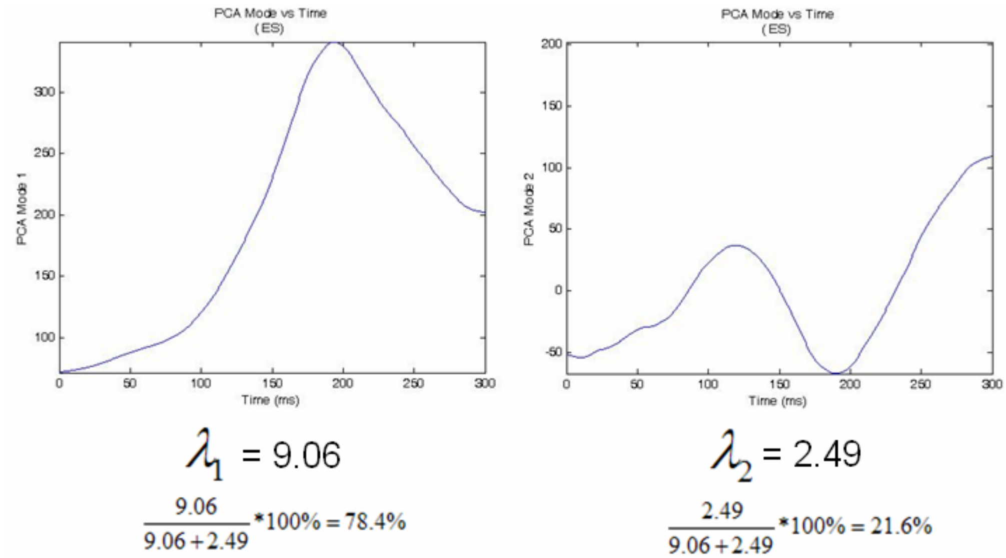

Figure 5. The eigenvectors of the PCA matrix are plotted to give the general curve characteristics. Each eigenvector corresponds to eigenvalues which are used to determine the relative percent variance explained by the eigenvectors.

$$Contribution(\%) = \frac{\lambda}{\sum_{i=1}^{n} \lambda_i} * 100\%$$

Calculating the relative contribution allows the focus of the analysis to be narrowed in on the general patterns that have the greatest amount of agreement with the data.

3.3.3.2 Analysis of PCA Results

Once the eigenvectors have been ranked in order of contribution towards explaining the data, the individual eigenvectors can be examined in further detail. The further examination was done in two different ways. The first way was to examine the shape of the eigenvectors. Based on the organization of the input matrix the shape represents a time series. Therefore, the shapes of the eigenvectors describe the timing and magnitude of peaks. These eigenvectors are then compared to the average activations to see if a common pattern is observed between these two methods.

The eigenvectors are common between all the input matrix columns. However, it is possible to identify differences in fall direction by converting the eigenvectors back to the original vector space. Coefficients (a_{ij}) are determined that describe the strength of agreement between the eigenvectors and the original column vector. The coefficients were determined by

$$\overline{D_i} = a_{i1}\lambda_1\overline{\zeta}_1 + a_{i2}\lambda_2\overline{\zeta}_2 + \ldots + a_{in}\lambda_n\overline{\zeta}_n = \sum_{j=1}^{n} a_{ij}\lambda_j\overline{\zeta}_j$$

where $\overrightarrow{D_i}$ is the original data column vector for the ith column, $\overline{\zeta}$ is the eigenvector, and n is the total number of eigenvectors. In this manner of calculation, all directions will share common eigenvectors, but differences in fall direction are determined by the strength of association with each eigenvector. Based on the PCA input matrix organization there is a single coefficient for every subject, in every fall direction, and for every side of the erector spinae (excluding removed data) that corresponds with each eigenvector.

3.3.3.3 Statistical Analysis

After identifying an eigenvector of interest, statistical tests were performed on the associated coefficients. The coefficients were analyzed using an ANOVA for a two-factor experiment with repeated measures on both. The main effects were the side of the ES and the fall direction. Due to exclusion of trials there were missing coefficients. In order to minimize the loss of entire subjects, missing coefficients were replaced for the ANOVA if no more than one directional coefficient is missing for the individual side of the ES. The method to estimate the missing data were an iterative averaging technique described in Research Design and Statistical Analysis 2nd Ed by Myers [6],

$$\hat{X}_{ij} = \frac{nT_i + bT_j - T}{(b-1)(n-1)}$$

where b is the number of conditions, T_i is the sum of the data points for i^{th} subject, T_j is the sum of points from all subjects, and T the sum of all points. The iterations are shown in the appendix (Appendix D). Significance was determined at an $\alpha = 0.05$ level.

3.4 References

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5. Perez, M.A. and M.A. Nussbaum, *Principal components analysis as an evaluation and classification tool for lower torso sEMG data.* J Biomech, 2003. **36**(8): p. 1225-9.
6. Myers, J.L., Well, Arnold D., *Research Design and Statistical Analysis*. Second ed. 2003: Mahwah, N.J. : Lawrence Erlbaum Associates.

Chapter 4 Results

4.1 Average Activation

Examination of the group's average activation of the erector spinae (ES) demonstrated interesting patterns. The group's average EMG activations for the ES were clearly different in the four different fall directions. This included a difference in average activation between the right and left sides of the ES (rES and lES respectively) on a single fall direction as well as an average activation difference between the same side ES in different fall directions. It was seen that during falls in the anterior direction both the rES and lES had a distinct activation peak occurring near 200ms (Figure 6). Following this activation peak both sides of the ES began to rise in magnitude. The posterior fall direction was different from the anterior falls for both the rES and lES. During posterior falls neither the rES nor the lES had a distinct activation peak but the rES had a relatively high activation throughout the trial.

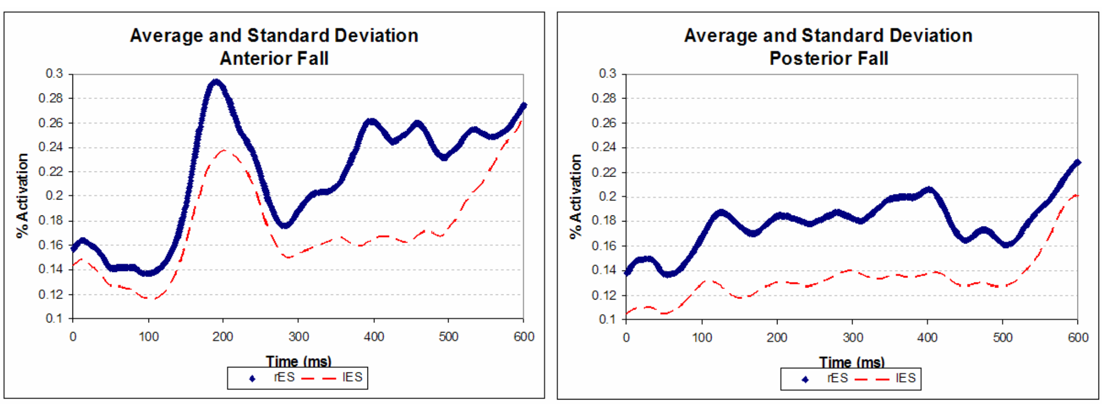


Figure 6. During anterior falls (left), both the right and left side of the erector spinae showed a distinct peak occurring near 200ms. Following they both return to a near baseline value. The right side then immediately begins an increasing activation as impact approaches, whereas the left side remains at baseline until much closer to the impact. During the posterior falls (right) there is no distinct peak, however the activation remains relatively high throughout the time period.

During the lateral falls it was observed that the contra lateral ES muscle group had a distinct 200ms activation peak whereas the ipsi lateral muscle group did not show the same pattern (Figure 7). During left lateral falls, it was seen that the average activation for the rES had a distinct activation peak while the lES did not. During the right lateral falls, the pattern switched such that the distinct activation peak was seen in the average activation of the lES and not seen by the activation of the rES. However, there was an increased activation of the rES beginning at 90ms that plateaus between 150ms and 350ms.

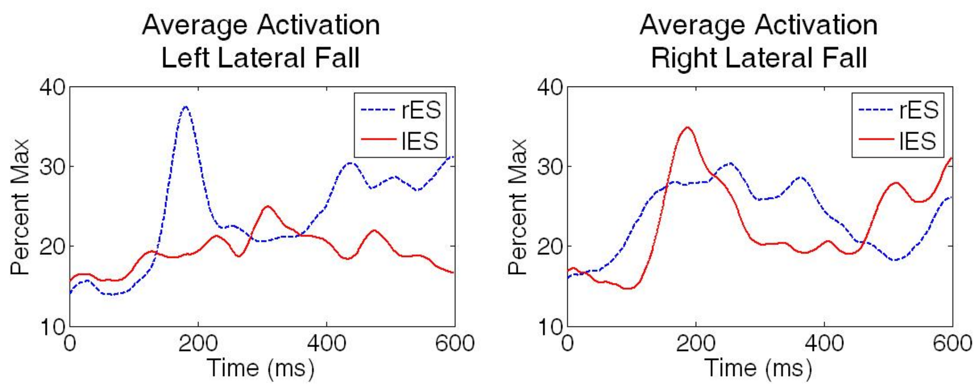


Figure 7. During the lateral falls it was observed that the contra-lateral muscle had a distinct activation occurring near the 200ms and no distinct activation for the ipsi-lateral muscle group. It was also observed that the rES had a higher activation during the left lateral falls when compared to the lES during right lateral falls. As nearly all subjects are believed to be right-handed, this increased activation may be due to a handedness effect.

4.2 Principal Component Analysis

To further examine the distinct 200ms peak and the contribution of this peak with different fall directions, principal component analysis (PCA) was applied. PCA was used to assess if the same directional patterns observed in the average activation would be observed. This analysis provides information in the form of eigenvectors ($\bar{\zeta}$), or PCA mode curves, that describe the general patterns within the data. The

PCA modes shapes that describe the general patterns within the data are common to all the input data. However, a coefficient (a_i) was calculated for each trial included in the input for each of the PCA modes. The coefficients identify the degree which the PCA mode shapes, or general patterns, agree between each directional fall.

The PCA was limited to the first 300ms following the platform release. This was done to minimize the effect of voluntary responses being included in the PCA [1]. Although this is shorter than the 600ms examined during the average activation method, comparisons can still be made to the average activation method over the initial 300ms. This was done to narrow the focus of the PCA on the 200ms peak which occurs when initial reflex responses would be present. The input for the PCA is a matrix of the average activations for each of the subjects, in each of the four fall directions. The column number of the input matrix is listed in the table (Table 1). All trials for a specific subject number, fall direction, and muscle group were averaged in order to have equal representation between different subjects in the analysis.

PCA Mantrix Input Columns

	rES				lES			
	Anterior	**Left Lateral**	**Posterior**	**Right Lateral**	**Anterior**	**Left Lateral**	**Posterior**	**Right Lateral**
s4	1	19	35	52	70	89	106	123
s5	2	20	36	53	71	90	107	124
s6	3	21	37	54	72	91	108	125
s7	4	22	38	55	73	92		126
s8	5	23	39	56	74	93	109	127
s9	6	24	40	57	75	94	110	128
s10	7	25	41	58	76		111	129
s11	8	26	42	59				
s12	9	27	43	60	77	95	112	130
s13	10				78	96	113	131
s14	11	28	44	61	79	97	114	132
s15	12	29	45	62	80	98	115	133
s16	13	30	46	63	81	99	116	134
s17	14	31	47	64	82		117	135
s18	15		48	65	83	100	118	136
s19	16	32	49	66	84	101	119	137
s20	17	33	50	67	85	102	120	138
s21	18	34	51	68	86	103	121	139
s22				69	87	104	122	140
s23					88	105		141

Removed Trials

Table 1. For the PCA analysis the average activation for each of the 20 subjects in each of the four fall directions for both the left and right erector spinae were included. This table shows data columns of the trials included as well as the subjects and directions which were not included due to removed data trials.

4.2.1 PCA Mode Shapes

The principal component analysis determined the eigenvectors, or PCA modes, which describe the general patterns of variance within the data. The relative percent of contribution for each PCA mode, or eigenvector ($\vec{\zeta}$), was determined by the ratio of the corresponding eigenvalue to the sum of all eigenvalues. Using the relative percent contribution the PCA modes ordered from the greatest to least amount of variance explained. The PCA modes with the greatest variance were reduced to those required to explain 95% of the variance (Table 2). This effectively reduced the PCA modes to examine from 141 to 5 PCA modes.

	Variance	Summation of Variance
PCA 1	63.6	63.6
PCA 2	17.5	81.1
PCA 3	9.4	90.5
PCA 4	4.0	94.5
PCA 5	2.5	97.0

Table 2. Table of variance explained by each eigenvector.

The first principal component mode (PCA 1) accounts for 63.6% of the variability within the input data. The general shape of PCA 1 consists of a distinct peak occurring at 200ms (Figure 8). The shape of this mode curve is similar to the shape of the curve seen in the first 300ms of the average activation analysis. The second principal component mode (PCA 2) accounts for 17.5% of the variability in the PCA. PCA 2 has an activation peak occurring slightly earlier at 125ms followed by an increasing slope until the end of the 300ms. In addition, to having a double peak curve, the curve transitions between negative and positive values. The transition from negative to positive values allows the second component to contribute both increases and decreases in the overall general pattern observed. The curve would contribute an increase in the overall general pattern of EMG activation at times when both the coefficient and the curve are the same sign. Conversely, when the signs of the coefficient and the mode curve are opposite the effect of PCA 2 on the overall pattern will be a decrease in the overall EMG activation. The third PCA mode explains 9.5% of the variability. The general shape of this curve is similar to PCA 1 with a single peak occurring at 150ms. The next two PCA modes describe 6.5% of the variability in the PCA. These two PCA modes have no distinct peak, but the shape is better explained as alternating peaks and valleys. These curves alternate above and below zero allowing the modes to contribute in the same manner as PCA 2.

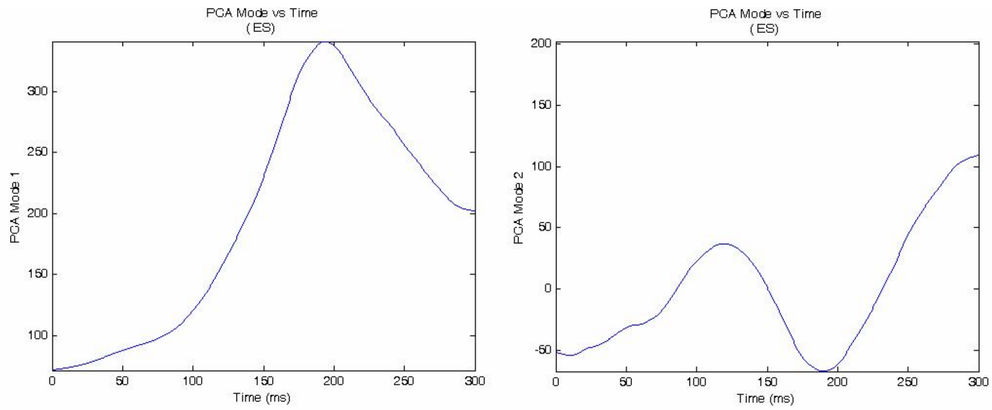


Figure 8. Examination of the PCA mode shapes gives the general patterns of the data. PCA 1 (left) shows a distinct peak occurring at 200ms. This peak represents data with 63.6% of data variability. PCA 2 (right) represents 17.5% accuracy of the data. The mode shape of PCA 2 changes from negative to positive. This transition allows for the addition and subtraction from the overall pattern based on the sign of the coefficient relating to this mode.

4.2.2 PCA Mode Shape 1 – The Activation Peak

As the first principal component mode (PCA 1) reflects the activation peak, additional analysis was performed on the coefficients of this mode. PCA 1 was selected for three reasons: 1) similar shape as the average EMG activation, 2) resembles a reflex-like response, and 3) describes the majority of the PCA variability (64%). A coefficient was determined for each subject in every combination of fall direction and muscle group that corresponds to a column in the PCA input matrix. The coefficients were determined by

$$\overline{D_i} = a_1\lambda_1\overline{\zeta}_1 + a_2\lambda_2\overline{\zeta}_2 + \ldots + a_n\lambda_n\overline{\zeta}_n = \sum_{j=1}^{n} a_j\lambda_j\overline{\zeta}_j$$

where $\overline{D_i}$ is the original data column vector for the ith column, $\vec{\zeta}$ is the eigenvector, and n is the total number of eigenvectors. Therefore a coefficient was determined for each column included in the PCA. Therefore it is possible to compare the coefficients between different fall directions.

The coefficient of the first principal component was found to vary with fall direction (Appendix E). The average coefficients for the rES show that anterior falls have a higher coefficient compared to posterior falls, 0.046 (SD 0.103) to 0.021 (SD 0.051) respectively. The average coefficient weights for the lES show a similar pattern with 0.039 (SD 0.074) for anterior falls compared to 0.007 (SD 0.043) for posterior falls. The coefficients for the contra-lateral muscle was 0.062 (SD 0.077) for the left lateral falls and 0.053 (SD 0.069) for the right lateral falls. The ills-lateral coefficients were less at 0.018 (SD 0.043) for left lateral falls and 0.016 (SD 0.115) for right lateral falls. Although the experiment was not specifically designed to examine differences in gender, a simple plot of the coefficients divided by gender displays that there may be a difference based on gender (Appendix F).

4.3 Analysis of Variance

Although the patterns appear clear from the plots, statistical tests were necessary to prove that there are significant differences in the erector spinae response based on the fall direction. The coefficients for PCA 1 were analyzed using an analysis of variance for a two-factor experiment, repeated measures on both side of the erector spinae and the fall direction, repeated measures on both. In order to maximize the number of subjects included in the analysis, missing values were estimated. The estimation method was an iterative averaging technique described in Research Design and Statistical Analysis 2^{nd} by Myers.

The ANOVA results demonstrated the overall effect of fall direction to be significant ($p<0.05$). However, the side of the erector spinae and the interaction of the side of the erector spinae and the fall direction were not significant (side $p=0.63$ and direction $p=0.14$). As there were strong patterns observed in the average activation the ANOVA was also used to examine pairs of opposite directions as these should have the greatest difference.

The first pair examined the difference between the anterior and posterior fall directions. It was found that the anterior and posterior fall directions were significantly different ($p<0.05$). As the coefficients are related to the eigenvector with the greatest explanation of variance within the data, it is concluded that this is representative to the data set. Furthermore as the eigenvector is similar to the curve determined by the average analysis, it is believed to also represent a reflex response. Therefore it is believed that the reflex response of the erector spinae is different based on the direction of the fall for falls in the anterior-posterior directions.

The second interaction pair examined the medial-lateral fall directions. The interaction of the side of the erector spinae and the fall direction were significant (side*direction $p < 0.05$). This agrees with the average analysis where the medial-lateral response was described by the contra-lateral and ipsi-lateral muscles. Both sides of the erector spinae respond appropriately to reduce the rate of descent of the upper body. The decrease in descent rate can then be taken advantage of by increasing the time allowed for active response or at a minimum to decrease the energy transferred at impact.

Along with the pair-wise comparisons listed above, there were a few other pairs that reached significance. There was a significant difference ($p < 0.05$) between falls in the posterior and right lateral fall direction for the left side of the erector spinae. This may be a result of fall strategy among other factors. It was also seen that there was a significant difference ($p < 0.05$) between the right and left side of the erector spinae during falls in the left lateral fall directions. It is interesting to note that this was not observed for the right lateral falls ($p = 0.21$). This difference is similar to the results seen by Sung et al. indicating that a difference in response time was observed between the dominant and non-dominant side of the erector spinae [2].

An analysis of variance for a two-factor experiment, repeated measures on both fall direction and side of erector spinae, was run on the PCA 1 coefficients. The results of the ANOVA showed the main effect of direction to be significant ($p \leq 0.05$) with $p = 0.02$ and a sample size of 16 subjects. The side of the erector spinae and the interaction of the side of the erector spinae and fall direction were not significant. Examination of the within subjects comparison showed that the interaction between the side of the erector spinae and fall direction was significant ($p < 0.05$) for the medial-lateral falls.

4.4 References

1. Radebold, A., J. Cholewicki, G.K. Polzhofer, and H.S. Greene, *Impaired postural control of the lumbar spine is associated with delayed muscle response times in patients with chronic idiopathic low back pain.* Spine, 2001. **26**(7): p. 724-30.
2. Sung, P.S., K.F. Spratt, and D.G. Wilder, *A possible methodological flaw in comparing dominant and nondominant sided lumbar spine muscle responses without simultaneously considering hand dominance.* Spine, 2004. **29**(17): p. 1914-22.

Chapter 5 Discussion

5.1 Data Removal

A large number of trials, 228 of 640 total trials, were removed from the analysis based on the aforementioned criteria (Table 3). This was simply based on the nature of the experiment. There was a great deal of motion during the experiment. The motion caused stretching of the skin in a variety of directions making it difficult for the tape to hold the EMG sensors onto the subject. This was exaggerated occasionally when the wires got snagged on equipment or subjects which added a force pulling the sensor off of the belly of the muscle. Additionally the effort of repeatedly returning to an upright position following each fall led many subjects to sweat which increased the difficulty of the tape to maintain the correct position.

Data Removed for right erector spinae (rES)

Subject	Anterior				Left Lateral				Posterior				Right Lateral			
s4	1	2	3	4	161	162	163	164	321	322	323	324	481	482	483	484
s5	5	6	7	8	165	166	167	168	325	326	327	328	485	486	487	488
s6	9	10	11	12	169	170	171	172	329	330	331	332	489	490	491	492
s7	13	14	15	16	173	174	175	176	333	334	335	336	493	494	495	496
s8	17	18	19	20	177	178	179	180	337	338	339	340	497	498	499	500
s9	21	22	23	24	181	182	183	184	341	342	343	344	501	502	503	504
s10	25	26	27	28	185	186	187	188	345	346	347	348	505	506	507	508
s11	29	30	31	32	189	190	191	192	349	350	351	352	509	510	511	512
s12	33	34	35	36	193	194	195	196	353	354	355	356	513	514	515	516
s13	37	38	39	40	197	198	199	200	357	358	359	360	517	518	519	520
s14	41	42	43	44	201	202	203	204	361	362	363	364	521	522	523	524
s15	45	46	47	48	205	206	207	208	365	366	367	368	525	526	527	528
s16	49	50	51	52	209	210	211	212	369	370	371	372	529	530	531	532
s17	53	54	55	56	213	214	215	216	373	374	375	376	533	534	535	536
s18	57	58	59	60	217	218	219	220	377	378	379	380	537	538	539	540
s19	61	62	63	64	221	222	223	224	381	382	383	384	541	542	543	544
s20	65	66	67	68	225	226	227	228	385	386	387	388	545	546	547	548
s21	69	70	71	72	229	230	231	232	389	390	391	392	549	550	551	552
s22	73	74	75	76	233	234	235	236	393	394	395	396	553	554	555	556
s23	77	78	79	80	237	238	239	240	397	398	399	400	557	558	559	560

	Trials Removed	Trials Removed	Trials Removed	Trials Removed
Power Spectrum	13	25	21	18
Max Activation	4	4	5	4
Initial Activation	5	3	4	4

Data Removed for left erector spinae (lES)

Subject	Anterior				Left Lateral				Posterior				Right Lateral			
s4	641	642	643	644	801	802	803	804	961	962	963	964	1121	1122	1123	1124
s5	645	646	647	648	805	806	807	808	965	966	967	968	1125	1126	1127	1128
s6	649	650	651	652	809	810	811	812	969	970	971	972	1129	1130	1131	1132
s7	653	654	655	656	813	814	815	816	973	974	975	976	1133	1134	1135	1136
s8	657	658	659	660	817	818	819	820	977	978	979	980	1137	1138	1139	1140
s9	661	662	663	664	821	822	823	824	981	982	983	984	1141	1142	1143	1144
s10	665	666	667	668	825	826	827	828	985	986	987	988	1145	1146	1147	1148
s11	669	670	671	672	829	830	831	832	989	990	991	992	1149	1150	1151	1152
s12	673	674	675	676	833	834	835	836	993	994	995	996	1153	1154	1155	1156
s13	677	678	679	680	837	838	839	840	997	998	999	1000	1157	1158	1159	1160
s14	681	682	683	684	841	842	843	844	1001	1002	1003	1004	1161	1162	1163	1164
s15	685	686	687	688	845	846	847	848	1005	1006	1007	1008	1165	1166	1167	1168
s16	689	690	691	692	849	850	851	852	1009	1010	1011	1012	1169	1170	1171	1172
s17	693	694	695	696	853	854	855	856	1013	1014	1015	1016	1173	1174	1175	1176
s18	697	698	699	700	857	858	859	860	1017	1018	1019	1020	1177	1178	1179	1180
s19	701	702	703	704	861	862	863	864	1021	1022	1023	1024	1181	1182	1183	1184
s20	705	706	707	708	865	866	867	868	1025	1026	1027	1028	1185	1186	1187	1188
s21	709	710	711	712	869	870	871	872	1029	1030	1031	1032	1189	1190	1191	1192
s22	713	714	715	716	873	874	875	876	1033	1034	1035	1036	1193	1194	1195	1196
s23	717	718	719	720	877	878	879	880	1037	1038	1039	1040	1197	1198	1199	1200

	Trials Removed	Trials Removed	Trials Removed	Trials Removed
Power Spectrum	19	31	26	20
Max Activation	2	2	5	3
Initial Activation	3	2	3	2

Table 3. Trials removed from the analysis of the erector spinae are shaded. Trials were removed based on three criterions: 1) Strong spike dominating power spectrum or 2) Maximum voluntary contraction more than two standard deviations above the average max in a particular direction, 3) Initial activation more than two standard deviations above the average initial activation in a particular direction. The number in each cell represents the identification number of the data trial. The power spectrum removed 173 trials, the maximum criteria removed 29 trials, and the initial criteria removed 26 trials. The break down for each individual side and direction can be seen in the table.

There were several consequences of removing the data trials. As the data trials were not replaced, there were missing trials from many subjects. This required that an average activation trial be determined such that each subject would be represented equally. Equal representation of subjects is important as a single subject could have drastically different responses from the other subjects and should not be allowed to over-power other subjects by having more trials present in the analysis.

The averaging brings the question of consistency of timing and magnitude within the subjects. This was examined by plotting the averages and standard deviations in the different fall directions (Appendix G).

Additionally, there is a pattern of some subjects having multiple trials removed. This could be due to specific subject conditions: sweating, repeatedly snagging wires on the subject, etc. The result of which is that there are some subjects with no trials in a particular fall direction. Following the first few subjects, this problem was identified and attempted to be prevented by having future subjects' complete more trials. Four trials were selected under the belief that this would provide at least one good trial without fatiguing the subject. The trials were block randomized in hopes that a good trial could be collected in each fall direction early during the data collection to have data in every direction before fatigue could become an issue.

5.2 Average Activation

The first attempt to interpret the general characteristics within the fall data was to examine the group average activation of the normalized EMG signals for each fall direction and side of the erector spinae (SD for each direction in Appendix G). The plots of the group averages demonstrate clear characteristics for comparison of the different fall directions (Figure 9).   The anterior fall direction exhibits a distinct peak occurring near the 200ms time whereas the posterior fall direction does not exhibit a distinct peak. Although previous studies have estimated reflex reactions to occur closer to 100ms [1, 2], this 200ms peak is still believed to be a reflex as this time is relative to the platform release and not the actual time the loss of balance was identified.

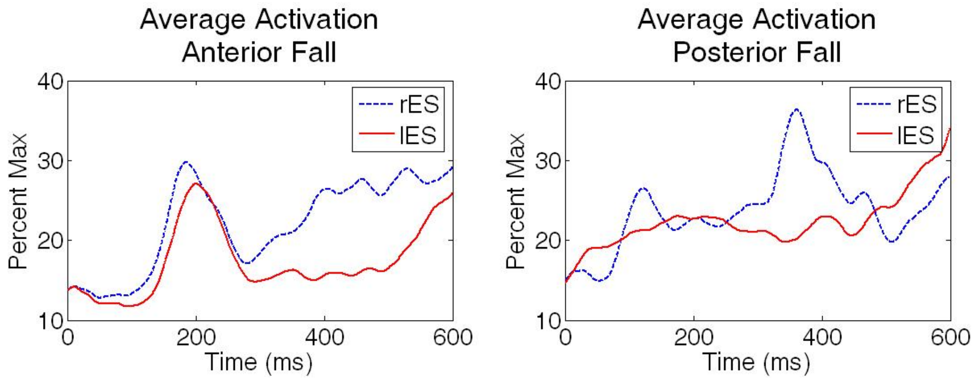


Figure 9. During anterior falls (left), both the right and left side of the erector spinae showed a distinct peak occurring near 200ms. Following they both return to a near baseline value. The right side then immediately begins an increasing activation as impact approaches, whereas the left side remains at baseline until much closer to the impact. During the posterior falls (right) there is no distinct peak, however the activation remains relatively high throughout the time period.

The difference in response between the anterior and posterior falls could be a neuromotor response to minimize or prevent injury. When falling in the posterior direction, a sharp activation of the erector spinae muscle would lead to a shortening of the muscle. As a result of this shortening, the back curvature would increase resulting in accelerating the head towards the ground. This action would increase the probability of receiving a head injury. Similarly, a sharp activation of the erector spinae during an anterior fall would help to decelerate the head, thus increasing the time for active response or at a minimum reducing the energy involved at impact. This reasoning could explain the benefits of having a different erector spinae response based on the fall direction which matches what is observed in the plots. This result is also similar to the results by Sabick et al. that demonstrated that active responses can decrease impact forces [3]. This automatic response, when appropriately applied, in addition to the active responses following could help explain the low percentage of falls (5%) that lead to fractures [4].

A different pattern is observed in the medial-lateral falls. Both the left and right lateral falls show a distinct peak. For the medial-lateral falls, the contra-lateral muscle exhibits a distinct activation peak occurring near the 200ms time whereas the ipsi-lateral muscle does not have a distinct peak (Figure 10). Again this response can be understood to be a neuromotor response to decrease the rate of descent of the head during the fall. Slowing the rate of descent gives the person more time to perform an active response to prevent injury or at the very least decrease the severity of injury that would be received from an impact. It is interesting to note that the right side of the erector spinae muscle has a higher activation than the left side erector spinae when each is the ipsi-lateral muscle. This might be explained by the dominant and non-dominant sides of the erector spinae [5]. It is possible that the right side of the erector spinae exhibits a greater activation than the left side erector spinae when located as the ipsi-lateral muscle during lateral falls due to being the dominant side. However this can not be examined as subjects were not questioned about handedness.

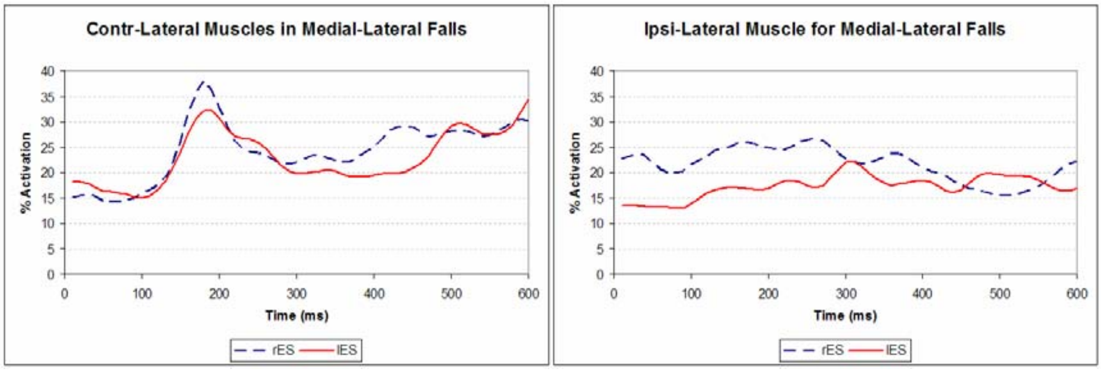
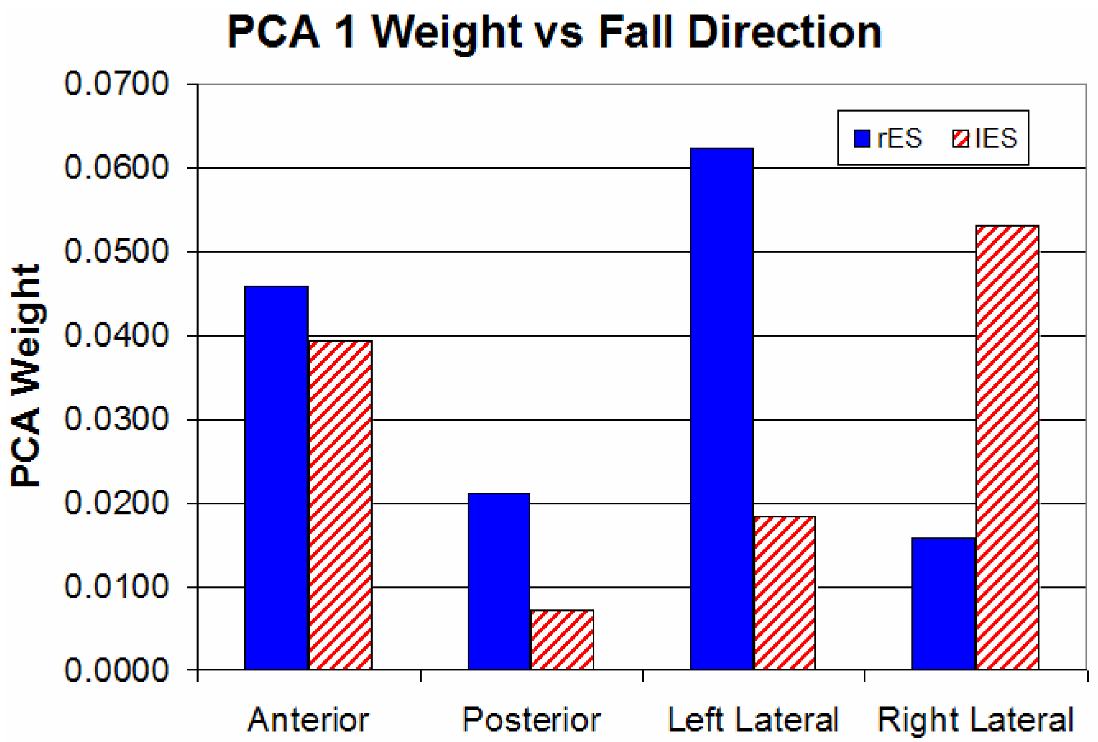


Figure 10. During the lateral falls it was observed that the contra-lateral muscle had a distinct activation occurring near the 200ms and no distinct activation for the ipsi-lateral muscle group. It was also observed that the rES had a higher activation during the left lateral falls when compared to the lES during right lateral falls. As nearly all subjects are believed to be right-handed, this increased activation may be due to a handedness effect.

5.3 Principal Component Analysis

After observing clear general characteristics within the average activations, a more in-depth examination of the data was performed. Using principal component analysis (PCA), the data was transformed into a new vector space representing the data set based on the similarities and differences within the data. The eigenvectors describing the fall data were organized by percent of agreement in explaining the fall data. It was found that the first eigenvector explained 63.6% of the variance and was examined further. This eigenvector was selected for further examination not only because it explained the majority of the variance within the data, but it also resembles the reflex like response that we are interested in.

The eigenvector explaining the greatest amount of variance was plotted versus time to show the same peak occurring at 200ms as observed in the average activation analysis. This is believed to be the reflex response of the erector spinae muscle. This reflex response was then translated back into the original data set to determine the coefficients relating the eigenvectors to the original data. The coefficients were then grouped based on fall direction and the following figure (Figure 11) was generated for the first eigenvector. The magnitude of the response decreases greatly from the anterior to the posterior falls. These patterns follow those observed with the average activation. Strong activation of the erector spinae when falling in the posterior direction would shorten the erector spinae. The shortening would accelerate the descent of the head increasing energy transferred at impact as well as decreasing the time available for active responses. However, strong activation of the erector spinae when falling in the anterior direction would decelerate the velocity of the head as it approaches the ground. This would result in less energy transfer to the head at impact or possibly the avoidance of impact by generating more time for active responses to prevent the head from impacting the ground.



	Anterior		Posterior		Left Lateral		Right Lateral	
rES	0.0458	(0.103)	0.0211	(0.051)	0.0624	(0.077)	0.0158	(0.115)
lES	0.0394	(0.074)	0.0071	(0.043)	0.0183	(0.043)	0.0532	(0.069)

Figure 11. Examining the average weight associated with the principal component shows differences based on fall direction. The anterior falls are higher than posterior falls. It is also seen that during the lateral falls, the contra-lateral muscle has a much higher weight than the ipsi-lateral muscle. Interestingly, the right side erector spinae muscle maintains a higher weight than the left side erector spinae when in the ipsi-lateral position. This may be explained by handedness.

Similar reasoning exists for medial-lateral falls. The contra-lateral side of the erector spinae is located such that shortening this side would decelerate the heads descent. However, the ipsi-lateral side of the erector spinae would accelerate the head towards the ground if it were to have a strong activation response. In this manner, the neuromotor response is based on the fall direction. The different responses are determined such that the head is decelerated as it approaches the ground.

5.4 *Study Limitations*

Although this study shows some interesting patterns it has some limitations. The largest limitation is the inability to have a completely unexpected and unknown fall direction. Due to the design of the fall apparatus, the fall direction was known. Previous work has shown that simulated slips differ from unexpected slips by limiting the direction of the slipping foot to the same direction of walking, which is associated with increased trunk extension and not increased lateral flexion and lateral flexion velocity [6]. By removing the walking element, we have reduced this concern, but it may still be present as the fall direction was known prior to the fall was initiated. It would be interesting to find out if the results would be more or less pronounced with unexpected fall directions. In addition to knowing the fall direction, the subject had an idea of when the platform was to be released. The subject heard the experimenter tell the operator when to release the platform, warning the fall would happen soon. Although there was a variable time delay between the instruction to release and the actual release, there was a click when the switch was turned off, releasing the platform.

Another limitation is the removal of a large amount of data due to questionable signals. The questionable signals were a result of the dynamic nature of the falls. The motion involved during the data collection was problematic for a couple of reasons. The motion of the subject repeatedly standing up after falls increased the body temperature. Many subjects began sweating as the body attempted to regulate the temperature of the body. This sweat however affected the adhesive holding the sensors in place. Additionally, the motion repeatedly stretched the skin. The stretching skin effectively applied a force peeling the sensor from the skin. The less secure sensors resulted in the EMG sensors being dislodged during some of the trials. As a consequence, examination of the data was necessary to remove the questionable trials from the analysis. There were three criterion determined to justify the removal of data trials. The criterion limited the data to trials that were reasonable and were based on: 1) the distribution of frequencies in the power spectrum, 2) the maximum normalized EMG values achieved during the trials, and 3) the initial activation level of the normalized EMG. The removal of trials produced unequal representation by different subjects and required the use of averages to allow evenly weighted subject comparisons. The missing data also created difficulties when performing statistical analysis. The analysis of variance removed all subjects that had any portion of the data missing. This reduced the analysis from 20 subjects to 12 subjects, where 4 subjects were re-included after an averaging technique for missing data was utilized. This was considered acceptable due to the averaging technique decreasing the likelihood of getting a Type I error, also known as a “False Positive”.

Another limitation is the determination of the input matrix to be included in the principal component analysis (PCA). There was a noticeable affect on the results of the PCA by increasing the length of time examined. Although the eigenvector discussed with the 200ms peak was consistently seen independent of the time length analyzed, the relative percent representation decreased when analyzing a longer time length. Additional eigenvectors were not consistently identified between the different time lengths examined. This demonstrates the ability of the PCA to focus on similarities within the data set and adjusting to best describe the entire data set. The shorter time periods reflect the common patterns during the reflex period [7] and the longer times have fewer commonalities as there is more variance between the subjects' active responses to the falls. This highlights the importance of identifying the time period of interest prior to defining the input for the PCA. There was also a noticeable affect on the results of the PCA by adjusting the number of trials included in the input. The inclusion of trials failing to meet the exclusion criterion discussed resulted in different eigenvectors. However, the eigenvector with the 200ms peak was still present due to the large number of trials demonstrating the pattern.

Although the motivation behind the study has been the incident of falls and the injuries sustained from falls by the elderly, due to safety concerns a younger population was examined in this study. This could be problematic as studies have shown that there is a significant difference between age groups in stepping responses [8]. There is an age related slowing in step response. Additionally, the force generation of the muscles generally decreases with age, which would affect the strength of the response. Therefore it is possible that there is also an age difference for neuromuscular dynamics within the torso.

Looking back now, I would do a few things differently. There would be more questions to collect information on the handedness of the subjects. Additionally there would be a question of whether the subject has had any training that has subjected them to repeated falls (i.e. gymnastics, martial arts, high jump, pole-vault, etc.) and the duration of such training. Additionally the subjects would be limited to a single gender as it appears gender may have a possible effect on the reflex responses observed.

5.5 Study Strengths

Although the current study does not provide information for the prevention of falls or even prevention of injuries from falls, it adds to the understanding of the neuromuscular dynamics during falls. The study demonstrates that erector spinae response is a function of fall direction. The average activation method displays characteristics curves that are visually different and the PCA supports this observed difference with coefficient values for statistical comparison. Analyzing the variance of the coefficients demonstrates that the difference observed in the average activation curves is significantly different.

Although the current study examined a younger population than the age group that motivated the study, the results can be extrapolated to the age group of interest. The population tested is similar to the age group that was examined by Fergusson et al. when he determined that compressive forces on the vertebrae, predicted by using an EMG-assisted biomechanical model developed over the past two decades in the Biodynamics Laboratory, to vary from 3400 to 7500N depending on the region within a bin that a 25lb weight was removed. These compressive forces nearly reach the 8000N average compressive failure force shown for 25 year old spines. However, muscle strength is known to generally decrease with age and usually can no longer attain these high compressive forces. At the same time, the forces required to fracture the vertebrae also decreases, reaching an average failure compression force of 2000N at 75 years of age. This compressive force is nearly half of the acceptable maximum compressive force (3400N) determined by the National Institute of Occupational Safety and Health, for working aged adults [9, 10]. This brings up the question: Does the decrease in muscle force generation decrease below the failure compressive force of the vertebrae or are the muscles contributing to fractures?

The similarities between the fall created and unexpected falls occurring in daily life is a strength. Slippery surfaces may be present any given day whether from a recently mopped floor or a sheet of ice. This may explain why 30-50% of falls are explained by slips and trips [11]. Not only are these types of falls easily possible, the cost associated with slips, even those not resulting in a fall, can result in a significant number of musculoskeletal injuries [12]. The similarities between the experimental and naturally occurring falls mean that the findings should apply to natural falls also.

Although using PCA without a basic understanding of the data and what is expected could be a limitation, the method itself identifies characteristics within the data very well. The time period selected for the current study was when the reflex response was expected to occur, however, if there is another period of interest the PCA would be able to identify common characteristics. This would be beneficial if it were desired to examine later voluntary response strategies occurring during the fall.

5.6 Future Work

The results from this study show that the initial reflex-like activation of the erector spinae muscle is dependent on the fall direction. However this is only one of the major muscles within the body's core. Additional muscles were also collected and should be examined to determine if a similar reflex pattern is observed. The rectus abdominus, internal oblique, and external oblique were collected during this experiment, although they have not been analyzed yet.

Although the data on the rectus abdominus muscles has not been analyzed yet, it can be speculated that a similar opposing pattern might emerge. The reflex response of the rectus abdominus muscle might work to decrease the acceleration of the head to decrease the probability of injury from a fall. Thus for anterior and posterior falls the pattern would be opposite that of the erector spinae muscle. The rectus abdominus could be expected to have a decreased activation for anterior falls as compared to the posterior falls. During the medial-lateral falls the rectus abdominus might be expected to have a reflex response similar to the erector spinae such that the contra-lateral muscle response is greater than the ipsi-lateral response.

Examination of the internal and external obliques could be analyzed using the same methods. The patterns expected may be similar to the patterns observed by the erector spinae and rectus abdominus muscles. However the obliques are also used more for trunk rotation instead of as primary muscles for trunk flexion and extension, suggesting an additional role.

Once the pattern has been determined for a few of the major muscles within the bodies core, the timing of the different muscles should be investigated to determine if the reflex responses of the various muscles are occurring in a specific pattern or activating simultaneously. Strong simultaneous muscle activation could lead to an injury through over loading of the spine. This could help explain why it has been reported that slips, not resulting in falls, have led to musculoskeletal injuries [12]. It could give evidence as to how muscles could at times increase the probability of an injury and explain how an injury occurs when an impact is not present.

Although the electromagnetic sensor data were collected, it has not been analyzed in this study. The electromagnetic sensor data could be analyzed to determine the kinematics of the fall. The electromagnetic sensor data could be used to examine hip or trunk rotation in each of the different fall directions and to further elucidate the relationship between the neuromotor response and motion.

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Chapter 6 Summary of Study

6.1 Introduction

Falls are a significant health problem, accounting for more than 14,000 deaths and 22 million medical visits in 1996 [1]. Although hip fractures are typically considered the greatest cause of morbidity following a fall, up to 15% of vertebral fractures are associated with falls and account for significant morbidity [2]. However, almost 50% of acute, symptomatic vertebral fractures in people 60 years and older occurred with a fall [3]. The symptoms that are associated with vertebral fractures include back pain, physical impairment, increased risk for fractures, and an increase in mortality [4]. Multiple vertebral fractures can lead to changes in spine shape such as increased kyphosis and spine torsion ("dowager's hump").

6.2 Methods

20 subjects (10M & 10F) between 19 and 34 years of age were tested with the approval of the human subjects committee at the University of Kansas. Controlled "slipping type" falls were created by a sliding platform to examine the trunk muscle activation response in different directions. Subjects were instructed to remain motionless and avoid attempting to maintain balance once the platform began moving. Eight surface electromyography (EMG) sensors (Delsys, Boston, MA) collected activation responses to assess the trunk muscle groups: erector spinae (ES), rectus abdominus (RA), internal oblique (IO) and external oblique (EO). EMG data was collected at 1500 HZ and synchronized with the electromagnet power circuit

used to identify the release of the sliding platform. In addition, electromagnetic sensors were collected for a separate study.

The activation response of the erector spinae was examined using two methods. The first method was examining the average of EMG data in four fall directions: anterior, posterior, left lateral, and right lateral. The second method utilized principal component analysis (PCA) to determine the general characteristics that best described the data. The principal eigenvector, which explained the greatest amount of variance within the data, was used to determine coefficients relating the eigenvector to the original data. These coefficients were used to analyze the variance for the two-factors, side of erector spinae and fall direction, repeated measures on both factors.

6.3 Results and Discussion

Both the average activation and the PCA methods demonstrated a difference in neuromotor response based on fall direction. The average method demonstrated a distinct peak activation occurring at 200ms for the anterior fall direction, but no distinct peak activation for the posterior fall direction. The lateral falls responded such that the contra-lateral muscle displayed a distinct activation and the ipsi-lateral muscle did not.

The PCA method demonstrated the same patterns as the average activation method when examining the eigenvector that explained the greatest amount of variance. This eigenvector explained 63.6% of the variability and resembled the reflex like response observed in the average activation method. The original data was related to the eigenvector through coefficients. The coefficients averaged in each fall direction displayed the activation of anterior falls was more than twice as high as posterior falls. The coefficients for the contra-lateral muscle were more than three times greater than the ipsi-lateral muscle for the medial-lateral falls. Analyzing the variance in coefficients of the two-factors, side of erector spinae and fall direction, demonstrated a significant difference ($p < 0.05$) between fall directions. The interaction of side of erector spinae and fall direction was also significant.

These responses are rationalized by the understanding that a strong distinct activation would lead to shortening of the muscle. In the anterior fall direction, shortening of the erector spinae would decelerate the descent of the head towards the ground, resulting in increased time for reaction or at least decreasing the energy transferred if the head does impact the ground. However the opposite would be true for strong activation for the posterior fall direction. Shortening would lead to decrease time available for voluntary response and increase the energy if the head did impact the ground. Similarly, the contra-lateral muscle increases available time to respond and decreases energy if impact occurs while the ipsi-lateral muscle would do the opposite.

6.4 Conclusions

While several possible contributors leading to injury from falls have been examined, the trunk muscle response during falls is currently unknown. This research demonstrates that the neuromotor reflex response is dependent on fall direction. The response appears to respond in a manner to minimize injury by allowing for an increase in time available for voluntary responses to recover from the fall.

PCA was useful in identifying the major contributing characteristics within the data during the time of interest where reflex responses were expected. Examining the data using PCA allowed for the reflex like response to be identified and statistical comparisons between the fall directions possible. The analysis of variance demonstrated a significant difference between the fall directions as well as the interaction of the side of the erector spinae and fall direction for the medial–lateral fall directions.

In addition to the reflex like response examined, PCA could also be used to identify active responses by adjusting the time frame of analysis. In the future, it would be possible to examine the active responses to the different fall directions to develop a better understanding of fall strategies.

6.5 References

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Appendix

A Subject Consent Form

The following consent form was created to inform the subject of the study. A copy of this form was retained by both the subject and the experimenter. This form provided information about the study as well as contact information should they have questions or concerns after leaving.

A Low-Cost Fall Monitor for Geriatric Subjects (Experiment 2)

INTRODUCTION

The Department of Mechanical Engineering at the University of Kansas supports the practice of protection for human subjects participating in research. The following information is provided for you to decide whether you wish to participate in the present study. You may refuse to sign this form and not participate in this study. You should be aware that even if you agree to participate, you are free to withdraw at any time. If you do withdraw from this study, it will not affect your relationship with this unit, the services it may provide to you, or the University of Kansas.

PURPOSE OF THE STUDY

We are interested in evaluating a device that monitors how and when a person falls. This device will be used to monitor if elderly people fall and will be able to call for medical assistance.

PROCEDURES

If you choose to participate, we will first give you a health questionnaire to make sure you do not have any heart problems or back injuries that might make it difficult to do the experiment.

Magnetic markers will be taped or strapped to your legs and at your hips. The markers are used to sense how you move. We will also put electromyographic sensors on the front and back of your torso that will measure what your muscles are doing. In addition we will have you wear a belt that also contains sensors that measure how you move. While wearing these markers, you will be asked to stand on a platform surrounded by soft mats. After a short period of time the platform will be released and you will feel the sensation of slipping. We will ask you to let yourself fall onto the soft mats. We will have you repeat these falls eight times with the slipping sensation in different directions. During this experiment you will be asked to wear protective headgear to avoid injury. Your participation is strictly voluntary and you can stop at anytime. We assure that your name will not be associated in any way with the research findings.

RISKS

Precautions will be taken to insure that falls are performed as safely as possible. While mats and protective headgear will reduce the risk of injury from a fall, it is still possible that someone may react to the fall with abnormal muscle activity and experience muscle soreness. In addition, like any physical activity there may be unforeseen injury risks. In addition, some people have allergies to adhesives such as in band-aids or in the tape we are using to attach the markers.

BENEFITS

With this research we hope to be able to develop a system that can track when someone falls. This will be used to monitor elderly people to determine how and when they fall. There is, however, no direct benefit for the subject of this study.

PAYMENT TO PARTICIPANTS

Subjects will receive $20 for participation in the study.

INFORMATION TO BE COLLECTED

To perform this study, researchers will collect information about you. This information will be obtained from a questionnaire that will assess if you have heart or musculoskeletal problems that might make exercise inadvisable. Also, information will be collected from the study activities that are listed in the Procedures section of this consent form. This includes information about movements, your height and your weight.

Your name will not be associated in any way with the information collected about you or with the research findings from this study. The researcher(s) will use a study number instead of your name.

In addition, Dr. Wilson and her team may share the information gathered in this study, including your information, with the National Institute of Health that is funding the study. Again, your name would not be associated with the information disclosed to these individuals.

In addition, the information collected about you will also be used by the company that created the measurement device (Barron Assoc.) to let them know how well their device works. Again, your name would not be associated with the information disclosed to these individuals. Some persons or groups that receive your information may not be required to comply with the Health Insurance Portability and Accountability Act's privacy regulations, and your information may lose this federal protection if those persons or groups disclose it.

The researchers will not share information about you with anyone not specified above unless required by law or unless you give written permission.

Permission granted on this date to use and disclose your information remains in effect indefinitely. By signing this form you give permission for the use and disclosure of your information for purposes of this study at any time in the future.

INSTITUTIONAL DISCLAIMER STATEMENT
In the event of injury, the Kansas Tort Claims Act provides for compensation if it can be demonstrated that the injury was caused by the negligent or wrongful act or omission of a state employee acting within the scope of his/her employment.

REFUSAL TO SIGN CONSENT AND AUTHORIZATION

You are not required to sign this Consent and Authorization form and you may refuse to do so without affecting your right to any services you are receiving or may receive from the University of Kansas or to participate in any programs or events of the University of Kansas. However, if you refuse to sign, you cannot participate in this study.

CANCELLING THIS CONSENT AND AUTHORIZATION

You may withdraw your consent to participate in this study at any time. You also have the right to cancel your permission to use and disclose information collected about you, in writing, at any time, by sending your written request to: Dr. Sara Wilson, Mechanical Engineering, University of Kansas, Lawrence, KS 66045. If you cancel permission to use your information, the researchers will stop collecting additional information about you. However, the research team may use and disclose information that was gathered before they received your cancellation, as described above.

PARTICIPANT CERTIFICATION:
I have read this Consent and Authorization form. I have had the opportunity to ask, and I have received answers to, any questions I had regarding the study and the use and disclosure of information about me for the study. I understand that if I have any additional questions about my rights as a research participant, I may call (785) 864-7429 or write the Human Subjects Committee Lawrence Campus (HSCL), University of Kansas, 2385 Irving Hill Road, Lawrence, Kansas 66045-7563, email dhann@ku.edu.

I agree to take part in this study as a research participant. I further agree to the uses and disclosures of my information as described above. By my signature I affirm that I am at least 18 years old and that I have received a copy of this Consent and Authorization form.

_______________________________ ______________________
Type/Print Participant's Name Date

____
Participant's Signature

Researcher Contact Information

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3013 Learned Hall
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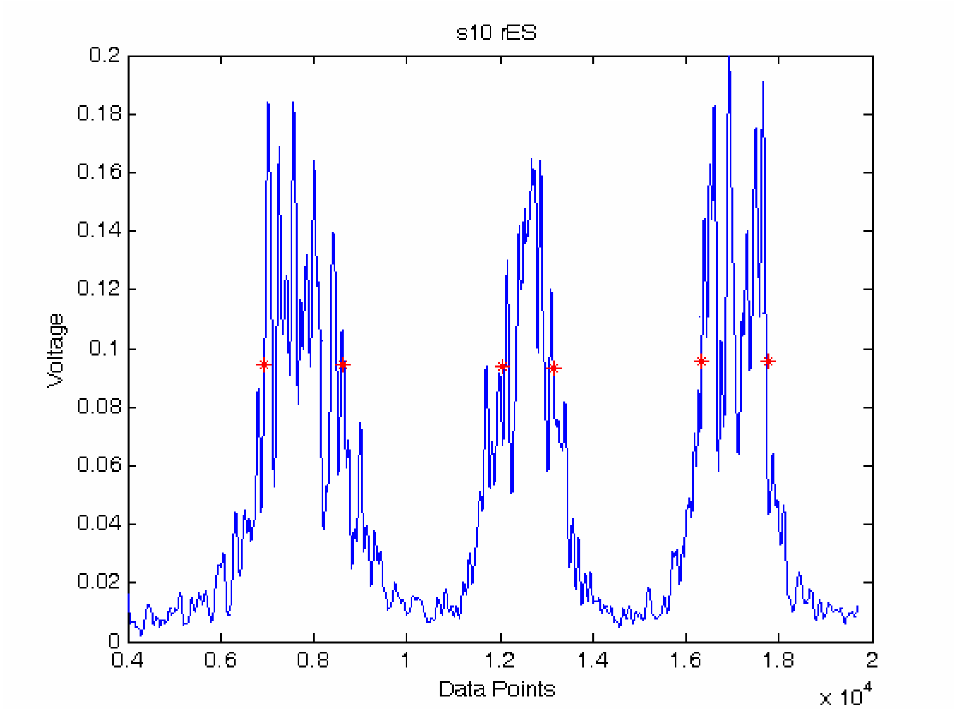
Lawrence, KS 66045
785 864-2103

B Maximum EMG Table

The maximum EMG activations were determined initially. This was accomplished by determining the mean peak activation of three activation peaks collected at the beginning of the experiment.

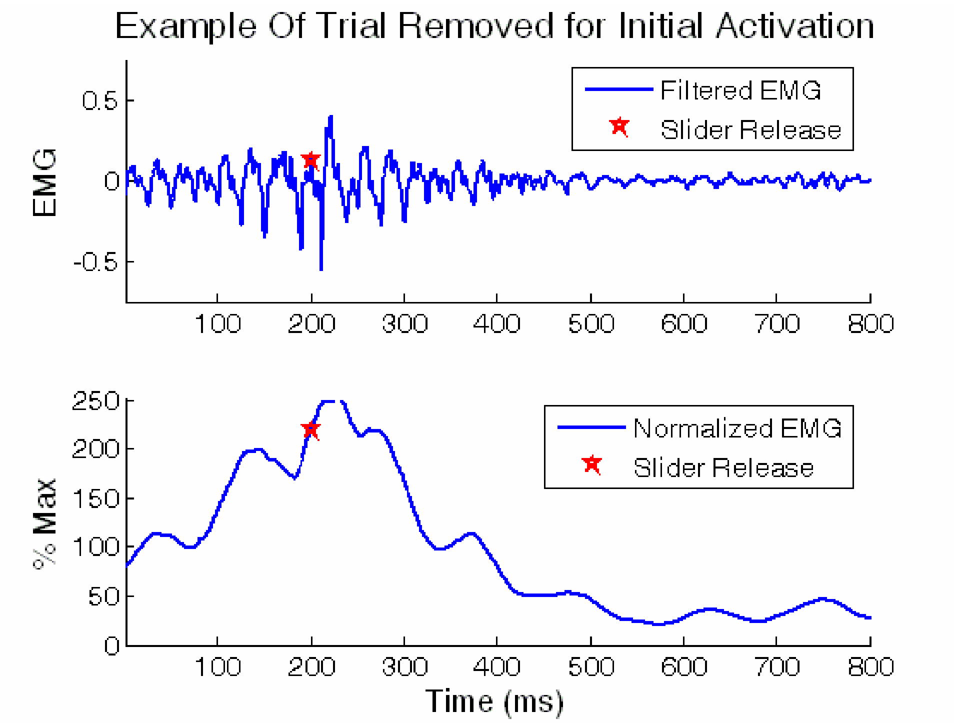


The average activations are included in the following table. Difficulty with equipment resulted in the left side erector spinae for subject 11 to be lost. In order to allow successive programs to run without interruption, an extremely low value was placed in the table. This was easily identified alter by having an extremely high activation and thus removed.

	rES	lES
s4	0.03099	0.03022
s5	0.01291	0.01984
s6	0.07495	0.08862
s7	0.07394	0.08987
s8	0.02670	0.00390
s9	0.02607	0.08755
s10	0.12688	0.18850
s11	0.03744	0.00001
s12	0.06034	0.06110
s13	0.01147	0.02662
s14	0.09656	0.09725
s15	0.04299	0.03039
s16	0.03437	0.03174
s17	0.06257	0.08986
s18	0.05533	0.05180
s19	0.12260	0.11921
s20	0.04048	0.05179
s21	0.04255	0.06997
s22	0.00748	0.13538
s23	0.02627	0.02560

C Initial Activation Test

The normalized EMG activations were plotted from 200ms prior to the platforms release for trials removed based on the second criterion. The platform release is marked by a star. The activation curve is increasing prior to the release, which would prevent the reflex response from the fall from being examined.



D Missing Data Estimation Iterations

Some of the missing PCA coefficients were estimated prior to analyzing the variance. The technique chosen to estimate the missing values is an iterative process. The average of the columns and rows were determined from the previous iteration and used in the calculation of the current values. The averages of the rows were also divided into two groups based on the side of the erector spinae muscle. The following tables show the iterations that were completed. The shaded cells were the missing coefficients. Of the shaded cells, five were estimated for the analysis and have the current estimation for the cell. All non-shaded cells remained constant.

				PCA Mode 1 - Iteration 1								
		rES				lES					rES	lES
	n	A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.65449	-0.32892
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.61394	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.02556	-0.22465
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0242	-0.0779		-0.45448	-0.1938
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.18378	0.009785
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.126076	-0.02736
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0218	-0.0127	-0.0323		-0.13591	-0.12268
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.25133	0
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.44492	-0.20457
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.14469	-0.49695
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.06313	-0.05094
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.11569	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.014389	-0.14272
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0055	0.0231	-0.0243		0.039158	0.033041
15	s18	0.1753	0.0204	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.368841	-0.12171
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.16604	0.057614
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.24344	-0.15441
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.129578	0.145006
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0	0.006193
20	s23					0.0782	0.0338	0.0426	0.0748		0	0.186852
											T	T
	Tj	-0.82511	-1.08723	-0.35924	-0.28508	-0.74881	-0.32504	-0.16168	-1.01136	T	-2.55666	-2.2469
	Missing data replaced by equation found in book - Research Design and Statistical Analysis											
n	a	Ti = sum of subject										
20	8	Tj = sum of Direction										

PCA Mode 1 - Iteration 2

		rES				IES					rES	IES
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0274	-0.0779		-0.4545	-0.2180
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0266	-0.0127	-0.0323		-0.1359	-0.1445
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0055	0.0231	-0.0243		0.0392	0.0386
15	s18	0.1753	0.0250	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3892	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0511	0.0748		0.0000	0.2294
											T	T
	Tj	-0.8251	-1.0669	-0.3592	-0.2851	-0.7488	-0.3413	-0.1433	-1.0114	T	-2.5363	-2.2448

PCA Mode 1 - Iteration 3

		rES				IES					rES	IES
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0276	-0.0779		-0.4545	-0.2212
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0278	-0.0127	-0.0323		-0.1359	-0.1493
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0052	0.0231	-0.0243		0.0392	0.0385
15	s18	0.1753	0.0261	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3939	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0530	0.0748		0.0000	0.2380
											T	T
	Tj	-0.8251	-1.0622	-0.3592	-0.2851	-0.7488	-0.3461	-0.1379	-1.0114	T	-2.5316	-2.2442

	PCA Mode 1 - Iteration 4											
		rES				IES					rES	IES
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0275	-0.0779		-0.4545	-0.2214
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0281	-0.0127	-0.0323		-0.1359	-0.1504
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0050	0.0231	-0.0243		0.0392	0.0382
15	s18	0.1753	0.0263	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3949	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0534	0.0748		0.0000	0.2398
											T	T
	Tj	-0.8251	-1.0611	-0.3592	-0.2851	-0.7488	-0.3476	-0.1363	-1.0114	T	-2.5306	-2.2441

	PCA Mode 1 - Iteration 5											
		rES				IES					rES	IES
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0275	-0.0779		-0.4545	-0.2213
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0281	-0.0127	-0.0323		-0.1359	-0.1507
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0050	0.0231	-0.0243		0.0392	0.0381
15	s18	0.1753	0.0264	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3952	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0535	0.0748		0.0000	0.2402
											T	T
	Tj	-0.8251	-1.0609	-0.3592	-0.2851	-0.7488	-0.3480	-0.1358	-1.0114	T	-2.5303	-2.2440

	PCA Mode 1 - Iteration 6											
		rES				IES					rES	IES
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0275	-0.0779		-0.4545	-0.2213
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0282	-0.0127	-0.0323		-0.1359	-0.1508
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0050	0.0231	-0.0243		0.0392	0.0380
15	s18	0.1753	0.0264	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3952	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0535	0.0748		0.0000	0.2403
											T	T
	Tj	-0.8251	-1.0609	-0.3592	-0.2851	-0.7488	-0.3482	-0.1357	-1.0114	T	-2.5303	-2.2440

	PCA Mode 1 - Iteration 7											
		rES				IES					rES	IES
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0275	-0.0779		-0.4545	-0.2213
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0282	-0.0127	-0.0323		-0.1359	-0.1508
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0050	0.0231	-0.0243		0.0392	0.0380
15	s18	0.1753	0.0264	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3952	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0535	0.0748		0.0000	0.2404
											T	T
	Tj	-0.8251	-1.0608	-0.3592	-0.2851	-0.7488	-0.3482	-0.1356	-1.0114	T	-2.5303	-2.2440

		PCA Mode 1 - Iteration 8										
		rES				IES					rES	IES
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0275	-0.0779		-0.4545	-0.2213
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0282	-0.0127	-0.0323		-0.1359	-0.1508
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0050	0.0231	-0.0243		0.0392	0.0380
15	s18	0.1753	0.0264	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3952	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0535	0.0748		0.0000	0.2404
											T	T
	Tj	-0.8251	-1.0608	-0.3592	-0.2851	-0.7488	-0.3482	-0.1356	-1.0114	T	-2.5303	-2.2440

		PCA Mode 1 - Iteration 9										
		rES				IES					rES	IES
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0275	-0.0779		-0.4545	-0.2213
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0282	-0.0127	-0.0323		-0.1359	-0.1508
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0050	0.0231	-0.0243		0.0392	0.0380
15	s18	0.1753	0.0264	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3952	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0535	0.0748		0.0000	0.2404
											T	T
	Tj	-0.8251	-1.0608	-0.3592	-0.2851	-0.7488	-0.3482	-0.1356	-1.0114	T	-2.5303	-2.2440

n		rES				IES					rES	IES
	PCA Mode 1 - Iteration 10											
n		A	C	E	G	A	C	E	G		Ti	Ti
1	s4	-0.1313	-0.1569	-0.1749	-0.1914	-0.0234	-0.1166	-0.0277	-0.1612		-0.6545	-0.3289
2	s5	-0.3343	-0.0254	-0.0935	-0.1608	-0.1907	-0.0226	-0.1189	-0.2012		-0.6139	-0.5334
3	s6	0.0000	-0.0133	-0.0118	-0.0005	-0.0743	-0.0784	-0.0125	-0.0593		-0.0256	-0.2246
4	s7	-0.0979	-0.1598	-0.0260	-0.1708	-0.0635	-0.0524	-0.0275	-0.0779		-0.4545	-0.2213
5	s8	-0.0661	-0.1021	-0.0076	-0.0080	0.0100	-0.0080	0.0051	0.0026		-0.1838	0.0098
6	s9	0.0333	0.0347	0.0328	0.0253	-0.0046	-0.0028	-0.0034	-0.0165		0.1261	-0.0274
7	s10	-0.0311	-0.0756	-0.0231	-0.0062	-0.0777	-0.0282	-0.0127	-0.0323		-0.1359	-0.1508
8	s11	-0.1119	-0.0826	-0.0252	-0.0316						-0.2513	0.0000
9	s12	-0.0755	-0.1536	-0.0365	-0.1794	-0.0857	-0.0352	0.0173	-0.1010		-0.4449	-0.2046
10	s13	-0.1447				-0.1348	-0.1015	-0.0979	-0.1628		-0.1447	-0.4969
11	s14	-0.0241	-0.0142	-0.0245	-0.0003	-0.0281	0.0009	-0.0119	-0.0119		-0.0631	-0.0509
12	s15	-0.0245	-0.0185	-0.0132	-0.0595	-0.0198	-0.0319	0.0027	-0.0342		-0.1157	-0.0833
13	s16	0.0281	-0.0607	0.0108	0.0362	-0.0139	0.0026	-0.0400	-0.0914		0.0144	-0.1427
14	s17	0.0051	0.0441	0.0010	-0.0110	0.0342	0.0050	0.0231	-0.0243		0.0392	0.0380
15	s18	0.1753	0.0264	0.0479	0.1456	-0.1822	0.0113	0.0377	0.0115		0.3952	-0.1217
16	s19	-0.0272	-0.0925	0.0054	-0.0518	0.0287	0.0057	0.0271	-0.0039		-0.1660	0.0576
17	s20	-0.0227	-0.2173	-0.0457	0.0422	-0.0694	0.0168	-0.0067	-0.0951		-0.2434	-0.1544
18	s21	0.0242	0.0064	0.0248	0.0743	0.0598	0.0402	0.0366	0.0084		0.1296	0.1450
19	s22				0.2627	0.0084	0.0131	0.0203	-0.0356		0.2627	0.0062
20	s23					0.0782	0.0338	0.0535	0.0748		0.0000	0.2404
											T	T
	Tj	-0.8251	-1.0608	-0.3592	-0.2851	-0.7488	-0.3482	-0.1356	-1.0114	T	-2.5303	-2.2440

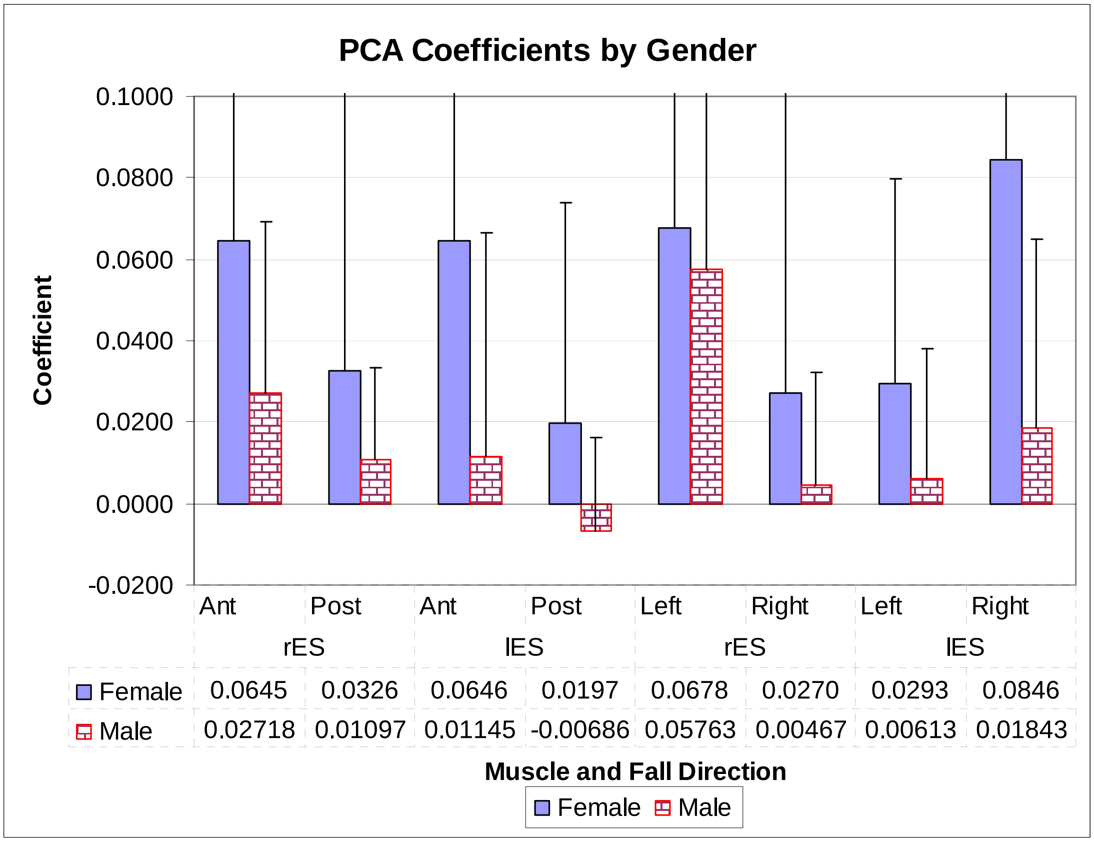
E PCA Coefficients

rES		lES		rES		lES	
Anterior	Posterior	Anterior	Posterior	Left	Right	Left	Right
-0.1313	-0.1749	-0.0234	-0.0277	-0.1569	-0.1914	-0.1166	-0.1612
-0.3343	-0.0935	-0.1907	-0.1189	-0.0254	-0.1608	-0.0226	-0.2012
0.0000	-0.0118	-0.0743	-0.0125	-0.0133	-0.0005	-0.0784	-0.0593
-0.0979	-0.0260	-0.0635	-0.0275	-0.1598	-0.1708	-0.0524	-0.0779
-0.0661	-0.0076	0.0100	0.0051	-0.1021	-0.0080	-0.0080	0.0026
0.0333	0.0328	-0.0046	-0.0034	0.0347	0.0253	-0.0028	-0.0165
-0.0311	-0.0231	-0.0777	-0.0127	-0.0756	-0.0062	-0.0282	-0.0323
-0.1119	-0.0252			-0.0826	-0.0316		
-0.0755	-0.0365	-0.0857	0.0173	-0.1536	-0.1794	-0.0352	-0.1010
-0.1447		-0.1348	-0.0979			-0.1015	-0.1628
-0.0241	-0.0245	-0.0281	-0.0119	-0.0142	-0.0003	0.0009	-0.0119
-0.0245	-0.0132	-0.0198	0.0027	-0.0185	-0.0595	-0.0319	-0.0342
0.0281	0.0108	-0.0139	-0.0400	-0.0607	0.0362	0.0026	-0.0914
0.0051	0.0010	0.0342	0.0231	0.0441	-0.0110	0.0050	-0.0243
0.1753	0.0479	-0.1822	0.0377	0.0264	0.1456	0.0113	0.0115
-0.0272	0.0054	0.0287	0.0271	-0.0925	-0.0518	0.0057	-0.0039
-0.0227	-0.0457	-0.0694	-0.0067	-0.2173	0.0422	0.0168	-0.0951
0.0242	0.0248	0.0598	0.0366	0.0064	0.0743	0.0402	0.0084
		0.0084	0.0203		0.2627	0.0131	-0.0356
		0.0782	0.0535			0.0338	0.0748

F Gender Comparison of PCA Coefficients

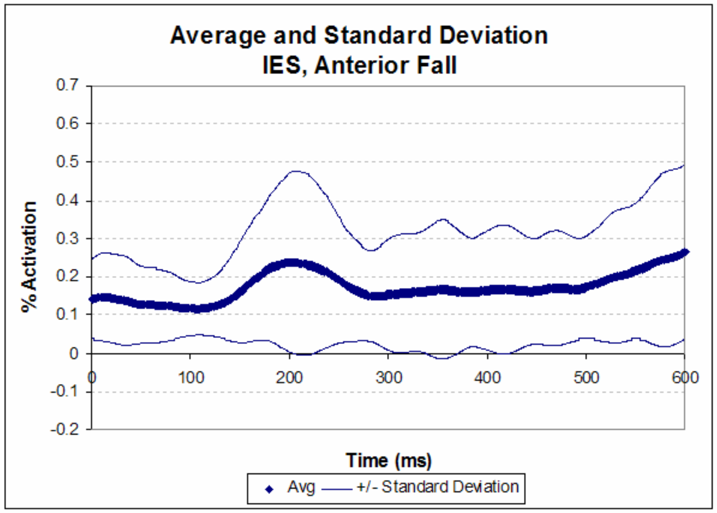
Following the determination of the coefficients, the separate genders were examined to determine if there was a difference. The table displays the averages and standard deviations based on gender. A bar plot was created to display the differences.

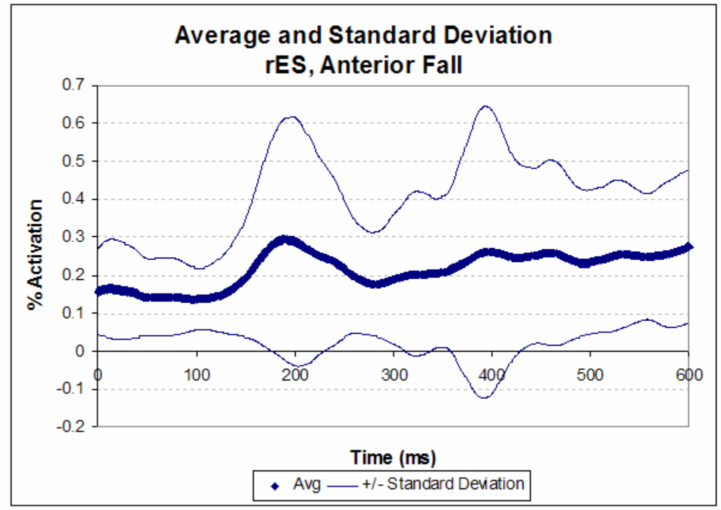
	FEMALE							
	rES		IES		rES		IES	
	A	E	A	E	C	G	C	G
	-0.1313	-0.1749	-0.0234	-0.0277	-0.1569	-0.1914	-0.1166	-0.1612
	-0.3343	-0.0935	-0.1907	-0.1189	-0.0254	-0.1608	-0.0226	-0.2012
	-0.0979	-0.0260	-0.0635	-0.0275	-0.1598	-0.1708	-0.0524	-0.0779
	-0.0755	-0.0365	-0.0857	0.0173	-0.1536	-0.1794	-0.0352	-0.1010
	-0.1447		-0.1348	-0.0979			-0.1015	-0.1628
	-0.0245	-0.0132	-0.0198	0.0027	-0.0185	-0.0595	-0.0319	-0.0342
	0.0281	0.0108	-0.0139	-0.0400	-0.0607	0.0362	0.0026	-0.0914
	0.1753	0.0479	-0.1822	0.0377	0.0264	0.1456	0.0113	0.0115
	0.0242	0.0248	0.0598	0.0366	0.0064	0.0743	0.0402	0.0084
			0.0084	0.0203		0.2627	0.0131	-0.0356
Avg	0.0645	0.0326	0.0646	0.0197	0.0678	0.0270	0.0293	0.0846
SD	0.1416	0.0718	0.0831	0.0542	0.0778	0.1650	0.0503	0.0736
	MALE							
	0.0000	-0.0118	-0.0743	-0.0125	-0.0133	-0.0005	-0.0784	-0.0593
	-0.0661	-0.0076	0.0100	0.0051	-0.1021	-0.0080	-0.0080	0.0026
	0.0333	0.0328	-0.0046	-0.0034	0.0347	0.0253	-0.0028	-0.0165
	-0.0311	-0.0231	-0.0777	-0.0127	-0.0756	-0.0062	-0.0282	-0.0323
	-0.1119	-0.0252			-0.0826	-0.0316		
	-0.0241	-0.0245	-0.0281	-0.0119	-0.0142	-0.0003	0.0009	-0.0119
	0.0051	0.0010	0.0342	0.0231	0.0441	-0.0110	0.0050	-0.0243
	-0.0272	0.0054	0.0287	0.0271	-0.0925	-0.0518	0.0057	-0.0039
	-0.0227	-0.0457	-0.0694	-0.0067	-0.2173	0.0422	0.0168	-0.0951
			0.0782	0.0535			0.0338	0.0748
Avg	0.0272	0.0110	0.0115	-0.0069	0.0576	0.0047	0.0061	0.0184
SD	0.0421	0.0226	0.0550	0.0231	0.0809	0.0277	0.0319	0.0463

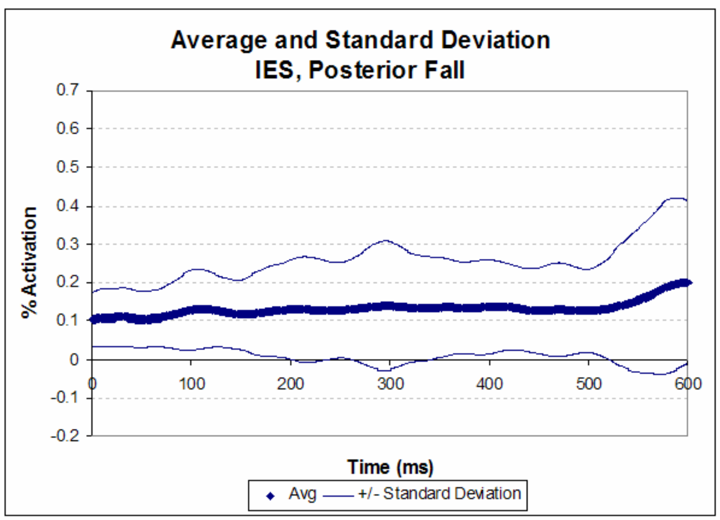


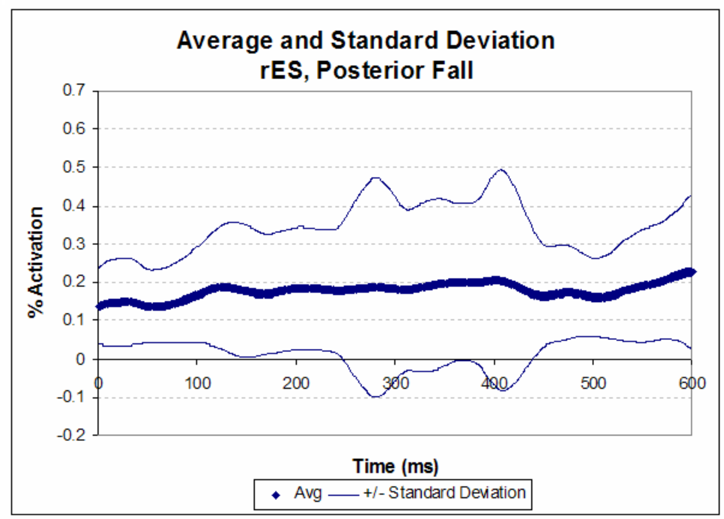
G Averages and Standard Deviations

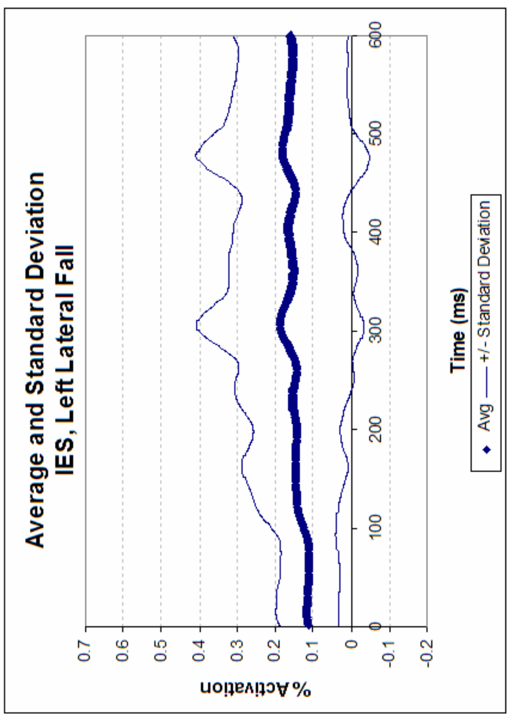
The removal of data caused the averaging of the subjects for equal representation. The overall averages were determined from these averages, but how well does this average represent the data (magnitude, timing)? The following plots examined the average activations and standard deviations to attempt to answer this question.

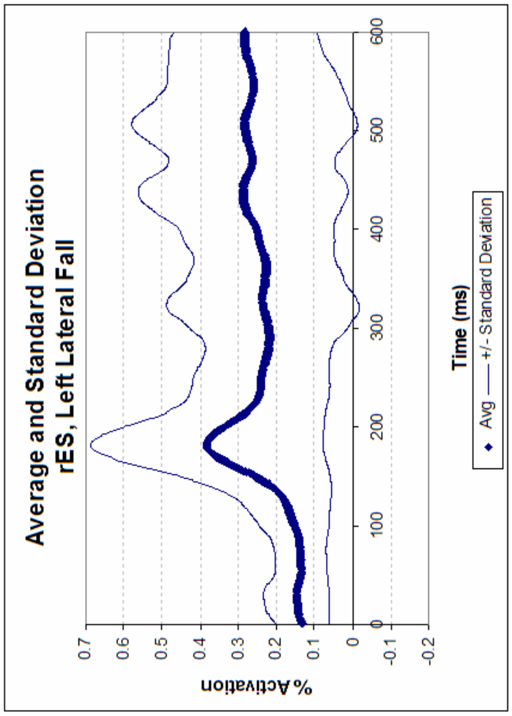

Average and Standard Deviation
lES, Anterior Fall
% Activation
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
0
100
200
300
400
500
600
Time (ms)
Avg
+/- Standard Deviation

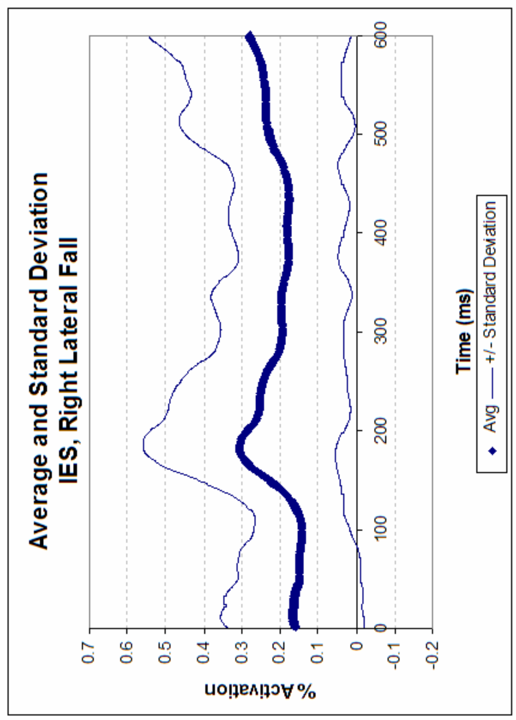

Average and Standard Deviation
rES, Anterior Fall
% Activation
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
0
100
200
300
400
500
600
Time (ms)
Avg
+/- Standard Deviation

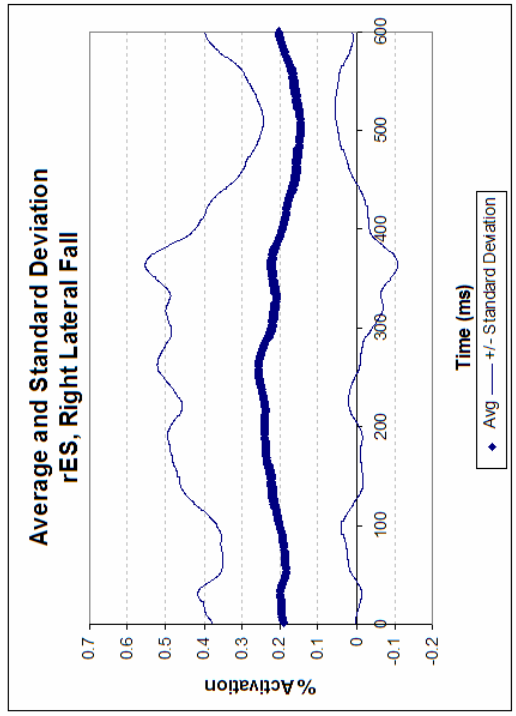

Average and Standard Deviation
lES, Posterior Fall
% Activation
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
0
100
200
300
400
500
600
Time (ms)
Avg
+/- Standard Deviation


Average and Standard Deviation
rES, Posterior Fall
% Activation
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
0
100
200
300
400
500
600
Time (ms)
Avg
+/- Standard Deviation


Average and Standard Deviation
lES, Left Lateral Fall
%Activation
Time (ms)
Avg
+/- Standard Deviation
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
0
100
200
300
400
500
600


Average and Standard Deviation
rES, Left Lateral Fall
%Activation
Time (ms)
Avg
+/- Standard Deviation
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
0
100
200
300
400
500
600


Average and Standard Deviation
lES, Right Lateral Fall
%Activation
Time (ms)
Avg
+/- Standard Deviation
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
0
100
200
300
400
500
600


Average and Standard Deviation
rES, Right Lateral Fall
%Activation
Time (ms)
Avg
+/- Standard Deviation
0.7
0.6
0.5
0.4
0.3
0.2
0.1
0
-0.1
-0.2
0
100
200
300
400
500
600