Spring 2010

Neuromuscular adaptations during perturbations in individuals with and without bilateral vestibular loss

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NEUROMUSCULAR ADAPTATIONS DURING PERTURBATIONS IN INDIVIDUALS WITH AND WITHOUT BILATERAL VESTIBULAR LOSS

by

Nora Havlik Riley

An Abstract

Of a thesis submitted in partial fulfillment
of the requirements for the Doctor of
Philosophy degree in Physical Therapy and Rehabilitation Science
in the Graduate College of
The University of Iowa

May 2010

Thesis Supervisor: Professor Richard K. Shields

ABSTRACT

Approximately 20% of the general population is affected by a vestibular disorder. Vestibular dysfunction is recognized as an important intrinsic factor leading to falls. Despite research on balance strategies with platform perturbations, limited information exists on neuromuscular performance of the knee with perturbations during functional tasks. Improved understanding of the effects of BVL on neuromuscular control of the knee will aid researchers and clinicians in developing rehabilitation programs that address the adaptations and balance deficits that occur with vestibular loss. Hence, the main purpose was to examine accuracy of performance, knee muscle activation patterns and long latency responses in response to unexpected perturbations during a controlled single leg squat in healthy individuals and those with bilateral vestibular loss (BVL).

The first study provided information about the ability to improve performance accuracy with perturbations based on the feedback available. It also showed concomitant changes in the LLR of quadriceps muscles with learning. In the second study, it was found that competent subjects with BVL show similar performance accuracy as healthy individuals during the SLS, with the exception of endpoint error. Muscle strategies are slightly different and vary on firm and foam surfaces. A significant finding was that the LLR is reduced in this group in response to unexpected perturbations, especially when visual feedback is absent. Rehabilitation and/or time living with bilateral vestibular deficiency can lead to a reorganization of the central nervous system, which may partly explain the alterations in neuromuscular control. More research is needed to determine the relationship between the long latency response and fall risk and if different training dosages with perturbations affect these in both healthy and vestibular-deficient populations.

Abstract Approved: ______________________________

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CERTIFICATE OF APPROVAL

PH.D. THESIS

This is to certify that the Ph.D. thesis of

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has been approved by the Examining Committee for the thesis requirement for the Doctor of Philosophy degree in Physical Therapy and Rehabilitation Science at the May 2010 graduation.

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For every fact there is an infinity of hypotheses.

Robert M. Pirsig

ACKNOWLEDGMENTS

The completion of this work was certainly not a simple endeavor and it is my pleasure to thank and recognize the many people that supported me during this process. The inspiration for my research came from previous work performed in this lab. I owe sincere thankfulness to Bryon Ballantyne and Sangeetha Madhaven for their work and guidance. In particular, special thanks to Bryon for his immense support throughout this research experience. Sincere thanks also to the individuals who rendered their valuable time and effort to participate in my study. Without their generosity, this project would never have been completed.

I am grateful to all of those with whom I have had the pleasure to work during the past years during this and other projects. In particular, special thanks to Masaki Iguchi, Andy Littmann, and Shuo-Hsiu Chang for their assistance with experimental set-up and general joviality. Thanks as well, to Dave Gerleman, for his technical assistance. I also owe sincere gratitude to the staff of the Physical Therapy program, especially Judy Biderman, who has the answer to every question. Her experience and support was invaluable.

My heartfelt gratitude goes to the St. Ambrose colleagues who have been supportive of my career goals and who worked actively to provide me protected academic time to pursue those goals. They are a truly wonderful group of people.

I am grateful to the esteemed members of my dissertation committee, Warren Darling, Tom Cook, John Yack, and Deema Fattal who enabled my dissertation experience to be satisfying as well as instructive. Each generously gave their time and expertise to better my work. I am thankful for their contribution and good-natured support.

My thanks and sincere appreciation to Richard K Shields for persevering with me as my advisor throughout the time it took me to complete this project. With him as my mentor, I was able to explore areas of great interest and learned more than I imagined. He has provided me extensive professional guidance and shown me, by his example, what a good scientist should be.

Nobody has been more important to me in the pursuit of this project than the members of my family. I would like to thank my parents, whose love and guidance are with me in whatever I pursue. They are the ultimate role models. My sisters and brothers, Havliks and Rileys alike, are an ongoing source of both support and entertainment. Most importantly, I want to thank my husband and children who did not complain (too loudly) as I worked many late nights and weekends. My husband, Joe, has grown infinitely supportive and patient during this graduate school experience. My three amazing children, Connor, Morgan, and Molly, are wise beyond their years and provide unending inspiration. Connor always seemed to understand the difficulty of my "homework," while Morgan and Molly finally know a mom who isn't always "working on her Ph.D." Their love and support provided me the energy to complete this endeavor.

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CHAPTER 1

INTRODUCTION

Overview

An estimated 20% of the general population is affected by a vestibular disorder.[1] Vestibular dysfunction is recognized as an important intrinsic factor leading to falls,[2-4] with a recent study finding approximately 80% of 428 fallers experience symptoms of vestibular impairment.[5] Maintenance of postural stability requires rapid processing of signals from visual, vestibular, and somatosensory systems, and balance may deteriorate when these systems fail individually or collectively.[6] Intact vestibular function is required during challenging activities or when visual and somatosensory information is unavailable or ambiguous.[7-10]

Contributors to Postural Control

Postural perturbations may be sensed by visual receptors sensitive to linear and angular motion of the optical array, by vestibular receptors sensitive to acceleration of the head, and by somatosensory receptors sensitive to contact forces and joint angle changes. The interaction of these systems has made it difficult to delineate the specific contribution of each subsystem.[9] The vestibular system, however, is a key contributor in the sensorimotor control of posture.[11-12] This was shown by Kaufman et al.[12] who demonstrated that the postural control system aligns with the gravitational vector. Vestibular input is also required for resolution of ambiguous visual and somatosensory reference.[11] In addition, vestibular influences on postural control include modulation of the body's postural tone and antigravity reflexes that are vital to maintaining stance of the body against gravity without conscious effort.[13]

While the otolith organs of the inner ear provide information about the orientation of the head to gravitoinertial acceleration, the ongoing sense of orientation also includes the alignment of the entire body, and all of its links and segments, relative to gravity. Position sense is influenced by the muscle spindle fibers interspersed in parallel with the extrafusal muscle fibers that perform muscle contraction.[14] The spindle sensory signal is interrelated to the ongoing pattern of gamma and alpha activation and other signals about body loading to compute the angle and rate of change of the joint controlled by the muscle.[15] Somatosensory stimulation influences the perception of being upright, and the control of posture as receptors in the soles of the feet are also important in the control of posture.[16]

Previous Methods of Studying Balance and
Response to Perturbations

Though it is difficult to imagine a physical condition in which a true threat to balance would disturb only a single sensory channel, scientists have attempted to manipulate this in order to determine the relative contributions of visual, vestibular, and sensory information to postural control.[17] Galvanic vestibular stimulation (GVS) is one technique used to activate the vestibular system and assess vestibulospinal reflexes.[18] Studies of the effects of galvanic vestibular stimulation on postural sway have produced important evidence of vestibular-somatosensory interaction. Computerized dynamic posturography (CDP) is another method used to examine the relative vestibular, visual, and sensory contributions to postural stability. The concepts underlying CDP were initially developed by Lewis M. Nashner in his doctoral thesis[11] and are commonly used to test postural control in a wide variety of individuals with various diagnoses.

A quantitative method, CDP, assesses how the balance system utilizes sensory and motor components in the standing human.[11] This test examines the ability of the patient to stand quietly under conditions in which visual and/or somatosensory feedback is altered, while recording balance corrections as forces created on a support surface, as well as electromyographic (EMG) signals from muscle responses. Numerous studies have examined postural control of patients with varying degrees of vestibular loss using dynamic posturography.[9-10, 19-20] Consistently, these studies demonstrate that patients with inadequate peripheral vestibular input show ineffective postural responses with perturbations, especially when both visual and somatosensory feedback are altered.

Although CDP is well-accepted as a reliable test of vestibulospinal function,[9, 11, 19, 21-23] limitations have been noted.[24-27] One limitation is the artificial manner in which balance is challenged.[24] Perturbations are delivered during static stance, using unidirectional movement of the support surface, such as a sudden translation or rotation. In addition, data are usually averaged over multiple trials. However, postural responses typically habituate (the response decreases with repetition).[27] The first trial response is generally very different from subsequent trials and is often excluded from analysis. The disadvantage of this analysis method is the potential loss of data which are only present in the initial trials. The trials that may provide the most input into the responses associated with truly unexpected perturbations are eliminated.[27] Despite its limitations, the posture platform continues to be employed as the primary method for obtaining quantitative measures of postural reactions in clinical populations.[6] Methodology that instead assesses feedforward, feedback, and volitional responses to perturbations, which are unexpected each and every trial ,would provide more accurate insights into the neuromuscular control strategies used.

Assessment of Neuromuscular Control

Neuromuscular control is believed to be fundamental to minimizing undue stresses and strains to tissues that stabilize the joint.[28-29] From a joint stability perspective, this is an operational definition that refers to the preparation and activation of controllable restraints of the joint to prepare and respond to joint motion and loading for the purpose of maintaining functional joint stability.[28] Neuromuscular control encompasses both preparatory and reflexive responses that are mediated by proprioceptive feedforward and feedback mechanisms.[29] More specifically, for the purpose of this study, neuromuscular control encompasses the task-specific activation of muscles to accurately perform a skilled task and effectively respond to perturbations. Any factor that impairs or inhibits one or more of the components of the neuromuscular control system may compromise joint stability and increase the risk of injury during functional activities.

One method to assess neuromuscular control of the knee has been the measurement of muscle response while subjecting the knee to sudden perturbation or stress.[30-31] However, these and other studies that investigate muscle activation strategies of the knee tend to measure response characteristics under resting conditions or while the joint is unloaded. These are static situations that do not approach real life weight-bearing conditions. In order to develop a broader understanding of postural responses used during functional activities, reflexive muscle activation patterns in response to an unexpected perturbation of the knee during dynamic weight bearing applications need to be studied.

Triggered Long Latency Responses

In response to perturbations, a typical recruitment of muscles (strategy) is noted which varies based on the speed, direction, and type of perturbation.[32] In general, postural responses include instantaneous torques due to mechanical stiffness, short-latency responses due to spinal reflexes, and a variety of long-latency responses.[33-34] A rapid stretch of an active muscle produces a series of reflex bursts of EMG.[35-36] The initial short latency stretch reflex, referred to as M1, has an onset latency of 30 – 40 ms and is mediated by the large diameter group Ia muscle spindle afferents of the muscle spindle through a monosynaptic segmental pathway.[37-40] The later M2 and M3 components are commonly referred together as the long latency response (LLR)[41] with the true origin of these less certain. M2 begins around 50 – 80 ms after the perturbation. Evidence suggests it is mediated by the slow conducting group II afferents and is considered to be an oligosynaptic spinal reflex.[42-44] The M3 component (onset latency 85 – 100 ms) is thought to be mediated by the group Ia afferents as well as supraspinal input through a partly transcortical pathway.[45-48] Supraspinal input, including visual and vestibular signals as well as proprioceptive signals, contribute to the determination of context and feedback responses, allowing for more complex control than would be possible with spinal reflexes alone.[32, 41, 48]

Long latency responses are important contributors to the regulation of muscle and joint stiffness.[41, 46, 49] The extent to which these may contribute to joint stability or to injury, however, is uncertain. In CDP and GVS studies, healthy adults exhibit a facilitation of this response with platform perturbations, which allows them to maintain stability without loss of balance (LOB). Therefore, in these conditions, the LLR helps to decrease the amount of postural sway, thus increasing stability.[17, 49-50] In contrast, individuals with bilateral vestibular loss (BVL) demonstrate a decreased LLR amplitude response and a subsequent increase in postural sway and/or loss of balance. This also leads to a reduced change in torque at the ankle joint so BVL subjects are not as effective in producing the necessary forward torque around the ankle joint that would restabilize the body.[51] Allum and Honegger (1998) studied perturbations using CDP with subjects with BVL. They concluded that ML/LLR are possibly gated or triggered by proprioceptive afferent signals elicited by muscle stretch in the lower leg and then their response amplitudes are modulated by vestibular signals.[52] These studies lend further support to the contribution of the vestibular system to these responses.

Factors That Influence Long Latency Responses

The potential contribution of supraspinal drive to the LLR provides greater flexibility in this triggered response than would be feasible in a purely segmental pathway.[53] Afferent information (vestibular, visual, and somatosensory) triggers long latency responses, and these responses are influenced by a variety of factors such as the task performed, [54-55] prior experience with a task,[7, 56-58] practice,[7, 56]intent,[59] environmental context,[60] and age.[7, 58, 61-64]

Long latency responses vary depending on the task performed.[54] In a study reported by Bawa and Sinkjaer (1999), the LLR of the flexor carpi radialis was found to be different during a movement tracking task of the wrist compared to an isometric contraction.[55] With prior experience, subjects are also able to scale responses accordingly.[7, 58] Winstein (2000) demonstrated that subjects were able to enhance the magnitude of anticipatory and triggered grip forces in the upper extremity based on previous experience in response to loads of predictable magnitudes.[58] Horak et al. (1989) also reported that LLRs are scaled according to the amplitude and velocity of the induced perturbation performed standing on a platform, and the magnitude of postural responses to the same stimulus was also shown to reduce with practice.[7, 56] Nardone and colleagues (1990) studied the effect of prior experience with certain surface perturbation velocities and amplitudes on long latency responses of young adults.[56] In their study, subjects were able to scale their reflex mediated muscle responses to the anticipated amplitude of the perturbation when they had immediate prior experience with the perturbation. Overshoot or undershoot of the response occurred when the perturbation velocity or amplitude changed unexpectedly; however, the magnitude of responses declined with practice.[56] While this previous work has examined the ability to adapt responses to perturbations, these have been done during static standing rather than during a more dynamic posture. Madhavan and Shields (2009), however, investigated the influence of age on the LLR by perturbing the knee during a dynamic weight-bearing exercise performed at different levels of resistance.[61] The amplitude of the responses increased with the resistance level used. Older adults scaled their LLR to perturbations at different resistances similar to younger adults, but the magnitude of these responses was almost 40% higher. Over 30% of the vastus medialis was activated within 150 ms of the perturbation, which greatly exceeded the 40% activation during the task itself. The authors suggested that excessive LLRs under these types of conditions may lead to soft tissue injury during everyday unexpected perturbations in the elderly. Therefore, quantifying muscle responses during movement patterns under pre-loaded weight-bearing conditions may assist researchers in developing innovative and quantifiable rehabilitation methods to track improvement in neuromuscular control strategies.

Influence of Vision, Vestibular, and Somatosensation

Each of the three sensory systems show increasing importance as the frequency and velocity of body sway increase.[56] Visual information is important for the control of head and trunk position in space and stabilizing the center of mass (COM), especially at fast sinusoidal surface translation frequency in the AP direction.[65] Somatosensory information about surface forces and joint motion is important for coordinating lower body motions to the oscillating support surface, especially in the absence of vision.[9] To maintain joint stability, automatic postural responses and LLR to surface translations are triggered by somatosensory information,[66] and they are scaled to the velocity and amplitude of the platform translation.[56, 66-67] This suggests that vision and vestibular information contributes to controlling the head and trunk position in space in a "top-down" manner, and that leg and foot somatosensory information may be used to control the lower body and to modify the trunk motion in a "bottom-up manner."[68-69] In this model, postural control consists of a proprioceptive loop (bottom up) to stabilize body motion relative to the support surface as well as a vestibular loop (top down) for stabilizing the trunk in space. On flat, firm surfaces, somatosensory information dominates in providing information to control postural adjustments (bottom up), while on unstable or moving surfaces, the vestibular system (top down) provides the most useful information to control postural adjustments.[70]

Whenever a sensory source is diminished (i.e., due to pathology or a change in environmental conditions), a corresponding increase in body sway is predicted, because the estimate of body dynamics is now less accurate.[10, 71-72] Thus, flexible balance control requires a continual updating of sensory weights to current conditions so that muscular commands are based on the most precise and reliable sensory information available. An inherent advantage of having at least three sensory sources available (visual, vestibular, and somatosensory) for posture is that as one sensory source is weighted down, the weighting of an alternative source can be increased to maintain a relatively constant sway level. Such a reweighting mechanism has been considered a crucial component of postural control ever since it was first suggested by Nashner and colleagues more than 30 years ago.[8, 20, 57] The responsiveness to vestibular signals appears to go up whenever somatosensory information from surface contact regarding body orientation in space is absent or uncertain[17, 20, 73] When both visual and somatosensory feedback is altered, subjects with BVL typically lose their balance.[20] The picture is one of a control process in which information from the three sensory channels is dynamically weighted to regulate balance. Each channel has direct access to the balance control process such that a perturbation delivered to one channel will always produce a response. However, the response is computed with reference to the current information available from the other sensory systems. In particular, the gain of a particular input-output relationship is updated as a function of the amount of information available in the other channels.[17]

Despite this knowledge, the contribution of vision to the long latency response during dynamic weight-bearing conditions remains largely unexplored. Using GVS, Britton (1993) and Welgampola (2001) each studied these reflexes in stance conditions with eyes open and closed, and found decreased responses as long as vision was present in healthy adults.[50, 74] Timmann et al. (1994) investigated the contribution of visual input in healthy adults to stabilization after sudden postural disturbances with fast transient platform movements and found that when the eyes were closed, there was a decrease in the latency of the triggered response and an increase in the integrated electromyographic activity of this response. This suggests that leg muscle responses are modulated according to the availability of visual input.[75] In 1985, Allum and Pfaltz examined the differences between the stabilizing reactions in the ankle muscles of normals and subjects with BVL.[76] Subjects with BVL consistently demonstrated smaller responses than normals with the eyes closed during perturbed ankle rotations, which resulted in more frequent falls during the task.

Although studies such as these have assessed the role of vision and proprioception during postural tasks, few have investigated the contribution in tasks that necessitate accurate performance under loaded weight-bearing conditions. Functional daily activities, such as gait and stair negotiation, require accuracy for success. Previous work in this lab has assessed the LLR to perturbations of the knee in weight bearing during a visually guided lower limb target matching task.[77-80] In 2007, this work was extended to include examination of the LLR when the task was performed by healthy females as well as those with anterior cruciate ligament reconstruction (ACLR) in the absence of vision. The absence of visual feedback, however, did not significantly influence the responses in this study.[81] To more completely understand the hypothesized vestibular contributions to the LLR, it is also necessary to examine the responses to knee perturbations in this dynamic weight-bearing condition when the task is performed without vision when vestibular information is also absent.

Single Leg Squat Exercise

Task specific, weight-bearing exercises are advocated over non weight-bearing exercises because these mimic functional activities such as moving from sit to stand, walking, and stair climbing, which must be performed each day in many different environments.[82-84] In addition, weight-bearing exercises are thought to be beneficial by promoting increased stability through lower extremity joint compression and muscle coactivation;[84] these encourage cocontraction between the quadriceps and hamstring muscles, which increases joint stiffness and contributes to joint stability.[82, 84]

One exercise that meets the above criteria and is prescribed for improving neuromuscular control in the rehabilitation of the LE is the single limb squat (SLS) exercise.[83, 85-86] This is performed by having a subject assume a unilateral stance position near the edge of a step. The subject begins with the hip and knee of the stance limb in full extension, then slowly lowers the body COM into hip and knee flexion while the opposite limb is unsupported. The subject then returns to the initial position by extending the hip and knee.

Shields and Madhavan used a custom device to assess muscle activation strategies during the SLS exercise providing bi-directional resistance to knee motion.[77]As the resistance to knee motion increased, biceps femoris activity increased significantly. In a subsequent study, the rate and excursion of the resisted SLS exercise was controlled by instructing subjects to track a sinusoidal target with knee displacement.[78] Young, healthy subjects demonstrated improved accuracy of performance within several sets of practice, which resulted in a reduction in the activity of hamstrings and increased activation of the quadriceps. Improvements in accuracy with this task were accompanied by a decrease in coactivation of selected musculature around the knee.[78] Ballantyne (2009) demonstrated that fatigue of the quadriceps reduced task accuracy and resulted in an increased activation of the quadriceps.[80] The amplitude of the long latency response of the vastus lateralis was also increased with quadriceps muscle fatigue.

In the previous studies using the device, a novel method was used to assess the strategies required to control the knee during higher resistance conditions under the guidance of visual feedback. Now, studies will be extended into non-visual conditions and on a compliant surface to examine changes in the LLR in varying conditions, with training, and include patients that have bilateral vestibular loss.

Purpose

This study aims to advance our understanding of neuromuscular control of the knee during the performance of a novel, dynamic, functional lower extremity task in adults with and without bilateral vestibular loss. The specific aims of this study were to (1) assess the contribution of visual feedback and surface type on accuracy, retention, and the associated muscle activation strategies (feedforward, feedback, volitional responses) and the LLR of the knee joint in response to unexpected perturbations in healthy adults; and (2) to compare the accuracy, retention, and the associated muscle activation strategies (feedforward, feedback, volitional responses) and the LLR of the knee during a weight-bearing task under different conditions of visual feedback and surface type in response to unexpected perturbations in individuals with and without bilateral vestibular loss. Male and female subjects with healthy vestibular systems and those with bilateral vestibular loss were recruited to perform a weight-bearing task that involved tracking a sinusoidal target under three conditions of visual feedback (eyes open following sinusoidal target, eyes open/no target, and eyes closed). As an equal number of males and females were recruited, testing was performed to determine if gender influenced performance or learning of this SLS task in any of the visual or surface conditions. Subjects performed the SLS exercise in a custom mechanical device.[77, 81] The amount of resistance applied to the knee joint and the rate and amplitude of knee excursion during the SLS exercise were monitored. The main task of the subjects was to match knee displacement to the sinusoidal target projected on a screen in front of them during the first condition with eyes open and to maintain accuracy to the best of their ability with unexpected perturbations caused by a sudden drop in resistance. This target feedback was not available during the second condition with eyes open or during the eyes closed condition. Subjects were provided some initial training with this task under each condition of feedback without perturbations. Matching the knee displacement to a target not only controlled the rate of movement and joint excursion but also provided feedback regarding the accuracy of performance. The accuracy of performance and electromyographic (EMG) responses of the knee during the weight-bearing exercise with and without visual feedback and in response to the unexpected perturbations were investigated.

Specific Aims

This study addressed the following broad specific aims in Chapters II and III.

Specific Aim 1

a. To compare the accuracy of performance (learning) of the perturbed SLS task, over a 2-day period, under visual and non-visual feedback conditions, and on firm and foam surfaces in healthy adults.

b. To compare the muscle activation patterns (feedforward, feedback, and volitional responses) and long latency responses (quadriceps and hamstrings) during the perturbed SLS task over 2 days under visual and non-visual feedback conditions, and on firm and foam surfaces in healthy adults.

Specific Aim 2

a. To compare the accuracy of performance (learning) of the perturbed SLS task, over a 2-day period, under visual and non-visual feedback conditions, and on firm and foam surfaces in healthy adults and those with BVL.

b. To compare the muscle activation patterns (feedforward, feedback, volitional responses) and long latency responses (quadriceps and hamstring muscle activity) during the perturbed SLS task under visual and non-visual feedback conditions, and on firm and foam surfaces in individuals with and without bilateral vestibular loss.

Hypotheses

Hypothesis 1a

Subjects will demonstrate learning and retention of the perturbed SLS task under both visual and non-visual feedback conditions. However, accuracy of tracking the target during the eyes open/no template and eyes closed conditions will be lower in these conditions than in the eyes open condition.

Rationale

Practice will result in improved accuracy of performance for both the nonperturbed and perturbed events, as well as increased efficiency of muscle responses in the healthy group as practice or prior experience with a postural task influences EMG output.[7, 62] Previous studies in this lab have demonstrated learning of the lower extremity weight-bearing exercise within 5 sets of 10 repetitions of the exercise with visual feedback and knowledge of results.[77] In addition, subjects can learn (as demonstrated through decreased error) the nonperturbed SLS task in a period of 2 days.[79] However, the ability to decrease error of the perturbed trials over 2 days has not been examined. This will be the first study to do so.

When vision is present and subjects are able to track the sinusoidal signal, accuracy will be best, as visual feedback provides important information about error and contributes to correction of error to improve motor performance.[61, 87-89] When visual feedback of the screen is not available, or if vision is absent, an open loop control is used to plan limb trajectory without online updates, which results in increased error.[89] This is consistent with results from a previous study in our lab which demonstrated that accuracy of tracking the target with the eyes closed was significantly less in young, healthy females and those with ACL reconstruction.[81] However, these studies have not examined the learning of this task under an eyes open/no template condition and performance while standing on a compliant surface, nor have we tested training and retention with unexpected perturbations. The ability to demonstrate retention of skilled behavior after a period of practice provides a true index of motor learning.[90]

<u>Hypothesis 1b</u>

Quadriceps activity will increase and hamstring activity will decrease in all conditions of feedback as learning occurs. However, quadriceps and hamstring muscle activity will each be higher in the no template and eyes closed conditions than during the eyes open condition. As accuracy of performance improves, there will be a concomitant decrease in the LLR with training.

Rationale

Madhavan and Shields demonstrated that improved performance of a skilled lower extremity weight-bearing exercise is accompanied by an increase in the quadriceps and decrease in hamstrings muscle activity, suggesting that decreased "stiffness" of the knee is necessary to achieve accuracy.[78-79] Increased efficiency of muscle responses is expected in this study as well, as practice or prior experience with a postural task influences EMG output.[7, 62] It is anticipated that the absence of visual input will be accompanied by greater quadriceps and hamstrings coactivation. The magnitude of postural responses to the same stimulus is shown to reduce with practice.[7] Taube et al. (2007) examined the LLR in the soleus and noted that it decreased with training of platform perturbations.[91] The authors suggested that since only changes in cortical excitability were correlated with improved stance stability, this indicated that supraspinal rather than spinal mechanisms are responsible for the postural improvement. This has not been examined, however, with perturbations during a dynamic weight-bearing task in which accuracy is the goal. This is the first study to examine the effect of training on the LLR during the SLS task.

<u>Hypothesis 2a</u>

Both groups will show equivalent error in the eyes open condition. However, performance error during the eyes closed condition will be greater in the BVL group.

Rationale

The preservation of accurate perception of body orientation despite loss of vestibular function is based on somatosensory, proprioceptive, efferent, and visual signals. Practice will result in improved accuracy of performance and increased efficiency of muscle responses in both groups as practice or prior experience with a postural task influences EMG output.[7, 62] Visual and somatosensory information can compensate, in part, for vestibular loss, allowing individuals with BVL to perform complex motor tasks.[9, 73] It can be expected that accuracy during the SLS task will deteriorate for both groups when visual input is removed. However, with eyes closed, subjects with BVL will have greater difficulty performing this task. By removing vision, subjects with BVL will have only proprioceptive sensory information available to perform this task, while healthy subjects will have vestibular and proprioceptive input. This will become increasingly difficult when standing on the foam cushion as subjects will then have ambiguous surface sensory input to perform the task, with the BVL group demonstrating greater error than the healthy controls in both the visual and nonvisual conditions.

Hypothesis 2b

Muscle activation strategies used to perform the controlled SLS exercise will be different between individuals with intact vestibular systems and those with bilateral vestibular loss. Individuals with BVL will exhibit reduced quadriceps activation and greater hamstrings coactivation to perform the task. Subjects with BVL will also demonstrate reduced LLR response compared to the controls, especially in the absence of vision. However, both groups will be able to adapt to the task as learning improves, as demonstrated by significant changes in the magnitude of responses with training.

Rationale

Previous studies in this lab have shown that accurate performance of the controlled SLS exercise requires an augmented quadriceps and attenuated hamstrings activation.[61, 77, 81] The vestibular system gives rise to the vestibulospinal tract which projects down through the lumbar spinal cord to assist in maintaining an upright and balanced posture by facilitating extensor motor neurons of the legs.[92] Individuals with BVL may adopt patterns of muscle activity such as decreased activation of the quadriceps and increased hamstrings activity with this task which may compromise accuracy of performance. Little is known at this time about patterns of LE muscle activation during dynamic weight-bearing tasks in this population.

With practice, as accuracy improves, the magnitude of the LLR to the same perturbation will be significantly reduced with repeated trials in both groups; however, the LLR will reduce to a greater degree in the control group. The LLR responses will be graded according to the amount of visual information available.[17] One of the properties considered essential for flexible control of upright stance is reweighting of the sensory information from the visual, vestibular, and somatosensory systems.[73] As individuals move about in the environment, sensory conditions continually change, potentially in ways that make certain sources of sensory information unreliable for the maintenance of upright stance.[73] During the experimental task, as the knee flexes against resistance, a sudden drop in resistance level will accelerate the knee joint into more flexion. This perturbation will create errors in target matching and will activate the LLR needed to stabilize the joint. However, vestibular dysfunction will affect the LLR which will result in different magnitudes of responses compared to the healthy adults. The BVL group is expected to show the same trend as healthy adults when vision is present. However, vestibular associated changes in function will result in an even lesser magnitude of response compared to the healthy subjects without vision.[50] Vision plays a large role in compensating for vestibular loss. For individuals with complete BVL, absent vision may cause a further disruption of the afferent information that is critical for responding to unexpected perturbations. As a result, unexpected perturbations may lead to even more decreased responses compared to healthy controls when the eyes are closed.

CHAPTER 2
INFLUENCE OF PRACTICE, VISUAL FEEDBACK, AND SURFACE ON ACCURACY, KNEE MUSCLE ACTIVATION PATTERNS, AND LONG LATENCY RESPONSES DURING A SINGLE LEG WEIGHT BEARING TASK

Introduction

Neuromuscular control is defined as the ability to produce controlled movement through coordinated muscle activity, in anticipation of and in response to motion or loading for the purposes of maintaining functional stability.[28-29]Vision, somatosensation, and vestibular input each contribute to and interact for neuromuscular control of movement. Control of posture during quiet standing and during platform perturbations has been studied extensively, with accurate, inaccurate, and absent input from one or more of those three systems provided.[10-11, 17, 24, 51, 69, 76, 93-99] Postural sway consistently increases when vision or proprioception is absent. Vision and proprioception are important not only for maintaining stability during these perturbed postural tasks but also for skill acquisition.

Motor skill learning refers to the acquisition of a complex movement sequence over several training sessions and retention of the skill after the period of practice.[100] The muscle activation strategies used to learn a task over a period of training are impacted by vision and proprioception. Both agonist and antagonist muscles are active when learning the movement patterns, increasing coactivation or “stiffness” of the limb. This is a strategy used to maintain accuracy during the early phases of learning. With practice, the body develops more efficient methods of task performance, and coactivation decreases.[7, 62, 78, 101-102] However, joint stiffness increases when visual feedback is reduced.[78, 81, 103] Proprioceptive sensation serves an important role in acquiring and performing movement patterns.[104] When proprioception is altered or absent, it results in impaired ability to learn new motor tasks. The ability to learn and adapt muscle activation strategies with perturbations during movement, however, is a less explored area.

Training with knee perturbations while in weight-bearing positions has been advocated to enhance functional stability about the joint and promote more advanced motor control skills.[105-108] Stretch of an active muscle after a perturbation results in several bursts of reflex activity, which can be categorized into short latency (0-50ms) and long latency responses (LLR) (50 – 150 ms). It is suggested that these long latency postural responses have greater potential for modification by supraspinal neural centers, as they occur more quickly than voluntary movements but not as quickly as spinal stretch reflexes.[38, 53, 109] Training exercises involving the use of perturbing forces applied to the lower extremity while performing a single leg squat (SLS) exercise in a controlled and progressive manner may provide the neuromuscular system the opportunity to develop successful compensatory muscle activation patterns in response to unexpected and potentially destabilizing forces at the knee. Although weight-bearing exercises such as these are used for training, relatively little is known about the motor learning ability of the CNS and the associated muscle activation patterns used for the exercise in response to perturbations and training these across days.

Previous studies in this lab used a novel method to assess the strategies required to control the knee during higher resistance conditions of the perturbed SLS exercise under the guidance of visual feedback. Shields and Madhavan (2005) standardized the task by having subjects track a sinusoidal target on the computer screen with resisted knee displacement, providing measures of accuracy.[77, 81] Using perturbations during this task provides measurements of muscle activation and latency.[24, 30, 57] With practice, the magnitude of postural responses to the same stimulus is generally attenuated.[7, 91] Changes to the LLR with training have been examined for the upper and lower extremities at rest,[110-111] and with dynamic posturography,[7] but not with perturbations during a dynamic weight-bearing task in which accuracy is the goal. Reflexive activation patterns in response to sudden stress of the knee during dynamic, weight-bearing activities, and the ability to modify these responses with practice need to be studied in a controlled manner. This will provide insight into the muscle activation strategies that contribute to joint stiffness and stability during functional activities in healthy adults.

The purpose of this study was to quantify performance accuracy, muscle activation strategies, and the LLR, while performing the SLS task during 3 conditions of graded visual feedback over a 2-day period: (1) eyes open/with visual feedback of the monitor, (2) eyes open/without visual feedback (no template), and (3) eyes closed. Performance was also compared in these visual conditions while standing on a compliant surface in order to alter somatosensory information from the support surface. Performance accuracy was measured by calculating subjects' error in tracking a sinusoidal target. Accuracy of tracking the target should be graded based on the visual condition, with greatest accuracy in the eyes open/monitor feedback condition and lesser accuracy in the eyes open/no template and eyes closed conditions. It was anticipated that the subjects would demonstrate improved accuracy during the controlled SLS task over the 2-day period with learning for both nonperturbed and perturbed trials. Error was expected to be increased during perturbations on the foam surface compared to the solid surface. It was also hypothesized that the task would be accompanied by increased quadriceps and decreased hamstrings activity in the eyes open conditions as learning occurred. Greater quadriceps and hamstrings muscle activity in the eyes closed condition, however, was expected in comparison to eyes open. The LLR was expected to decrease as performance accuracy improves across each visual condition. The foam surface was not expected to affect the LLR.

Methods

Subjects

The study sample consisted of 16 healthy female and male subjects between the ages of 19 and 26 years. Selection was a sample of convenience. Inclusion criteria also included regular physical activity and the ability to climb stairs without any difficulty. Exclusion criteria included body mass index greater than 29, history of neurological deficits, musculoskeletal disorders, degenerative joint diseases, cardiovascular diseases, previous knee injury or surgery, previous fractures of the lower extremity, patellar dislocations, and past or current knee pain during activity or rest. Demographic data of the subjects are shown in Table 2.1.

The test side for the subjects was the dominant lower extremity. Prior to participation, subjects were given a brief description of the protocol and possible risks and benefits of participation, and were required to sign an informed consent statement approved by the University of Iowa's Human Subjects Review Board.

Screening Examination

All subjects completed the following questionnaires: a general medical history form, the Short Form Medical Outcome Survey (SF 36) which assesses perception of quality of life,[112] the Baecke Questionnaire of Habitual Physical Activity,[113] the Tegner Activity Rating Scale, [114] the Marx Activity Scale,[115] and the IKDC Subjective Knee Evaluation form.[116]

Balance Assessment

Assessment of static standing balance was measured with the modified clinical test of sensory interaction and balance (modCTSIB)[117] as well as the single leg standing balance test by means of a strain gauge force plate (24 inches x 15 inches).[118] For the modCTSIB, subjects were required to stand erect with feet together without moving in the center of the force plate with shoes donned, looking straight ahead as long as possible or until the trial of 30 sec was over. This was performed up to three trials, repeated with eyes closed, and then while standing with feet together on a 24- x 24-inch piece of medium density foam cushion. For single leg standing balance, the opposite leg was flexed to 90 degrees at the knee joint, legs apart, with both arms hanging relaxed at the sides. The subjects were instructed to stand as motionless as possible and were allowed to practice this position for 30 seconds before two measurements were taken.

Experimental Task

The main task of the subjects was to perform the SLS task standing on one leg in a custom mechanical device that enabled the subject's knee flexion and extension excursion to follow a sinusoidal target (0.4 Hz, T = 2500 ms) projected on a computer screen in front of them (see Figure 2.1). Resistance, which was normalized to each subject and set at 17% body weight (BW), was provided to both knee flexion and extension. Each SLS lasted for 2500 ms with equal times in flexion and extension.

While performing the SLS task, subjects are secured in a custom device which allows perturbations to occur without risk of injury. Since participants are standing and performing a dynamic, weight-bearing task, subjects experience the sensory redundancy of neural components as in daily life. It is not equivalent, however, to the unilateral stance balance test. Pilot data support that body sway (center of pressure) during unilateral stance with eyes closed is reduced nearly 80% when subjects are attached to the weight-bearing tracking system designed for this study. The degrees of freedom when attached to this instrumentation are reduced because only sagittal plane motion is permitted. Accordingly, the limited degrees of freedom permit the safe delivery of perturbations in weight bearing in an intact system (visual, vestibular, somatosensory integration) without an emphasis on maintaining balance. [61] This SLS task provides a novel approach not only to train the ability to effectively respond to perturbations of the knee but also to emphasize accuracy of performance. With this method, we can assess certain integrated responses of vestibular, proprioceptive, and vision under a preloaded condition in a safe environment.

SLS Exercise Instrumentation

Subjects performed the resisted and controlled SLS task in a lower extremity perturbation device that has been described previously (Figure 2.1).[77-78] The device consisted of a modified standing frame with a rack and pinion gearbox mounted to the frame between the two side supports. At one end of the horizontal shaft of the gearbox was a padded vertical plate, the height of which was adjusted via a spring-loaded mechanism. During the SLS exercise, the plate was positioned against the anterior aspect of the knee with the patella located at the center of the pad. The lower extremity was secured to the pad by a velcro strap that extended around the popliteal fossa. The knee pad was spring loaded to prevent shearing as the subject performed the task. As the SLS was performed, knee flexion and extension was translated into horizontal linear displacement of the shaft of the gear system. The horizontal position of the knee joint was measured by a precision potentiometer mounted to the shaft of the pinion gear of the device. In a previous substudy of subjects performed in this lab,[81] kinematic data of the lower extremity during the SLS task were analyzed using a video motion analysis system. The angular motion and velocity of the knee was found to have an excellent correlation ($r = 0.99$) with the linear displacement of the horizontal shaft.

An electromagnetic braking system that was mounted to the shaft of the pinion gear provided the resistance during the SLS exercise. Resistance level was controlled by a custom software program that provided a constant current to the brake through the analog output channel of the computer’s A-D board. The brake had a resistance range of 0 – 45 pounds which remained constant as the horizontal shaft of the device was displaced forward and backward during the exercise, thus providing resistance to both the flexion and extension phases of the SLS task. The software that controlled the braking system also allowed for the instantaneous release of the resistance when an event marker was received at the digital input of the computer’s A-D board. This provided the desired perturbations. The user could specify the duration of the release of resistance at the time that the program was initiated.

The timing of the drop in resistance level was dependent upon the position of the horizontal shaft of the device. Output of the potentiometer was cabled to a Schmitt trigger set to produce an event marker (5 V square-wave TTL pulse, duration = 30 ms) whenever a threshold voltage of the potentiometer was exceeded. This threshold was adjusted so that the release of resistance occurred at a consistent point in the range of motion. The duration of the perturbation lasted for 300 ms and occurred at approximately one-third of the distance into knee flexion (about 400 ms). Following the period of release, the resistance of the brake returned immediately to the previously specified level.

Perturbations (release of the brake) were delivered during 2 of the 10 repetitions within each set of the SLS exercise. To determine the repetitions during which a release of resistance occurred, the computer program generated two random numbers between 1 and 10 prior to each repetition of the program. The brake was then automatically released to produce a perturbation during the exercise, during those repetitions corresponding to the two random numbers. Resistance was maintained at a constant level throughout the rest of the exercise, with no perturbations occurring during the other 8 repetitions.

Light Touch Force

The amount of resistance acting horizontally at the knee during the SLS task is quite substantial and necessitates some light touch support, so subjects were permitted to place one finger for support on a load sensor (Wafer Load cell, Model 872, Loadstar Sensor Inc) mounted on the side of the device contralateral to the test leg. The output from the load cell was passed through a differential amplifier into a voltage-controlled oscillator and then amplified with an audio amplifier to produce a sound signal that was delivered to the subjects via headphones. This device provided a progressive auditory warning as the applied force increased, peaking when it exceeded 5 N. The amount of touch force was found to be similar between males and females over days with testing and across conditions, and ranged from 0 – 3 N across subjects; however, one male subject’s peak amplitude was consistently between 5 and 6 N.

Electromyography Recordings

Bipolar silver-silver chloride surface electromyography (EMG) electrodes (8 mm in diameter, fixed inter-contact distance of 20 mm) with internal pre-amplification (gain*35) were used to record the activity of vastus medialis oblique (VM), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH), and lateral hamstrings (LH) during the SLS task. The gain of the EMG signals was also further adjustable at the main amplifier (gain*10,000). This amplifier used a high impedance circuit with a common mode rejection ratio of 87 dB at 60 Hz and a bandwidth of 15 – 4000 Hz. EMG electrodes for the quadriceps were placed at 4/5 the distance along a line from the anterior superior iliac spine (ASIS) to the medial joint line for VM, at 1/2 the distance along a line from the ASIS to the superior pole of the patella for RF, and at 2/3 the distance along a line from the ASIS to the lateral joint line for VL.[119] For the hamstring muscles, EMG electrodes were placed at 50% of the distance along a line from the ischial tuberosity to the medial and lateral femoral condyles for the MH and LH, respectively.[80]

Maximum Voluntary Isometric Contractions

A dynamometer was used to obtain maximum voluntary isometric contractions (MVICs) of each muscle to normalize EMG data. Subjects were seated on the chair with hip and knee in 90 degrees of flexion. The pelvis, hip, thigh, and foot were firmly secured to minimize other movement. Three MVICs of the quadriceps and hamstrings were obtained in this position. No significant difference in the torque during the quadriceps and hamstrings MVIC was seen, suggesting that the male and female groups generated similar MVICs.

Experimental Setup

The SLS task was performed with the subject in unilateral stance on a strain gauge force plate (24 inches x 15 inches), with the other leg unsupported and flexed. The subject was positioned with the supporting foot placed on the center of the force plate. The foot was then rotated slightly outwards so that the great toe was approximately 5 cm away from the center of the step. Foot position was traced for each subject and used to ensure that the foot remained in the same position throughout the experiment and across days. Subjects were permitted to place one finger of the contralateral hand on the load sensor mounted on the contralateral side of the device. Subjects were instructed to place very little load through their fingers, with an auditory warning given through the headphones if the force exceeded 5N. The knee pad on the horizontal shaft was aligned with the patella and the knee secured to the pad with straps around the popliteal fossa. Subjects were instructed to follow, as accurately as possible, a sinusoidal target (0.4 Hz) projected on a 17-inch computer screen placed approximately 40 cm in front of them. The real time visual display of the output of the potentiometer also indicated the horizontal displacement of the shaft of the exercise device, providing feedback regarding the subject's ability to accurately follow the target. The peak-to-peak amplitude of this sine wave form corresponded to 15 cm of linear displacement of the horizontal shaft. Previous work has shown that this 15 cm displacement corresponds to approximately 30 degrees of knee flexion.[80]

Experimental Protocol

Training Session (Day 1)

Each subject attended two sessions separated by 24-48 hours. The first portion of the first session allowed the subjects to become familiar with the device and to ensure that the SLS task was performed correctly. Before the start of the training session, subjects performed a standardized warm-up protocol on an exercise bike for 5 minutes. After completing the warm up, bipolar surface EMG electrodes were attached to the skin overlying the VM, RF, VL, MH and LH muscles with double-sided adhesive washers and secured with adhesive tape. Subjects were then positioned to perform MVICs of knee extension and flexion to normalize the EMG signals obtained during the SLS task. Subjects performed three maximum isometric contractions in extension followed by three maximum contractions in flexion while seated with the hip and knee in 90 degrees of flexion. Each contraction was held for 3 seconds with a 1-minute rest between contractions. The trial with the highest recorded peak EMG was used to normalize the activity of each muscle during the resisted SLS task.

During the initial portion of the training session, the subjects learned to follow the target pattern seen on the computer screen with knee displacement and the resistance of the device set to 12% of body weight. Subjects were instructed to avoid leaning or rotating during the task and were given verbal cues if there were any deviations in the technique or form during the learning sessions. Ten repetitions were considered one set. The SLS task was first performed with eyes open (EO) for 5 sets of 10 repetitions until they were accustomed to the task. Subjects who did not reach the minimum criterion for learning (within 5 cm of endpoint error for the EO condition) at the end of training were considered ineligible for the testing sessions. All subjects recruited met the minimum criterion for learning.

Following the initial training, testing was performed with the device resistance set to 17% body weight under three conditions of visual feedback: (1) EO, (2) EO/No Template (NT), and (3) eyes closed (EC). In the EO condition, the subject obtained continuous visual feedback of knee displacement and the sinusoidal target on the screen. In the NT condition, the subject kept the eyes open but the template was turned off, so no visual feedback regarding knee position was provided. In the EC condition, the subject kept the eyes closed during performance of the task. One-minute rest intervals separated each set of 10 repetitions to avoid fatigue during the training session. After every set, subjects were asked to report their rating of perceived exertion (RPE) of their quadriceps on the Borg Scale (Borg, 1982). This scale is found to be sensitive to perceived levels of exertion in isolated muscle.[120-121] If subjects perceived any fatigue of the quadriceps, they took additional time to rest. Reported RPE averaged 11 (on 0-20 scale) which was consistent with subjects reporting they were not fatigued. After the initial 5 sets were performed with EO, the remainder of the session consisted of sets of 10 repetitions of the

SLS exercise performed in the following order: 1 set EO→ 1 set NT→ 1 set EC→ 1 set EO →2 sets EC →1 set EO →2 sets NT – 1 set EC – 1 set EO, 1 set NT. EC and NT trials were interspersed between EO trials to help acquisition and consolidation of learning.

Perturbations were delivered during 20% of the repetitions. The threshold level of the Schmitt trigger was adjusted such that the perturbations occurred when the horizontal shaft of the exercise device translated 5 cm from the start of the flexion phase of the exercise. When the perturbations were encountered, subjects were instructed to react as quickly as possible to restore the knee to its original trajectory and minimize the error between the target waveform and the actual knee position trace.

Testing Session (Day 2)

The testing on Day 2 was conducted 24-48 hours after Day 1. Prior to data collection, subjects performed a standardized warm-up protocol on an exercise bike for 5 minutes. After completing the warm up, bipolar surface EMG electrodes were attached and MVICs were completed as described earlier.

After the obtaining the MVICs, the subjects were placed in the exercise device in the same manner as for the previous session. Subjects then performed a series of the SLS task in the EO, NT, and EC conditions. Two sets of 10 repetitions for each condition were performed. A 1-minute rest interval separated each set of 10 repetitions. The order of retention trials was as follows: 1 set EO→ 1 set NT - 1 set EC→ 1 set EO →I set EC - 1 set NT. Subjects then performed the trials while standing on the foam cushion in the following order: 2 sets EO at 12% resistance (no perturbations); the remainder at 17% resistance with unexpected perturbations - 1 set EO →1 set NT→1 set EC→1 set EO→1 set EC→1 set NT.

<u>Data Reduction</u>

All experimental data were collected online and subsequently analyzed using Datapac 2K2 software (version 3.14; Run Technologies Inc., CA). Electromyographic activity of the quadriceps and hamstring muscles was sampled at a rate of 2000 Hz. MVICs were analyzed by finding the peak RMS EMG during each of the three contractions and calculating the mean RMS EMG for 200 ms on either side of the peak EMG. All EMG derivates are expressed as percentage of MVIC.

All other signals (linear potentiometer, target waveform, brake and touch force) were digitized at a rate of 1000 Hz. For the modCTSIB and single leg balance assessment, movements of the center of pressure (COP) in the frontal and sagittal plane were sampled at a frequency of 500 Hz. Antero-posterior and medio-lateral amplitude of the COP was analyzed for each subject.

During both perturbed and nonperturbed repetitions of the task, a TTL pulse was recorded from the Schmitt trigger as the potentiometer attached to rack and pinion gear of the exercise device passed the threshold voltage. For perturbed trials, the TTL pulse corresponded to the onset of the perturbation. For nonperturbed trials, the TTL pulse served as a marker to indicate the point in the range of motion where the perturbation would have occurred if the software had allowed for it.

Dependant Variables

Dependant variables analyzed in this study included:

1. Absolute Error (absolute value of the difference between the target waveform and the actual position of the knee): To obtain absolute error of performance during the learning and retention trials, the sinusoidal target was subtracted from the linear knee displacement to calculate error. The error signal was then rectified and averaged in 10% bins within each flexion and extension cycle of the SLS exercise. The time frame 50-150 ms post-perturbation was analyzed as well as overall absolute error.

2. Variable Error (a measure of precision in tracking the visual target): The variable error was obtained by calculating the standard deviation about the error signal of each subject. This was also averaged in 10% bins within each flexion and extension cycle

of the SLS exercise. The time frame 50-150 ms post-perturbation was analyzed as well as overall absolute error.

3. Endpoint Error (overshoot): The endpoint error is a measure of the deviation of the endpoint of the subject's flexion from the endpoint of sine wave template to calculate the overshoots of the target during the perturbation trials.

4. Peak Velocity: The linear velocity of the knee was obtained by differentiating the displacement signal. The peak of this velocity signal was measured in the time bins 200 ms prior to perturbation and 0-200 ms and 200 – 400 ms post perturbation to examine anticipatory (Pre), reflex (Post 1), and voluntary (Post 2) phases, respectively.

5. Cycle EMG activity: RMS muscle activity of the VM, RF, VL, MH, and LH were processed with a time constant of 50 ms and then averaged in 10% bins for each flexion and extension cycle of the single leg squat task to recognize synergies that were being utilized to perform the task. All values are expressed as a percentage of MVIC.

6. Mean EMG: Muscle activity (expressed as % MVIC) in the time bins 200 ms prior to perturbation and 0-200 ms and 200 – 400 ms post perturbation was analyzed to examine anticipatory (Pre), reflex (Post 1) and voluntary (Post 2) activity, respectively. In the no perturbation trials, the same bins were analyzed with respect to the time when the perturbation could have occurred.

7. Normalized LLR activit*y*: To filter out the effect of background activity in the mean long latency response, the normalized muscle activity of the quadriceps and hamstrings muscles in the time bin 50 – 150 ms after the onset of the perturbation was calculated for each subject using the following formula:

$$\text{Normalized LLR} = \frac{\text{Mean EMG of Perturbation Trials} - \text{Mean EMG of Unperturbed Trials}}{\text{Mean EMG of Unperturbed Trials}}$$

Statistical Analysis

All data were analyzed using a three-factor repeated measures analysis of variance, and a separate analysis was performed for each of the dependant variables. Flexion and extension cycles were analyzed separately. Nonperturbed and perturbed events were also analyzed separately. The subjects recruited in this study were evenly distributed between males and females, so gender was included in the model. The three within-subject factors were Condition (EO, NT, and EC), Day (Day 1, Day 2), and Gender (Male, Female). To compare performance on firm versus foam surface, a separate three-factor repeated measures analysis of variance was performed with the following factors: Condition (EO, NT, and EC), Surface (Firm, Foam), and Gender (Male, Female).

Absolute error (AE) and variable error (VE) measures were used to quantify performance. Muscle activity during the SLS task was quantified using EMG from each of the five muscles sampled. When necessary, significant main effects and interactions were further analyzed using Bonferroni correction. The level of significance for all tests was established at $\alpha \leq 0.05$. All statistical analyses were performed using SPSS software (version 17.0).

Results

Subjects

Demographic data of the subjects are shown in Table 2.1. A t-test was performed to compare differences in descriptive variables between males and females. No significant difference was found for age. However, a significant difference between the two groups was noted for Body Weight ($p = 0.002$) and Height ($p = 0.001$). Males had greater body weight and height than females.

The Short Form 36 Medical Outcome Questionnaire was used to compare the subject's perception of general health, and IKDC and the KOOS and were used to assess knee symptoms and ability (Table 2.1, Figure 2.3). There was no difference between male and female scores in any of the categories for either test. No difference in activity level was found as assessed with the Marx, Tegner Activity, or the Baecke Habitual activity scales (Table 2.1). There was also no difference between groups in antero-posterior or medio-lateral movement amplitude of the COP with standing balance assessment on the firm or foam surface, with feet together or with single limb stance.

Effect of Vision, Surface, and Training on
Accuracy of Performance

The accuracy of tracking the target waveform was assessed during each repetition of the SLS exercise by calculating the average absolute and variable errors of both the flexion and the extension phases, as well as the endpoint error at the transition between flexion and extension phases. Comparing these errors across Days 1 and 2 as well as during eyes open (EO), no template (NT), and eyes closed (EC) on firm and foam surfaces provided information regarding learning and the effect of vision and surface, respectively. There was no effect of Gender for AE, VE, or endpoint error. In addition, the NT condition was no different than the EC condition for these variables. Hence, the results and graphs will depict all subjects pooled together with EO and EC conditions only.

Absolute Error

The mean absolute errors (AE) for the nonperturbed and perturbed events of the SLS task from Days 1 and 2 performed in each visual condition are shown in Figures 2.4-2.9. A three-way repeated measures ANOVA (Condition x Gender x Day) on the absolute errors for both types of events was performed. For the nonperturbed events, the greatest errors were seen in the mid portions of the flexion and extension phases of the task during the EO sets and in the earlier intervals for the NT and EC sets. The greatest AE during the perturbed events extended from the mid to the late portions of the flexion phase.

A significant main effect of Condition was seen during both the flexion and the extension phases of the nonperturbed events. Absolute error for the EO condition was less than both the NT and EC conditions ($p<.001$). As hypothesized, AE during EC was greater than when visual feedback was provided (Flexion – EC 4.60 cm, EO 1.43 cm, Extension – EC 4.77 cm, EO 1.37 cm). Multiple pairwise comparisons showed that error decreased significantly from Day 1 to Day 2 during the EO Condition of the flexion phase. This was not the case for the extension phase. In addition, there was no significant difference in absolute error from Day 1 to Day 2 in the NT or EC conditions for flexion or extension phases, suggesting that subjects were not able to improve performance in these conditions.

For the AE of perturbed events (Figures 2.7-2.9), a significant main effect of Condition was seen during both the flexion and the extension phases. Post-hoc analysis revealed absolute error for the EO condition was less than both the NT and the EC conditions ($p<.001$). As with the nonperturbed events, AE during the perturbed NT and EC conditions was doubled than when visual feedback was provided (Flexion – EC 5.39 cm, EO 2.23 cm, Extension – EC 5.37 cm, EO 1.79 cm). Absolute error during these perturbed trials was nearly 1.5 times the AE during the nonperturbed events. Multiple pairwise comparisons showed that with EO, Day 1 absolute error was greater than Day 2 during flexion ($p = .003$) for the females only and during extension ($p = .005$) phases for all subjects.

A three-way repeated measures ANOVA (Condition x Group x Surface) was then performed. When comparing AE on the firm versus foam surface, the trend was the same across visual conditions as on the firm surface. AE on the foam surface was greater than on the firm surface during the flexion phase ($p = .026$) of the perturbation events, with no difference seen during the extension phase of these events.

Variable Error

Variable errors showed that subjects were very consistent in the performance of the SLS task with the greatest variability in errors occurring as subjects transitioned between flexion and extension. A three-way repeated measures ANOVA (Condition x

Gender x Day) on the variable errors showed similar trends as seen for absolute error. For the nonperturbed and perturbed events and firm and foam surfaces, VE during EO was less than both NT and EC conditions ($p < .001$) during both phases, while there was no difference between NT and EC conditions. As with AE, VE during EC was doubled than during EO for nonperturbed events (Flexion – EC .39 cm, EO .16 cm, Extension – EC .43 cm, EO .17 cm) and perturbed events (Flexion – EC .52 cm, EO .32 cm, Extension – EC .22 cm, EO .45 cm). No main effect of surface was found for variable error during either flexion or extension phases during nonperturbed or perturbed events.

A three-way repeated measures ANOVA (Condition x Group x Surface) found no effect of foam when compared to the firm surface.

Endpoint Error

Endpoint error (Epe) measures the deviation of the endpoint of the subject's flexion and was determined for both the nonperturbed and perturbed events (Figure 2.10). A three-way repeated measures ANOVA (Condition x Group x Gender) was done, with no effect of Day, Condition, or Gender found. Endpoint error did not significantly change from Day 1 to Day 2, indicating subjects were unable to lessen overshoot over the testing days.

To examine the effect of the foam surface, a three-way repeated measures ANOVA (Condition x Group x Surface) was done. There was a main effect of surface for perturbed events ($p<.001$) with Epe greater on the foam surface.

<u>Effect of Training and Vision on Peak Velocity</u>

Peak linear velocity of the knee was measured within the anticipatory (Pre: -200-0 ms), reflex (Post 1: 0-200 ms), and volitional (Post 2: 200-400 ms) time bins (Figure 2.10). A three-way repeated measures ANOVA (Condition x Group x Day) on peak velocity in each of these bins showed significant differences between conditions. Again, there was no effect of Gender. For the nonperturbed events, there was a significant main effect of Condition ($p<.001$) and Day ($p = .002$) for only the volitional time bin. Peak velocity was higher in the NT and EC Conditions than in the EO and was also higher on Day 2 than on Day 1 in this time frame. No difference was seen between the NT and EC conditions over either day.

For the perturbed events, a significant main effect of Condition ($p<.001$) was seen, with peak velocity being higher in the EC ($p = .003$) conditions than in the EO conditions during the volitional time bin. No effect of Gender or Day was seen for the perturbed trials.

A three-way repeated measures ANOVA (Condition x Group x Surface) on this variable found a significant interaction effect of Surface and Condition during the Pre time bin for both the nonperturbed and perturbed events ($p = .016$, $p = .02$, respectively). During nonperturbed events, velocity was less for the solid compared to the foam condition for the EC trials ($p = .004$). For perturbed events, this was also the case for both the EO ($p = .04$) and the EC ($p = .022$) trials. At the Post 1 time bin, velocity was increased for the foam surface ($p = .003$) during the nonperturbed trials but not during the perturbed trials. There was no effect of Surface in the volitional time bin (Post 2).

Effect of Training, Vision, and Surface on Muscle Activation Patterns

Activity of the vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH), and lateral hamstrings (LH) during the SLS task was determined by averaging EMG activity in 10% interval bins of knee displacement (Figures 2.12-2.18; only perturbed events shown). Flexion and extension phases were analyzed separately and muscle activity was expressed as a percentage of maximum voluntary isometric contraction (MVIC). A three-way repeated measures ANOVA (Condition x Gender x Day) was used to examine the influence of each of these factors on muscle activity during the SLS task. For each of the muscles examined, there was no main effect of Gender.

Average muscle activity of the VM ranged from 10% to 38% MVIC with increased activity during the extension phase (Figures 2.12, 2.17). There was a significant effect of Condition during the extension phase for both the nonperturbed ($p = .002$) and the perturbed events ($p = .002$). For both groups of events, VM activity was increased in the EC condition compared to EO ($p = .009$ for nonperturbed trials; $p = .015$ perturbed). There was no difference found between the EO condition and NT or EC for the flexion phase of the nonperturbed or perturbed events. During the flexion phase for nonperturbed and perturbed events, RF activity remained relatively uniform (Figures 2.13, 2.17). Average muscle activity of RF ranged from 5% to 24% MVIC. The only significant effect found was for Day (p =. 043) during the extension phase of the nonperturbed events: RF activity was greater on Day 1 than on Day 2.

VL was the most active of the quadriceps during the SLS task for all Conditions and across both Days (Figures 2.14, 2.17), with activity ranging from 18% to 55% MVIC. For both the nonperturbed and the perturbed events in the flexion phase, there was a significant main effect of Day, with greater activity seen on Day 1 than on Day 2 ($p = .029$ nonperturbed events, $p = .03$ perturbed events). There was no main effect for Day during the extension phases for either nonperturbed or perturbed events; however, there was a main effect of Condition. VL activity during EO was less than that for EC for both groups of events ($p = .023$ nonperturbed events; $p = .003$ perturbed events).

MH was least affected by vision or training on the firm surface (Figures 2.15, 2.18). At the initiation of the flexion phase of the SLS, the mean amplitude was around 20% MVIC; otherwise, it ranged from approximately 6-12% MVIC. It was most active during flexion, but there was no main effect of Day or Condition. Similarly, LH was little affected by vision or training (Figures 2.16, 2.18). Activity ranged from 8% to 17% MVIC, with no significant main effects of Condition, Day, or Surface.

A three-way repeated measures ANOVA (Condition x Group x Surface) on this variable revealed that VM and VL each had lower EMG activity during both flexion and extension phases on the firm (p = .002) surface as compared to foam (p = .027) for perturbed events only. RF showed lower activity during EO than EC during nonperturbed (p = .006) and perturbed (p = .03) events. MH was similarly effected by foam, showing decreased activity on solid surface for the flexion phase of nonperturbed (p<.001) and perturbed (p = .005) events and during the extension phase of the nonperturbed events (p = .028).

Muscle Activity in the Anticipatory, Reflex, and Volitional Time Bins

Muscle activity of the quadriceps and hamstrings was analyzed in the Pre, Post 1, and Post 2 time bins during the perturbed and nonperturbed trials (Figures 2.19-2.20). A three-way repeated measures ANOVA (Condition x Group x Day) found no main effect of Gender for any of the muscles examined. VM showed an effect of Vision in the volitional (Post 2) time bin for both nonperturbed and perturbed events. Activity was greater with EC than with EO (p = .009 nonperturbed; p = .015 perturbed). RF showed the least effects of the quadriceps for Day or Condition in any of the time bins for either nonperturbed or perturbed events. There was no main effect noted for Condition or Day. Analysis of VL during the nonperturbed events showed an effect of Day (p = .037) during the reflex time bin (Post 1) which was not seen for the perturbed events. VL activity was greater on Day 1 than on Day2 in this bin of the nonperturbed trials; however, this effect was not seen in the perturbed trials. A similar effect of Day was also seen in the Post 2 bin for both the nonperturbed events (p = .018) and the perturbed events (p = .003). VL activity was greater on Day 1 than on Day 2. For the perturbed events in the Post 2 bin, there was also a significant main effect of Condition (p = .002). VL activity with EO was less than with EC (p = .023). As with RF, MH and LH did not have any significant main effects of Day, and no difference for EO versus EC condition for either muscle in nonperturbed or perturbed events.

A three-way repeated measures ANOVA (Condition x Group x Surface) on this variable found that on the foam surface, VM at Post 1 and Post 2 of the perturbed events showed higher activity compared to the solid surface ($p = .04$ Post 1, $p = .008$ Post 2). A similar effect of Condition was also seen for VM as when comparing Days. VM activity during EC condition was greater than EO for nonperturbed ($p = .012$) and perturbed ($p = .022$) events of the extension phase. VL showed this same effect at Post 2 only ($p = .003$). MH, however, showed this effect at Post 1 and Post 2 for the nonperturbed ($p = .043$ Post 1; $p = .014$ Post 2) and perturbed events ($p = .016$ Post 1; $p = .019$ Post 2). RF and LH were not modulated by surface during the time bins examined.

Normalized Long Latency Responses

Difference in the EMG activity between the perturbed and nonperturbed trials in the 50-150 ms time bin normalized to the background activity (nonperturbed trials) was compared between Conditions and over Days (Figure 2.21). A three-way repeated measures ANOVA (Condition x Group x Day) on the normalized long latency response revealed a significant effect of Day for VM ($p = .034$) and RF ($p = .027$), with a greater response on Day 2 than on Day 1. The response was also greater in the EC condition for VM than in the EC condition ($p = .014$). Similar to VM, MH response was increased in the EC condition compared to EO ($p = .035$). LH did not show any significant main effects or interactions. No significant effects of Surface were found for any of the muscles.

Time of Peak LLR

No significant differences in the latencies of the peak LLR of any muscles was found between males and females, in the different visual feedback conditions, and over the 2 days or surfaces. On average, the peak LLR was 116 ms for VM, 120 ms for RF, and 117 for VL. The MH peak LLR occurred at 81 ms whereas LH at around 77 ms.

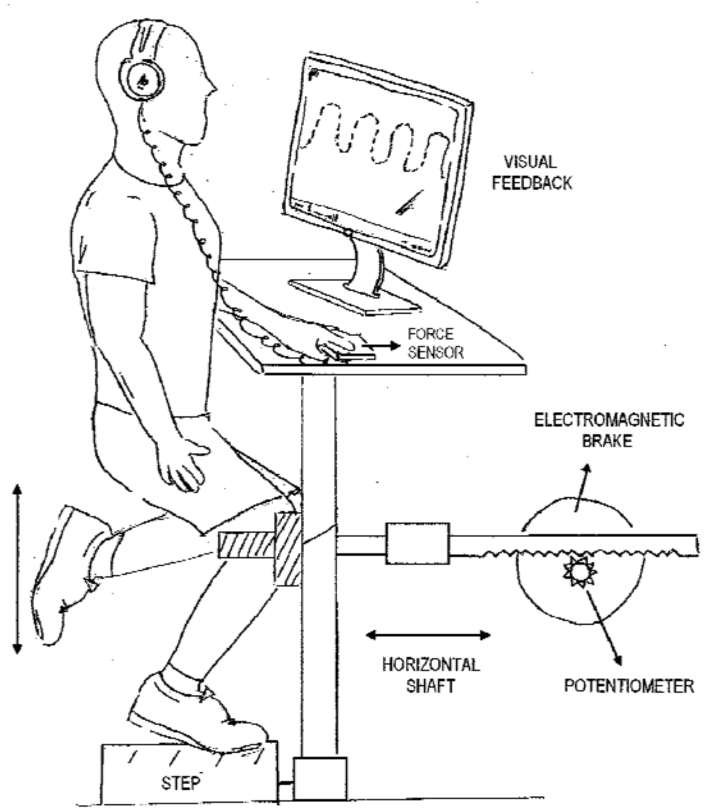

Figure 2.1. Schematic diagram of the lower extremity perturbation device

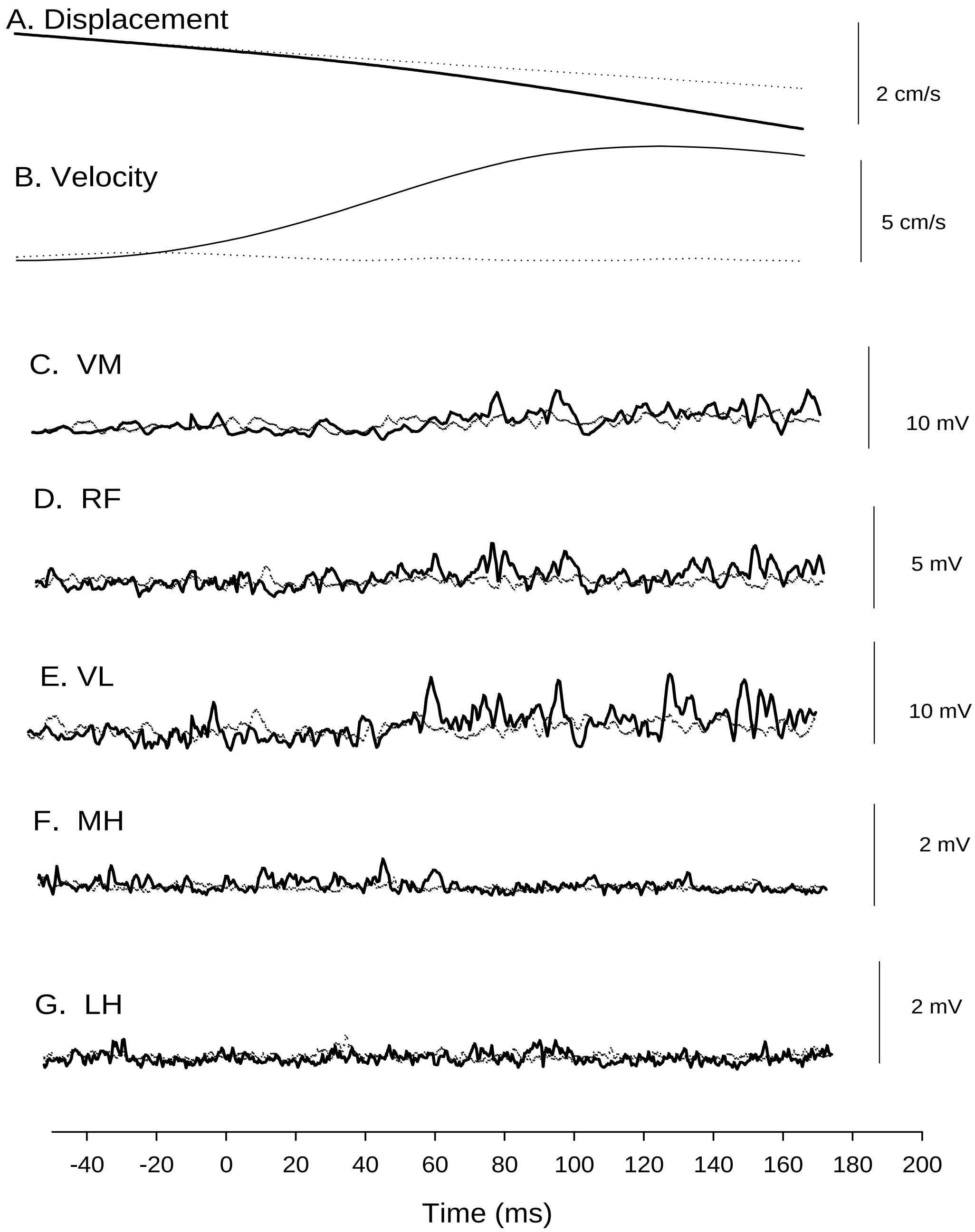

Figure 2.2 Representative example of **a**. linear displacement, **b**. linear velocity, EMG traces of **c**. vastus medialis, **d**. rectus femoris, **e**. vastus lateralis, **g**. medial hamstrings and **h**. lateral hamstrings of a single subject (average of 8 trials). *Dotted* lines represent the unperturbed trials and *solid* lines are perturbed trials. Time frame is from -50 to 200 ms post perturbation. X axis represents time (ms); the release of the brake occurred at 0 ms. EMG traces are root mean square averaged.

Table 2.1 Characteristics of male and female subjects

	Males (n=8)	**Females (n=8)**
Age (yrs)	23.6 (1.7)	22.3 (2.1)
Weight (lb)*	179.4 (24.5)	143.6 (16.1)
Height (cm)*	178.6 (6.1)	167.8 (5.4)
IKDC score	96.3 (4.9)	98.3 (2.4)
Marx Activity Scale	12.4 (3.2)	9.9 (3.3)
Tegner Activity Scale (current)	7.9 (1.8)	8.3 (1.5)
Five Time Sit to Stand (sec) (best trial)	8.2 (1.6)	9.0 (1.2)

Note: Values are Mean (SD)

*indicates significant difference between Groups

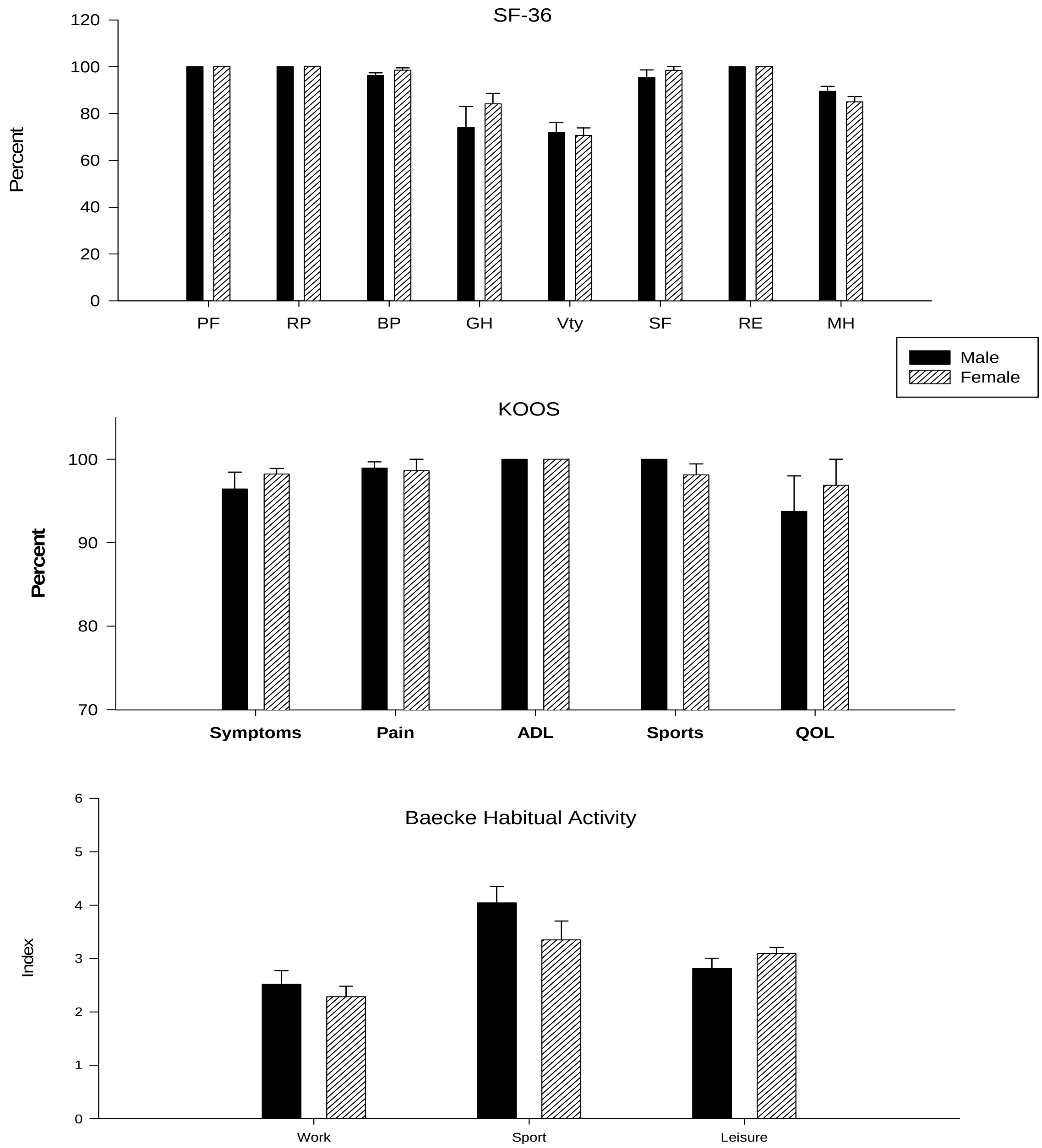

Figure 2.3 Mean SF36, KOOS, and Baecke scores of Male (*dark bars*) and Females (*striped bars*). The X-axis represents the various domains of the SF-36 and KOOS (PF – Physical Function, RP – Role Physical, BP – Bodily Pain, GH – General Health, Vty – Vitality, SF – Social Function, RE – Role Emotional, MH – Mental Health. Values are means $\pm$ SE of all 8 Males and 8 Females.

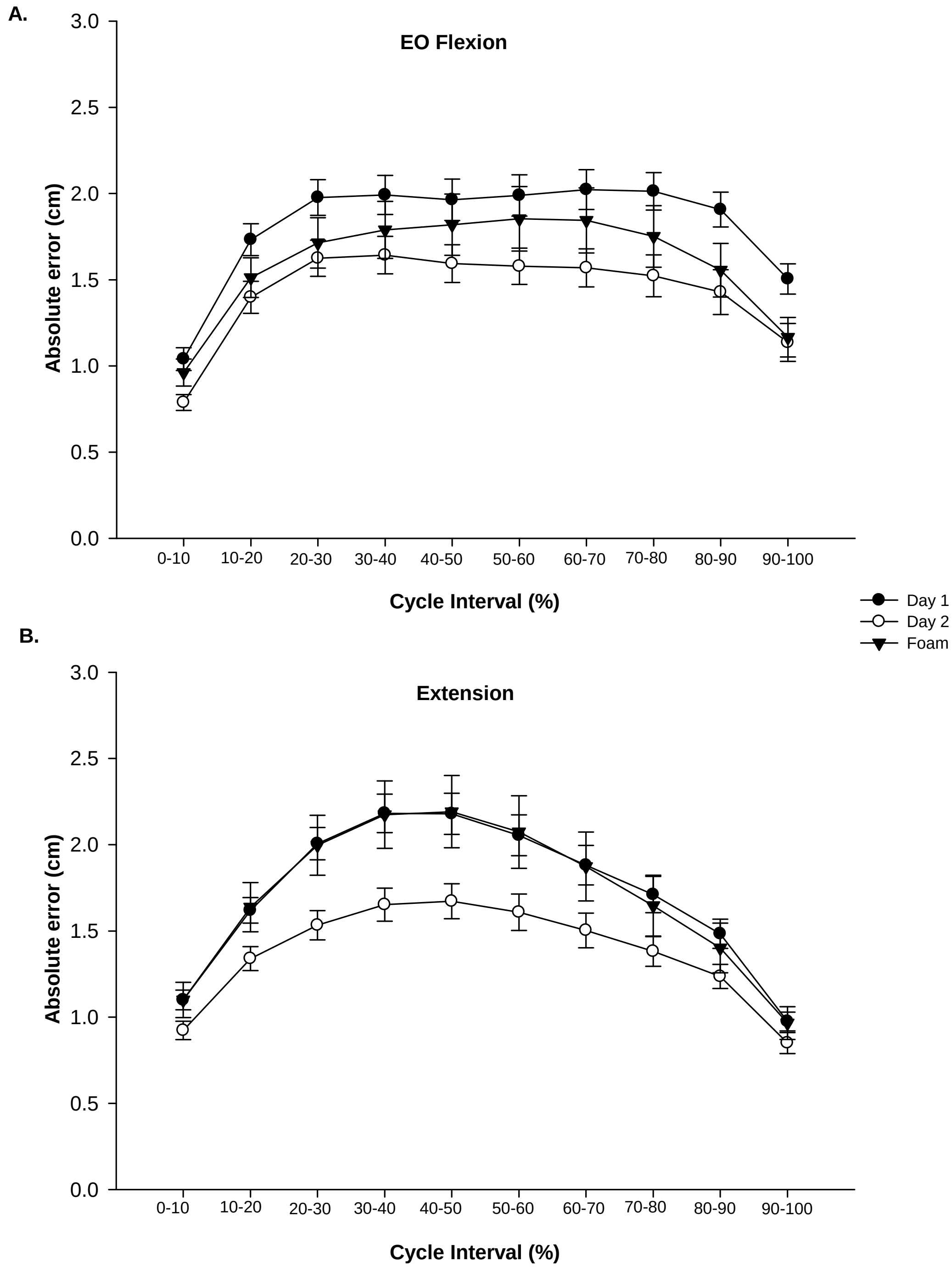

Figure 2.4 Absolute errors of performance in the Eyes Open (EO) condition No Perturbation events within the flexion (A) and extension (B) phases of the single leg squat task. Mean error values are presented for Day 1 (*dark circles*), Day 2 (*open circles*), and Day 2 - Foam (*triangles*). Values are means ± SE of all 16 subjects.

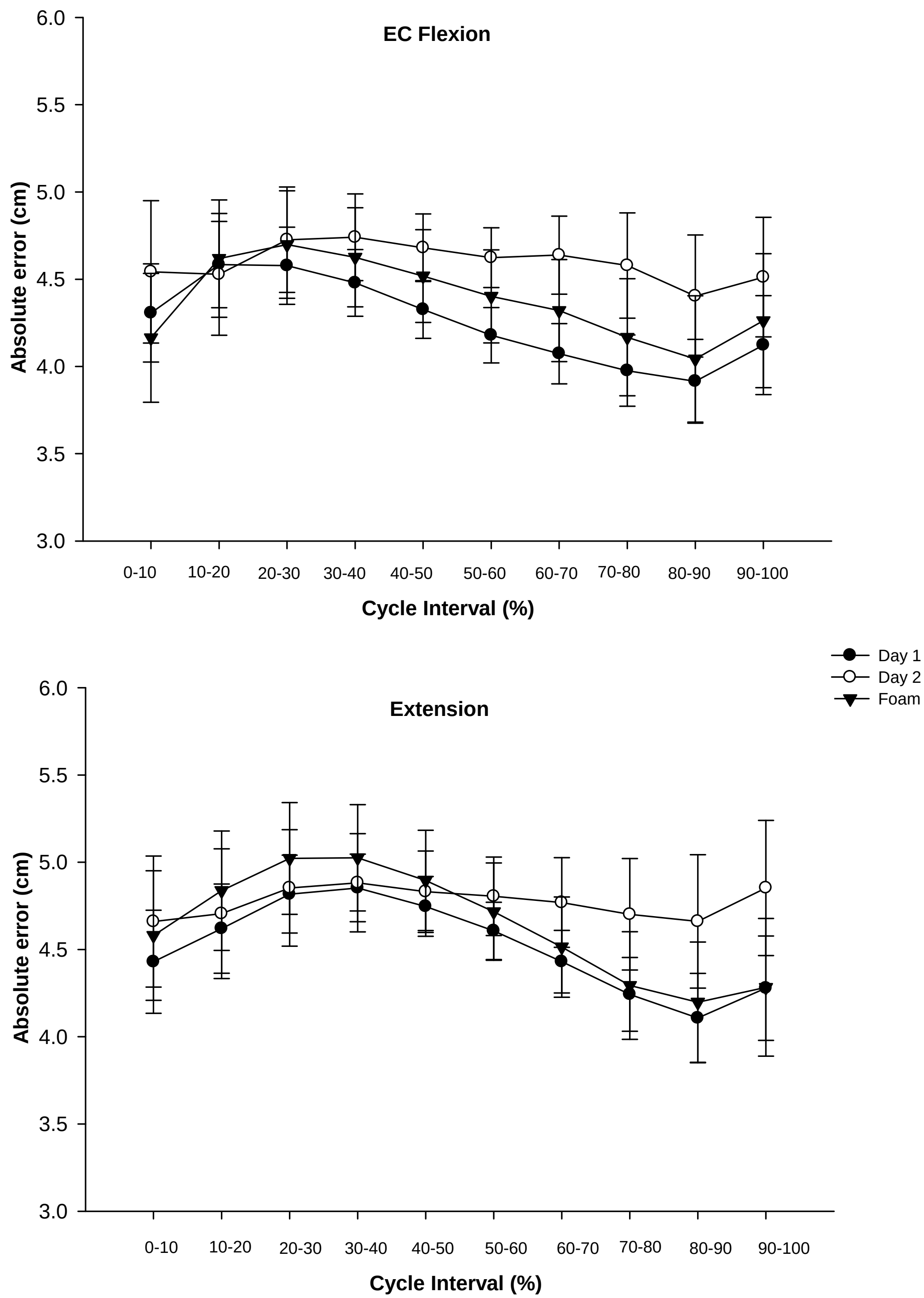

Figure 2.5 Absolute errors of performance in the Eyes Closed (EC) condition No Perturbation events within the flexion (A) and extension (B) phases of the single leg squat task. Mean error values are presented for Day 1 (*dark circles*), Day 2 (*open circles*), and Day 2 - Foam (*triangles*). Values are means ± SE of all 16 subjects.

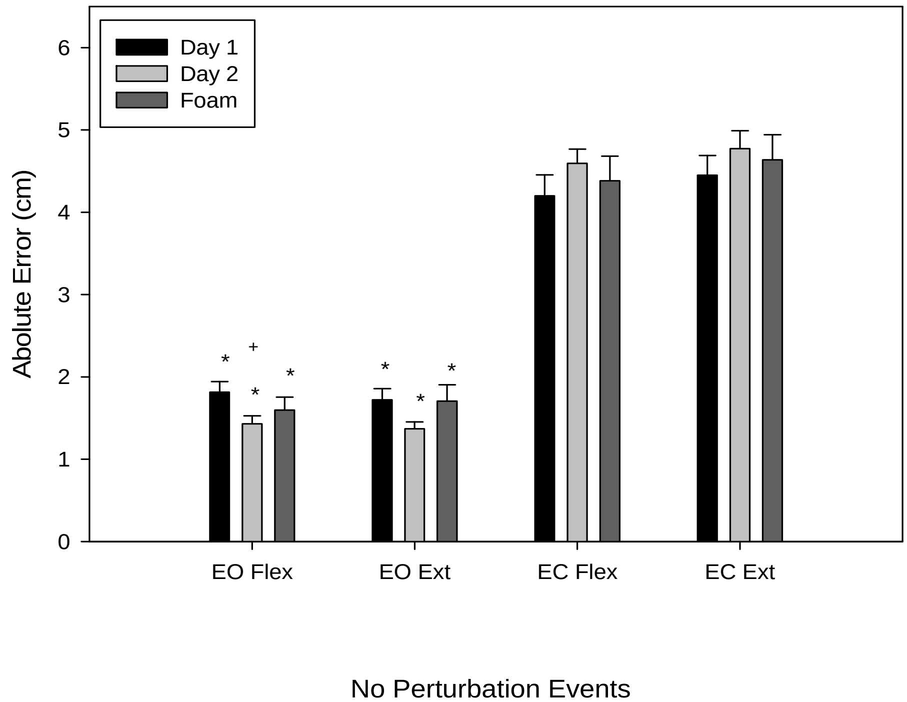

Figure 2.6 Mean absolute error during the flexion and extension phases of the SLS task in the eyes open (EO) and eyes closed (EC) conditions of the No Perturbation events. The vertical bars represent the average errors over Day 1, Day 2, and Day 2 - Foam. The Y-axis represents absolute error of performance in cm. Values are means + SE of all 16 subjects.

* indicates significant difference from Eyes Closed

+ indicates significant difference from Day 1

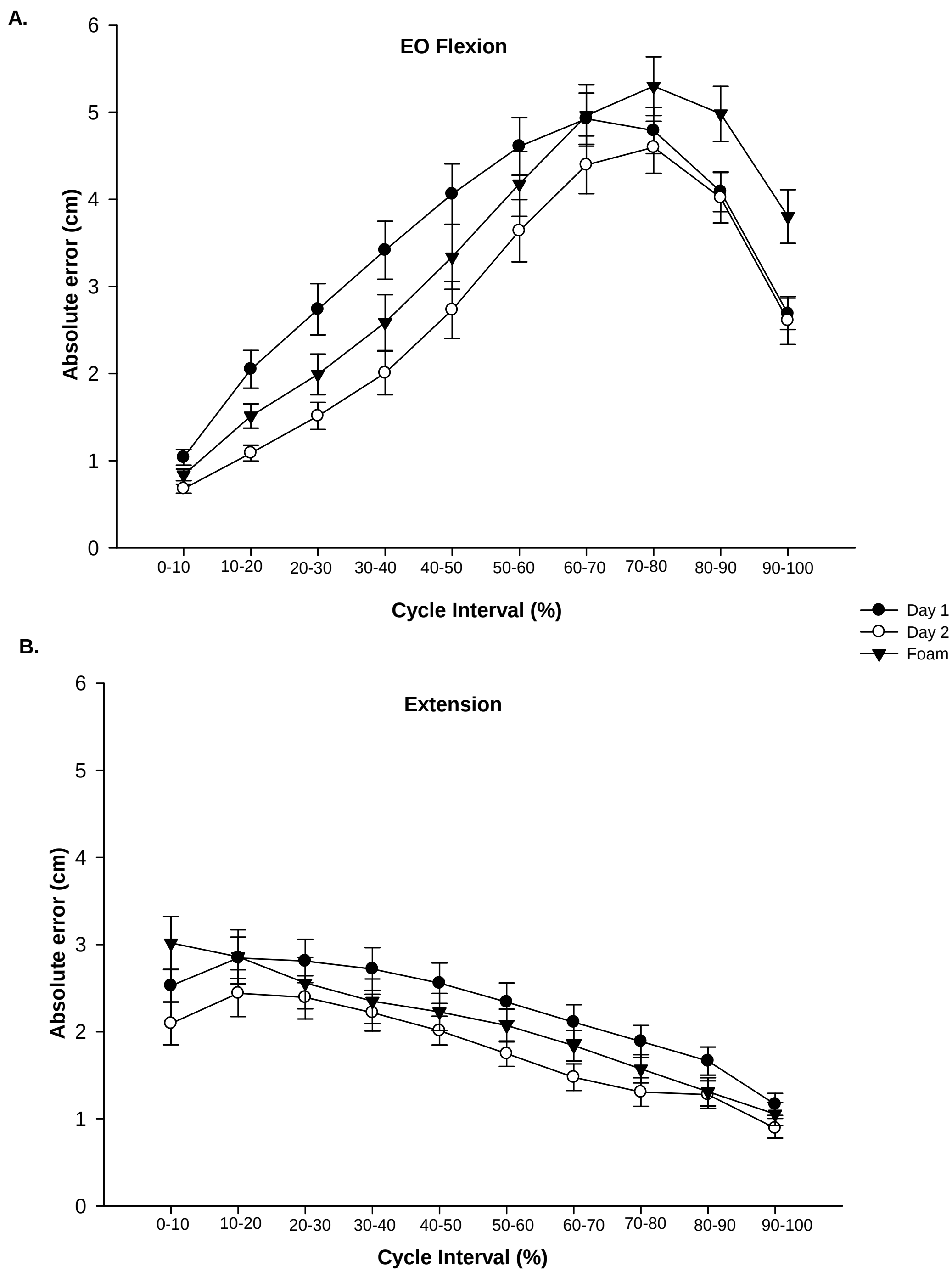

Figure 2.7 Absolute errors of performance in the Eyes Open (EO) condition Perturbation events within the flexion (A) and extension (B) phases of the single leg squat task. Mean error values are presented for Day 1 (*dark circles*), Day 2 (*open circles*), and Day 2 - Foam (*triangles*). Values are means ± SE of all 16 subjects.

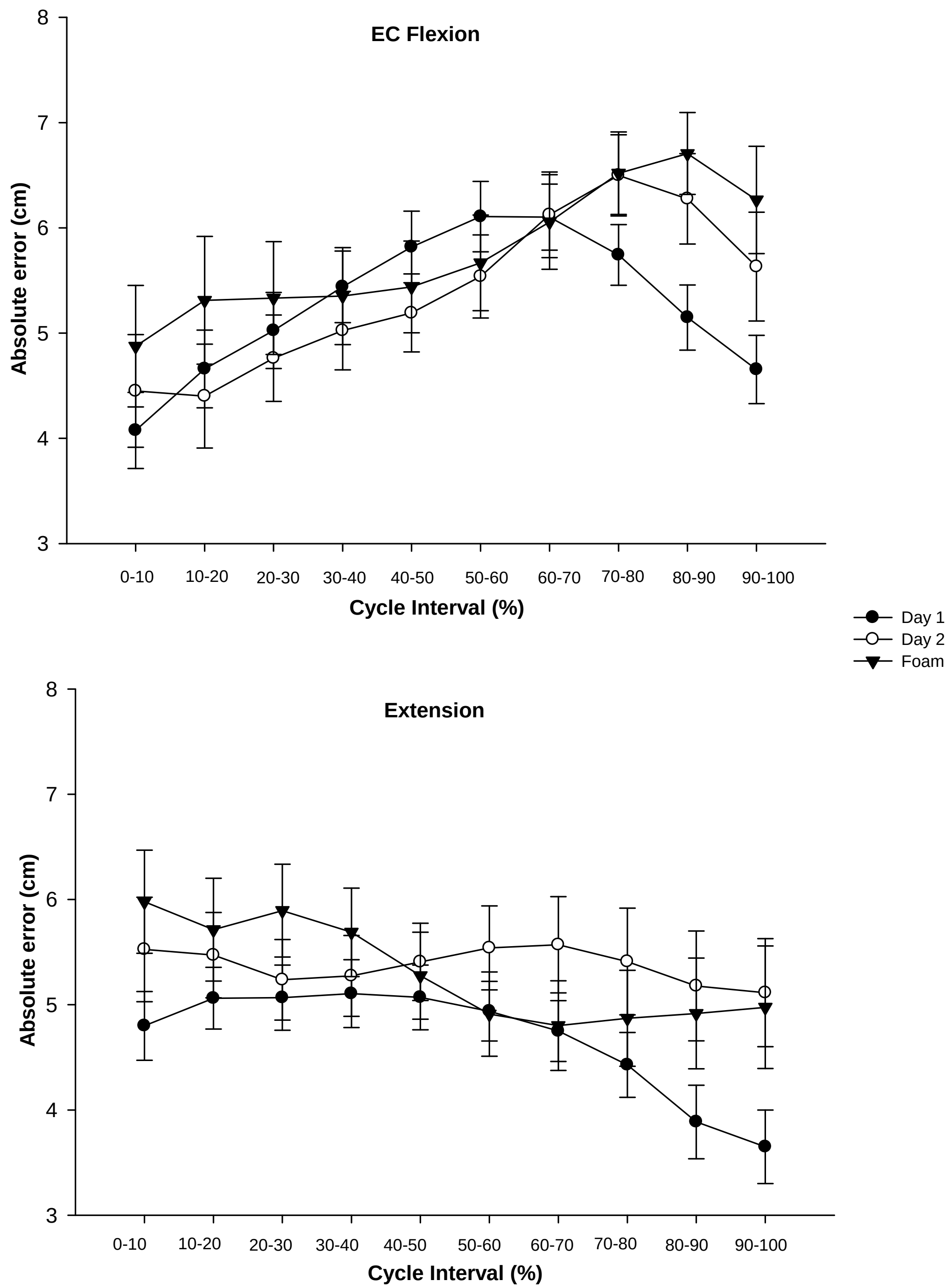

Figure 2.8 Absolute errors of performance in the Eyes Closed (EC) condition Perturbation events within the flexion (A) and extension (B) phases of the single leg squat task. Mean error values are presented for Day 1 (*dark circles*), Day 2 (*open circles*), and Day 2 - Foam (*triangles*). Values are means ± SE of all 16 subjects.

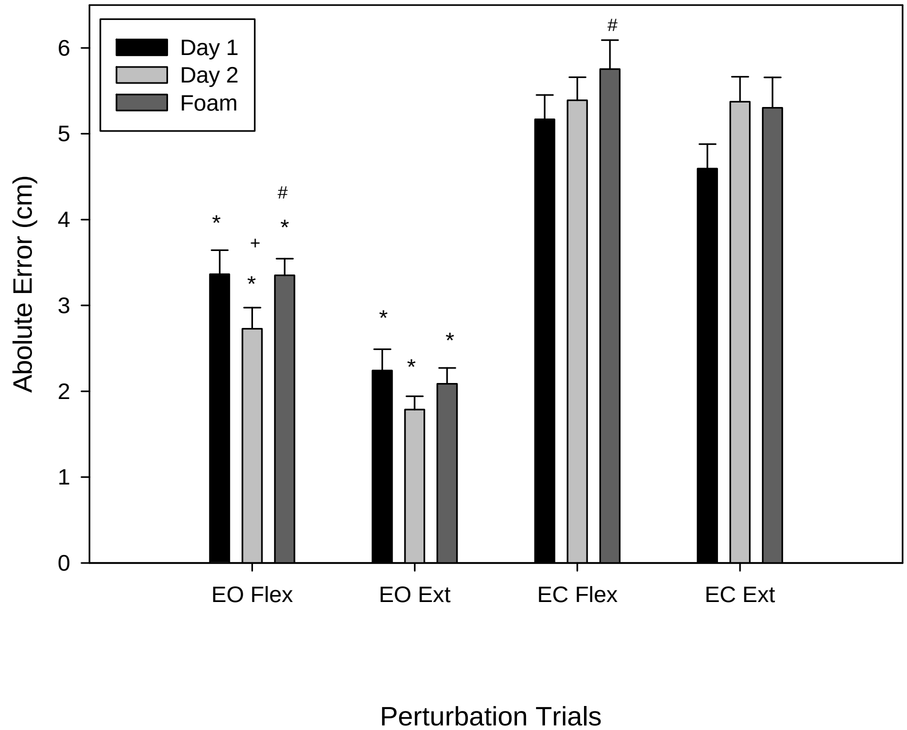

Figure 2.9 Mean absolute error during the flexion and extension phases of the SLS task in the eyes open (EO) and eyes closed (EC) conditions of the Perturbation events. The vertical bars represent the average errors over Days 1, 2, and Day 2- Foam surface. The Y-axis represents absolute error of performance in cm. Values are means ± SE of all 16 subjects.

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* indicates significant difference from Eyes Closed

\+ indicates significant difference from Day 1

\# indicates significance difference from firm surface

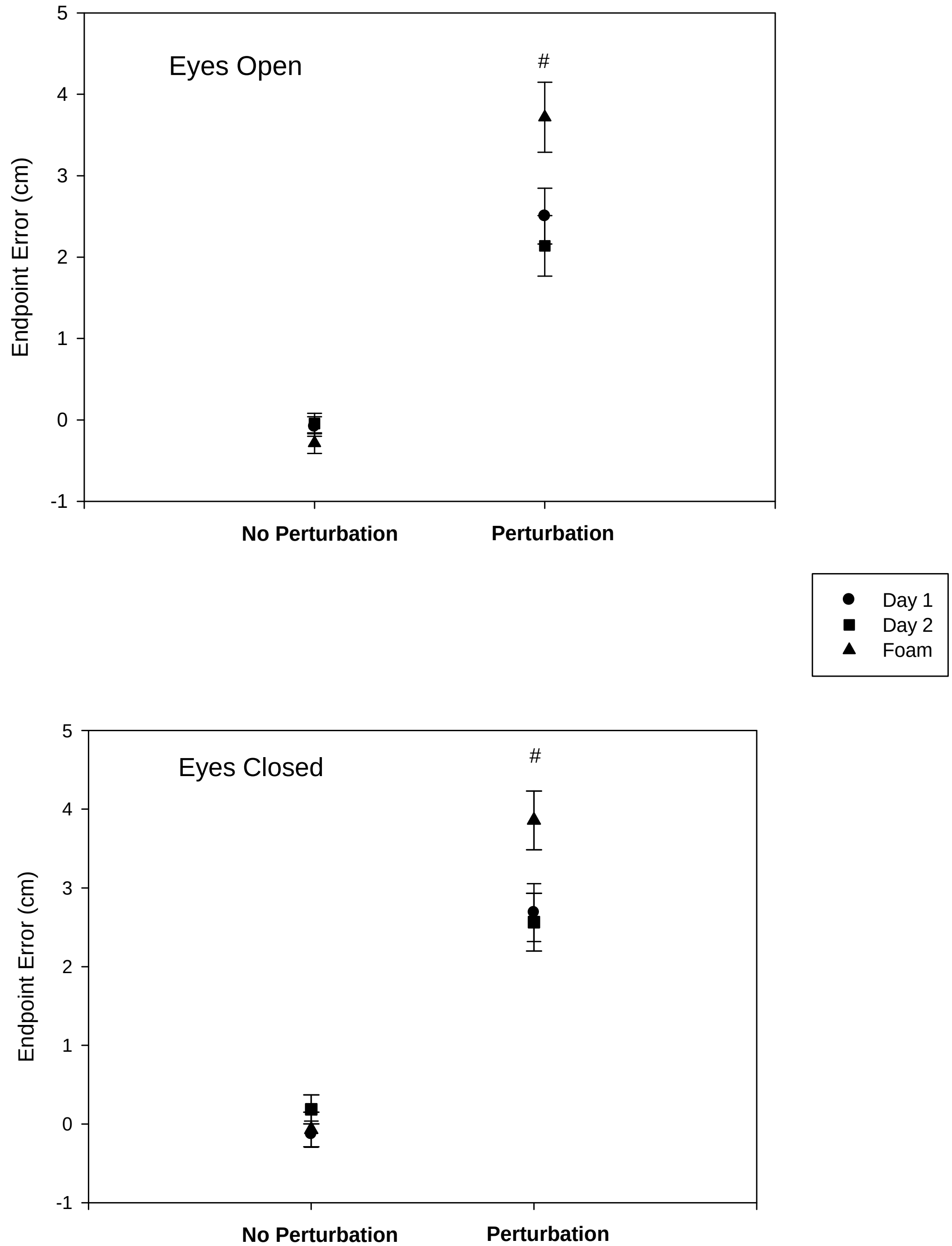

Figure 2.10 Average overshoot error for the eyes open and eyes closed conditions during unperturbed and perturbed trials across Day 1 (*circle)*, Day 2 *(square),* and Day 2- Foam (*triangle*). Values are means ± SE of all 16 subjects.

\# indicates significant difference between Solid and Foam surfaces

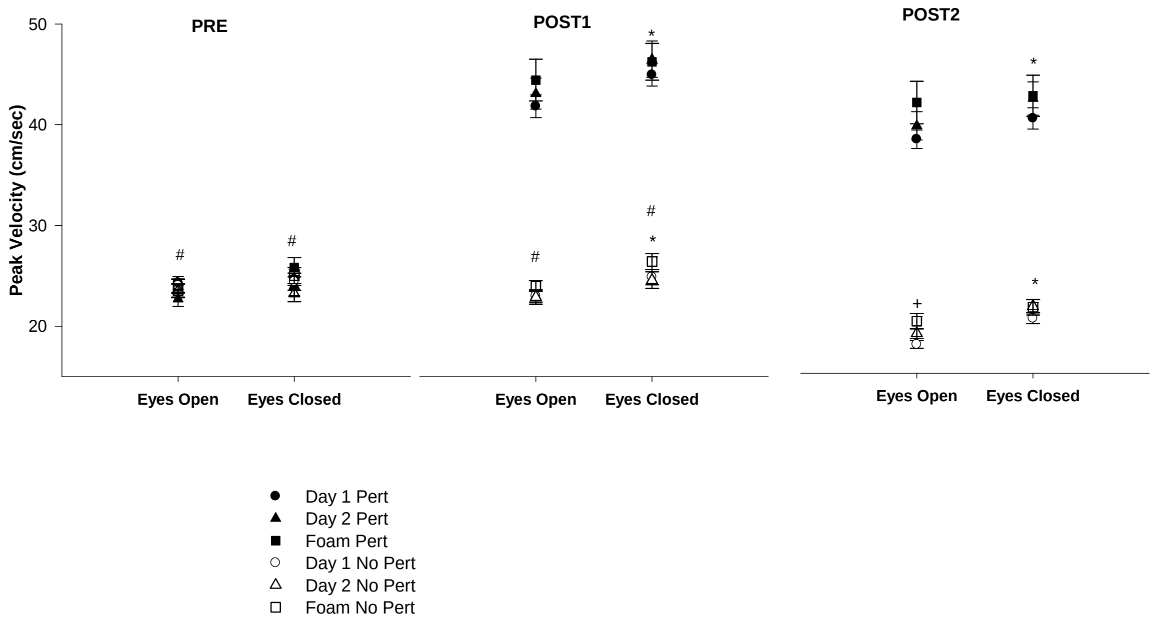

Figure 2.11 Average peak velocity for the non perturbed (*open symbols*) and perturbed (*closed symbols*) events for the eyes open and eyes closed conditions. Data are represented in 200 ms bins -Pre (anticipatory), Post 1 (reflex) and Post 2 (volitional) from the time perturbation was supposed to occur (non-perturbed trials) or occurred (perturbation trials).Values are means $\pm$ SE of all 16 subjects.

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* indicates significant difference from Eyes Open Condition

\+ indicates significant difference between Days

\# indicates significant difference between Surfaces

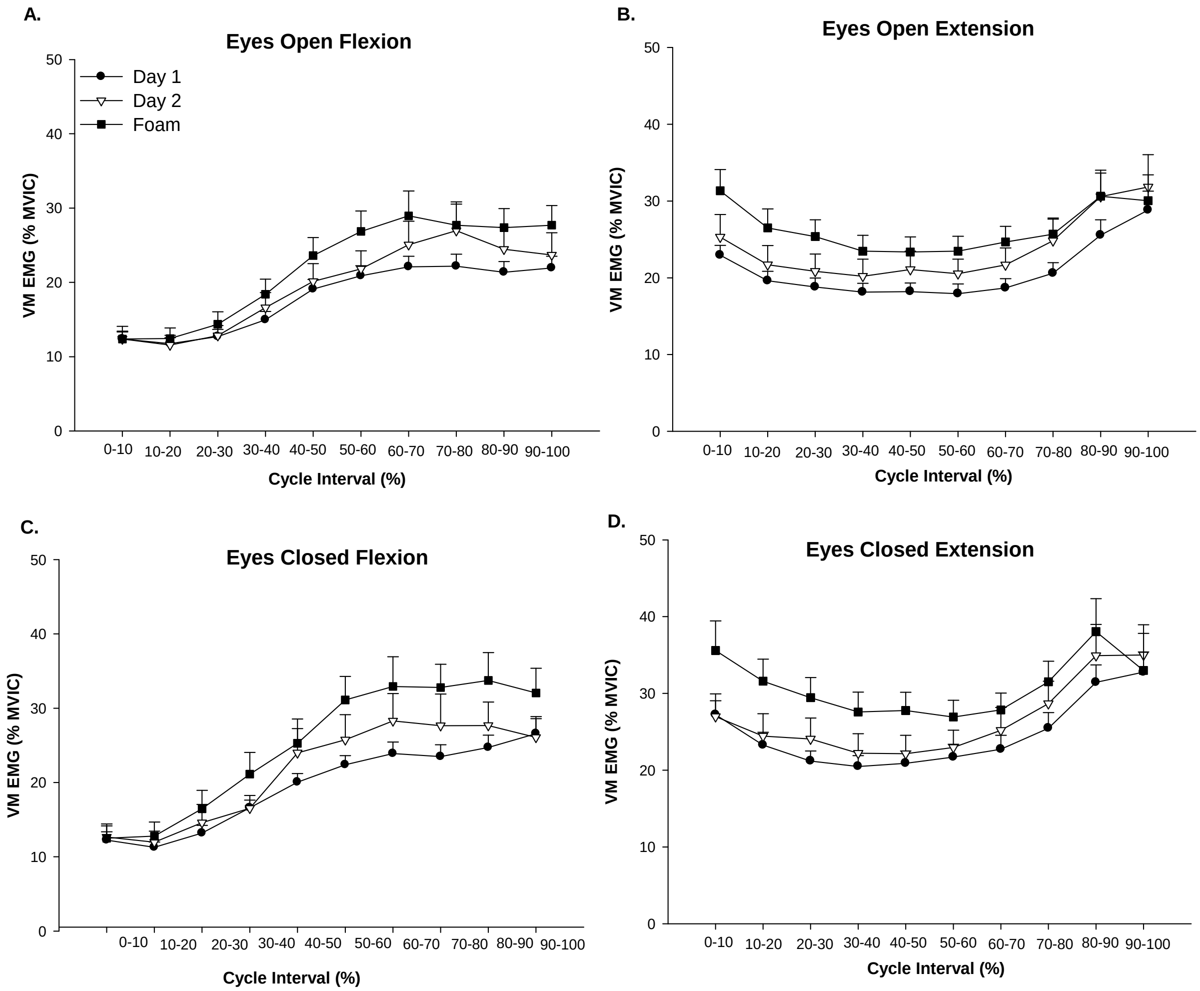

Figure 2.12 Effect of training on the pattern of activation of Vastus Medialis EMG during the flexion (A and C) and extension (B and D) phases, Perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for Day 1 (*circles*), Day 2 (*triangles*) and Day 2 - Foam (*squares*). Values are means ± SE of all 16 subjects.

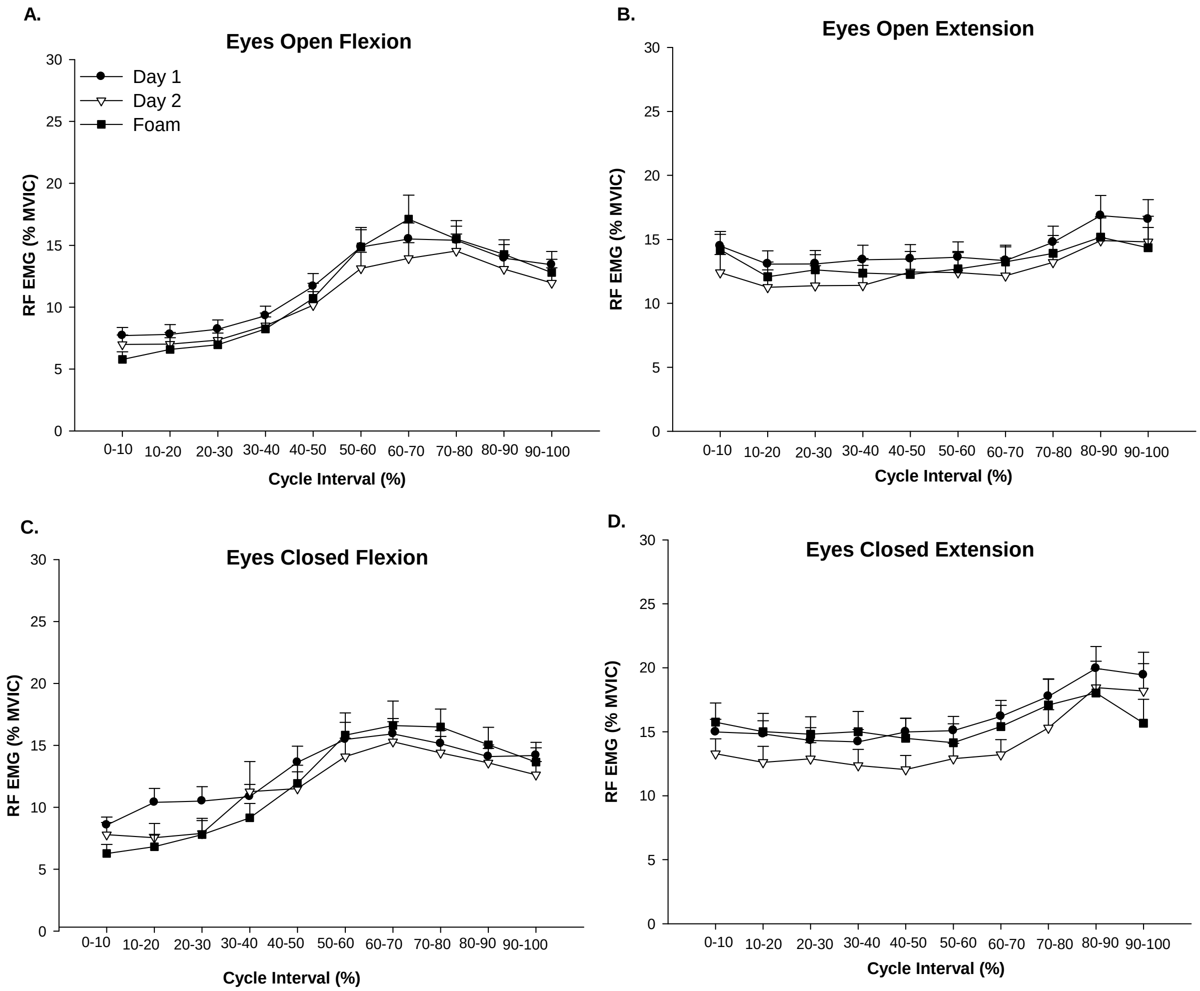

Figure 2.13 Effect of training on the pattern of activation of Rectus Femoris EMG during the flexion (A and C) and extension (B and D) phases, Perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for Day 1 (*circles*), Day 2 (*triangles*) and Day 2 - Foam (*squares*). Values are means $\pm$ SE of all 16 subjects.

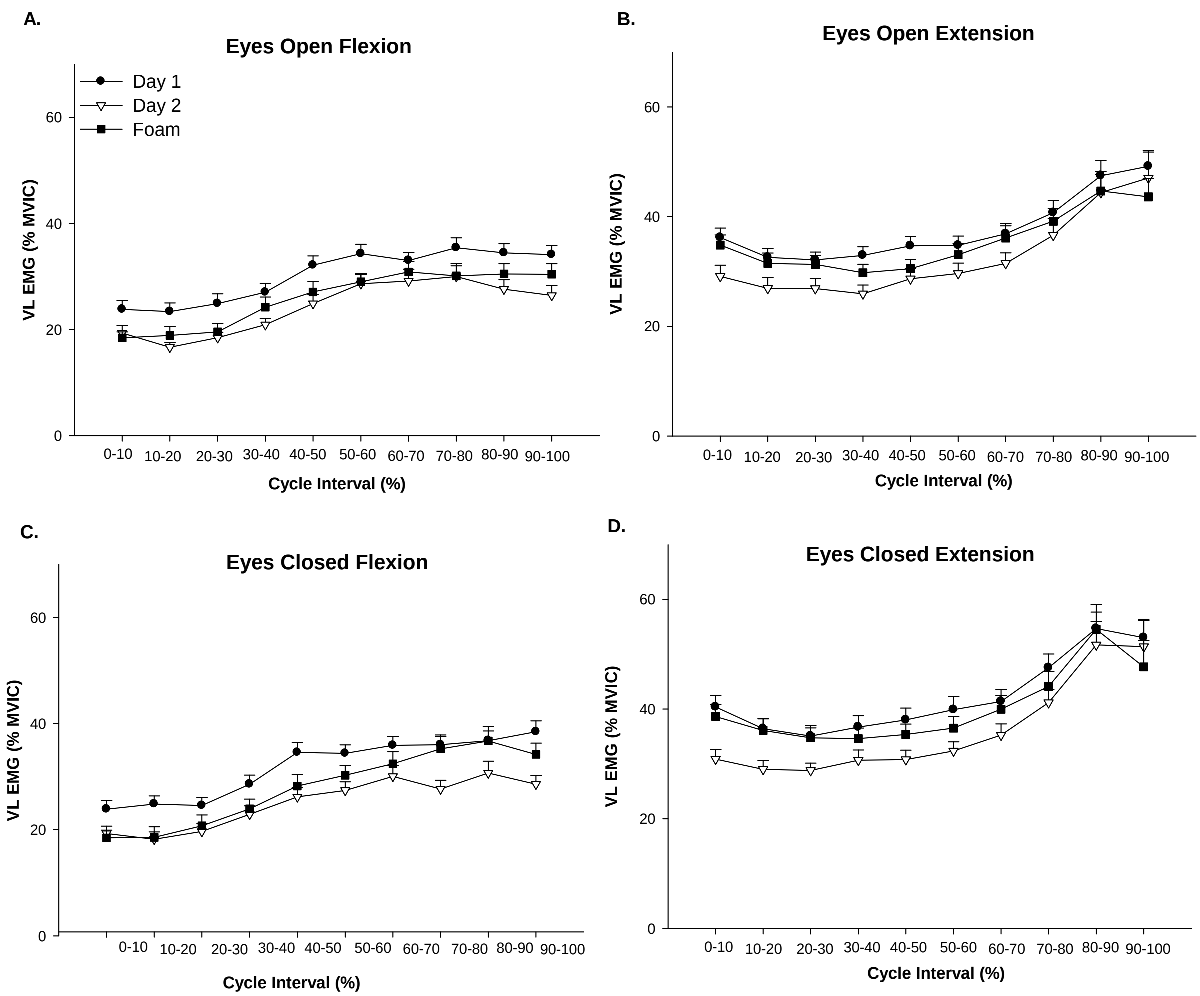

Figure 2.14 Effect of training on the pattern of activation of Vastus Lateralis EMG during the flexion (A and C) and extension (B and D) phases, Perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for Day 1 (*circles*), Day 2 (*triangles*) and Day 2 - Foam (*squares*). Values are means $\pm$ SE of all 16 subjects.

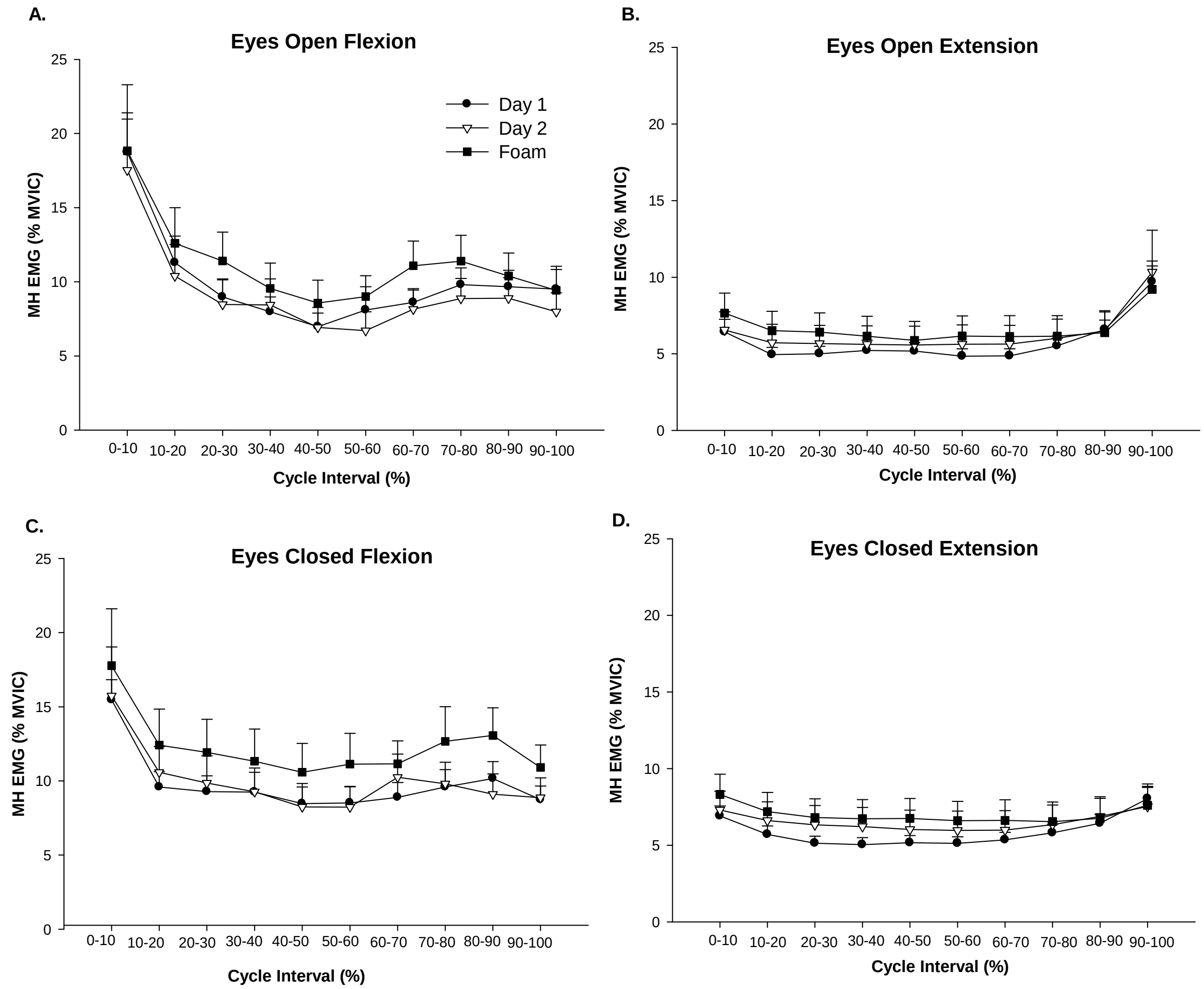

Figure 2.15 Effect of training on the pattern of activation of Medial Hamstrings EMG during the flexion (A and C) and extension (B and D) phases, Perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for Day 1 (*circles*), Day 2 (*triangles*) and Day 2 - Foam (*squares*). Values are means $\pm$ SE of all 16 subjects.

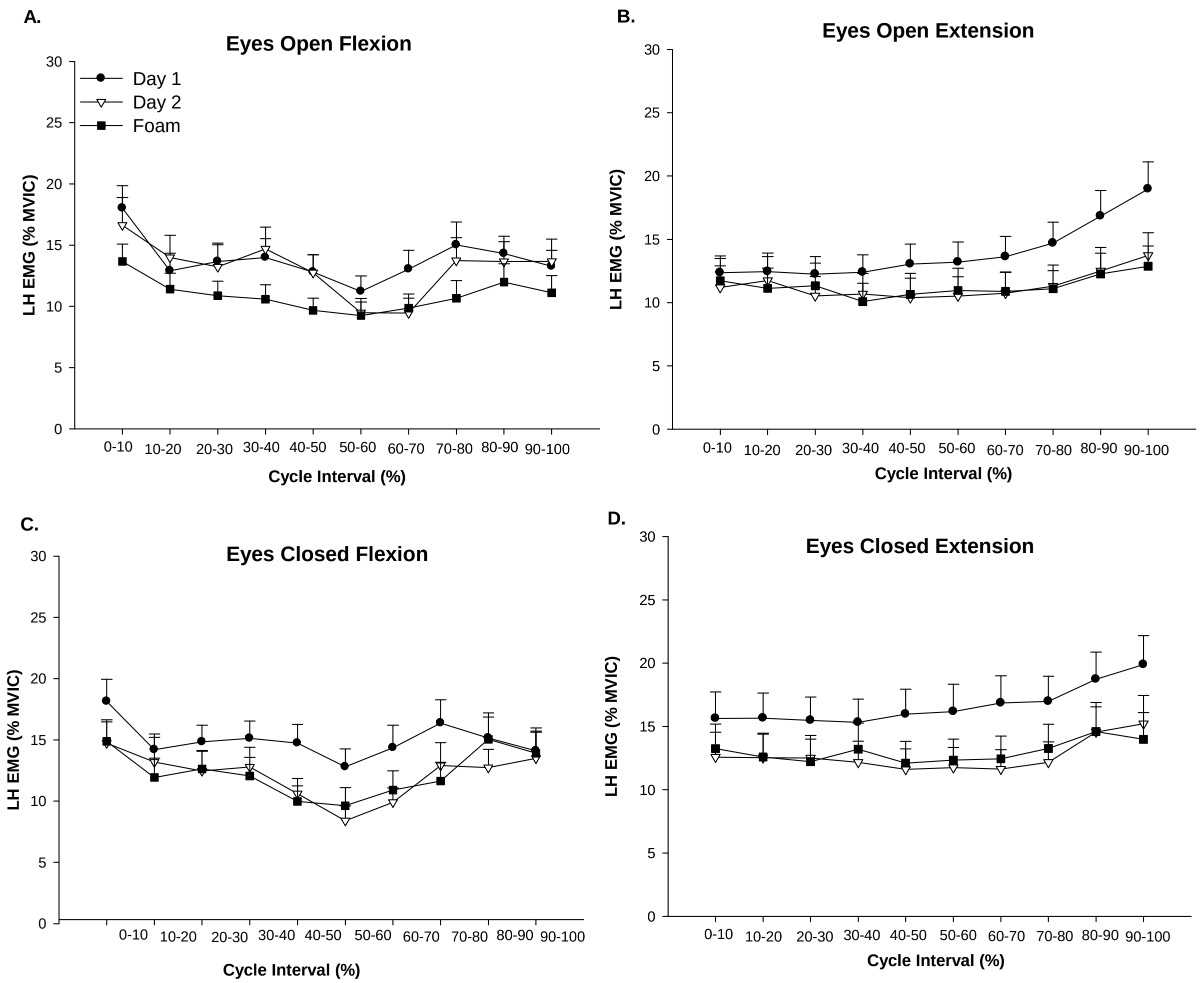

Figure 2.16 Effect of training on the pattern of activation of Lateral Hamstrings EMG during the flexion (A and C) and extension (B and D) phases, perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for Day 1 (*circles*), Day 2 (*triangles*) and Day 2 - Foam (*squares*). Values are means $\pm$ SE of all 16 subjects.

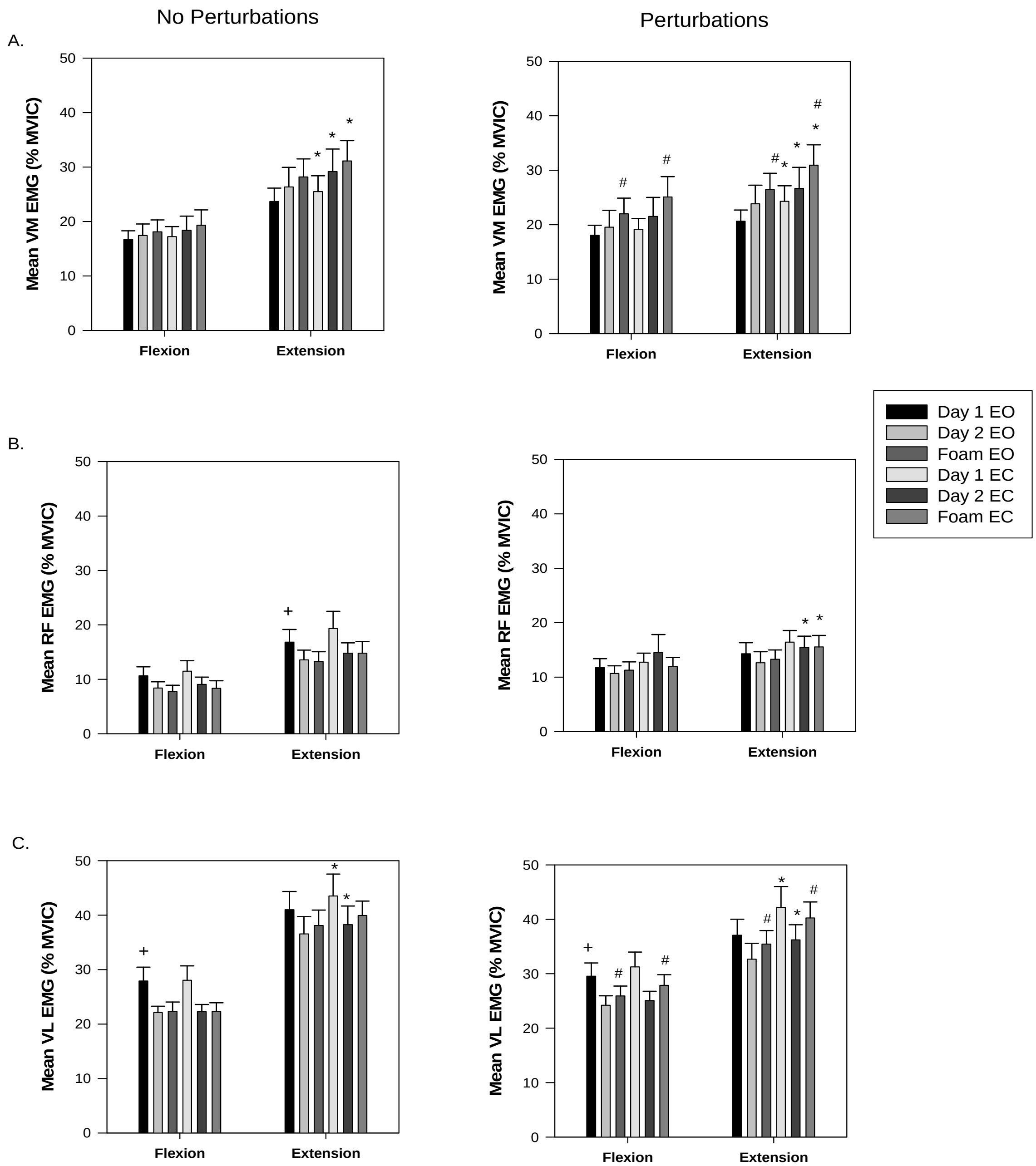

Figure 2.17 Effect of training on the mean EMG during the flexion and extension phases of the nonperturbed and perturbed events for the quadriceps muscles. Normalized EMG for the Eyes Open and Closed are presented for Day 1, Day 2, and Day 2 - Foam. Values are means ± SE of all 16 subjects.

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* indicates significant difference from Eyes Open

\+ indicates significant difference from Day 2

\# indicates significant difference between Surfaces

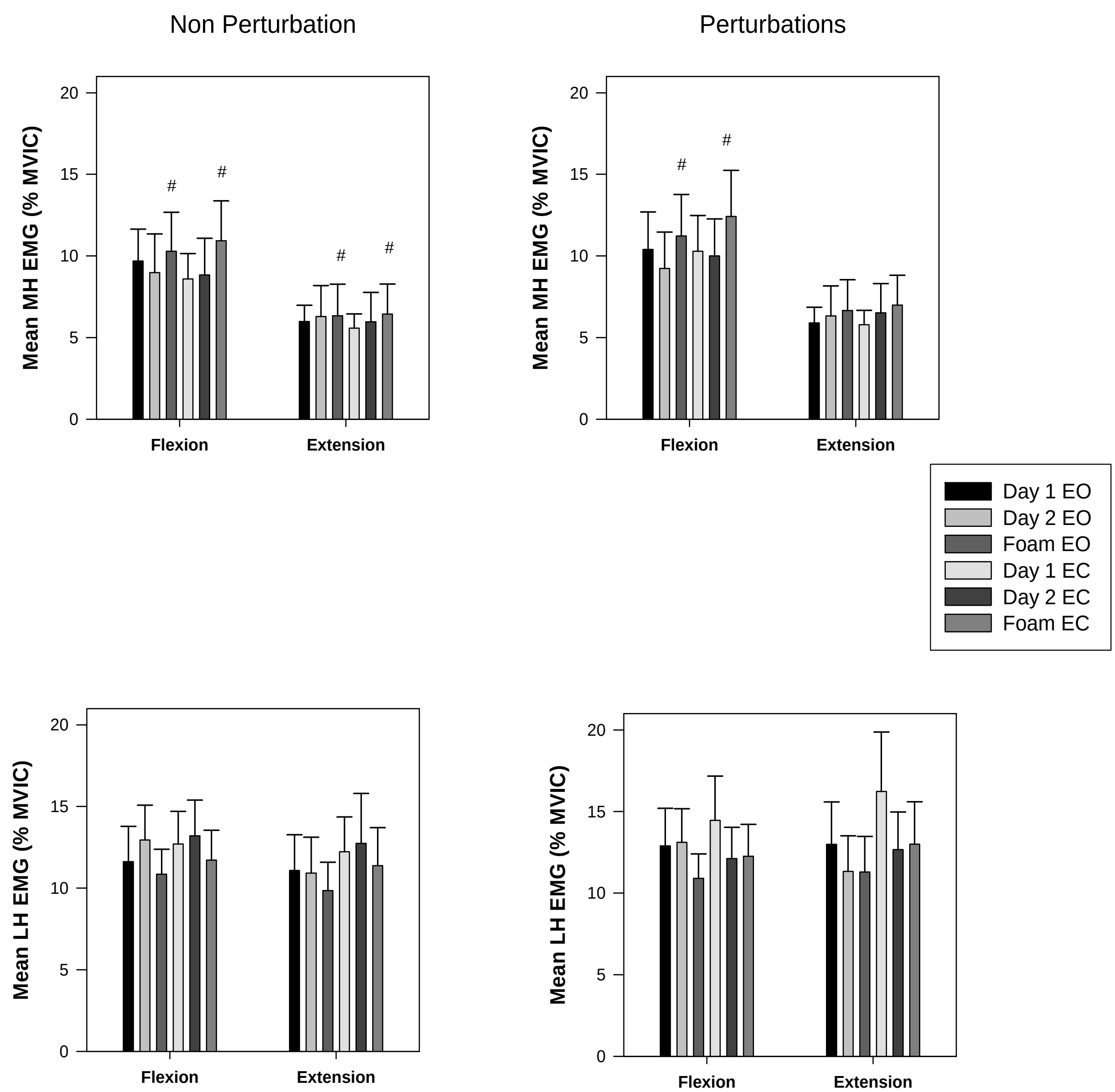

Figure 2.18 Effect of training on the mean EMG during the flexion and extension phases of the nonperturbed and perturbed events for the hamstring muscles. Normalized EMG for the Eyes Open and Closed are presented for Day 1, Day 2, and Day 2 - Foam. Values are means ± SE of all 16 subjects.

\# indicates significant difference between Surfaces

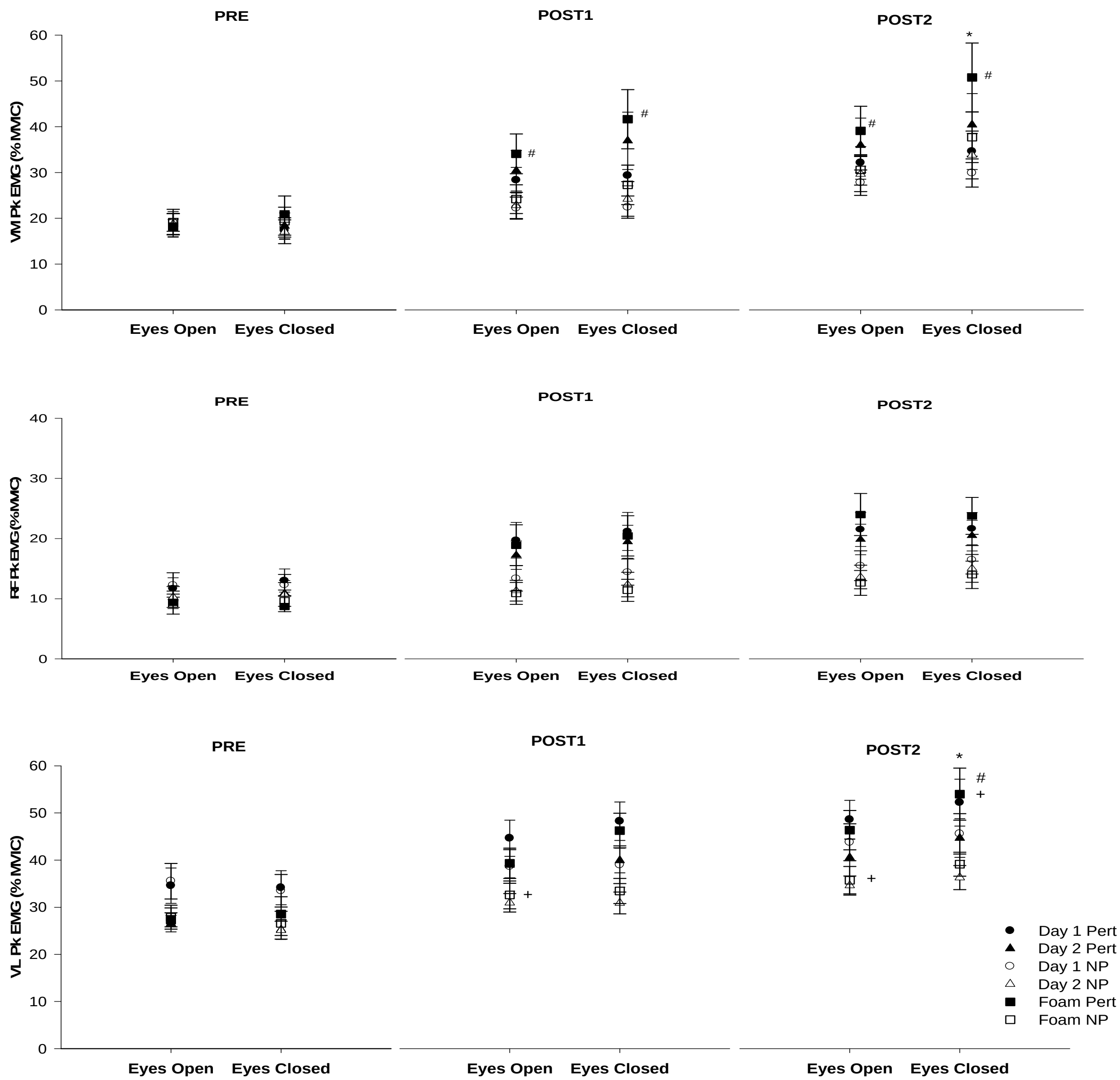

Figure 2.19 Average muscle activity of the Vastus Medialis, Rectus Femoris, and Vastus Lateralis for subjects during the Eyes Open and Eyes Closed conditions during unperturbed (*open symbols*) and perturbed (*closed symbols*) trials. Data are represented in 200 ms bins - Pre (anticipatory), Post 1 (reflex) and Post 2 (volitional) from the time perturbation was supposed to occur (non-perturbed trials) or occurred (perturbation trials) for Day 1 (*circles)*, Day 2 (*triangles)*, and Day 2 - Foam *(squares)*.Values are means $\pm$ SE of all 16 subjects.

+ indicates significant difference between Days when Conditions are combined

# indicates a significant difference between Surfaces when Conditions are combined

* indicates a significant difference from Eyes Open Condition

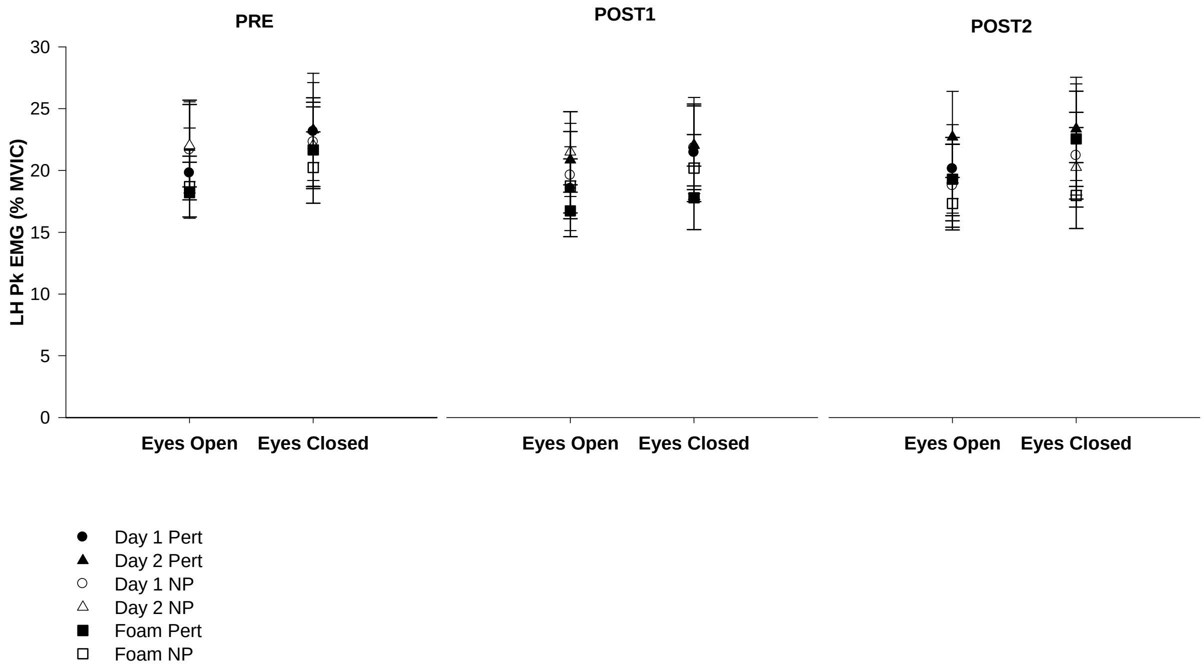

Figure 2.20 Average muscle activity of the Medial and Lateral Hamstrings for subjects during the Eyes Open and Eyes Closed conditions during unperturbed (*open symbols*) and perturbed (*closed symbols*) trials. Data is represented in 200 ms bins -Pre (anticipatory), Post 1 (reflex) and Post 2 (volitional) from the time perturbation was supposed to occur (non-perturbed trials) or occurred (perturbation trials) for Day 1 (*circles)*, Day 2 (*triangles)*, and Day 2 - Foam *(squares)*.Values are means ± SE of all 16 subjects.

\# indicates a significant difference between Surfaces when Conditions are combined

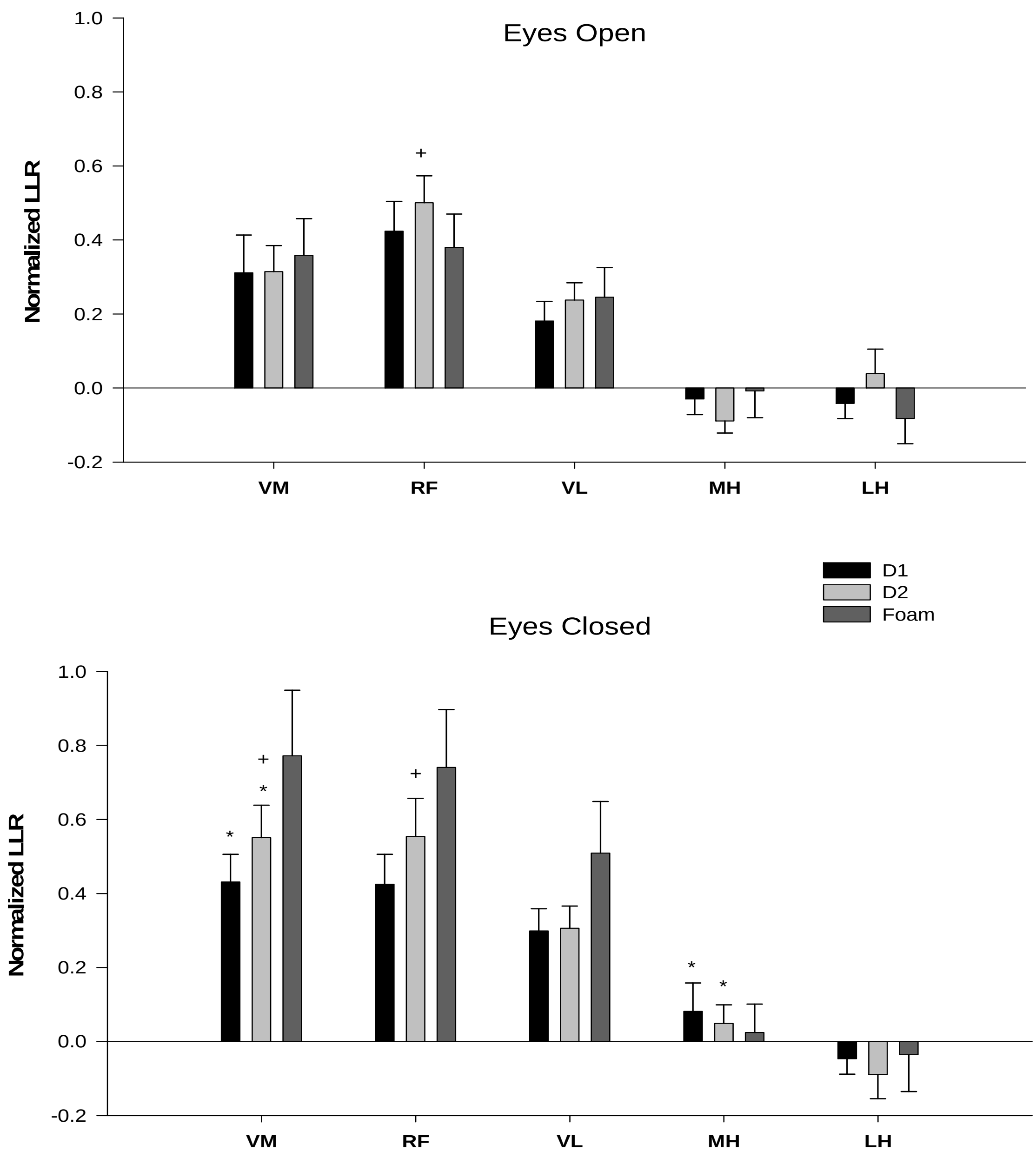

Figure 2.21 Normalized LLRs of the Vastus Medialis (VM), Rectus Femoris (RF), Vastus Lateralis (VL), Medial Hamstrings (MH), and Lateral Hamstrings (LH) during the 50 – 150 ms time bin following the perturbation, for Day 1 (*dark bars*), Day 2 (*light gray)*, and Day 2 - Foam (*dark gray bars*) conditions in the Eyes Open and Eyes Closed conditions. Values are means $\pm$ SE of all 16 subjects.

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* indicates a significant difference from Eyes Open when days are combined
+ indicates a significant difference between Days when conditions are combined

Discussion

The purpose of this study was to examine accuracy of performance, muscle activation strategies, and long latency responses (LLR) during a lower extremity weight-bearing task performed under different conditions of visual feedback and surface type across a 2-day training session. Though perturbation tasks are commonly used in rehabilitation of the lower limb after injury, little information about the learning patterns of this training is available. The sinusoidal target used in this study provided feedback about motor performance, showing the rate and amplitude of subjects' motion. Subjects in this study were focused on accurately tracking a sinusoidal target during a single limb squat (SLS), so were unable to anticipate unexpected perturbations that occurred.

Tasks that require tracking a target have been used to both evaluate and treat individuals with coordination deficits, post ACL reconstruction (ACLR), as well as individuals post stroke.[78-79, 122-124] This method provides constant feedback and, thus, constant self-correction during repeated series of a motor task pattern. As a result, motor planning and performance becomes more efficient.[123, 125] Maffiuletti studied trajectory training of the lower extremity in healthy adults while supine on a leg press machine.[126] The authors found that healthy subjects improved their tracking error within the second day of learning. Madhavan (2009) similarly found that healthy female subjects were able to improve accuracy of performance within 2 days of training a standing, knee tracking task.[79] In 2005, Perez et al. determined that selective presynaptic inhibition of group Ia afferents occurs with training, which contributes to the modulation of sensory inputs during the learning process.[127] Similarly in 2006, Perez et al. demonstrated that visuomotor skill learning in humans is accompanied by increased corticospinal drive to the motor neurons.[128] Training during tracking tasks such as these not only improves motor skill, but also induces cortical changes.[124-125] After 4 weeks of training knee flexion/extension in a seated position, subjects post stroke demonstrated not only improvements in accuracy, walking speed, and motor scores, but also primary sensory motor cortex cortical activation shift from the unaffected to the affected hemisphere in the training group.[124] Consistent with these studies, the young, healthy subjects in this study were also able to improve their tracking accuracy over a period of 2 days for both the nonperturbed and perturbed trials.

Although EMG and kinematic analysis has been performed for this SLS activity,[82, 129] few studies have examined accuracy and how it changes with unexpected perturbations. The present study contributes to the existing knowledge of motor learning literature by adding information about learning during lower extremity perturbations with dynamic, weight-bearing task in healthy knees. It also suggests that gender does not affect performance of this task. The training protocol was performed over 2 days, under three different conditions of visual feedback, and over two different surface conditions. Overall accuracy for nonperturbed events did improve from Day 1 to Day 2 for both males and females, reinforcing previous work suggesting that 2 days of training is sufficient for learning this task.[79] Absolute error was decreased across days for perturbed events as well. However, endpoint error was not different over days, suggesting that subjects were not truly able to learn to adjust to the perturbations. This reinforces the assumption that even with training, participants are unable to anticipate the unexpected perturbations. Although absolute error during perturbed trials improved, ultimately subjects were not able to decrease the overshoot error associated with perturbations over this 2-day protocol.

As accuracy of performance improved with training, muscle activity of the quadriceps was modulated across days. Although all muscles recorded were active during the training exercise, RF and VL were modulated most with training. The change over days, however, was not consistent with the hypothesis that an increase in quadriceps activity would occur with training. For nonperturbed events, RF activity during the extension phase and VL activity during flexion decreased across days, while no significant difference was seen in VM. VL was the only muscle that demonstrated a change in activity level across days with perturbations, decreasing during the flexion phase as it did with nonperturbed events. VL is required to prevent excessive knee flexion and falling forward with the perturbations. One explanation for the decrease over days may be that on Day 1, when subjects were first introduced to the task, VL activity was increased as a strategy to prevent falling forward too quickly with the perturbation. On Day 2, as subjects were more knowledgeable about the extent of perturbations, this strategy was not used to perform the task. In addition, compared to previous work in this lab with the SLS task,[61, 79] the knee pad device was spring loaded to prevent sliding within the device. As a result, there was less shearing and superior/inferior motion within the device which may have influenced the strategy used. Subjects may have perceived greater instability with this instrumentation modification. No difference was seen over days for MH or LH, for either nonperturbed or perturbed events. This lack of modulation of hamstrings activity with training is also consistent with previous work.[81]

Though the single limb squat is a common exercise prescribed in lower extremity rehabilitation, the effect of training with perturbations on the anticipatory (feedforward), reflex (feedback), and volitional activation of musculature has not been examined. Subjects were challenged with this task to maintain accuracy to the best of their ability, even though unexpected perturbations were presented. As a result, subjects could not anticipate or prepare for the perturbations in advance. Muscle activity measures at 200 ms pre-perturbation, 200 ms, and 400 post-perturbation provided information on the anticipatory, reflex, and volitional activation of these muscle groups during the task. Since there was no effect of training for any muscle during the anticipatory time bin, this supports the assumption that subjects were not able to “prepare” for the perturbations with training over 2 days. VL was the only muscle that demonstrated any change in activation across days, with decreased activity on Day 2 during the reflex and volitional time bins of the nonperturbed events. During the perturbed events, VL had decreased activity on Day 2 during the volitional time bin only. This suggests that of the muscles examined during this task, only VL was modified with perturbation training, in its volitional response 200-400 ms post-perturbation.

Another measure of response to perturbations is to determine the long latency reflex (LLR). Since these occur earlier than voluntary responses, but not as quickly as spinal stretch reflexes, they may have greater potential for modification by supraspinal neural centers.[38, 53, 109] Muscle spindle, visual input, and vestibular sensory systems modify long latency responses during perturbations.[10, 46-47] This lends more credence to the hypothesis that the LLR during a weight-bearing task may be flexible and context specific.[7, 48, 62, 130] Previous work suggests that when perturbations are unpredictable, the LLR is increased[131-132] while there is an attenuation of the response with predictable perturbations.[7, 133] Unpredictable perturbation of the knee during flexion of the SLS task during this study stretched the quadriceps muscle, resulting in LLRs in the VM, RF, and VL. The normalized LLR response was modulated in the VM and RF across days, increasing for both on Day 2 while no significant changes were seen in the VL or the hamstrings.

Studies utilizing operant conditioning have shown changes in the LLR in the biceps brachii. After 6 baseline and 24 training sessions, the stretch reflex was downtrained in the training group, while it increased in the control group.[110] Wolf (1996) used a 16-week training session, with subjects attending two to three times per week for 1.5 hours and experiencing up to 250 stretches per session.[134] Subjects were trained to decrease their response to a stretch, with reductions in both the M1 and M2 response amplitude noted in the training group which was not seen in the control group. The reduction in the LLR was noted at Session 5 and persisted through Session 30. The perturbation in the study was given to the biceps contracting at only 5% of maximal effort, and subjects were not performing a skilled tracking task. In addition, the subjects were not standing, requiring vestibular input and modulation.

In a balance training study, Taube's (2007) subjects trained for 16 sessions over 4 weeks performing various balance activities during hour-long sessions. Horak (1989) also found that with practice, the magnitude of postural responses to the same stimulus is attenuated with platform perturbations.[7] Post training, this group noted a reduction in the LLR in the soleus. Perturbations performed during computerized posturography may not have truly been unpredictable, however, as subjects were not focused on a tracking task or other skilled activity while tested. The actual training performed in that study, however, did not include perturbations.[91]

The results of the current study suggest that the LLR actually increased across days for VM and RF during this SLS tracking task. One potential explanation for this finding may be that as subjects learned the task, there was a change in cortical excitability. As a result, the LLR was facilitated on Day 2 in relation to this change in excitability. Many authors have suggested that there are transcortical components to the LLR.[40, 45-46, 135-136] In 2006, Mrachacz-Kersting et al. suggested that the RF muscle may have a differential neural control because it is a two-jointed muscle.[136] During a seated task, this group quantified the LLR to an imposed knee flexion delivered in combination with a transcranial magnetic stimulus in quadriceps and hamstring muscles. Their results indirectly supported the notion that a transcortical pathway is involved in shaping the RF LLR. As there were no effects for VM and VL, they suggested that these results were specific to the RF. The neural control of the RF may be different since it is both a hip flexor and knee extensor.

Non-invasive brain stimulation has been used to identify the functional relevance of particular brain regions in motor learning and to facilitate activity in specific cortical areas involved in motor learning in an attempt to improve motor function.[137] The acquisition of new motor skills is accompanied by changes in neuronal activity and excitability.[138] As shown by functional imaging techniques and transcranial magnetic stimulation, the primary motor cortex transiently displays enhanced activity and excitability during learning of sequential finger movements.[139-142] Pascual-Leone et al. (2005) used TMS to map the cortical motor areas targeting the contralateral long finger flexor and extensor muscles in subjects learning a one-handed, five-finger exercise on the piano.[142] Over the course of 5 days, as subjects learned the skilled task through daily practice, the cortical motor areas targeting the long finger flexor and extensor muscles enlarged, and their activation threshold decreased. They also studied the effect of increased hand use without specific skill learning in subjects who played the piano at will for 2 hours each day using only the right hand. In those control subjects, the changes in cortical motor outputs were similar but significantly less prominent than in those occurring in the test subjects, who learned the new skill. If cortex excitability was increased on Day 2 of the current SLS study with learning (indicated by decrease in absolute error during the task), this could help to explain the increase in the LLR. It cannot be assumed, however, that subjects were truly “trained” with perturbations in this study. Each subject experienced approximately 20 perturbations during 100 repetitions of the tracking task. To truly attempt to train the LLR, the dosage used in this study was likely inadequate. The results, however, merely identify differences seen in the LLR with this particular protocol.

Effect of Vision and Surface

Visual, somatosensory, and vestibular information working together is required for efficient neuromuscular control of movement. When one of these systems is removed, control is affected, as demonstrated in many studies with posturography.[20, 66, 72, 98] However, this remains a less explored area in the modulation of muscle activity during dynamic, weight-bearing activities. One difference between this study and others is that a challenging task that emphasized performance accuracy was used, and how subjects reacted to unexpected perturbations of this task under different conditions of visual feedback and surface was examined. Since vision and proprioception play important roles in functional tasks, the effect of vision and surface type during weight bearing was studied by having subjects perform the knee joint tracking task under three visual conditions and repeating these on a compliant foam surface. The goal was to determine whether there was a difference from a visual condition in which the subject does not have the template to view but is still able to use vision to assist in orientation versus no vision at all to orient self to upright. We also wanted to see if there was an effect of the compliant surface on task accuracy, muscle activation patterns, and the LLR.

There was no significant difference between the NT and EC conditions as both conditions showed equivalent error. Absolute error during both NT and EC was almost three times higher than during the EO condition. Error was significantly less when visual feedback was available to detect errors online and correct the movement trajectory immediately. Without visual feedback in either condition, a large positional error accumulated despite reasonable ability to follow the remembered target waveform. As expected, no improvements in absolute error were seen with training under the no screen or eyes closed condition, suggesting that visual feedback is required in this 2-day protocol for learning this task.

As subjects improved performance with training, muscle activity was modulated under the various conditions of visual feedback. VM and VL demonstrated increased activity with eyes closed during the extension phase of both the nonperturbed and perturbed events. In the absence of visual feedback, quadriceps activity was increased during this phase of the task to return the subject to upright.

The long latency response was also modulated by vision, with VM and MH showing an increased response with eyes closed. Previous work examining the effect of vision on the LLR during CDP studies also show an increase in the LLR response in the absence of vision.[50, 75, 143] Timmann (1994) demonstrated this increase in LLR as well as a decrease in latency in this condition. Welgampola (2001) determined via galvanic stimulation and CDP that larger responses are seen when one of the balance systems is lost.[50]

The foam surface provided a method to alter proprioceptive input as subjects performed the task. Especially with eyes closed, this foam condition (it was theorized) would provide ambiguous sensory information, perhaps affecting the LLR. Standing on the cushion depresses the range needed to move in order to still reach the target. As the tibia advances forward, the foam gives way to assist in forward weight shift without as much excursion at the ankle required. As expected, error post perturbations was increased in this condition compared to the firm surface. Absolute error during the flexion phase of the perturbed events was significantly increased compared to the firm surface. Endpoint error was also increased on the foam surface for both nonperturbed and perturbed events.

Muscle activation strategies were altered in the foam condition. VM and VL were the most affected of the quadriceps, demonstrating increased activity on the foam for both flexion and extension phases of the perturbed events compared to the firm surface. MH also showed increased activity in both phases for nonperturbed and perturbed events, suggesting that an increase in knee flexor activity is required to maintain accuracy on the foam. This created greater stiffness around the knee joint to perform the task in this condition. The mechanical effectiveness of muscle contraction is potentially reduced on a more compliant surface, which may explain this increase in muscle activity.[50] To flex the knee against resistance while on the foam, the hamstrings needed to increase their contribution to the task. If increased hamstring activation is desired, the foam cushion will facilitate this.

Although there was a difference in LLR response with eyes closed for VM, RF, and MH on the firm surface, no differences were seen for any muscle on the foam surface. Previous studies using CDP have found increased LLR when visual and surface input is altered.[50, 143] Welgampola (2001) and Bacsi (2005) used galvanic stimulation to affect the vestibular system while subjects stood on compliant surfaces. Inglis (1994) and Bloem (2002) examined patients with complete proprioceptive loss of the lower extremities and each saw reduced LLR of the ankle musculature, while quadriceps LLR remained unchanged compared to controls. The foam surface in this study provided ambiguous sensory information rather than removing it entirely and left the vestibular system intact.

While the normalized LLR was not affected by foam in this study, there were changes in individual muscle responses when examined during the reflex and volitional time bins after perturbations. VM and MH showed an increase in activity during the reflex and volitional time bins of the perturbation events, and VL similarly was increased, but only during the later volitional bin. These findings suggest that different muscle activation strategies are used in different conditions.

One possible reason the foam surface did not affect LLR is that the foam altered the mechanics of the task so that this condition cannot be accurately compared to the firm surface data. On the compliant surface, one has a greater distance to “sink” into the surface with a perturbation overshoot. Subjects moved through decreased excursion, as suggested by COP measurements taken during the task. In addition, velocity was different on the foam surface compared to the firm surface in the anticipatory time bin. This may indicate that subjects were using different strategies to track the target during the first 1/3 of the range of the SLS task on the foam. While a subject is standing on foam performing the SLS, there is no solid surface providing a shear force on the bottom of the foot as the knee flexes and pushes the bar forward. Instead, the subject may sink down into the surface while advancing the bar, resulting in less knee flexion required to stay on the target.

Limitations

A limitation of this study is the constrained environment in which perturbations were delivered during the SLS task. The reason for confining motion to the sagittal plane was to assess the neural response to movement including visual, somatosensory, and vestibular contributions, while maintaining a safe environment for the subject.

Conclusion

The controlled SLS task detected differences in neuromuscular control of the knee within 2 days of training in different visual and surface conditions with respect to performance accuracy, muscle activation strategies, and long latency response. Specifically, subjects had statistically significant improved accuracy of performance with training for both nonperturbed and perturbed events. Performance was improved with visual feedback and when on a firm surface. Though endpoint error, with and without perturbations, did not change over the training days, absolute error and muscle activation strategies did. The lack of change in endpoint error for both nonperturbed and perturbed events may be due to the shortened training protocol. Training also affected the mean RMS EMG of RF and VL as well as the LLR of VM and RF. The exact etiology of changes in muscle synergy and long latency response is not yet established.

CHAPTER 3

BILATERAL VESTIBULAR LOSS INFLUENCES PERFORMANCE ACCURACY, KNEE MUSCLE ACTIVATION PATTERNS, AND LONG LATENCY RESPONSES DURING A SINGLE LEG WEIGHT BEARING TASK

Introduction

Bilateral vestibular loss (BVL), a condition that results in significant functional disability and handicap, was first described by Dandy in 1941.[144-146] Approximately 1% to 2% of all patients undergoing electronystagmography studies have this condition.[147-149] Origins of BVL include ototoxicity, autoimmune inner ear disease, bilateral vestibular neuritis, bilateral endolymphatic hydrops, bilateral vestibular schwannoma, and idiopathic vestibular loss.[19, 147, 150-151]

Postural control dysfunction is well documented for patients with BVL, including instability in stance, during ambulation, and with transitional activities such as moving from sitting to standing.[4, 20, 150, 152-156]As a result, falls are a common problem among persons with BVL.[4, 150, 157] To address the impairments, activity limitations, and participation restrictions, vestibular rehabilitation is considered the treatment of choice for this population.[158-159] In these programs, patients are taught to improve gaze stabilization and eye/head coordination, utilize sensory information and movement strategies for balance, utilize sensory substitution, and learn adaptation of strategies based on task demand.[153, 159-160] The neurophysiologic basis for improvement in individuals with BVL is believed to be adaptation of the central nervous system, sensory substitution, or reweighting of the sensory systems.[6] Sensory reweighting is the brain's ability to change the relative reliance on a specific sensory modality for orientation depending on the environment, task, or pathology.[161] Visual and/or somatosensory information must play dominant roles in compensating for the decreased contribution of the vestibular system.

A majority of studies investigating postural control in individuals with BVL have utilized computerized dynamic posturography (CDP).[9-10, 19-20] Consistently, these studies demonstrate that patients with inadequate and/or absent peripheral vestibular input fall when both visual and somatosensory feedback are altered.[11] Falls may be more related to changes in postural control during movement, however, which has not been well examined in this population.[4] Despite research pertaining to BVL via platform posturography, there is limited evidence available on neuromuscular function as it relates to movement accuracy and muscle activation strategies used during functional activities in this patient population. Identifying the neuromuscular strategies used by individuals with BVL to perform functional tasks will provide better insight into the functional stability that rehabilitation provides and may help advance rehabilitation techniques.

One such task, which is both dynamic and challenging, is the single limb squat (SLS). This weight-bearing task mimics functional activities such as moving from sit to stand, walking, and stair climbing, which must be performed each day in many different environments.[82-84] In addition, weight-bearing exercises such as this promote increased stability through lower extremity joint compression and cocontraction of the quadriceps and hamstring muscles, which in turn increases joint stiffness and contributes to joint stability.[82, 84] Performing this task while tracking a sinusoidal target with knee displacement provides a method to train dynamic stability of the knee while emphasizing accuracy of performance.[61, 77-78] Without vision, individuals must use greater coactivation of the quadriceps and hamstrings.[79, 81] Though studies have assessed the contribution of visual feedback to learning this task in healthy adults and college-aged women post anterior cruciate ligament reconstruction, the impact of BVL on performance in this condition is not yet known. When vestibular information is absent, the lack of visual feedback during this task may impact accuracy and activation strategies to an even greater degree.

Adding perturbations to the SLS task provides opportunities for the neuromuscular system to potentially develop compensatory muscle activation patterns in response to destabilizing forces at the knee. Voluntary and reflex components of muscle activity need to be considered when evaluating neuromuscular control. Studying the response to perturbations also provides a measurement of muscle activation and latency.[24, 30, 57] A rapid stretch of an active muscle produces a series of reflex bursts of EMG.[35-36] The initial short latency (SL) stretch reflex has an onset latency of less than 50 ms and is mediated by group Ia muscle spindle afferents of the muscle spindle through a monosynaptic segmental pathway.[37-40] Muscle responses occurring after 50 ms but prior to volitional activity at 200 ms are termed long latency responses (LLR). Evidence suggests the LLR have a polysynaptic pathway that includes supraspinal input through a partly transcortical pathway.[45-47] These are also believed to be mediated by the group Ia and/or group II afferents.[42-44] The supraspinal input, including visual, vestibular, and proprioceptive signals, contribute to the determination of context and feedback responses, allowing for more complex control than would be possible with spinal reflexes alone.[32, 41] If vestibular signals do contribute to the LLR, individuals with BVL should have altered responses to perturbations as a result. Despite any sensory reweighting that may occur after BVL, these individuals still lack important vestibular input for these responses. Reflexive activation patterns in response to sudden stress of the knee during dynamic, weight-bearing activities need to be studied in a controlled manner in varying sensory environments to help therapists better understand the neuromuscular strategies that contribute to joint stability during functional activities in this patient population. Understanding this may provide more insight into the contribution of visual and vestibular input to the neuromuscular control of functional activities. In addition, identifying the strategies used by individuals with BVL to perform functional tasks will provide better insight into the functional stability that rehabilitation provides and may help advance rehabilitation techniques that will reduce the risk for falls.

Therefore, the purpose of this study was to compare accuracy of performance and the muscle activation strategies used to perform a controlled weight-bearing task among subjects with BVL and those with healthy vestibular systems. In this study, the SLS task was performed under controlled conditions by performing the exercise at a standardized resistance applied to knee flexion and extension. Subjects followed a sinusoidal target on the screen with knee motion to enable monitoring of rate and amplitude of movement. The effect of visual feedback on the accuracy of performance and muscle strategies of the controlled SLS task in these two groups was examined. Subjects with BVL were expected to have equivalent levels of accuracy as the control group when the eyes were open (both with and without feedback of the monitor); however, with the eyes closed, the magnitude of the difference in performance accuracy between the visual and non-visual feedback conditions was expected to be greater in the BVL group. It was expected that individuals with BVL would also show altered muscle activation strategies (decreased quadriceps and increased hamstrings muscle activity) during the controlled SLS task in the eyes closed condition, but would be equivalent to the healthy groups when vision was present. The LLR was expected to be reduced in the BVL group compared to the healthy matched controls in the eyes closed condition. This will be the first study to examine neuromuscular control of the knee during a dynamic functional task in individuals with BVL.

Methods

Subjects

The study sample consisted of 10 subjects: 5 subjects with BVL (1 male, 4 females) and 5 gender- and age-matched controls between the ages of 22 and 65 years (Table 3.1). Selection was based on similar body mass, height, and leg length. Inclusion criteria also included regular physical activity without participation in any physical training program designed to increase/maintain fitness and ability to climb stairs without any difficulty. Eligibility requirements for BVL were based on clinical and vestibular function tests. Vestibulopathy was classified as BVL based on bilaterally absent caloric responses and vestibular ocular reflex (VOR) gains >2.5 standard deviations (SD) below normal during whole-body sinusoidal vertical axis rotation (SVAR) tests at a frequency range of 0.01 to 1.0 Hz.[162] Reduced or absent horizontal canal function in this population is generally indicative of vertical canal dysfunction.[163] Each subject included met these criteria. In addition to having had BVL more than 2 years, subjects had clinically stable symptoms (no substantial change in functional abilities) for at least 2 months before study entry as documented by the referring physician. All subjects were able to walk without an assistive device. Exclusion criteria included body mass index greater than 35, history of neurological deficits, musculoskeletal disorders, degenerative joint diseases, cardiovascular diseases, previous knee injury or surgery, previous fractures of the lower extremity, patellar dislocations, and past or current knee pain during activity or rest. Also excluded were individuals with Benign Paroxysmal Positional Vertigo, Meniere's disease, or other unstable vestibulopathies. If subjects were unable to perform the SLS task, they were also excluded. Six potential subjects who initially met the criteria for BVL were excluded from the study (4 males, 2 females). Three males were excluded due to inability to perform the task completely and within the required accuracy criteria. In addition, one of these males had low back discomfort during initial screening. Another male was excluded as he also had a history of a CVA. Two females were excluded as the task caused too much knee discomfort. Prior to participation, subjects were given a brief description of the protocol and possible risks and benefits of participation, and signed an informed consent statement approved by the University of Iowa's Human Subjects Review Board.

Screening Examination

All subjects completed the following questionnaires: a general medical history form, the Short Form Medical Outcome Survey (SF 36) that assesses perception of quality of life,[112] Duke Activity Status Index,[164] Baecke Questionnaire of Habitual Physical Activity,[113] Tegner Activity Rating Scale,[114] and Marx Activity Scale.[115]

Balance Assessment

Assessment of static standing balance was measured with the modified clinical test of sensory interaction and balance (modCTSIB)[117] as well as the single leg standing balance test by means of a strain gauge force plate (24 inches x 15 inches).[118] For the modCTSIB, subjects were required to stand erect with feet together without moving in the center of the force plate with shoes donned, looking straight ahead as long as possible or until the trial of 30 sec was over. This was performed up to three trials, repeated with eyes closed, and then while standing with feet together on a 24- x 24-inch piece of medium density foam cushion. For single leg standing balance, the opposite leg was flexed to 90 degrees at the knee joint, legs apart, with both arms hanging relaxed at the sides. The subjects were instructed to stand as motionless as possible and were allowed to practice this position for 30 seconds before two measurements were taken. Anterior/Posterior (A/P) sway, Mediolateral (M/L) sway, as well as A/P and M/L amplitude, velocity, and range were measured both statically and while performing the SLS task. Static COP was compared in a subset of BVL and control subjects while standing free and while strapped into the podium.

The five time sit to stand test was performed[165-166] to quantify the subject's ability to perform transitional movements. Dynamic balance was also assessed with the Timed Up & Go (TUG).[167-168] The TUG measures, in seconds, the time taken by an individual to stand up from a standard arm chair (approximate seat height of 46 cm, arm height 65 cm), walk a distance of 3 meters (approximately 10 feet), turn, walk back to the chair, and sit down again. The subjects wore their regular footwear and performed the test without an assistive device. No physical assistance was given. Each subject walked through the test once before being timed in order to become familiar with the test.

Experimental Task

The main task of the subjects was to perform the SLS task standing on one leg in a custom mechanical device that enabled the subject's knee joint excursion to follow a sinusoidal target (0.4 Hz, T = 2500 ms) projected on a computer screen in front of them. Resistance, which was normalized to each subject and set at 17% body weight (BW), was provided to both knee flexion and extension. Each SLS lasted for 2500 ms with equal times in flexion and extension.

While performing the SLS task, subjects are secured in a custom device which allows perturbations to occur without risk of injury. Since participants are standing and performing a dynamic, weight-bearing task, subjects experience the sensory redundancy of neural components as in daily life. It is not equivalent, however, to the unilateral stance balance test. Pilot data support that body sway (center of pressure) during unilateral stance with eyes closed is reduced nearly 80% when subjects are attached to the weight-bearing tracking system designed for this study. This is an important consideration because the device used in this study is designed to study perturbations in weight bearing in subjects with BVL who have balance impairments. The degrees of freedom when attached to this instrumentation are reduced because only sagittal plane motion is permitted. Accordingly, the limited degrees of freedom permit the safe delivery of perturbations in weight bearing in an intact system (visual, vestibular, somatosensory integration) without an emphasis on maintaining balance. [61] This SLS task provides a novel approach not only to train the ability to effectively respond to perturbations of the knee but also to emphasize accuracy of performance. With this method, we can assess certain integrated responses of vestibular, proprioceptive, and vision under a preloaded condition in a safe environment.

SLS Exercise Instrumentation

Subjects performed the resisted and controlled SLS task in a lower extremity perturbation device that has been described previously.[61, 77-78] The device consisted of a modified standing frame with a rack and pinion gearbox mounted to the frame between the two side supports. At one end of the horizontal shaft of the gearbox was a padded vertical plate, the height of which was adjusted via a spring-loaded mechanism (Figure 2.1). This prevented shearing of the knee within the pad during flexion and extension of the knee. During the SLS exercise, the plate was positioned against the anterior aspect of the knee with the patella located at the center of the pad. The lower extremity was secured to the pad by a velcro strap that extended around the popliteal fossa. As the SLS was performed, knee flexion and extension was translated into horizontal linear displacement of the shaft of the gear system. The horizontal position of the knee joint was measured by a precision potentiometer mounted to the shaft of the pinion gear of the device. In a previous substudy of subjects performed in this lab,[81] kinematic data of the lower extremity during the SLS task were analyzed using a video motion analysis system. The angular motion and velocity of the knee was found to have an excellent correlation ($r = 0.99$) with the linear displacement of the horizontal shaft.

An electromagnetic braking system that was mounted to the shaft of the pinion gear provided the resistance during the SLS exercise. Resistance level was controlled by a custom software program that provided a constant current to the brake through the analog output channel of the computer’s A-D board. The brake had a resistance range of 0 – 45 pounds which remained constant as the horizontal shaft of the device was displaced forward and backward during the exercise, thus providing resistance to both the flexion and extension phases of the SLS task. The software that controlled the braking system also allowed for the instantaneous release of the resistance when an event marker was received at the digital input of the computer’s A-D board. The user could specify the duration of the release of resistance at the time that the program was initiated.

The timing of the drop in resistance level was dependent upon the position of the horizontal shaft of the device. Output of the potentiometer was cabled to a Schmitt trigger set to produce an event marker (5 V square-wave TTL pulse, duration = 30 ms) whenever a threshold voltage of the potentiometer was exceeded. This threshold was adjusted so that the release of resistance occurred at a consistent point in the range of motion. The duration of the perturbation lasted for 500 ms and occurred at approximately 1/3 of the distance into knee flexion (about 400 ms). Following the period of release, the resistance of the brake returned immediately to the previously specified level.

Perturbations (release of the brake) were delivered during 2 of the 10 repetitions within each set of the SLS exercise. To determine the repetitions during which a release of resistance occurred, the computer program generated two random numbers between 1 and 10 prior to each repetition of the program. The brake was then automatically released to produce a perturbation during the exercise, during those repetitions corresponding to the two random numbers. Resistance was maintained at a constant level throughout the rest of the exercise, with no perturbations occurring during the other 8 repetitions.

Light Touch Force

The amount of resistance acting horizontally at the knee during the SLS task is quite substantial and necessitates some light touch support, so subjects were permitted to place one finger for support on a load sensor (Wafer Load cell, Model 872, Loadstar Sensor Inc) mounted on the side of the device contralateral to the test leg. The output from the load cell was passed through a differential amplifier into a voltage-controlled oscillator and then amplified with an audio amplifier to produce a sound signal that was delivered to the subjects via headphones. This device provided a progressive auditory warning as the applied force approached 5 N. The amount of touch force was found to be similar between the groups. Though the average peak touch was below 3 N for subjects in each group, several individuals did have peak touch that exceeded this level. Subjects 3 and 5 of the BVL group and Subjects 2 and 4 of the control group consistently had peak touch greater than 3 N.

Electromyography Recordings

Bipolar silver-silver chloride surface electromyography (EMG) electrodes (8 mm in diameter, fixed inter-contact distance of 20 mm) with internal pre-amplification (gain*35) were used to record the activity of vastus medialis oblique (VM), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH), and lateral hamstrings (LH) during the SLS task. The gain of the EMG signals was also further adjustable at the main amplifier (gain*10,000). This amplifier used a high impedance circuit with a common mode rejection ratio of 87 dB at 60 Hz, and a bandwidth of 15 – 4000 Hz. EMG electrodes for the quadriceps were placed at 4/5 the distance along a line from the anterior superior iliac spine (ASIS) to the medial joint line for VM, at 1/2 the distance along a line from the ASIS to the superior pole of the patella for RF, and at 2/3 the distance along a line from the ASIS to the lateral joint line for VL.[119] For the hamstring muscles, EMG electrodes were placed at 50% of the distance along a line from the ischial tuberosity to the medial and lateral femoral condyles for the MH and LH, respectively.[80]

Maximum Voluntary Isometric Contractions

A dynamometer was used to obtain maximum voluntary isometric contractions (MVICs) of each muscle to normalize EMG data. Subjects were seated on the chair with hip and knee in 90 degrees of flexion. The pelvis, hip, thigh and foot were firmly secured to minimize other movement. Three MVICs of the quadriceps and hamstrings were obtained in this position. No significant difference in the torque during the quadriceps and hamstrings MVIC was seen, suggesting that the two groups generated similar MVICs.

Experimental Setup

The SLS task was performed with the subject in unilateral stance on a strain gauge force plate (24 inches x 15 inches), with the other leg unsupported and flexed. The subject was positioned with the supporting foot placed on the center of the force plate. The foot was then rotated slightly outwards so that the great toe was approximately 5 cm away from the center of the step. Foot position was traced for each subject and used to ensure that the foot remained in the same position throughout the experiment and across days. Subjects were permitted to place one finger of the contralateral hand on the force platform mounted on the contralateral side of the device. Subjects were instructed to place very little load through their fingers with a progressively loud auditory warning given through the headphones as the force increased. The knee pad on the horizontal shaft was aligned with the patella and the knee secured to the pad with straps around the popliteal fossa. Subjects were instructed to follow, as accurately as possible, a sinusoidal target (0.4 Hz) projected on a 17-inch computer screen placed approximately 40 cm in front of them. The real time visual display of the output of the potentiometer also indicated the horizontal displacement of the shaft of the exercise device, providing feedback regarding the subject's ability to accurately follow the target. The peak-to-peak amplitude of this sine wave form corresponded to 15 cm of linear displacement of the horizontal shaft. Previous work has shown that this 15 cm displacement corresponds to approximately 30 degrees of knee flexion.[80]

Experimental Protocol

Training Session (Day 1)

Each subject attended two sessions separated by 24-48 hours. The first portion of the first session allowed the subjects to become familiar with the device and to ensure that the SLS task was performed correctly. Before the start of the training session, subjects performed a standardized warm-up protocol on an exercise bike for 5 minutes. After completing the warm up, bipolar surface EMG electrodes were attached to the skin overlying the VM, RF, VL, MH, and LH muscles with double-sided adhesive washers and secured with adhesive tape. Subjects were then positioned to perform MVICs of knee extension and flexion to normalize the EMG signals obtained during the SLS task.

Subjects performed three maximum isometric contractions in extension followed by three maximum contractions in flexion while seated with the hip and knee in 90 degrees of flexion. Each contraction was held for 3 seconds with a 1-minute rest between contractions. The trial with the highest recorded peak EMG was used to normalize the activity of each muscle during the resisted SLS task.

During the initial portion of the training session, the subjects learned to follow the target pattern seen on the computer screen with knee displacement and the resistance of the device set to 12% of body weight. Subjects were instructed to avoid leaning or rotating during the task and were given verbal cues if there were any deviations in the technique or form of exercise during the learning sessions. Ten repetitions were considered one set. The SLS task was first performed with eyes open (EO) for 5 sets of 10 repetitions until they were accustomed to the task. Subjects who did not reach the minimum criterion for learning (within 5 cm of endpoint error for the EO condition) at the end of training were considered ineligible for the testing sessions. All subjects included in the study met the minimum learning criterion.

Following the initial training, testing was performed with the device resistance set to 17% body weight under three conditions of visual feedback: (1) EO, (2) EO/No Template (NT), and (3) eyes closed (EC). In the EO condition, the subject obtained continuous visual feedback of knee displacement and the sinusoidal target on the screen. In the NT condition, the subject kept the eyes open but the template was turned off, so no visual feedback regarding knee position was provided. In the EC condition, the subject kept the eyes closed during performance of the task. One-minute rest intervals separated each set of 10 repetitions to avoid fatigue during the training session. After the initial 5 sets were performed with EO, the remainder of the session consisted of sets of 10 repetitions of the SLS exercise performed in the following order: 1 set EO→ 1 set NT→ 1 set EC→ 1 set EO →2 sets EC →1 set EO →2 sets NT – 1 set EC – 1 set EO, 1 set NT. EC and NT trials were interspersed between EO trials to help acquisition and consolidation of learning. Perturbations were delivered during 20% of the repetitions. The threshold level of the Schmitt trigger was adjusted such that the perturbations occurred when the horizontal shaft of the exercise device translated 5 cm from the start of the flexion phase of the exercise. When the perturbations were encountered, subjects were instructed to react as quickly as possible to restore the knee to its original trajectory and minimize the error between the target waveform and the actual knee position trace. Subjects were asked to rate their perceived exertion (RPE) on a Borg Scale after every other set of the exercise. This was used to monitor fatigue during the experimental protocol.

Testing Session (Day 2)

The testing on Day 2 was conducted 24-48 hours after Day 1. Prior to data collection, subjects performed a standardized warm-up protocol on an exercise bike for 5 minutes. After completing the warm up, bipolar surface EMG electrodes were attached and MVICs were completed as described earlier.

After the obtaining the MVICs, the subjects were placed in the exercise device in the same manner as for the previous session. Subjects then performed a series of the SLS task in the EO, NT, and EC conditions. Two sets of 10 repetitions for each condition were performed. A 1-minute rest interval separated each set of 10 repetitions. The order of retention trials was as follows: 1 set EO→ 1 set NT → 1 set EC→ 1 set EO → 1 set EC → 1 set NT. Subjects then performed the trials while standing on the foam cushion in the following order: 2 sets EO at 12% resistance (no perturbations); the remainder at 17% resistance with unexpected perturbations: 1 set EO →1 set NT→1 set EC→1 set EO→1 set EC→1 set NT.

<u>Data Reduction</u>

All experimental data were collected online and subsequently analyzed using Datapac 2K2 software (version 3.14; Run Technologies Inc., CA). Electromyographic activity of the quadriceps and hamstring muscles was sampled at a rate of 2000 Hz.

MVICs were analyzed by finding the peak RMS EMG during each of the three contractions and calculating the mean RMG EMG for 200 ms on either side of the peak EMG. All EMG derivates are expressed as percentage of MVIC on the corresponding days.

All other signals (linear potentiometer, target waveform, brake and touch force) were digitized at a rate of 1000 Hz. For the modCTSIB and single leg balance assessment, movements of the center of pressure (COP) in the frontal and sagittal plane were sampled at a frequency of 500 Hz. Antero-posterior and medio-lateral amplitude of the COP was analyzed for each subject.

During both perturbed and nonperturbed repetitions of the task, a TTL pulse was recorded from the Schmitt trigger as the potentiometer attached to rack and pinion gear of the exercise device passes the threshold voltage. For perturbed trials, the TTL pulse corresponded to the onset of the perturbation. For nonperturbed trials, the TTL pulse served as a marker to indicate the point in the range of motion where the perturbation would have occurred if the software had allowed for it.

Dependant Variables

Dependant variables analyzed in this study included:

1. Absolute Error (absolute value of the difference between the target waveform and the actual position of the knee): To obtain absolute error of performance during the learning and retention trials, the sinusoidal target was subtracted from the linear knee displacement to calculate error. The error signal was then rectified and averaged in 10% bins within each flexion and extension cycle of the SLS exercise. The time frame 50-150 ms post-perturbation was analyzed as well as overall absolute error.

2. Variable Error (a measure of precision in tracking the visual target): Variable error is obtained by calculating the standard deviation about the error signal of each subject. This was also averaged in 10% bins within each flexion and extension cycle of

the SLS exercise. The time frame 50-150 ms post-perturbation was analyzed as well as overall absolute error.

3. Endpoint Error (overshoot): The endpoint error is a measure of the deviation of the endpoint of the subject's flexion from the endpoint of sine wave template to calculate the overshoots of the target during the perturbation trials.

4. Peak Velocity: The linear velocity of the knee was obtained by differentiating the displacement signal. The peak of this velocity signal was measured in the time bins 200 ms prior to perturbation, 0-200 ms, and 200 – 400 ms post perturbation to examine anticipatory (Pre), reflex (Post1), and volitional (Post2) phases respectively.

5. Cycle EMG activity: Muscle activity of the VM, RF, VL, MH and LH were RMS processed with a time constant of 50 ms and then averaged in 10% bins for each flexion and extension cycle of the single leg squat task to recognize synergies that were being utilized to perform the task. All values are expressed as a percentage of MVIC.

6. Mean EMG: Muscle activity (expressed as % MVIC) in the time bins 200 ms prior to perturbation, 0-200 ms and 200 – 400 ms post perturbation was analyzed to examine anticipatory (Pre), reflex (Post 1) and volitional (Post 2) activity respectively. In the no perturbation trials, the same bins were analyzed with respect to the time when the perturbation would have occurred.

7. Normalized LLR activity: To filter out the effect of background activity in the mean long latency response, the normalized muscle activity of the quadriceps and hamstrings muscles in the time bin 50 – 150 ms corresponding to the onset of perturbation was calculated for each subject using the following formula:

$$\text{Normalized LLR} = \frac{\text{Mean EMG of Perturbation Trials} - \text{Mean EMG of Unperturbed Trials}}{\text{Mean EMG of Unperturbed Trials}}$$

Statistical Analysis

All data were analyzed using a three-factor repeated measures analysis of variance, and a separate analysis was performed for each of the dependant variables. Flexion and extension cycles were analyzed separately. Perturbed and nonperturbed events were analyzed separately. The three within-subject factors were Condition (Eyes Open, No Template, and Eyes Closed), Day (Day 1, Day 2) and Group (BVL, Ctrl). To compare performance on firm surface versus foam, a separate three-factor repeated measures analysis of variance was performed with the following factors: Condition (Eyes Open, No Template, and Eyes Closed), Surface (Firm, Foam), and Group (BVL, Ctrl).

Absolute error (AE) and variable error (VE) measures were used to quantify performance. Muscle activity during the SLS task was quantified using EMG from each of the five muscles sampled. When necessary, significant main effects and interactions were further analyzed using Bonferroni correction. The level of significance for all tests was established at $\alpha \leq 0.05$. All statistical analyses were performed using SPSS software (version 17.0).

Results

Subjects

Demographic data of the subjects are shown in Tables 3.1-3.2. Five subjects with BVL were included in the study and compared to five gender- and age-matched controls. A t-test was performed to compare differences in descriptive variables between the included BVLs and controls. No significant difference was found between age, height, or weight when comparing the groups. The Short Form 36 Medical Outcome Questionnaire was used to compare the subject’s perception of general health (Figure 3.2). Subjects with BVL scored significantly lower in the general health ($p = .04$), vitality ($p = .03$), and social function ($p = .03$) areas compared to the controls. Control subjects reported no dizziness or balance deficits with the dizziness handicap inventory (DHI), while BVL subjects scores ranged from 20-52 points out of 100 possible (Figure 3.3). No significant differences were found between groups on the KOOS, IKDC, Marx Activity Scale, Tegner Activity Scale, Five Time sit to Stand, or Timed Up and Go (Table 3.1). However, BVL subjects scored lower than controls on the Duke Activity Skills Index ($p = .008$). Static balance times for all subjects are shown in Tables 3.4 and 3.5. For the subset of BVL and control subjects tested while strapped in the podium, all were able to stand for 30 sec with eyes closed, when they were unable to do so when tested standing unsecured.

Effect of Vision, Surface, and Training
on Accuracy of Performance

The accuracy of tracking the target waveform was assessed during each repetition of the SLS exercise by calculating the average absolute and variable errors of both the flexion and extension phases, as well as the endpoint error at the transition between flexion and extension phases. Nonperturbed and perturbed events were analyzed separately. Comparing these errors across Days 1 and 2 as well as during eyes open (EO), no template (NT), and eyes closed (EC) on firm and foam surfaces provided information regarding learning and the effect of vision and surface, respectively. Only 3 of the 5 subjects with BVL could perform the SLS task with eyes closed on the firm surface, and 2 could perform the task with eyes closed on the foam surface. All control subjects could perform the task with eyes closed on both surfaces.

Absolute Error

The mean absolute errors (AE) for the nonperturbed and perturbed events of the SLS task from Days 1-2 performed in each visual condition are shown in Figures 3.4-3.9. For the nonperturbed events, the greatest errors were seen in the mid portions of the flexion and extension phases of the task during the EO sets and in the earlier intervals for the NT and EC sets. The greatest AE during the perturbed events extended from the mid to the late portions of the flexion phase. A three-way repeated measures ANOVA (Condition x Group x Day) on the absolute errors for both types of events showed a significant main effect of Condition for both the nonperturbed and the perturbed events ($p<.001$) during flexion and extension phases. AE was lower in the EO condition than in the NT ($p < .001$) or EC ($p <.001$) conditions for both phases. As hypothesized, AE during EC was greater than when visual feedback was provided for both groups. There was no difference between the NT and EC conditions. There was also no difference between Groups or between Days.

A three-way repeated measures ANOVA (Condition x Group x Surface) on the absolute error showed a significant interaction effect of Surface x Group ($p = .025$). On the foam surface, AE was less for BVLs than Controls (Ctrls) ($p = .004$) during the flexion phase of the no perturbation events; on the solid surface, there was no difference between groups. This surface effect was not seen during extension, or during the perturbation events of either phase.

Variable Error

Variable error (VE) showed that subjects were very consistent in the performance of the SLS task with the greatest variability in errors occurring as subjects transitioned between flexion and extension. For the nonperturbed and perturbed events on firm and foam surfaces, AE during EO was less than both NT and EC Conditions ($p< .001$) during both phases while there was no difference between NT and EC Conditions (Figures 3.10-3.11). As with AE, VE during EC was greater than when visual feedback was provided for nonperturbed events.

A three-way repeated measures ANOVA (Condition x Group x Surface) on VE showed a significant interaction effect of Surface x Group ($p = .03$). On the foam surface, VE was less for BVLs than for Ctrls ($p = .035$) during the extension phase of the no perturbation events; on the solid surface, there was no difference between groups. This surface effect was not seen during flexion or during the perturbation events of either phase.

Endpoint Error

Endpoint error (Epe) measures the deviation of the endpoint of the subject's flexion and was determined for both the nonperturbed and perturbed events across days and on different surfaces (Figures 3.12-3.13). Epe did not significantly change from Day 1 to Day 2 for either group, indicating neither BVL nor Control subjects were unable to lessen overshoot over the testing days. However, during perturbation events, there was a significant main effect of Group with BVLs showing increased error compared to controls ($p = .004$).

A three-way repeated measures ANOVA (Condition x Group x Surface) on Epe showed a significant effect of Surface. Epe was greater on firm versus foam surface ($p<.001$). There was also a significant interaction between Surface and Group, with BVL endpoint error greater than Controls on a firm surface ($p = .002$), but less than Controls on the foam surface ($p =.009$).

As there were no significant differences found between the NT and EC conditions, results from this point will focus on the EO and EC conditions.

<u>Effect of Training and Vision on Peak Velocity</u>

Peak linear velocity of the knee was measured within the anticipatory (Pre: -200-0 ms), reflex (Post 1: 0-200 ms), and volitional (Post 2: 200-400 ms) time bins (Figure 3.14). A three-way repeated measures ANOVA (Condition x Group x Day) on peak velocity at each of these bins showed significant differences between conditions. Post 2 velocity of the nonperturbed events was greater in the EC condition compared to EO ($p = .004$). For perturbed events, this was the case during the Post 1 ($p = .023$) time bin, with no difference at Post 2.

A three-way repeated measures ANOVA (Condition x Group x Surface) on this variable found significant main effects of Condition ($p <.001$) and Surface ($p = .019$). Both groups showed increased velocity on foam compared to firm in the anticipatory time bin as well ($p = .019$ NP events, $p = .007$ Perturbed events). Velocity was increased in the EC condition compared to EO ($p = .001$) and when on foam ($p = .019$) in the Post 1 time bin of nonperturbed events. Analysis at Post 1 found a significant main effect of Condition ($p = .002$) and Surface ($p = .048$) for nonperturbed events and for perturbation events (Condition $p = .034$; Surface $p = .007$).

Effect of Training, Vision, and Surface on Muscle Activation Patterns

Activity of the vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH), and lateral hamstrings (LH) during the SLS task was determined by averaging EMG activity in 10% interval bins of knee displacement (Figures 3.15-3.21). Flexion and extension phases were analyzed separately, and muscle activity was expressed as a percentage of maximum voluntary isometric contraction (MVIC). A three-way repeated measures ANOVA (Condition x Group x Day) was used to examine the influence of each of these factors on muscle activity during the SLS task.

No Group or Day effects were seen for any of the muscles. No significant effects were seen for the no perturbation events. There was a significant effect of Condition during the extension phase of the perturbed events for both VM ($p =.002$) and VL ($p <.001$). Post hoc testing revealed muscle activity was increased in the EC condition compared to EO (VM: $p =.018$; VL: $p = .009$). RF, MH, and LH were not modulated by vision during flexion or extension of either event type.

A three-way repeated measures ANOVA (Condition x Group x Surface) on this variable revealed a significant main effect of Surface only for VL. Post hoc testing revealed that VL activity was greater when on the foam cushion compared to the firm surface for both flexion ($p = .002$) and extension ($p = .013$) phases of perturbed events. Otherwise, there were no other main effects seen for the other quadriceps or hamstring muscles.

Normalized Long Latency Responses

Difference in the EMG activity between the perturbed and nonperturbed trials in the 50-150 ms time bin normalized to the background activity (nonperturbed trials) was compared between conditions and over days (Figure 3.22-3.23). A three-way repeated measures ANOVA (Condition x Group x Day) on the normalized long latency response revealed some main and interaction effects. There was no effect of Day, so data from the firm surface on D1 and D2 were pooled and then differences compared between groups. For MH, the response was greater in the EC condition compared to EO ($p = .034$). A significant Group x Condition interaction was found for RF ($p = .043$). With EC, RF LLR was less for BVLs than for Controls. It is interesting to note that at the $p<.1$ level, there were several other effects noted. For RF, a Group x Condition interaction was seen; with EO, BVL LLR was less than Ctrl ($p = .072$). Also at this level, VL showed a main effect of Condition with VL LLR greater in the EC condition compared to EO ($p = .07$). MH also showed a Group x Condition interaction; with EC, BVL LLR was greater than Ctrls ($p = .093$).

A three-way repeated measures ANOVA (Condition x Group x Surface) on Nm LLR found no significant main effects or interactions.

Time of Peak LLR

The time at which the Peak LLR occurred was determined for all conditions, days, and both groups (Figure 3.24). A three-way repeated measures ANOVA (Condition x Group x Day) on this variable found a significant effect of Group ($p = .030$) for VL. Peak time was lower for BVL subjects. A significant interaction of Condition x Day ($p = .014$) was found for RF. In the EC condition, peak time for RF was higher on Day 2 ($p = .017$). A three-way repeated measures ANOVA (Condition x Group x Surface) on this variable was also performed, with no significant effects or interactions found.

Muscle Activity in the Anticipatory, Reflex,

and Volitional Time Bins

Muscle activity of the quadriceps and hamstrings was analyzed in the Pre, Post 1, and Post 2 time bins during the perturbed and nonperturbed trials (Figures 3.28-3.29). A three-way repeated measures ANOVA (Condition x Group x Day) found no significant effects or interactions for VM or RF over days. However, during the anticipatory time bin, there was a significant main effect of Group. VL activity was greater for BVLs than Controls for VL ($p = .036$) for both no perturbation and perturbation events. During the reflex and volitional time bins for MH, there was a significant main effect of Day; Day 2 was greater than Day 1 during the Post 1 ($p =.035$) and Post 2 ($p = .020$) time bins. LH was also modulated by Condition. During the anticipatory ($p = .027$) and reflex time bins ($p = .024$), LH activity was greater with EO for nonperturbed events as well as for perturbed events ($p = .012$, $p = .042$, respectively). This was also the case for the Post 2 time bin of perturbed events ($p =.042$) and the volitional time bin of the nonperturbed events ($p = .003$).

A three-way repeated measures ANOVA (Condition x Group x Surface) on this variable found a significant main effect of Surface for the volitional time bin of VM during perturbed events ($p = .026$). This was also the case for VL during both nonperturbed and perturbed events of the reflex (nonperturbed: $p = .011$, perturbed: $p = .050$) and volitional (nonperturbed: $p = .011$; perturbed: $p = .007$) time bins. Post hoc testing revealed that in each case, activity was greater on the foam versus the firm surface.

Table 3.1 Characteristics of BVL and control subjects

	BVL (n=5)	Ctrl (n=5)
Age (yrs)	53.6 (19.3)	53.2 (18.7)
Weight (lb)	149.7 (30.4)	157.8 (47.6)
Height (cm)	163.3 (7.2)	162.3 (7.2)
IKDC score	80.3 (17.5)	94.5 (7.9)
Marx Activity Scale	3.6 (4.9)	5.6 (6.1)
Tegner Activity Scale (current)	3.8 (1.3)	5.6 (2.4)
Baecke (total score)	36.6 (4.0)	46.3 (9.6)
Duke Activity Skills Index (DASI)*	48.2 (6.3)	58.2 (0)
Five Time Sit to Stand (sec) (best trial)	8.2 (1.6)	9.0 (1.2)
Timed Up and Go (sec) (best trial)	8.6 (1.1)	7.7 (1.1)

Note: Values are Mean (SD)

* indicates significant difference between Groups

Table 3.2 Characteristics of BVL subjects

	Gender	Age	BVL Cause	BVL duration	DASI score (max = 58.2)
BVL 01	Male	23	Autoimmune	4 yrs	58.2 (ec foam)
BVL 02	Female	65	Gentamycin	10 yrs	44.7
BVL 03	Female	47	Gentamycin	8 yrs	50.7 (ec foam)
BVL 04	Female	62	Gentamycin	16 yrs	42.7 (ec firm)
BVL 05	Female	71	Gentamycin	10 yrs	44.7

Note: Table includes which subjects were able to perform the SLS with eyes closed (ec) on the foam surface and the one who could perform eyes closed on firm but not on foam surface.

DASI = Duke Activity Skills Index

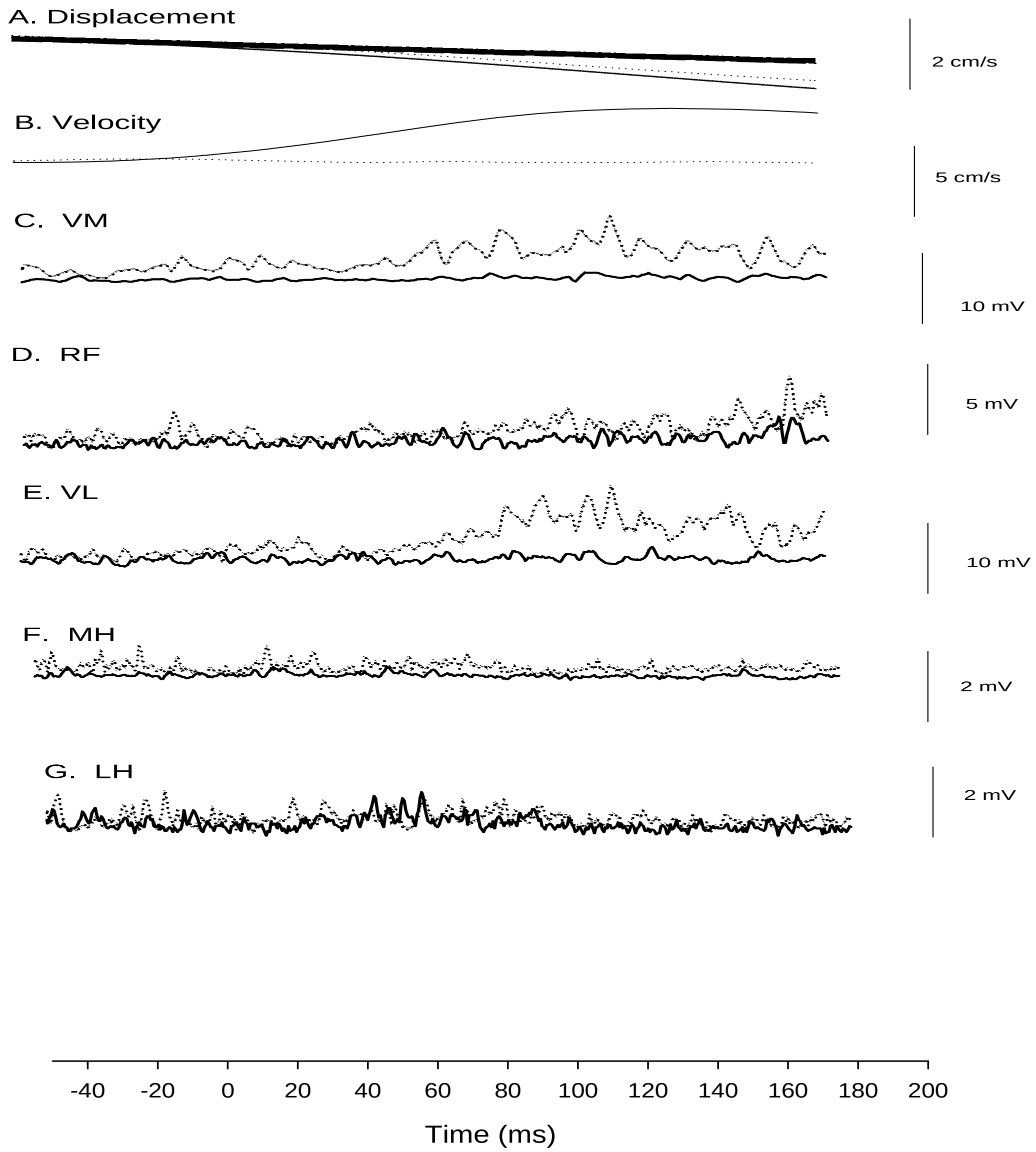

Figure 3.1 Representative example of **a**. linear displacement, **b**. linear velocity, EMG traces of **c**. vastus medialis, **d**. rectus femoris, **e**. vastus lateralis, **g**. medial hamstrings and **h**. lateral hamstrings of a single subject and the matched control (average of 6 perturbed trials each). The *Thick solid* line represents the template, the *Dotted* line represents the Control subject and the *Thin Solid* lines is the BVL subject. Time frame is from -50 to 200 ms post perturbation. X axis represents time (ms); the release of the brake occurred at 0 ms. EMG traces are root mean square averaged.

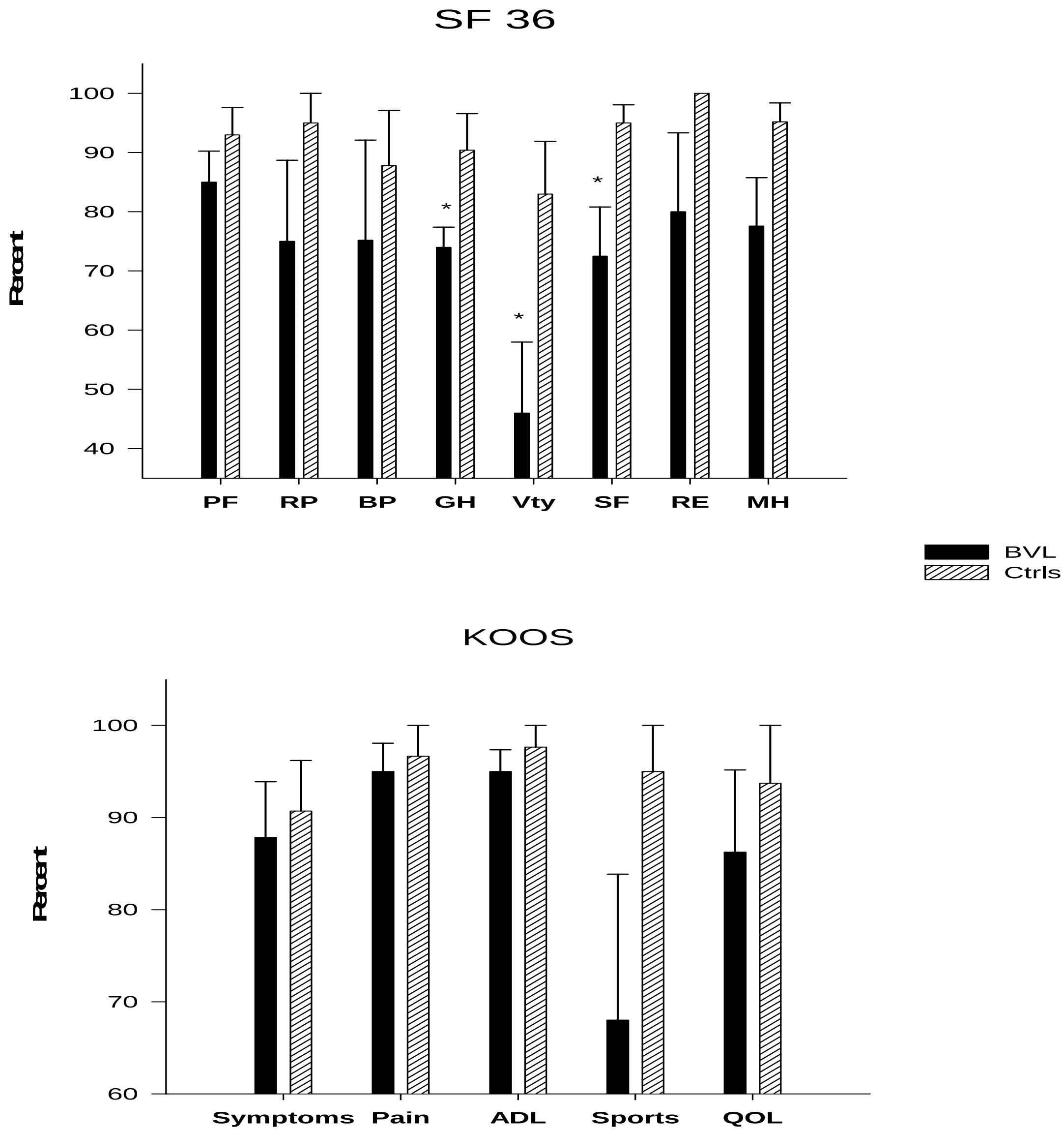

Figure 3.2 Mean SF36 and KOOS scores of BVLs (*dark bars*) and Ctrls (*striped bars*). The X-axis represents the various domains of the SF-36 and KOOS (PF – Physical Function, RP – Role Physical, BP – Bodily Pain, GH - General Health, Vty –Vitality, SF – Social Function, RE – Role Emotional, MH – Mental Health. Values are means ± SE of all subjects.

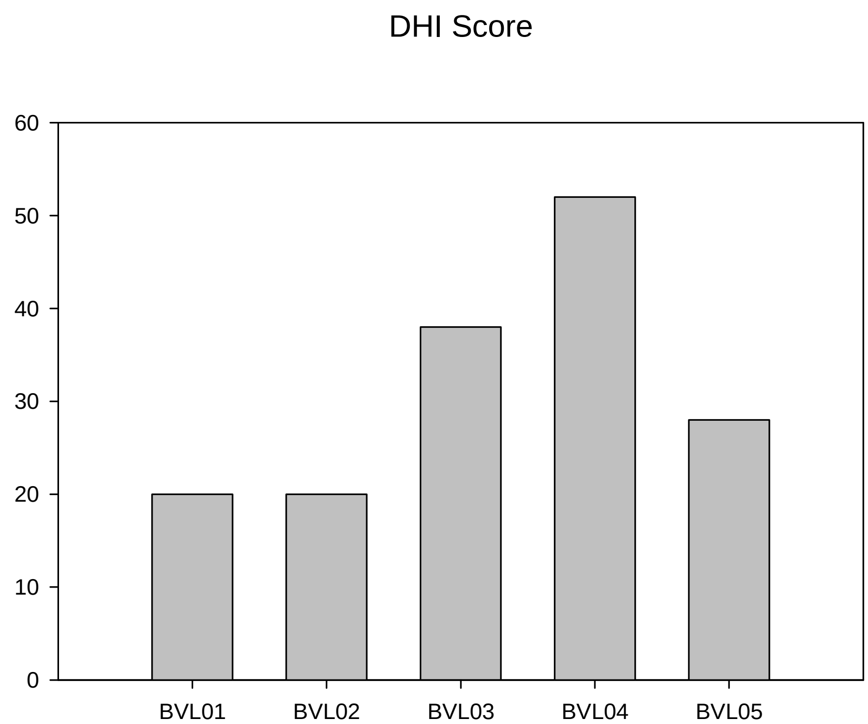

Figure 3.3 Dizziness Handicap Inventory (DHI) scores for each BVL subject

Table 3.3 Time (in sec) for each subject's static balance while on firm surface when free and when strapped into podium

	FT EO	FT EC	SLSEO	SLSEC	SLSEC (in podium)
BVL01	30	30	30	3	NT
BVL02	30	30	3	0	NT
BVL03	30	30	30	0	NT
BVL04	30	30	7	0	30
BVL05	30	30	18	0	30
Ctrl01	30	30	30	30	NT
Ctrl02	30	30	30	8	NT
Ctrl03	30	30	30	20	NT
Ctrl04	30	30	30	5	30
Ctrl05	30	30	30	11	30

Note: FT = Feet Together; SLS = Single Leg Stance; EO = Eyes Open; EC = Eyes Closed

Table 3.4 Time (in sec) for each subject's static balance while on foam surface when free and when strapped into podium

	FT EO	FT EC	SLSEO	SLSEC	SLSEC (in podium)
BVL01	30	5	9	1	NT
BVL02	30	0	0	0	NT
BVL03	30	0	30	0	NT
BVL04	30	0	0	0	30
BVL05	30	0	0	0	30
Ctrl01	30	30	30	30	NT
Ctrl02	30	30	24	3	NT
Ctrl03	30	30	30	7	NT
Ctrl04	30	30	30	0	30
Ctrl05	30	30	30	6	30

Note: FT = Feet Together; SLS = Single Leg Stance; EO = Eyes Open; EC = Eyes Closed)

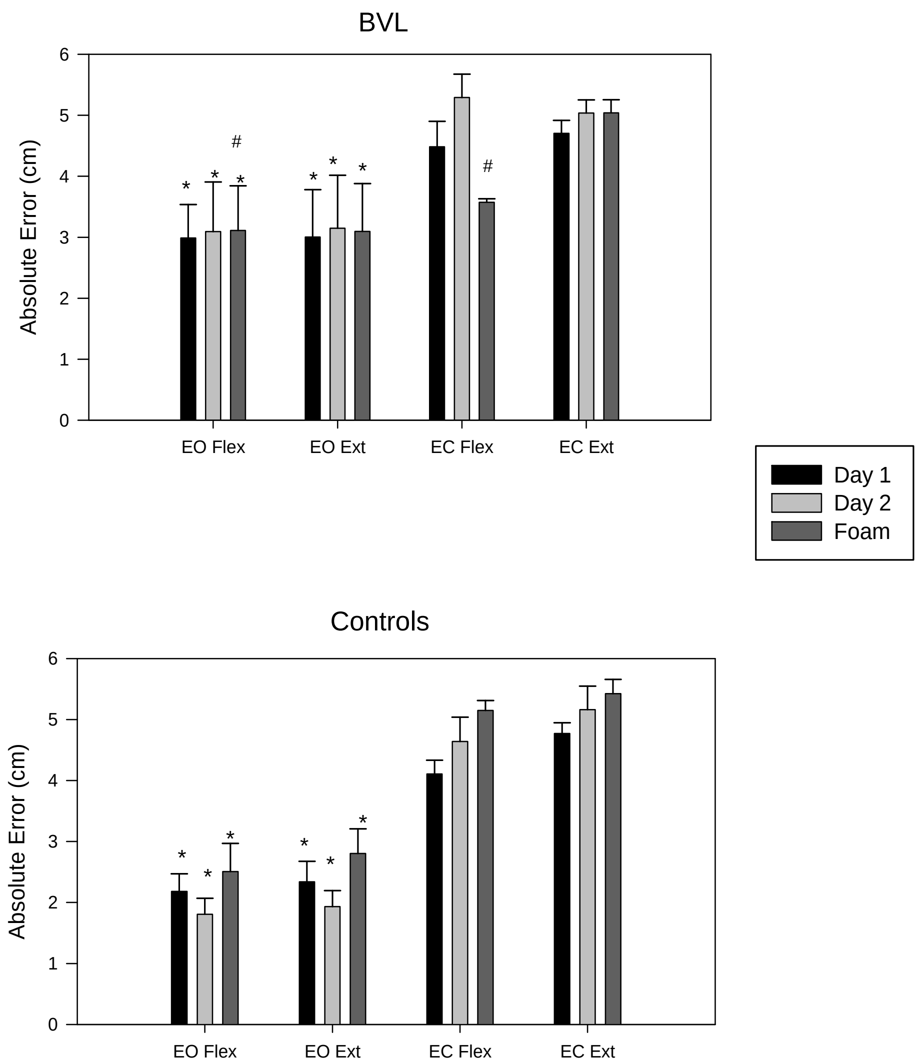

Figure 3.4 Mean absolute error during the flexion and extension phases of the SLS task in the Eyes Open (EO) and Eyes Closed (EC) conditions of the No Perturbation events. The vertical bars represent the average errors over Days 1 and 2, and when on the foam surface. The Y-axis represents absolute error of performance in cm. Values are means $\pm$ SE. Control = 5 sub for EO and EC; BVL = 5 sub EO, 3 sub EC on firm, 2 sub EC on foam.

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* indicates significant difference from Eyes Closed Condition (EC)

\# indicates significant difference between Groups on Foam Surface

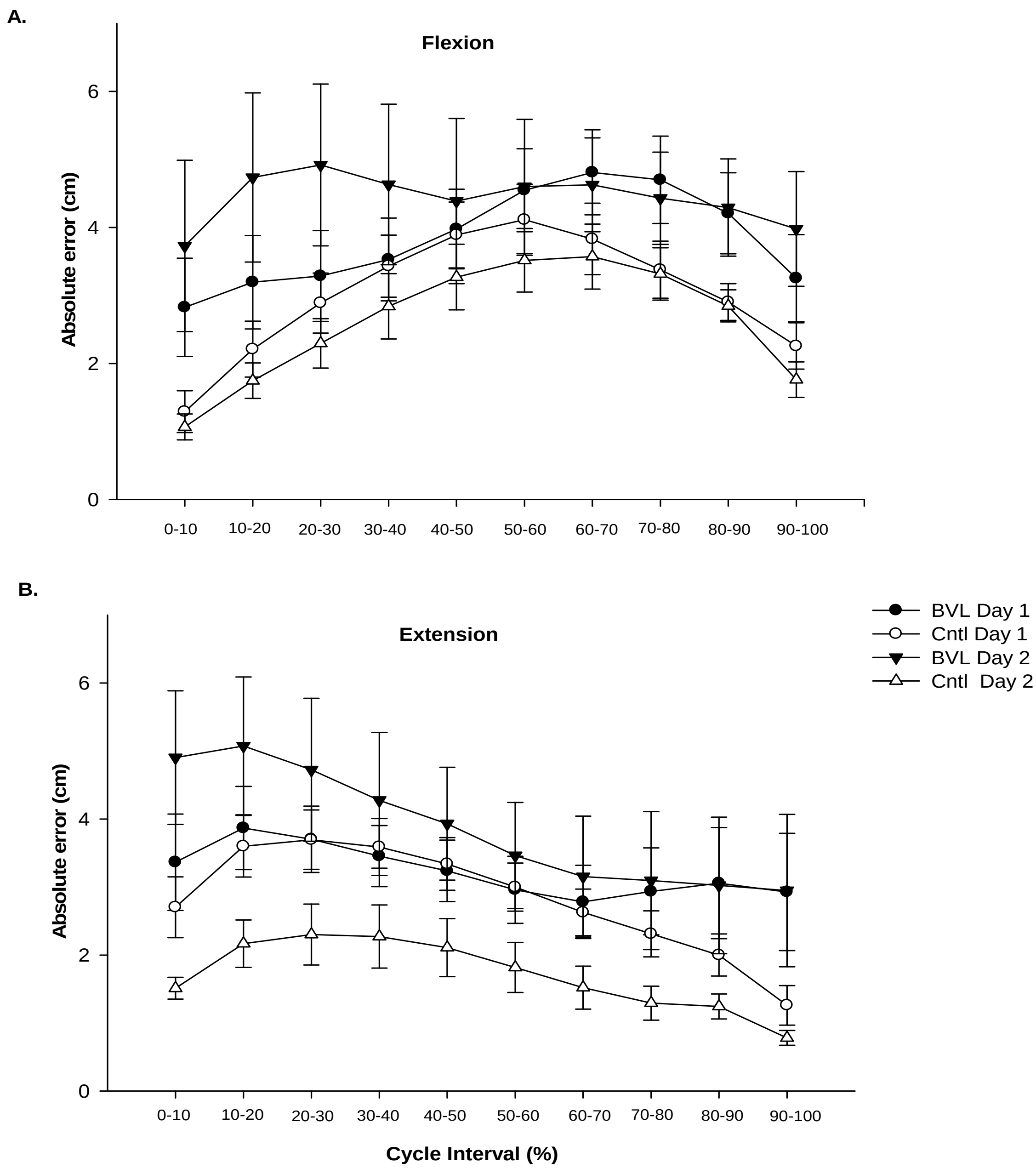

Figure 3.5 Absolute errors of performance in the Eyes Open (EO) condition Perturbation Events within the flexion (A) and extension (B) phases of the single leg squat task for Day 1 *(circles)* and Day 2 *(triangles)*. Mean error values are presented for BVLs (*dark symbols*) and Controls (*open symbols*). Values are means ± SE. Ctrl = 5 sub; BVL = 5 sub.

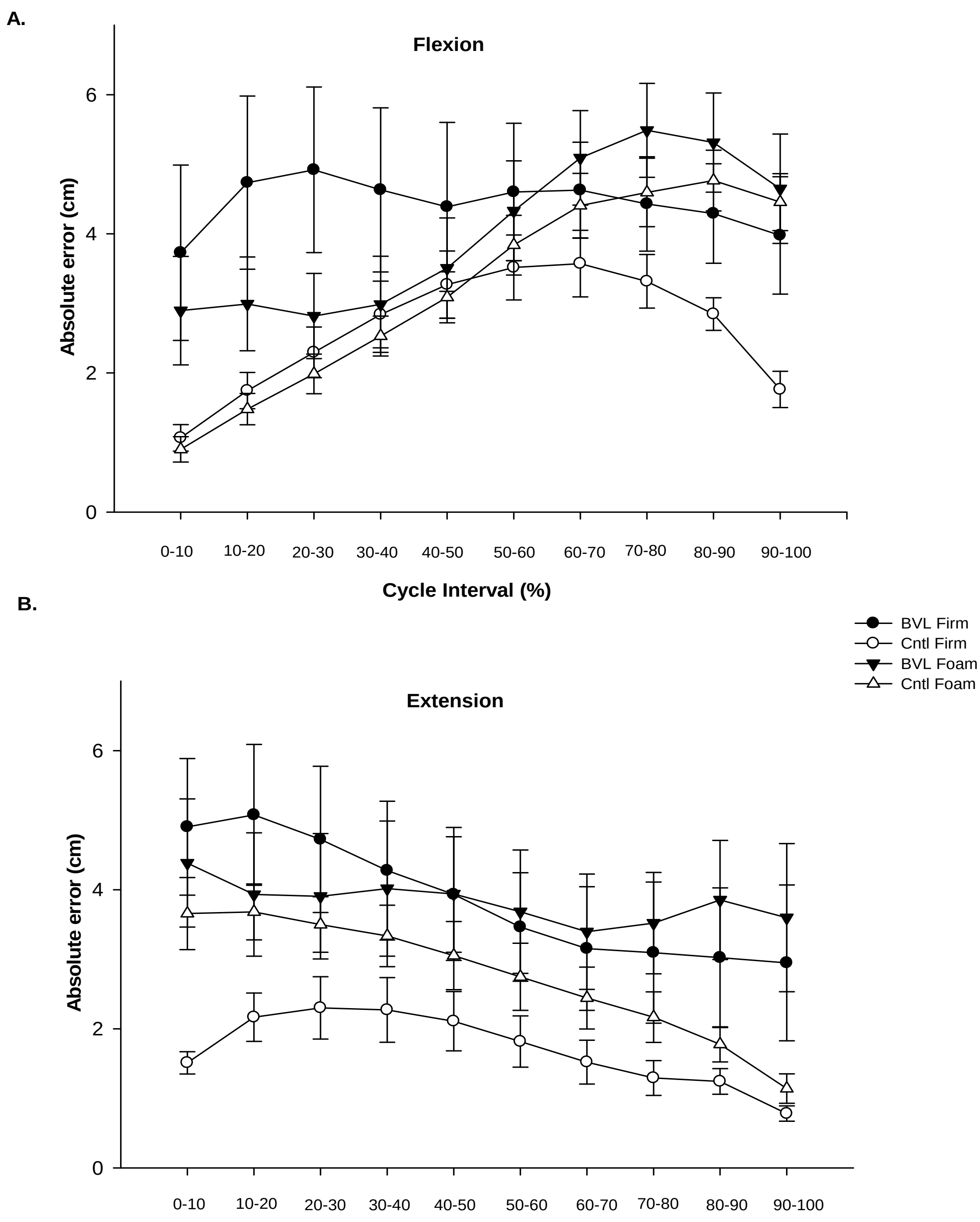

Figure 3.6 Absolute errors of performance in the Eyes Open (EO) condition Perturbation Events within the flexion (A) and extension (B) phases of the single leg squat task on the Firm *(circles)* and Foam *(triangles)* surface. Mean error values are presented for BVLs (*dark symbols*) and Controls (*open symbols*). Values are means ± SE. Ctrl = 5 sub; BVL 5 sub.

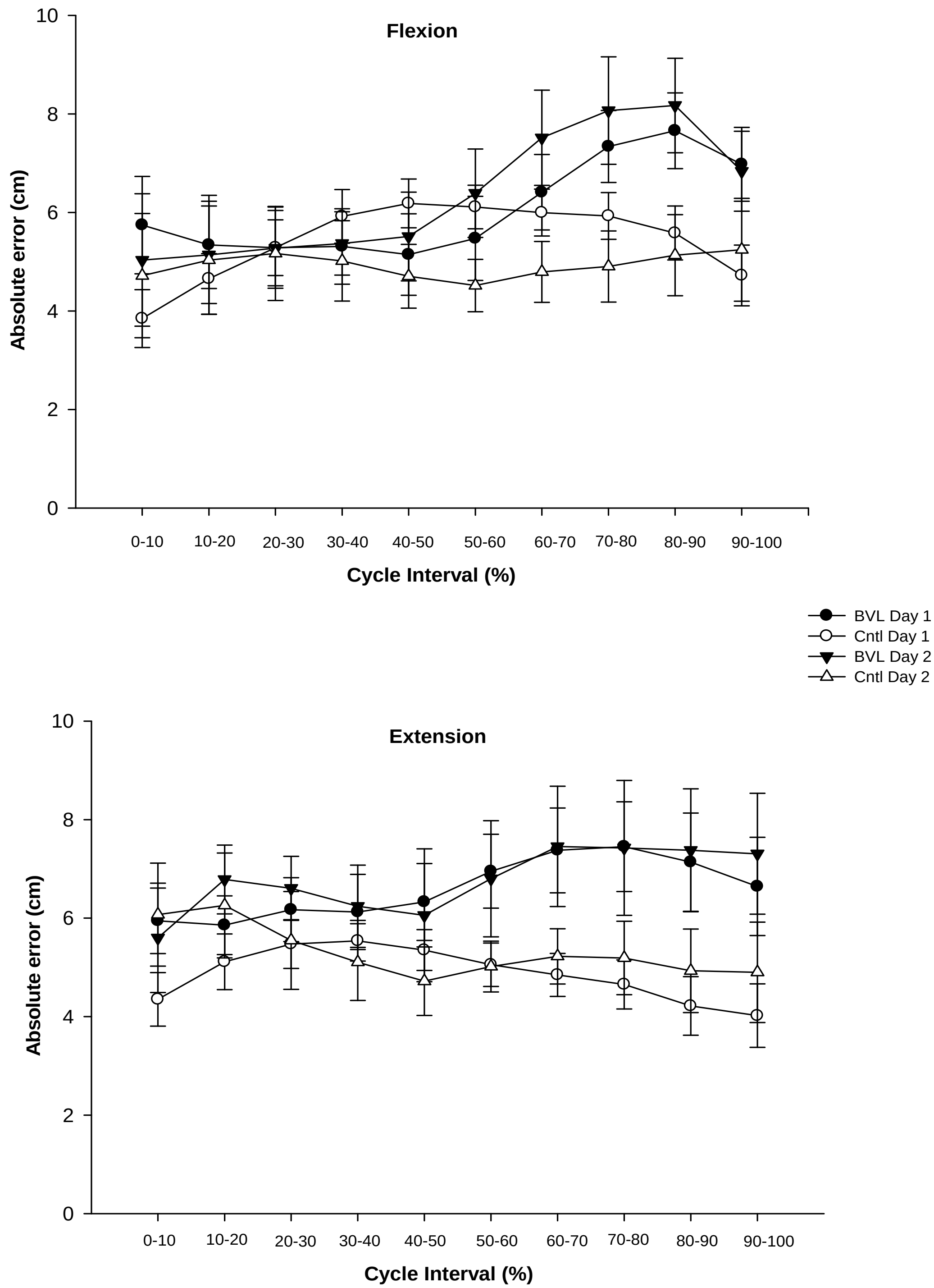

Figure 3.7 Absolute errors of performance in the Eyes Closed (EC) condition Perturbation Events within the flexion (A) and extension (B) phases of the single leg squat task for Day 1 (*circles*) and Day 2 (*triangles*). Mean error values are presented for BVLs (*dark symbols*) and Ctrls (*open symbols*). Values are means $\pm$ SE. Ctrl = 5 sub, BVL = 3 sub firm, 2 sub foam.

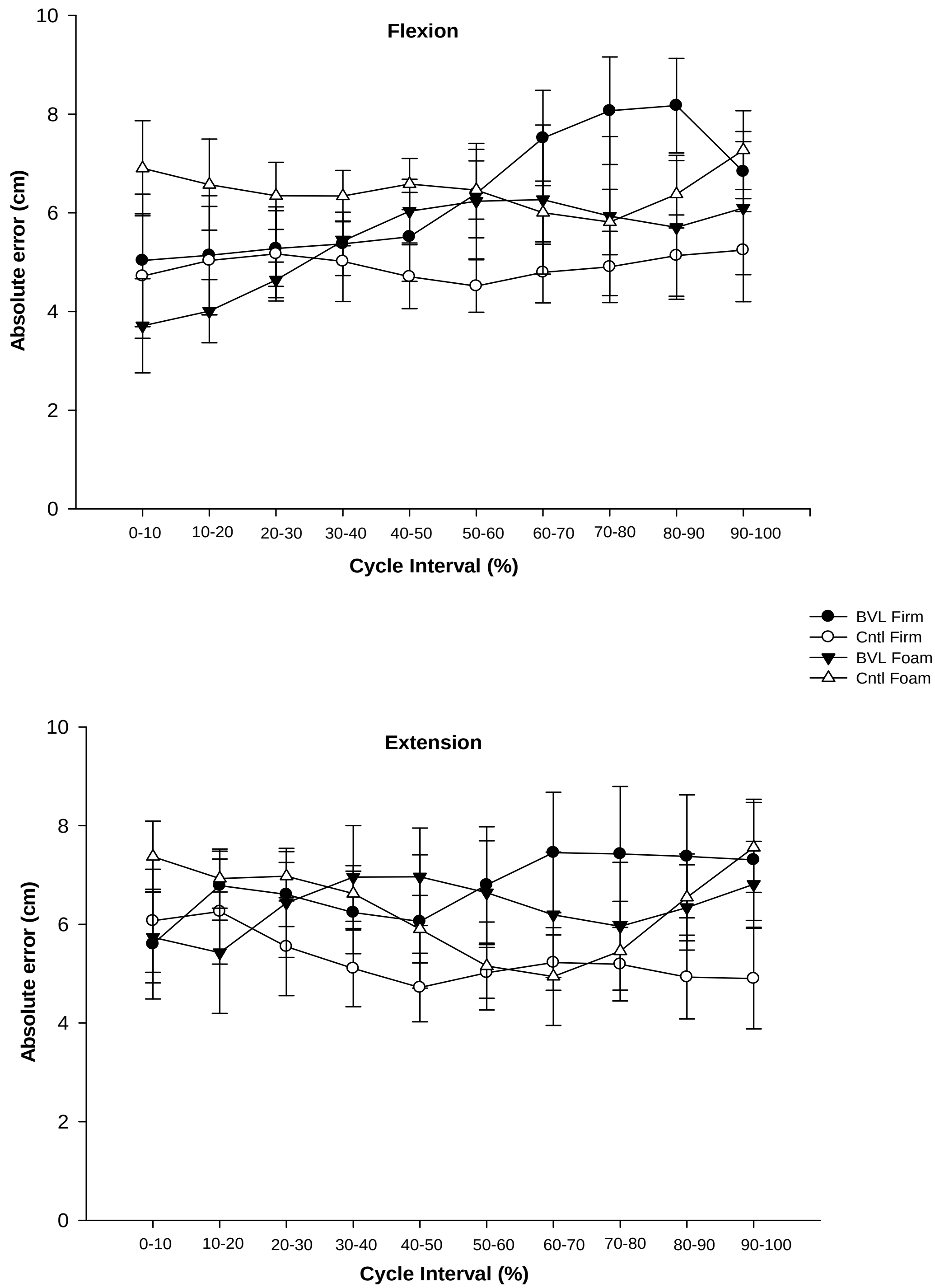

Figure 3.8 Absolute errors of performance in the Eyes Closed (EC) condition Perturbation Events within the flexion (A) and extension (B) phases of the single leg squat task on the Firm and Foam surface. Mean error values are presented for BVLs (*dark circles*) and Ctrls (*open circles*). Values are means ± SE. Ctrl = 5 sub, BVL = 3 sub EC firm, 2 sub EC foam.

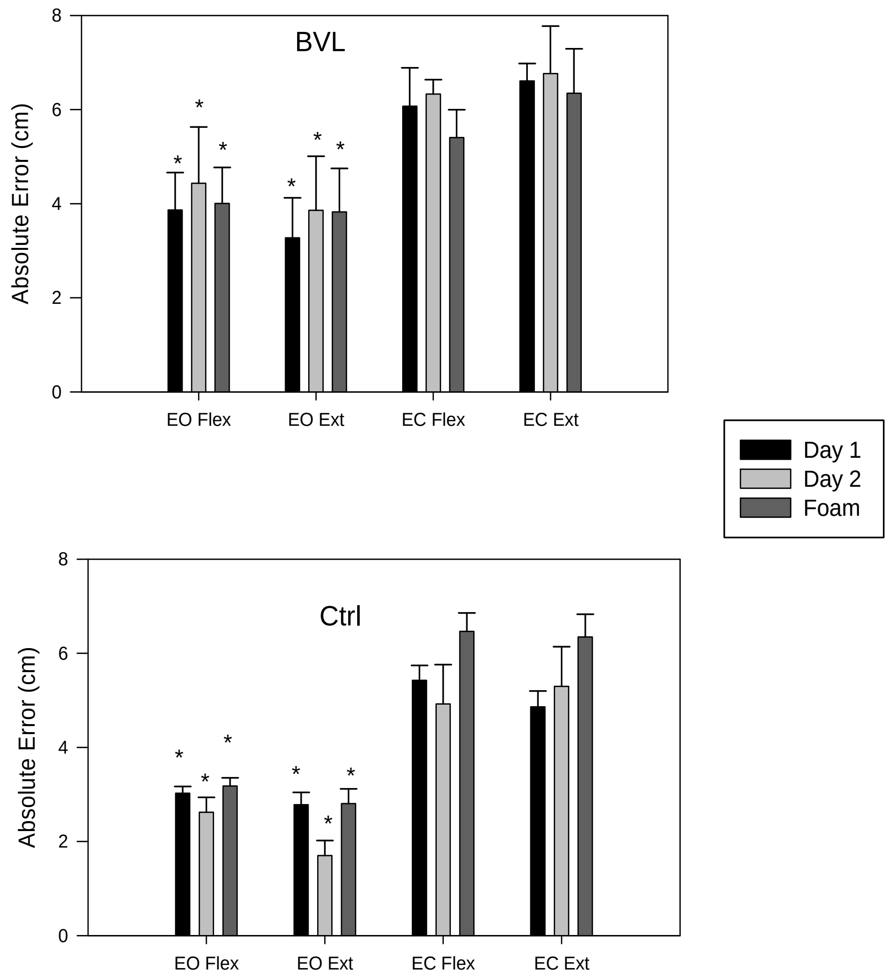

Figure 3.9 Mean absolute error during the flexion and extension phases of the SLS task in the Eyes Open (EO) and Eyes Closed (EC) conditions of the Perturbation Events. The vertical bars represent the average errors over Days 1, 2, and the foam surface. The Y-axis represents absolute error of performance in cm. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

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* indicates significant difference from Eyes Closed Condition

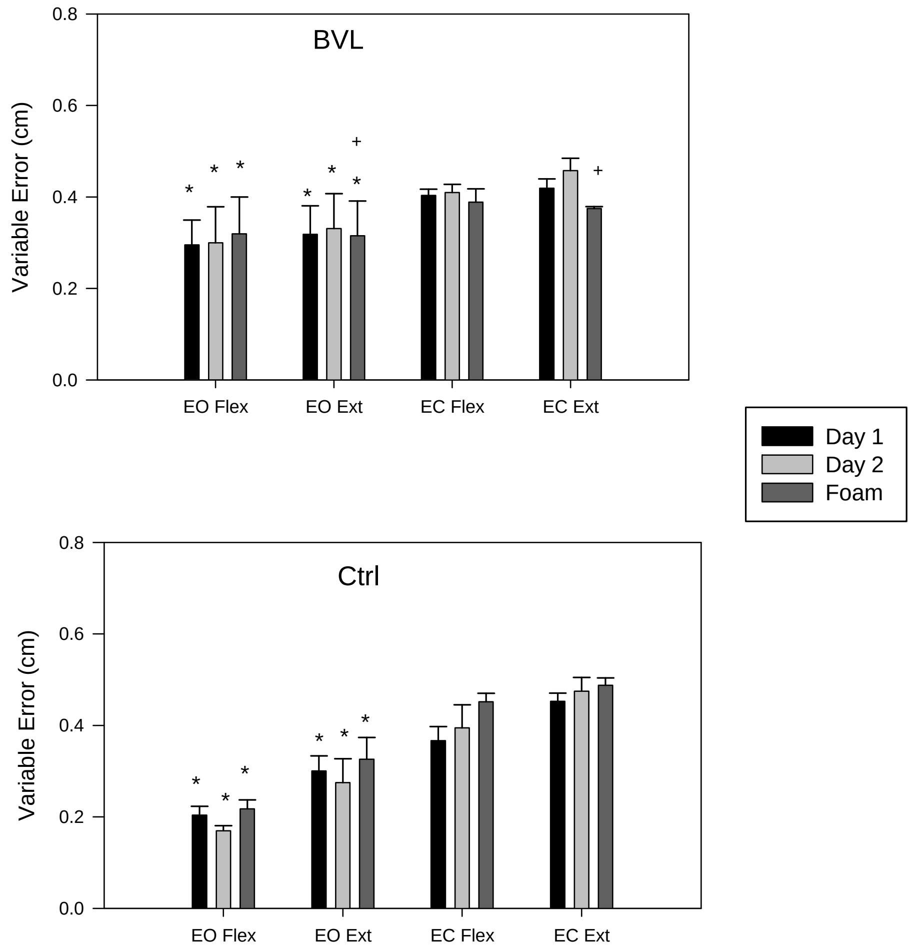

Figure 3.10 Mean variable error during the flexion and extension phases of the SLS task in the Eyes Open (EO) and Eyes Closed (EC) conditions of the No Perturbation Events. The vertical bars represent the average errors over Days 1, 2, and the foam surface. The Y-axis represents absolute error of performance in cm. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

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* indicates significant difference from Eyes Closed Condition (EC)

\+ indicates significant difference between Groups on Foam surface

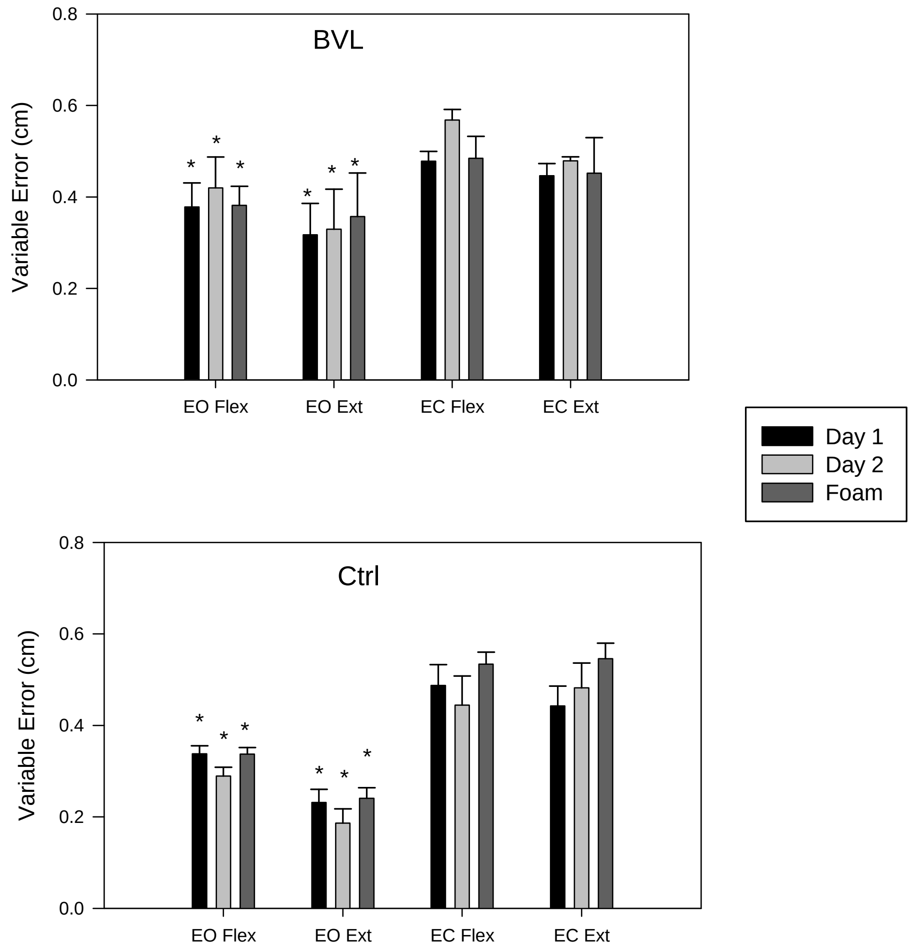

Figure 3.11 Mean variable error during the flexion and extension phases of the SLS task in the Eyes Open (EO) and Eyes Closed (EC) conditions of the Perturbation Events. The vertical bars represent the average errors over Days 1, 2, and the foam surface. The Y-axis represents absolute error of performance in cm. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

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* indicates significant difference from Eyes Closed Condition (EC)

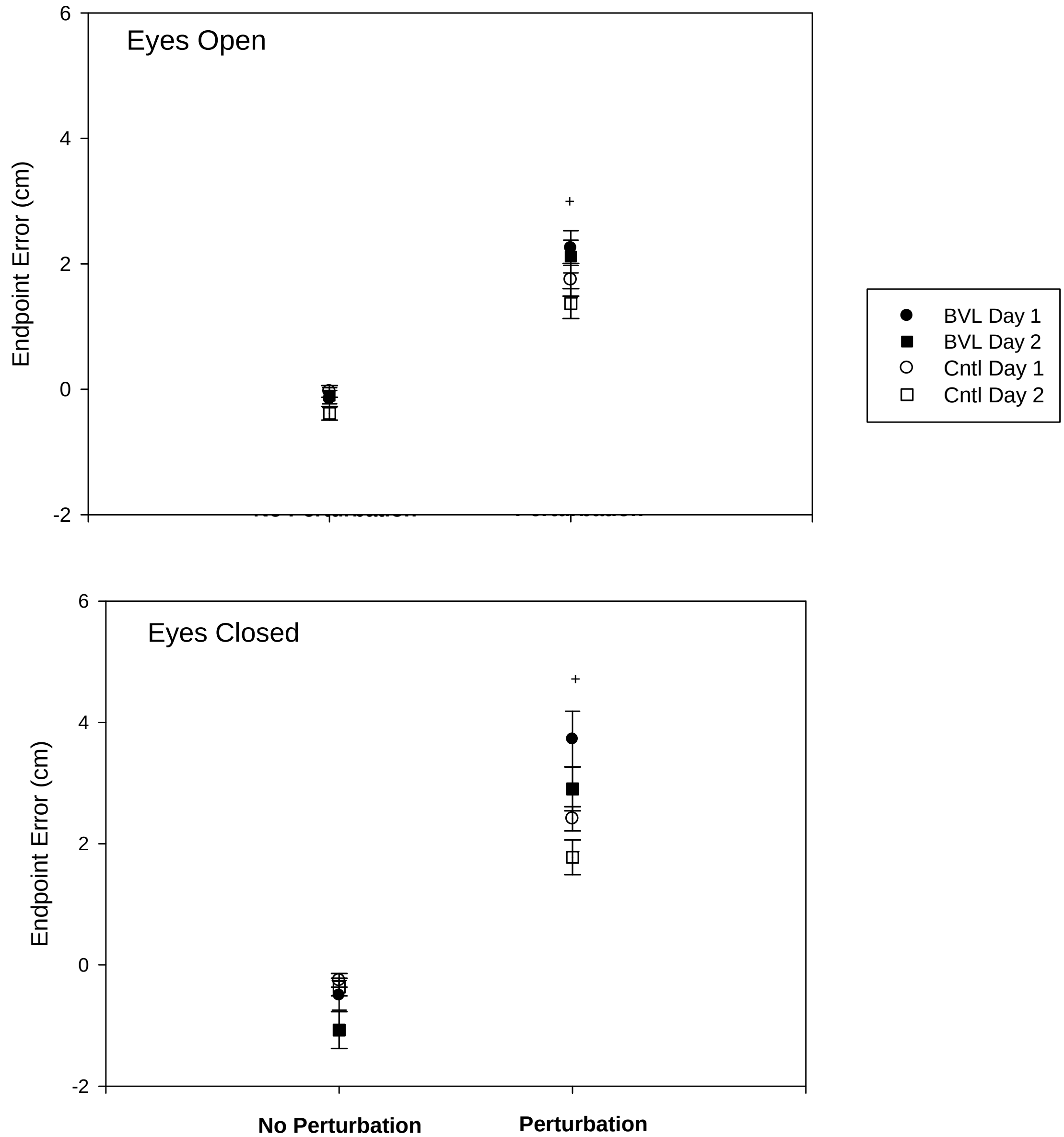

Figure 3.12 Average overshoot error for the Eyes Open and Eyes Closed conditions during unperturbed and perturbed trials across Day1 (*BVL = closed circles, Ctrls = open circles)* and Day 2 *(BVL = closed squares; Ctrls = open squares)*. Values are means ± SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC.

\+ indicates significant difference between Groups

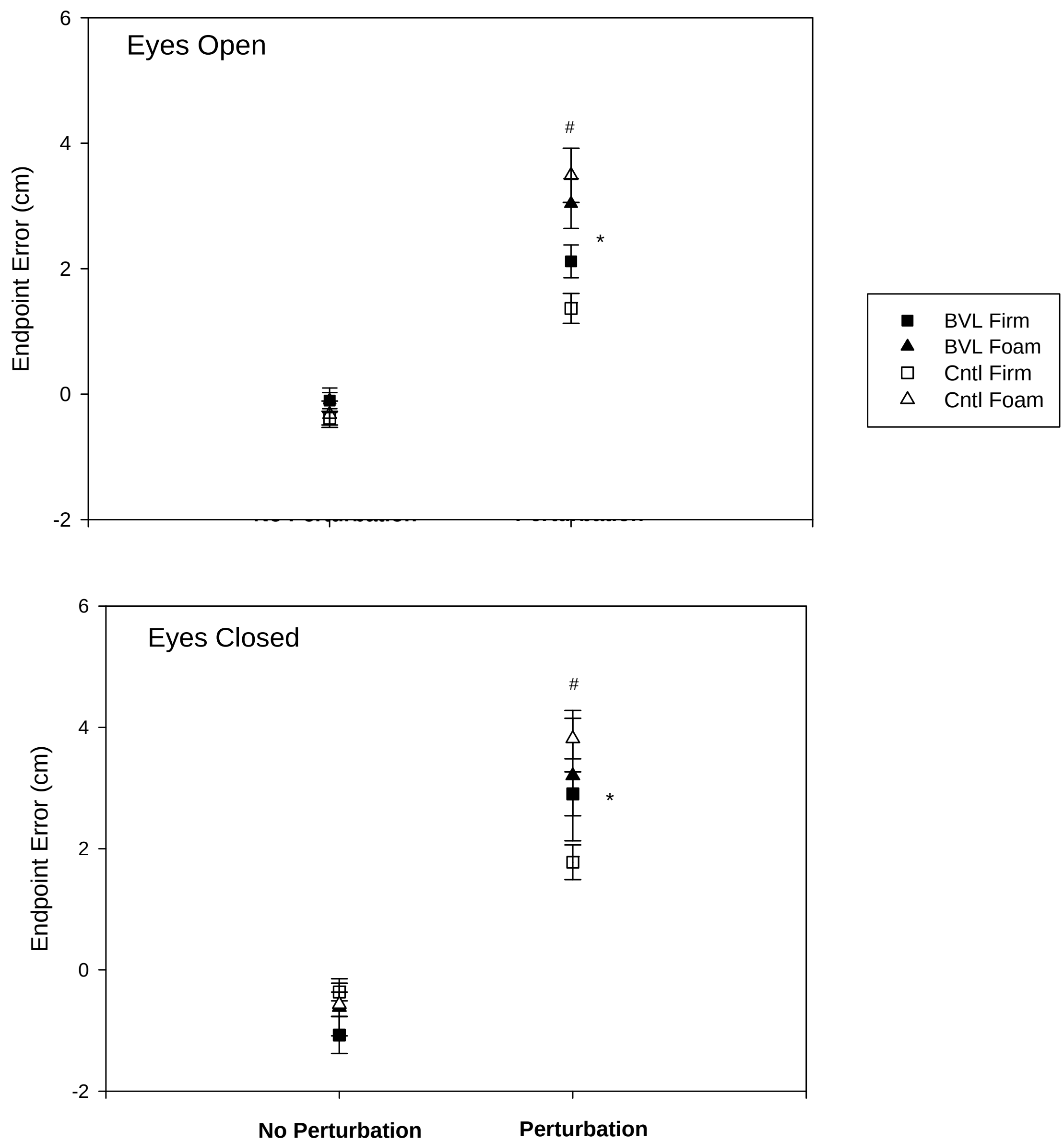

Figure 3.13 Average overshoot error for the eyes open and eyes closed conditions during unperturbed and perturbed trials across Firm (*BVL = closed circles, Ctrls = open circles)* and Foam surfaces *(BVL = closed squares; Ctrls = open squares).* Values are means $\pm$ SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 EC.

\# indicates significant effect of Surface

* indicates significant Group x Surface interaction

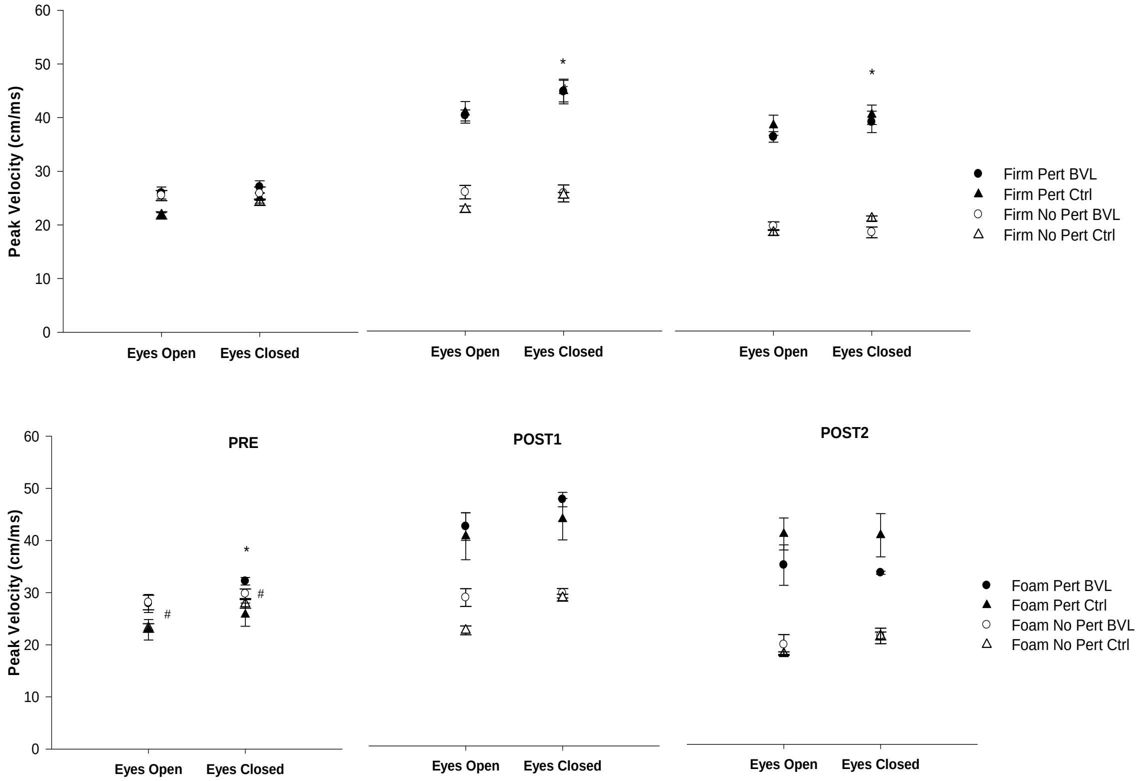

Figure 3.14 Average peak velocity for the non perturbed (*open symbols*) and perturbed (*closed symbols*) events for the Eyes Open and Eyes Closed conditions on Firm and Foam surfaces. Data is represented in 200 ms bins -Pre (anticipatory), Post1 (reflex) and Post2 (volitional) from the time perturbation was supposed to occur (non-perturbed trials) or occurred (perturbation trials).Values are means $\pm$ SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

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* indicates significant difference from Eyes Open Condition when groups are combined

# indicates significant difference from Firm Surface

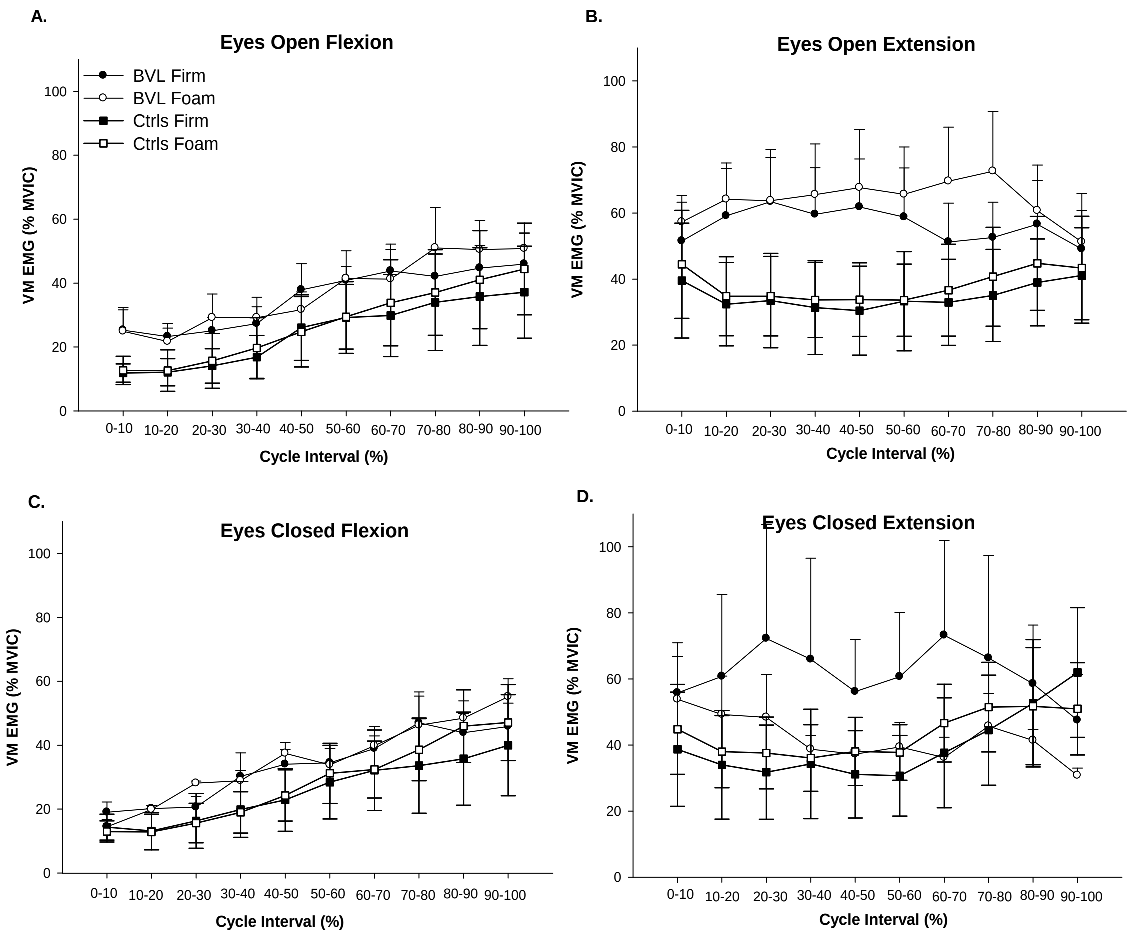

Figure 3.15 Effect of training on the pattern of activation of Vastus Medialis EMG during the flexion (A and C) and extension (B and D) phases, perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for BVL Day 2 (Firm surface) (*filled circles*), BVL Foam (*open circles*), Ctrl D2 (Firm Surface) *(filled squares,)* and Ctrl Foam (*open squares*). Values are means $\pm$ SE. Values are means $\pm$ SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

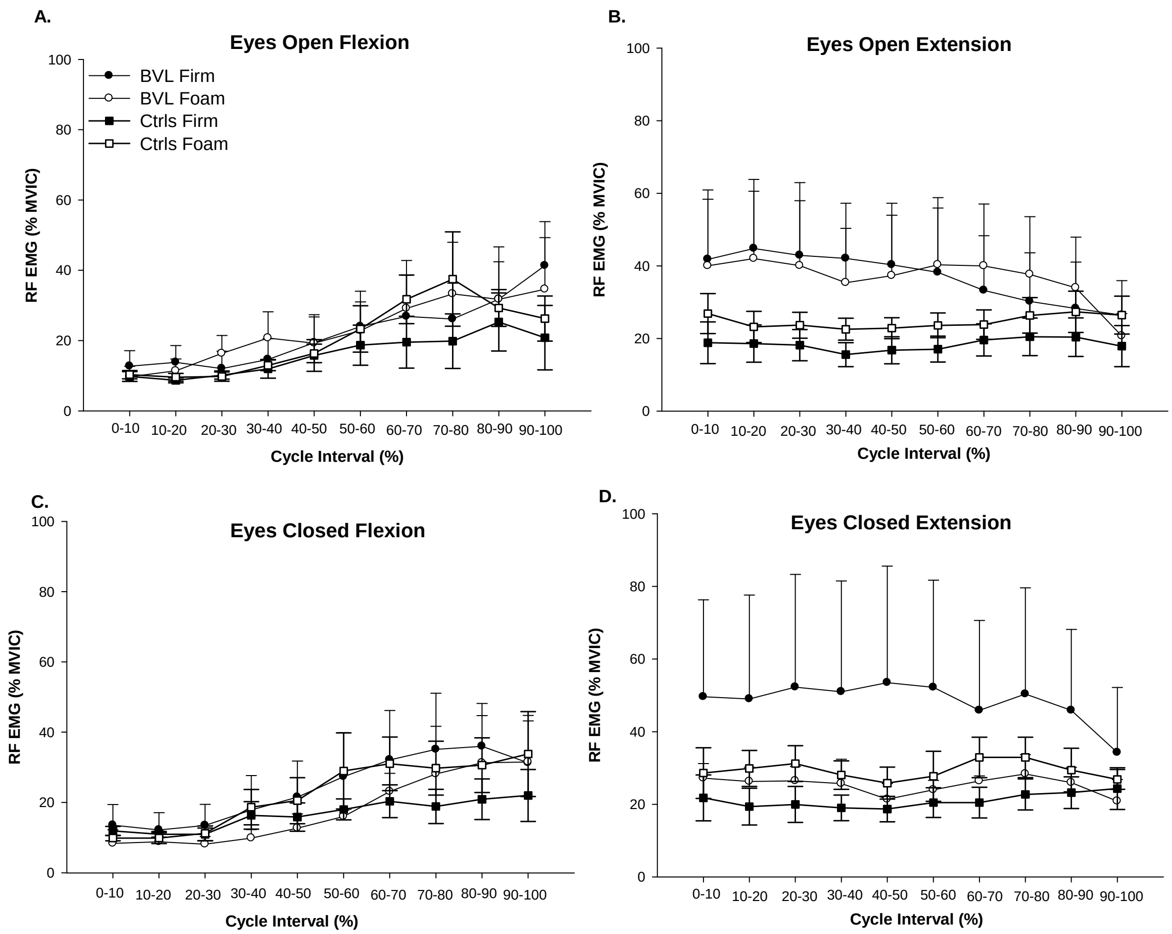

Figure 3.16 Effect of training on the pattern of activation of Rectus Femoris EMG during the flexion (A and C) and extension (B and D) phases, perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for BVL Day 2 (Firm surface) (*filled circles*), BVL Foam (*open circles*), Ctrl D2 (Firm Surface) *(filled squares,)* and Ctrl Foam (*open squares*). Values are means $\pm$ SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

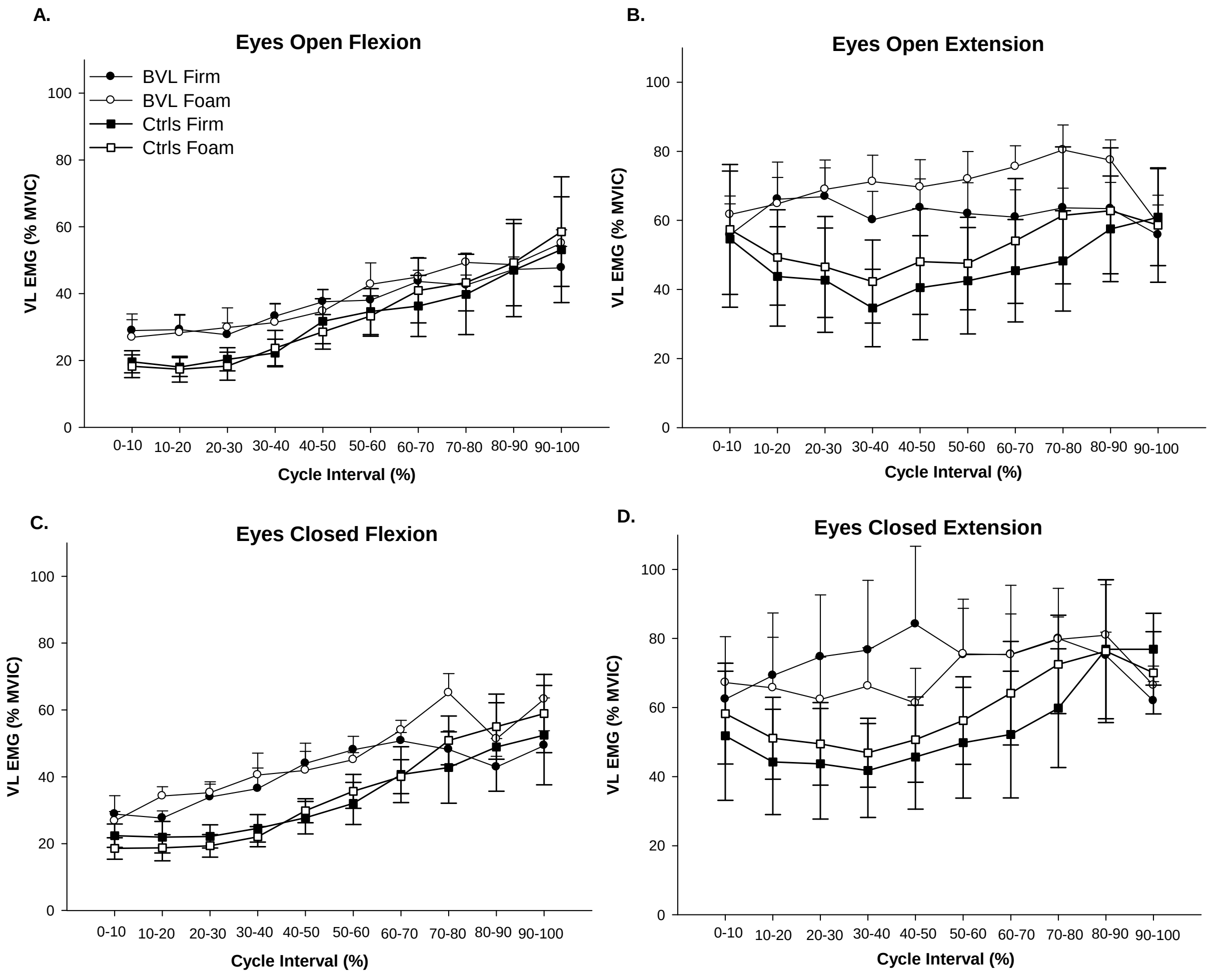

Figure 3.17 Effect of training on the pattern of activation of Vastus Lateralis EMG during the flexion (A and C) and extension (B and D) phases, perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for BVL Day 2 (Firm surface) (*filled circles*), BVL Foam (*open circles*), Ctrl D2 (Firm Surface) *(filled squares,)* and Ctrl Foam (*open squares*). Values are means ± SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

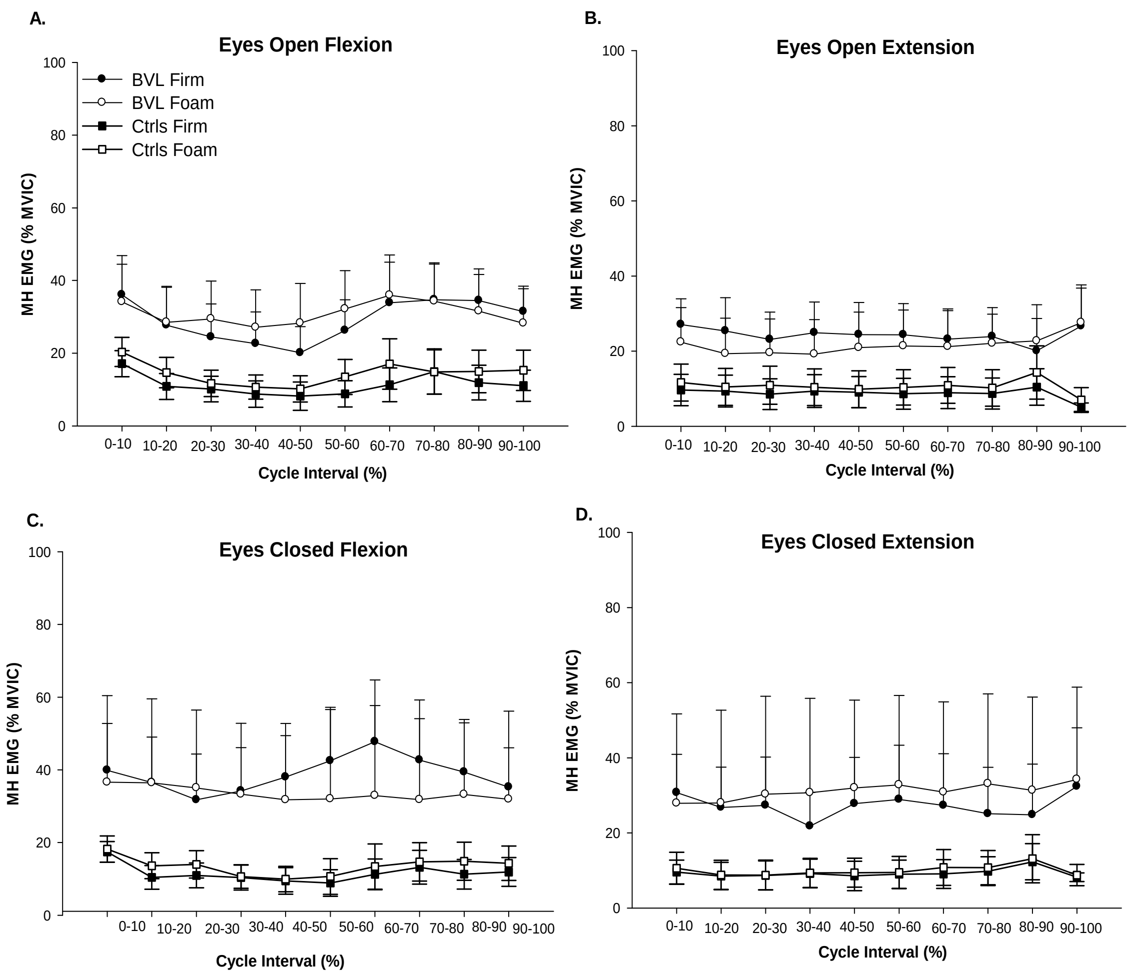

Figure 3.18 Effect of training on the pattern of activation of Medial Hamstring EMG during the flexion (A and C) and extension (B and D) phases, perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for BVL Day 2 (Firm surface) (*filled circles*), BVL Foam (*open circles*), Ctrl D2 (Firm Surface) *(filled squares,)* and Ctrl Foam (*open squares*). Values are means $\pm$ SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

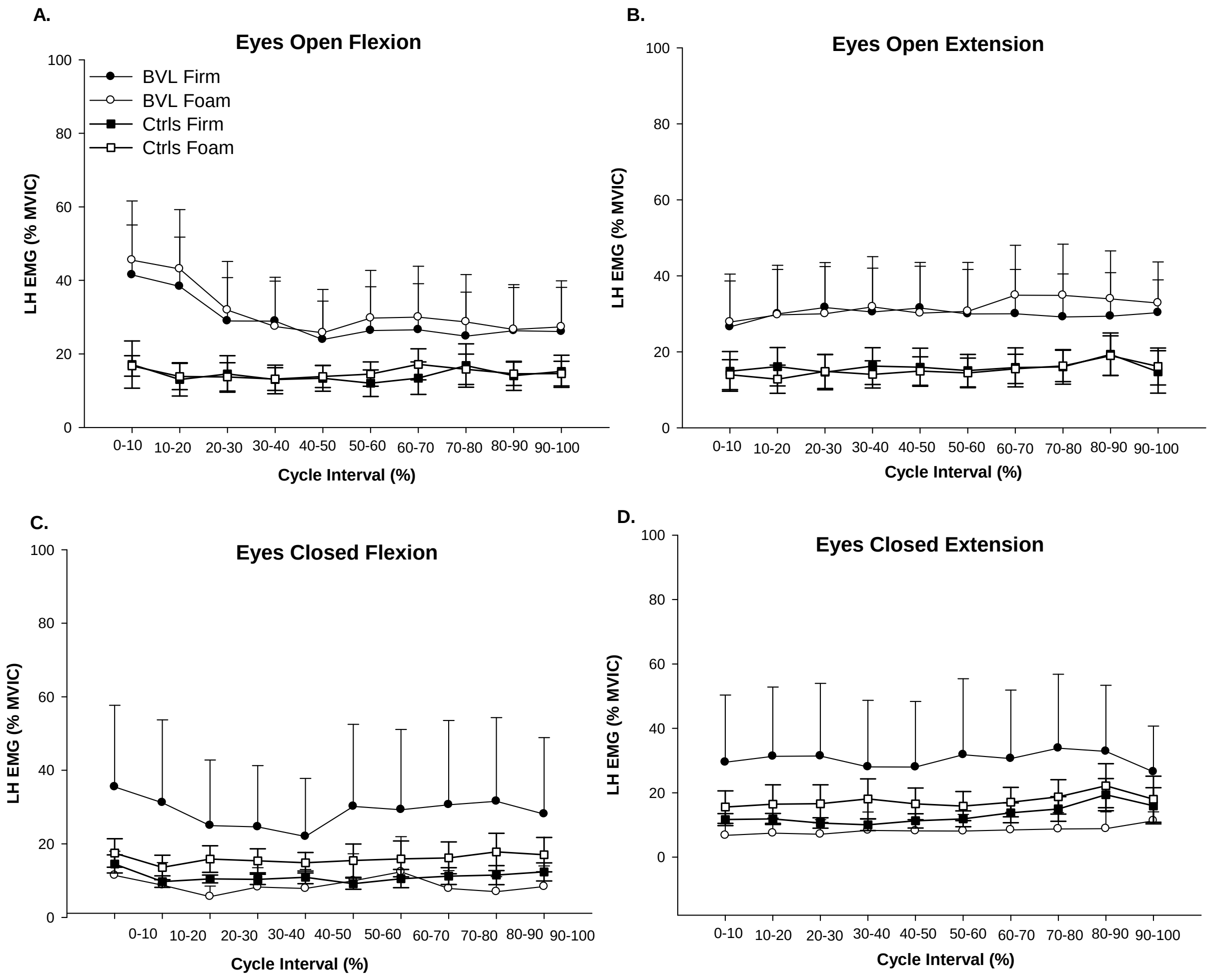

Figure 3.19 Effect of training on the pattern of activation of Lateral Hamstring EMG during the flexion (A and C) and extension (B and D) phases, perturbed events. Normalized EMG for the Eyes Open (A and B) and Eyes Closed (C and D) are presented for BVL Day 2 (Firm surface) (*filled circles*), BVL Foam (*open circles*), Ctrl D2 (Firm Surface) *(filled squares,)* and Ctrl Foam (*open squares*). Values are means $\pm$ SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

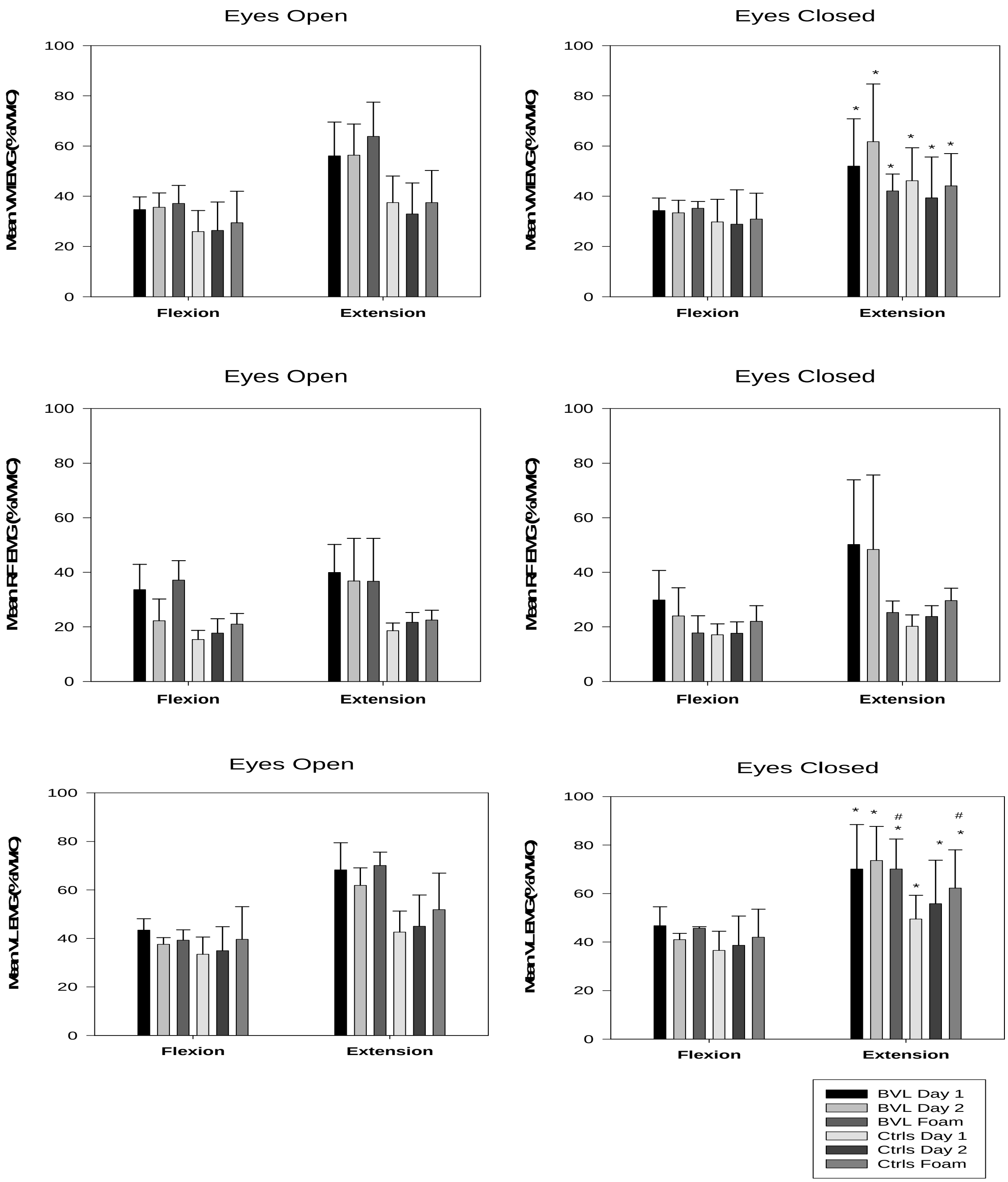

Figure 3.20 Effect of training on the mean EMG during the flexion and extension phases of the Perturbed events for the quadriceps muscles. Normalized EMG for the Eyes Open and Closed are presented for Day 1, Day 2, and Foam. Values are means + SE. Ctrl= 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

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* indicates significant difference from Eyes Open Condition

\# indicates significant difference from the Firm Surface

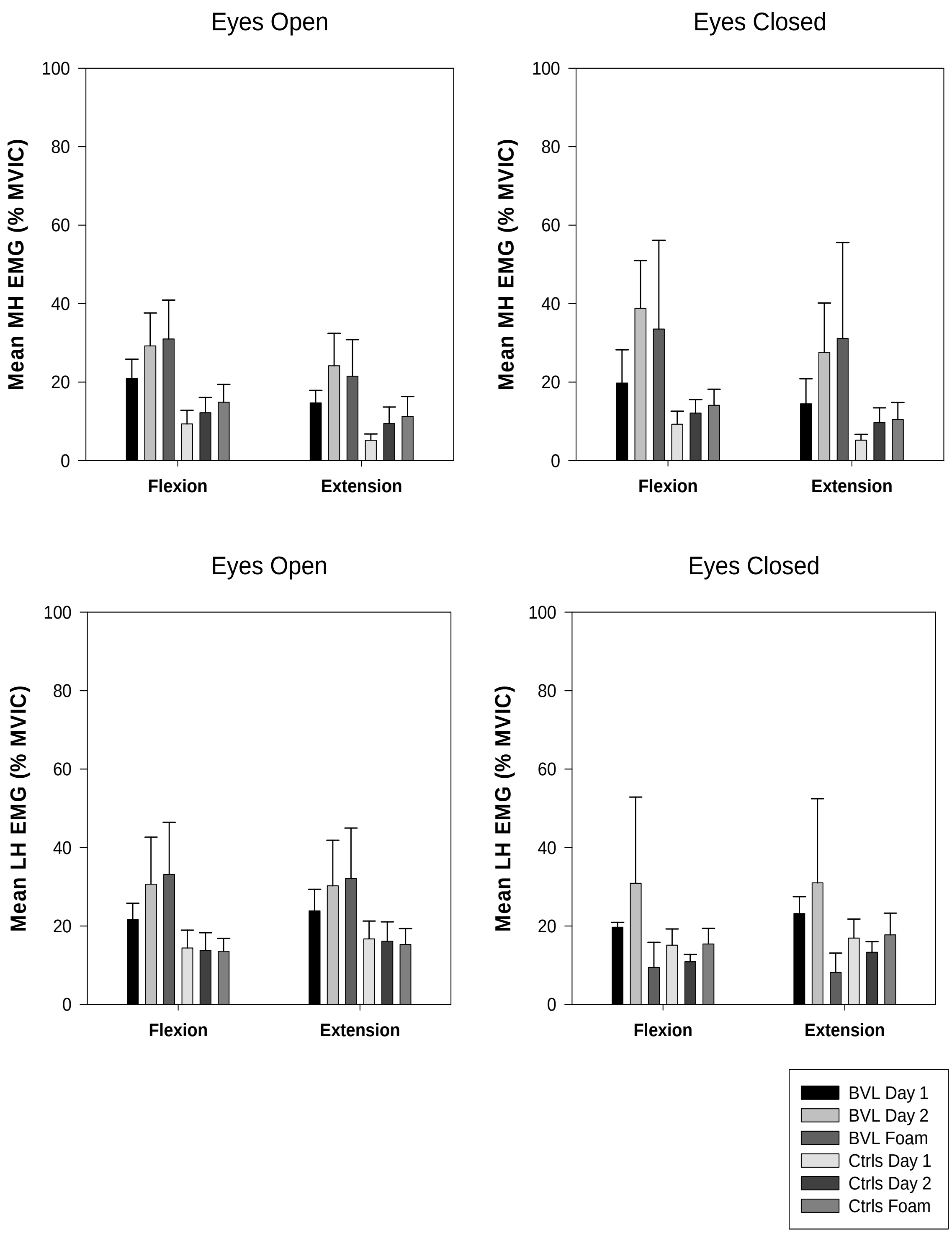

Figure 3.21 Effect of training on the mean EMG during the flexion and extension phases of the Perturbed events for the hamstring muscles. Normalized EMG for the Eyes Open and Closed are presented for Day 1, Day 2, and Foam. Values are means ± SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

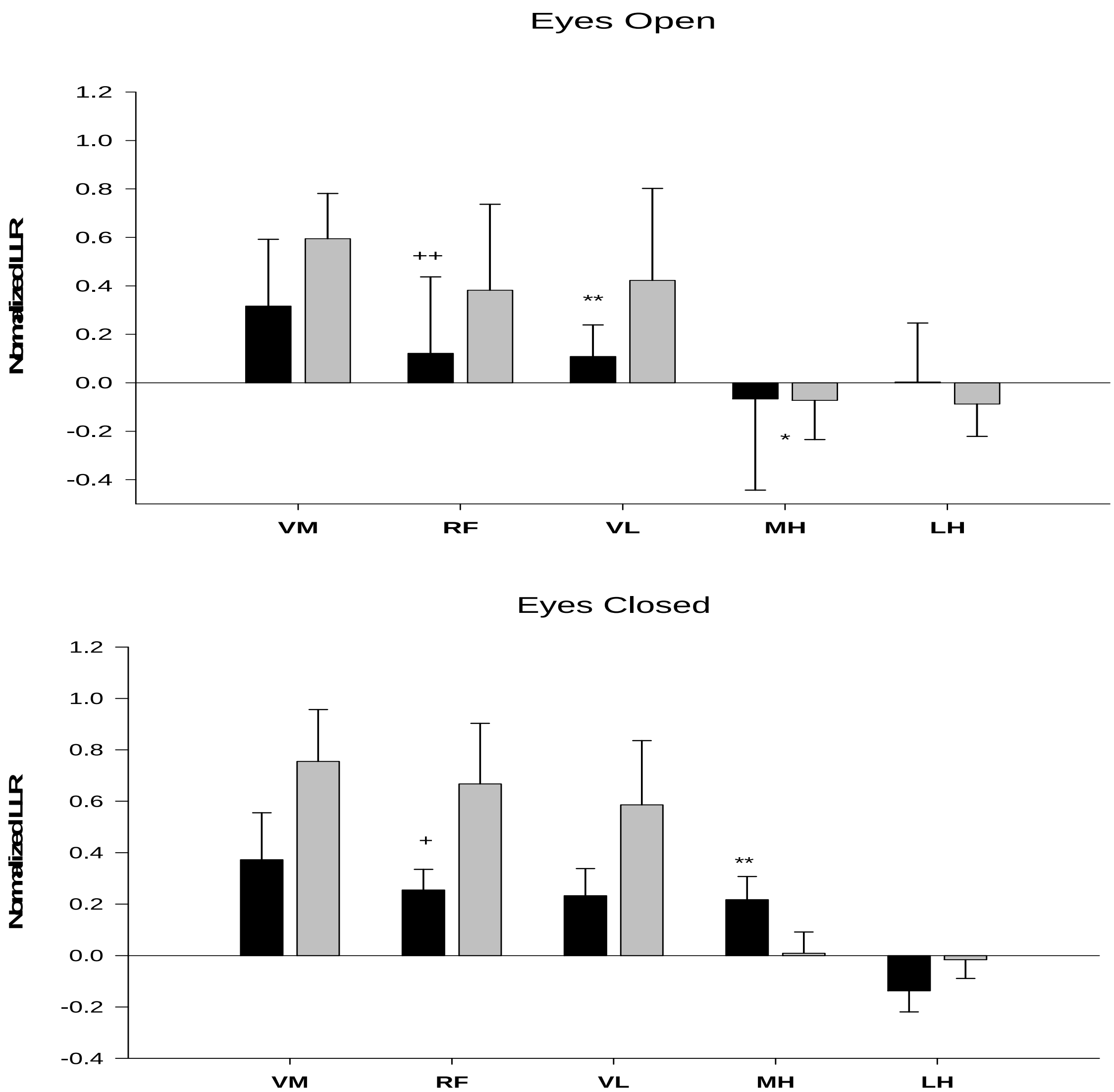

Figure 3.22 Normalized LLRs of the vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH) and lateral hamstrings (LH) during the 50 – 150 ms time bin following the perturbation, for BVLs (*dark bars*), and Ctrls (*gray bars*) on the Firm surface, eyes open and eyes closed conditions. Values are means ± SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC.

\+ indicates a significant difference between Groups in the EC Condition

* indicates a significant difference from the EC Condition when groups are combined

++ indicates a significant difference between Groups at p < .1

** indicates a significant difference from the EO condition at p < .1

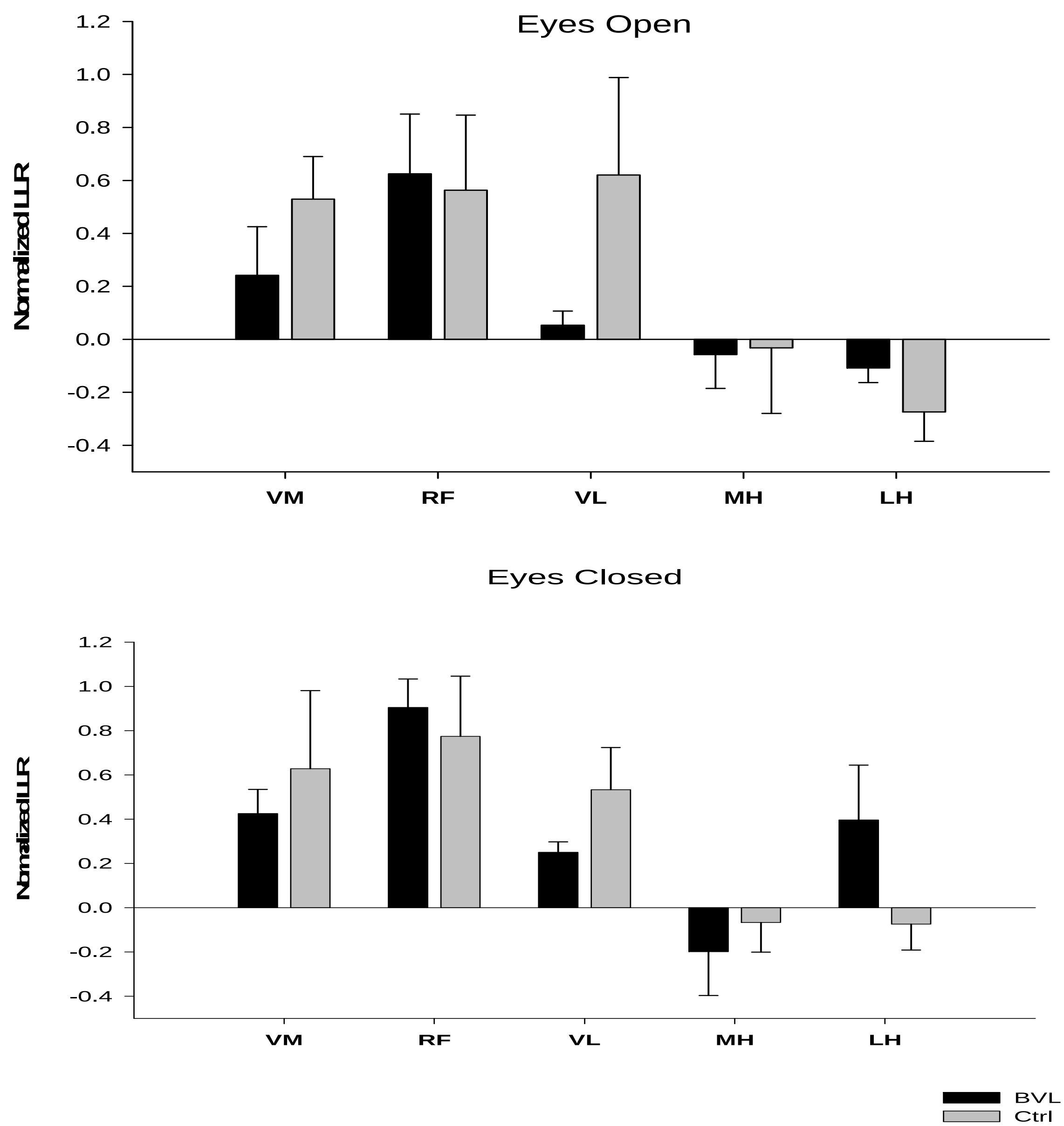

Figure 3.23 Normalized LLRs of the vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH) and lateral hamstrings (LH) during the 50 – 150 ms time bin following the perturbation, for BVLs (*dark bars*), and Ctrls (*gray bars*) on the Foam surface, eyes open and eyes closed conditions. Values are means ± SE. Ctrl = 5 sub; BVL = 5 EO; 2 sub EC foam.

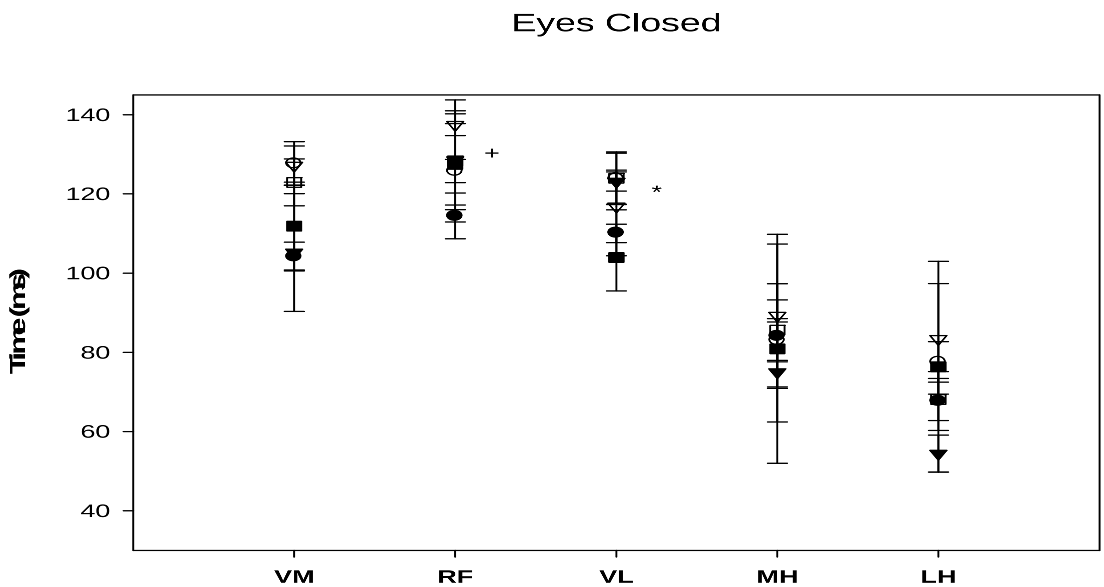

Figure 3.24 Time at which the long latency response peaked for each of the muscles tested [vastus medialis (VM), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH) and lateral hamstrings (LH)] during the unexpected perturbations of the single leg squat task in the eyes open and eyes closed conditions for BVLs (*filled circles*) and Controls (*open circles*). Values are means ± SE. Ctrl = 5 sub; BVL = 5 EO; 3 sub EC firm, 2 sub EC foam.

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* indicates significant difference between Groups
\+ indicates significant difference between Days

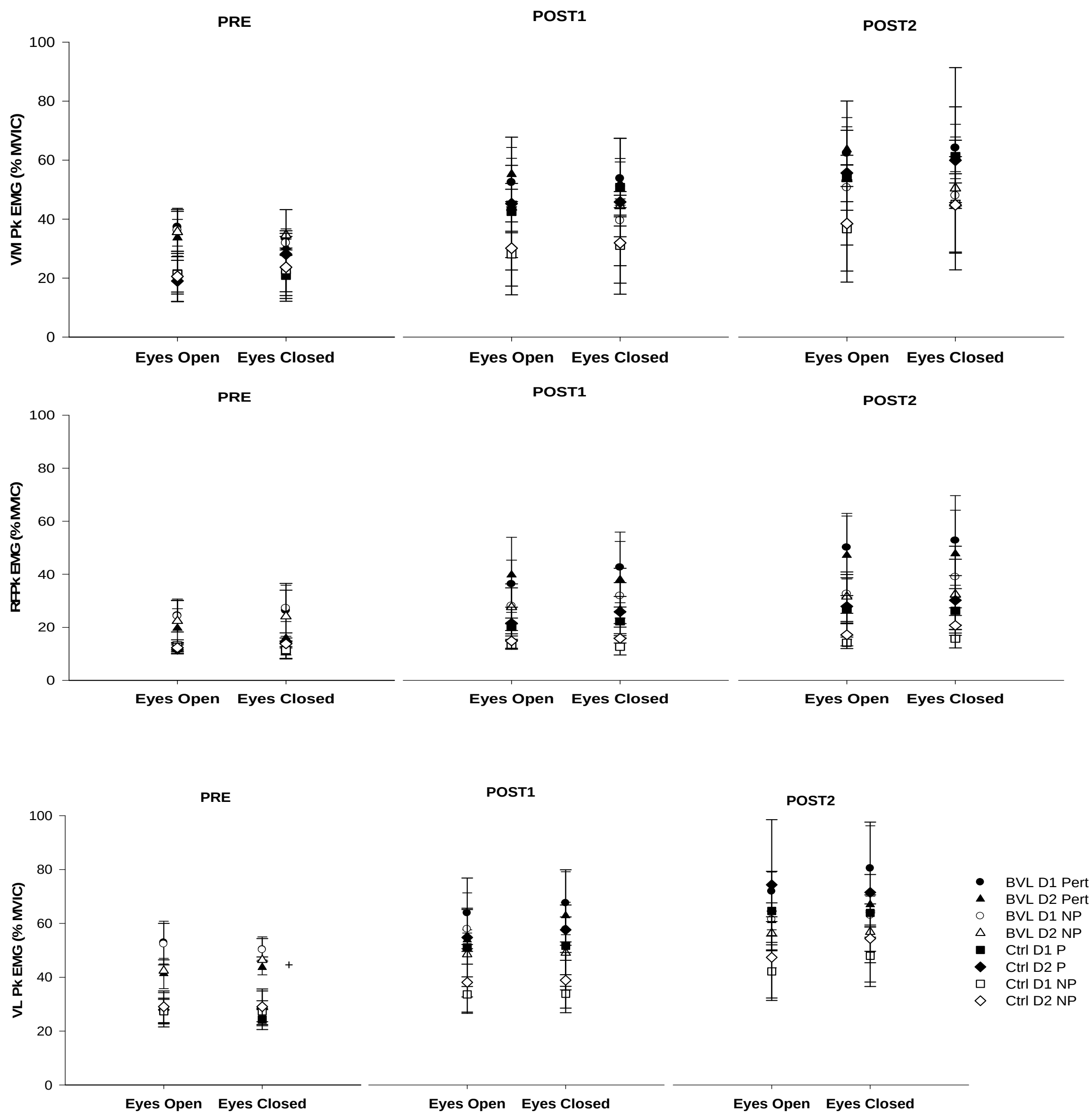

Figure 3.25 Average muscle activity of the vastus medialis, rectus femoris and vastus lateralis for subjects during the eyes open and eyes closed conditions during unperturbed (*open symbols*) and perturbed (*closed symbols*) trials. Data are represented in 200 ms bins -Pre (anticipatory), Post 1 (reflex) and Post 2 (volitional) from the time perturbation was supposed to occur (non-perturbed trials) or occurred (perturbation trials) for Day 1 (*BVL = circles; Ctrls = Squares)* and Day 2 (*BVL = triangles; Ctrls = Diamonds)* on the Firm surface. Values are means $\pm$ SE.

\+ indicates a significant difference between Groups

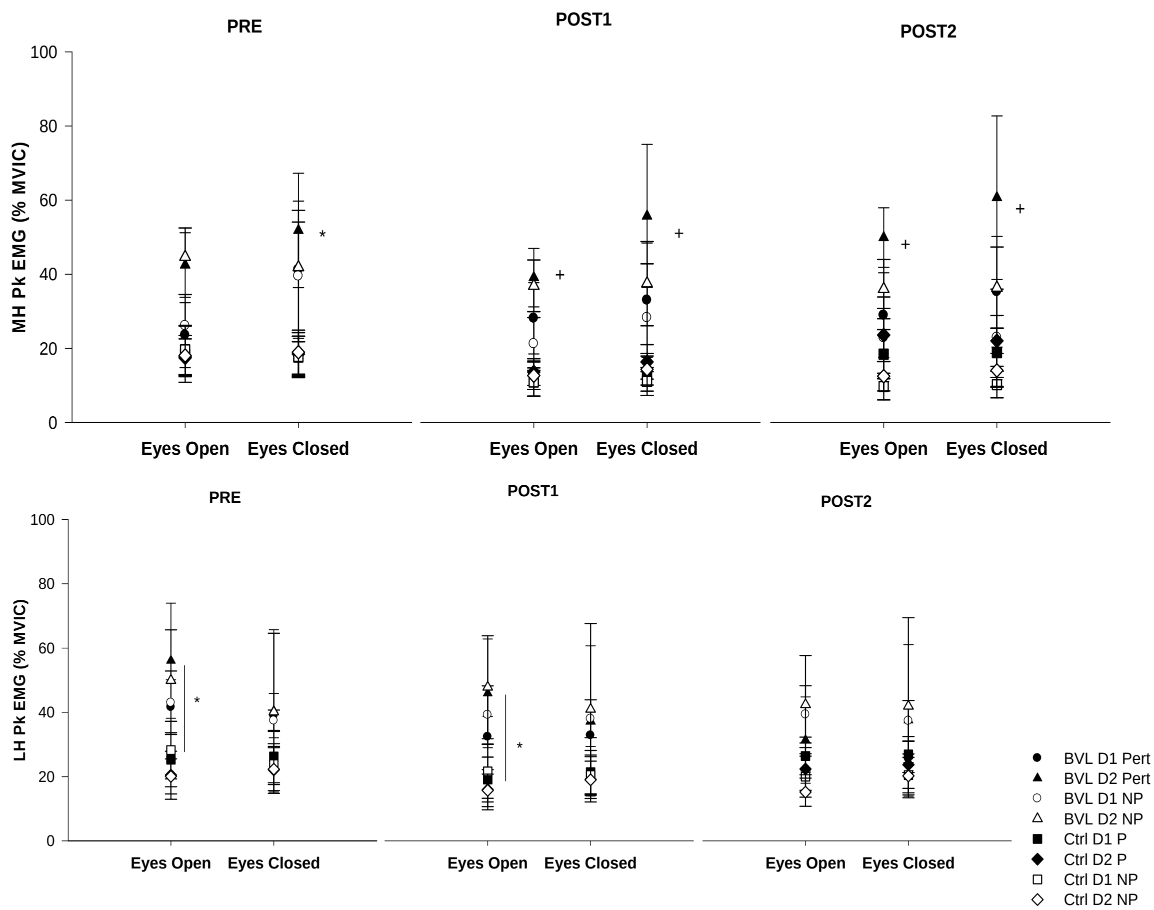

Figure 3.26 Average muscle activity of the medial and lateral hamstrings for subjects during the eyes open and eyes closed conditions during unperturbed (*open symbols*) and perturbed (*closed symbols*) trials. Data is represented in 200 ms bins -Pre (anticipatory), Post1 (reflex) and Post2 (volitional) from the time perturbation was supposed to occur (non-perturbed – trials-NP) or occurred (perturbation trials - Pert) for Day 1 (*BVL = triangles; Ctrls = Circles)* and Day 2 (*BVL = triangles; Ctrls = Diamonds)*.Values are means ± SE.

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* indicates a significance difference from Eyes Closed (EC)

\+ indicates a significant difference from Day 1

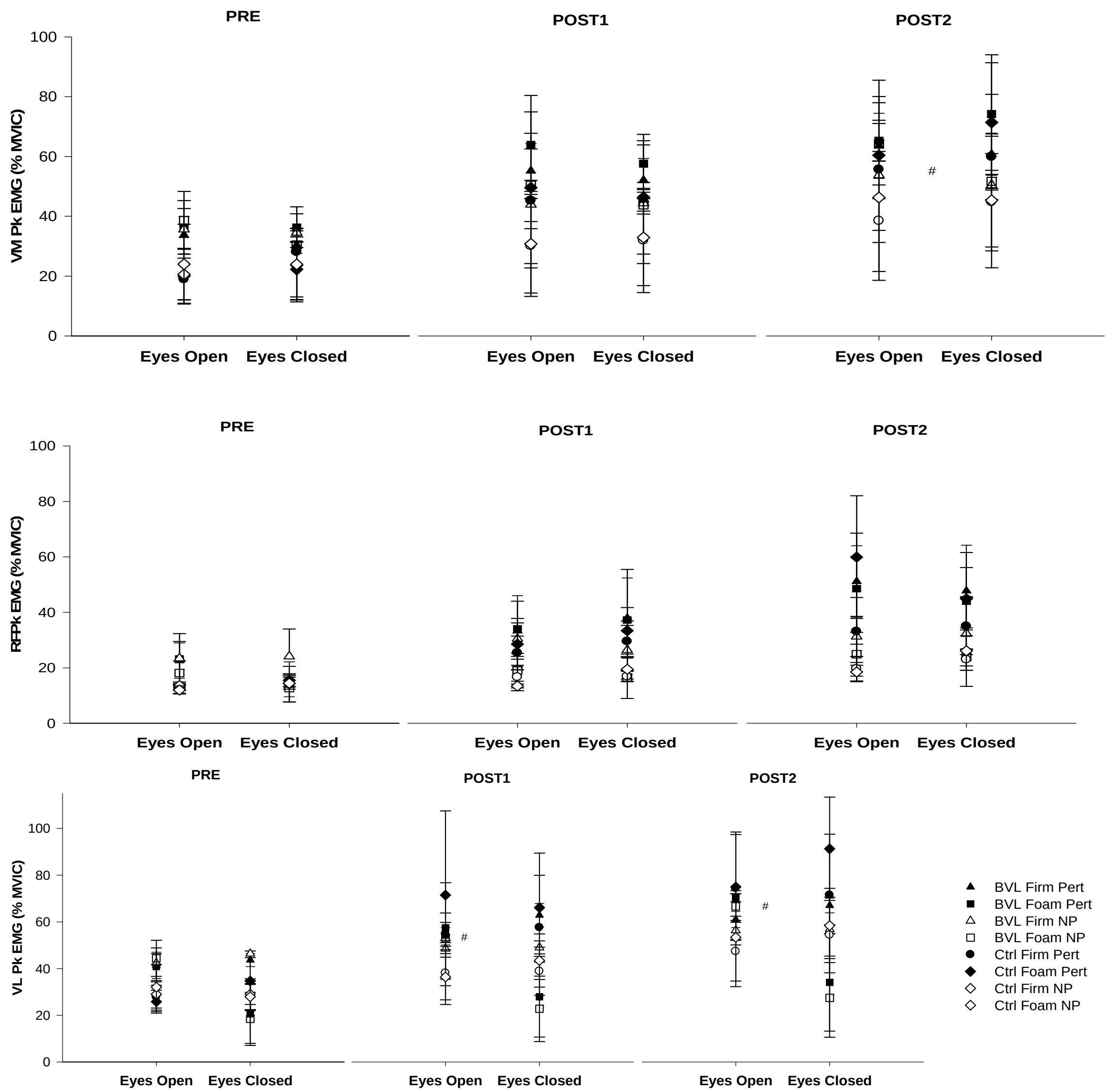

Figure 3.27 Average muscle activity of the vastus medialis, rectus femoris and vastus lateralis for subjects during the eyes open and eyes closed conditions during unperturbed (*open symbols*) and perturbed (*closed symbols*) trials. Data are represented in 200 ms bins -Pre (anticipatory), Post 1 (reflex) and Post 2 (volitional) from the time perturbation was supposed to occur (non-perturbed trials) or occurred (perturbation trials) for Firm (*BVL triangles; Ctrl circles)* and Foam *(BVL squares; Ctrl Diamonds)* surfaces. Values are means $\pm$ SE.

# indicates a significant difference from Firm Surface

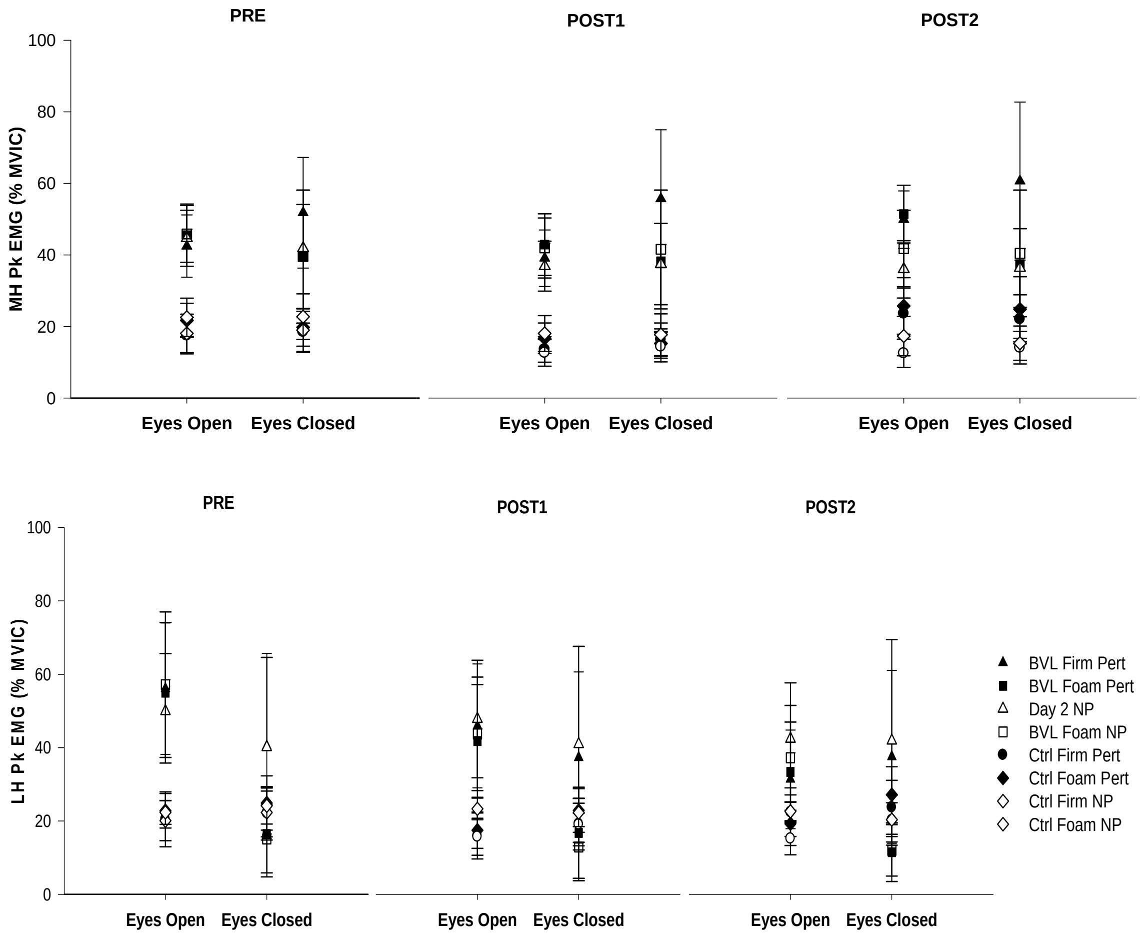

Figure 3.28 Average muscle activity of the medial and lateral hamstrings for subjects during the eyes open and eyes closed conditions during unperturbed (*open symbols*) and perturbed (*closed symbols*) trials. Data is represented in 200 ms bins -Pre (anticipatory), Post1 (reflex) and Post2 (volitional) from the time perturbation was supposed to occur (non-perturbed trials) or occurred (perturbation trials) for Firm (*BVL triangles; Ctrl circles)* and Foam *(BVL squares; Ctrl Diamonds)* surfaces. Values are means ± SE.

Discussion

The purpose of this study was to examine accuracy of performance, muscle activation strategies, and LLR during a lower extremity weight-bearing task performed under different conditions of visual feedback and surface type across a 2-day training session between individuals with bilateral vestibular loss (BVL) and healthy age- and gender-matched controls. Though perturbation studies have been performed via platform posturography with this group, little is known about their ability to respond to unexpected perturbations during a dynamic weight-bearing task such as the single limb squat (SLS). The sinusoidal target used in this study provided feedback about motor performance, showing the rate and amplitude of subjects' motion. Subjects in this study were focused on accurately tracking a sinusoidal target during a single limb squat, so were unable to anticipate the unexpected perturbations that occurred.

Computerized dynamic posturography (CDP) has studied subjects with BVL and suggests that this group performs differently than controls in terms of postural stability and strategies for maintaining balance with perturbations.[9, 24, 51, 69, 93, 169-171] The difference between this current study and those using CDP, however, is that a more challenging, dynamic task that emphasized performance accuracy was used. This was not a balance study, but rather a reflex study. Subjects were secured in a podium and were able to balance statically in unilateral stance for 30 seconds with eyes closed on both firm and foam surfaces with baseline testing. How both groups were able to accurately track a target and react to unexpected perturbations of the SLS task under different conditions of visual feedback was examined.

<u>Firm Surface</u>

One of the important findings of this study was that the BVL subjects and controls did not differ in terms of accuracy of performance during either phase of the SLS task in any visual condition on the firm surface. The BVL group responded as the controls to practice, with no significant improvements in performance and consistency from Day 1 to

Day 2. Error of performance in the absence of visual feedback for both groups was greater than when vision was available. This is consistent with previous observations (Chapter 2), as well as studies of young, healthy females and those with ACL injury.[81] However, only 3 of the 5 BVL subjects could perform the task with eyes closed on the firm surface and only 2 with eyes closed on the foam cushion. Even though the BVL subjects responded similarly to the controls in terms of accuracy with eyes open and closed, not all subjects with BVL could even perform the task with absent vision. This suggests that although the vestibular deficit did not alter their ability to perform this task with eyes open, it was too challenging for some to perform with eyes closed, even when secured in the podium. The inability to perform the SLS with eyes closed may be related to differences in age and/or activity levels. Although absolute and variable error were no different between groups, a difference was noted in the BVL response to perturbation. Endpoint error was greater for this group with perturbations on the firm surface, which suggests that BVL does affect the ability to respond to perturbations during this task; this group overshoots to a greater degree than controls. Muscle activation strategies were then assessed to determine any differences between groups that may have contributed to the increase in Epe.

Very few studies have reported EMG recordings during a dynamic task in the BVL population. The present study differs from posturography studies in that the loads used during the task were sufficiently high enough for detection of relevant changes to the neuromuscular system, and accuracy of performance of the task was quantified. These loads might be comparable to what an individual would encounter when faced with unexpected limb perturbations under weight-bearing conditions. Perhaps using tasks during rehabilitation that require performance accuracy would help prepare these individuals for high stress situations like perturbations when fully weight bearing. In disagreement with our primary hypothesis, we found that neuromuscular strategies used to perform the SLS task were not different between individuals with BVL and healthy controls on the firm surface. As there was similar task performance in both groups, muscle activity was also similar. There were no statistically significant group differences found in muscle activation strategies for cycle EMG data and no day effect. There was an effect of vision for both groups, however, with VM and VL activity greater during the extension phase of perturbation events with eyes closed. The BVL group did show greater activity of the VL compared to the controls during the anticipatory (feedforward) phase of nonperturbed and perturbed events. The latency of the peak VL LLR in the eyes open condition was also faster in the BVL group. This may be a strategy used to adhere to performance requirements.

After BVL, the ability to effectively respond to perturbations is disrupted. The vestibular system plays a vital role in maintaining the orientation of the whole body to vertical with the body properly aligned, parallel to gravity and directly over the feet.[172] In addition, motor output from this system contributes to dynamic postural movements, which assist in controlling the center of body mass within its limits of stability.[172] This motor output, via the vestibulospinal tracts, originates in the vestibular nuclei in the medulla and projects bilaterally down the spinal cord activating cervical spinal circuits that control neck and back muscles. Another component projects ipsilaterally down through the lumbar spinal cord, facilitating extensor motor neurons of the lower extremities.[92] Postural control dysfunction is well documented for patients with BVL, including instability in stance, during ambulation, and with transitional activities such as moving from sitting to standing.[4, 20, 150, 152-156]

While others have used platform posturography, we utilized the perturbed SLS task to assess reflex responses to perturbations in a vestibular-deficient population. Stretch of an active muscle after a perturbation results in several bursts of reflex activity, which can be categorized into short latency (0-50 ms) and long latency responses (LLR) (50 – 150 ms). Supraspinal input, including visual, vestibular, and proprioceptive signals, contribute to the determination of context and feedback responses, allowing for more complex control than would be possible with spinal reflexes alone.[32, 41] To maintain joint stability with platform perturbations, automatic postural responses and LLR to surface translations are triggered by somatosensory information and are scaled to the velocity and amplitude of the platform translation.[56, 66-67] The LLR to a perturbation serves as an important contributor to the stiffness of the limb, thereby assisting with stabilization and postural control.[41, 46] Long latency postural responses may have greater potential for modification by supraspinal neural centers, as they occur more quickly than voluntary movements, but not as quickly as spinal stretch reflexes.[38, 53, 109, 173] Afferent information (vestibular, visual, and somatosensory) is essential for triggering long latency responses, and these responses are influenced by a variety of factors like the task performed,[54-55] prior experience with a task,[7, 56-58] practice,[7, 56] intent,[59] environmental context,[60] and age.[7, 58, 61-64] Though LLR to surface perturbations are likely triggered by somatosensory information, vision and vestibular input also play key roles in maintaining stability. Visual information is important for the control of head and trunk position in space and stabilizing COM, especially at fast sinusoidal surface translation frequency in the AP direction.[65] With CDP testing, healthy adults exhibit a facilitation of the LLR with platform perturbations that allows them to maintain stability without loss of balance. Therefore, in these conditions, the LLR helps to decrease the amount of postural sway, thus increasing stability.[143] In contrast, individuals with BVL demonstrate a decreased LLR amplitude response and subsequent increase in postural sway and/or loss of balance.[24, 51, 68, 76, 171] In addition, when vision is absent during these experiments, subjects with BVL demonstrate significant decreases in the LLR and resultant instability and/or falls compared to healthy controls.[10, 68, 171] Allum et al.(1998) also saw reductions in BVL LLR in the tibialis anterior and quadriceps with eyes closed compared to healthy matched controls with surface translations. Based on results such as these, authors have concluded that LLRs are likely gated or triggered by proprioceptive afferent signals elicited by muscle stretch in the lower leg and their response amplitudes are modulated by visual and vestibular signals.[51, 171]

Despite several investigations on the balance strategies that patients post BVL use to maintain posture on a moving platform, the effects of BVL on performance of a dynamic weight-bearing task is limited. It is essential to perturb the joint in weight-bearing during motions that mimic functional activities to better understand these reflexes with different tasks. In addition, the interaction of loss of vestibular system input and vision on muscle responses during a functional, dynamic weight-bearing activity has not yet been studied. This is the first experiment to compare LLRs in subjects with BVL to healthy controls during a novel, dynamic, weight-bearing task. Subjects with BVL showed a pattern of decreased normalized LLR response compared to the controls for VM and VL with eyes open and closed, though this was not statistically significant. For RF, the difference between groups was significant with eyes closed ($p < .05$). BVLs had a significantly lower response in this muscle when vision was absent. The LLR with eyes closed for MH with groups combined was less than the response with eyes open. This suggests that vision contributes to and modulates the LLR during the SLS task differently for each group. Previous work in this lab examining college-aged females with and without ACL injury found similar effects of the eyes closed condition. Madhavan (2007) examined the LLR during the SLS task and found significant difference in the RF LLR between the ACL and the Control group in the EC condition (and not in EO).[81] With EC, the ACLRs had greater activation of the RF compared to the controls. The rectus femoris muscle may have a differential neural control because it is a two jointed muscle.[54] Mrachacz-Kersting et al. (2006) used transcortical magnetic stimulation and determined indirectly that a transcortical pathway is likely involved in shaping the RF LLR. There were no statistically significant effects for VM and VL in that study, suggesting that these results were specific to the RF. Of the quadriceps, RF is the only muscle that crosses two joints, working as both a hip flexor and a knee extensor. Different neural control of the

RF compared to VM and VL may be a way to functionally uncouple the hip and the knee during the SLS task, especially during eyes closed when the consequences of the perturbation are perceived to be greater.[54, 81] In the current study, the BVL groups had a significantly lower RF LLR response compared to the controls in the EC condition. With absent or significantly reduced vestibular function, perceptual and reflex responses to perturbations are impaired, which is more evident when vision is also removed.

There was no difference between groups in the background EMG of any muscle during the anticipatory time bin. Thus, the long latency response was task dependent rather than just automatic gain compensation from background central drive.[174] The underlying mechanisms for the attenuation of these long latency responses are not clear; however, loss of vestibular function is a plausible explanation.[24, 51, 69, 76] Long latency responses have been suggested to make effective contributions in protecting the limb against dynamic unpredictable force changes (Marsden, Rothwell, & Day, 1983). Subjects with BVL are not able to use vestibular input to detect changes in velocity and respond to perturbations adequately. The absence (or at least, major reduction) of vestibulospinal input in the BVL subjects resulted in a LLR at the knee joint that was inadequate to successfully restabilize, resulting in increased endpoint error when perturbed during the SLS task. The instability was even more profound when visual feedback for the vestibular deficit was absent.

Another possible contributor to the LLR is light touch. Light touch has been shown to influence the amount of sway during challenging balance tasks.[15] It may also influence the LLR when available. Welgampola (2001) noted that the LLR decreased when vision and light touch were allowed during platform posturography.[50] In his study, LLRs were largest when healthy subjects stood with a narrow base or on a compliant surface, deprived of vision and external support. This is consistent with the importance of vestibular function under these conditions. Long latency responses were preserved in most subjects even when vision and external support were available and a wider stance width was adopted, but were attenuated. This is similar to the pattern of responses seen in this study. Though not different at a statistically significant level between conditions for all muscles in this study, descriptively LLRs appear less with eyes open on firm surface than with eyes closed for VL, RF, and MH. This was also the case for foam, but again, not significantly so. Both groups were allowed light touch using similar levels, so the LLR reduction in the BVL group is not likely explained by the addition of touch. This study provides additional information about the vestibular contributions to the LLR. BVL subjects had absent vestibular function. Since the LLR was reduced in these individuals, it suggests that the vestibular system plays an integral role in the LLR during the perturbed weight-bearing SLS.

<u>Influence of Foam</u>

Foam was included as a condition in this study to determine if a compliant surface, in addition to vestibular loss, affected the variables studied. When the eyes were closed, this condition (it was theorized) would provide ambiguous sensory information, leaving BVLs at a greater disadvantage than controls. This was partially the case for performance accuracy and muscle activation strategies, but not for the LLR. On foam, Epe was different between surfaces and between groups. Overall, EPE was greater on foam than on the firm surface. This is likely due to the compliance of the surface; one has a greater distance to “sink” into the surface with a perturbation overshoot. On foam, however, BVLs showed less Epe than controls in contrast to their increase in Epe on the firm surface. This may be because BVLs moved through less excursion than controls on foam, as suggested by COP measurements taken during the task. There was no effect of foam on the LLR. One explanation for this may be that the foam condition altered the mechanics of the task so that it cannot be adequately compared to the firm surface. While standing on foam performing the SLS, a subject has no solid surface providing a shear force on the bottom of the foot as the knee flexes and pushes the bar forward. Instead, the subject sinks down into the surface and may adopt a forward lean strategy (vs. knee flexion) to advance the bar. Subjects may have sunk into the foam while flexing the knee, resulting in less knee flexion required to remain on the target. In addition, velocity was increased for both groups on the foam in the anticipatory bin for both nonperturbed and perturbed events. This may also indicate that both groups were using different strategies to track the target during the first 1/3 of the range of the SLS task.

Accuracy

Accuracy was an important component of this study. With CDP, subjects may anticipate subsequent perturbations and stiffen their legs to prevent a fall when they occur. This strategy is not possible during this task without compromising accuracy while tracing the computer template. Subjects in both groups demonstrated overshoot error over both days with perturbations, which reinforces that the perturbations were indeed, unexpected. They were unable to anticipate subsequent perturbations, which provides a condition that can adequately study the LLR without expectation confounding the results. The CNS is often “fooled” by perturbations in daily life, such as when anticipating a load will be heavy, then discovering when lifting it that the load is very light. The CNS prepares muscle activity to stabilize the body and to adequately tune muscles to lift the heavy load, then experiences a perturbation when the load does not require the force anticipated. Injury also can occur when individuals are focused on other tasks and fail to accurately anticipate perturbations. For example, when a person is preoccupied carrying groceries or talking while stepping off a curb, the CNS may incorrectly program the height necessary to negotiate the step. During the SLS, subjects focused on tracking a target and tuning the CNS to activate muscles in a pattern to maintain accuracy. When resistance suddenly was released, the perturbation fooled the nervous system that was unable to anticipate it. This provides important information into how the CNS responds when accuracy goals are incorporated into the task.[61] The data also suggest that the vestibular system plays a role in the response post perturbation during the SLS task, as demonstrated by differences seen in adults with BVL for endpoint error and the LLR.

Determining if the CNS can be trained to assume a mode that is able to efficiently and effectively respond to unexpected perturbations is an important area of study. Incorporating perturbations into training programs may help those with vestibular impairments respond more effectively to real-life disruptions.

Vestibular Rehabilitation

Vestibular rehabilitation has helped individuals with bilateral vestibular disorders improve self-perception of health, gaze stability, and balance, as well as make decreased double limb support during walking, improved gait speed, and gain higher levels of activity.[150, 154-155, 175] Due to the severity of vestibular loss bilaterally, full recovery of function is rare, and many are left with residual impairments.[150, 154-155, 157, 175] As this study suggests, altered muscle activation of the lower extremity may also be a consequence of BVL. Rehabilitation has not, however, focused on training responses to perturbations during functional movements. Exercises incorporating perturbing forces to the lower extremity during a single leg squat (SLS) exercise provided in a controlled and progressive manner can provide the neuromuscular system the opportunity to develop successful compensatory muscle activation patterns in response to unexpected and potentially destabilizing forces at the knee.[81] This may enhance neuromuscular responses, which in turn may lead to increased function for the individual.

Limitations

Limitations of this study include the small sample size. This is a clearly defined patient group and, as such, they are not frequently encountered clinically. Despite finding 11 subjects who met the BVL inclusion criteria to participate in this study, the task was too challenging a task for several individuals due to other musculoskeletal impairments. Subjects who did participate each had complete BVL as determined via medical testing, although there was variability in performance throughout the dependent measures examined. Differences in age and activity level may have contributed to this variability. This task is more challenging than CDP and, as such, a higher degree of competence is required to perform the SLS. These BVL subjects and their matched controls were on level ground with regard to the other physical requirements required to participate, but it remains that the BVLs were different in that they lacked vestibular input.

Conclusion

In summary, we found that well-compensated individuals with BVL have similar absolute error as healthy controls when performing the SLS task but greater endpoint error with perturbations on a firm surface. They also have lower long latency responses of the rectus femoris with eyes closed, despite similar levels of background activity, suggesting pure gain compensation (background quadriceps activity) is not the mechanism. Future studies with varied training schedules may help to determine if accuracy and long latency responses are trainable with practice in this group. A limitation of the available data on individuals with BVL is that few measures of actual changes in motor behavior (i.e., accuracy measurements) are available during functional tasks. Future work including skill training is recommended.

The results of our study also have implications for rehabilitation. Control of postural orientation and equilibrium can be significantly improved in patients with bilateral vestibular loss as long as it is considered a complex, sensorimotor skill that must be learned with appropriate feedback and active, context-specific training (Horak, 2009). Because neuromuscular function is influenced by the appropriate training programs, rehabilitation after BVL should focus on using behaviorally relevant tasks that are challenging to the motor and sensory systems. Data suggest that the vestibular system contributes to the LLR and was likely responsible for the altered muscle responses to the unexpected perturbation. In addition, further research is needed to determine if these LLRs can be trained over time to improve functional stability in this patient population.

CHAPTER 4

CONCLUSIONS

Approximately 20% of the general population is affected by a vestibular disorder,[1] with reports as high as 1% of clients treated in vestibular clinics living with bilateral vestibular loss.[176] Despite research on balance strategies with platform perturbations, limited information exists on neuromuscular performance of the knee with perturbations during functional tasks. Improved understanding of the effects of BVL on neuromuscular control of the knee will aid researchers and clinicians in developing rehabilitation programs that address the adaptations and balance deficits that occur with vestibular loss.

These studies presented a novel method to assess neuromuscular performance in healthy subjects and those with BVL using the SLS task. Using a specially designed apparatus that provided controlled resistance to knee motion and perturbations during the task, [61, 77, 79, 81] voluntary and reactive components of the neuromuscular control system during the performance of a functional, dynamic, weight-baring activity were examined. Using visual feedback initially, the rate and amplitude of the movement was controlled and errors in performance were quantified, along with the patterns of muscle activation. In addition, the EMG responses to sudden unexpected perturbations were characterized during three different visual conditions. By also examining individuals with BVL, we anticipated that a greater appreciation of compensatory knee stabilization strategies in this population would be gained. The hypotheses and conclusions from each study are discussed.

Hypotheses

Hypothesis 1a

Subjects will demonstrate learning and retention of the perturbed SLS task under both visual and non-visual feedback conditions. However, accuracy of tracking the target during the eyes open/no template and eyes closed conditions will be lesser than the eyes open condition.

In support of this hypothesis, subjects showed improvements in performance with training for both nonperturbed and perturbed trials under the eyes open condition of feedback from Day 1 to Day 2. No improvements in accuracy were noted over days for the no template or eyes closed conditions, however. Performance accuracy was lesser during the non-visual feedback conditions than when feedback was available. Although performance accuracy improved with training, endpoint error remained unchanged with perturbations. These results suggest that that the perturbations in this study were truly unexpected. Weight bearing tasks such as the SLS can be used as a tool to evaluate and quantify motor learning of the lower extremity.

Hypothesis 1b

Quadriceps activity will increase and hamstring activity will decrease in all conditions of feedback as learning occurs. However, quadriceps and hamstring muscle activity will each be higher in the no template and eyes closed conditions than during the eyes open condition. As accuracy of performance improves, there will be a concomitant decrease in the LLR with training.

Our results partially supported this hypothesis. Improved performance (decreased error) with this task has been associated with increased quadriceps and decreased hamstring coactivation. Over the 2-day period, RF during flexion and VL during extension demonstrated increased activity for the nonperturbed events, while VL showed an increase only during flexion of the perturbed events. In the absence of visual feedback, however, both VM and VL demonstrated increased activity during the extension phase of the non perturbed and perturbed events. No change was seen in hamstring activity over days. Hence, with learning of the task, quadriceps activity did change, while hamstrings did not. However, on the foam surface, MH activity was greater as well as VM and VL compared to the firm surface. The results of this study showed that muscle coactivation is altered according to demands of the task and the feedback/type of surface available.

The normalized LLR was increased on Day 2 for VM and RF. The absence of vision also affected VM and MH, with larger responses seen in the eyes closed condition. There was no change in the anticipatory muscle activation over days, without vision, or on a compliant surface, indicating that subjects were not able to “prepare” for the perturbation. VL was the only muscle modulated with perturbation trials, decreasing on Day 2 during the volitional time bin. On the foam surface, however, activity was increased for VM in the reflex time bin, VL in the volitional time bin, and MH during both reflex and volitional time bins of the perturbed events, indicating that the compliant surface did affect muscle activation strategies in the time frame 200-400 ms post perturbation.

Hypothesis 2a

Both groups will show equivalent error in the eyes open condition. However, performance error during the eyes closed condition will be greater in the BVL group.

In agreement with the primary hypothesis, BVL and control subjects performed equivalently on the SLS task with eyes open on the firm surface as measured with absolute and variable error. Accuracy of performance was expected to be similar in both groups, but the magnitude of difference in performance accuracy between the visual and non-visual feedback conditions was anticipated to be greater in the BVL group than in the control group. This was partially supported by the results as no group effect was seen in absolute or variable error for nonperturbed or perturbed events in the eyes open or closed conditions. In addition, there was no significant decrease in error over training days, indicating that neither group was able to adequately learn the task. However, endpoint error was increased in the BVL group on the firm surface regardless of the visual condition.

Hypothesis 2b

Muscle activation strategies used to perform the controlled SLS exercise will be different between individuals with intact vestibular systems and those with bilateral vestibular loss. Individuals with BVL will exhibit reduced quadriceps activation and greater hamstrings coactivation to perform the task. Subjects with BVL will also demonstrate reduced long latency responses compared to the controls, especially in the absence of vision. However, both groups will be able to adapt to the task as learning improves, as demonstrated by significant changes in the magnitude of responses with training.

VL showed greater activity in BVLs during the anticipatory time frame prior to perturbation. There were no other differences in muscle activation between groups or over days. Vision affected performance, as only 3 subjects with BVL could perform the task on the firm surface with eyes closed and only 2 could do so on the foam cushion without vision. However, those who could perform the task in this condition were not significantly different from controls in terms of error. Muscle activity was greater when the task was performed in the absence of visual feedback, however, for VM and VL. In addition, the LLR was reduced in MH for both groups with eyes closed and for RF for the BVLs with eyes closed. No significant effect of foam was seen on LLR for either group.

The findings of greater overshoot error and change in the LLR for the BVL subjects suggest that bilateral vestibular loss contributes to deficits in neuromuscular control of the knee in this group. Both groups showed comparable performance (in terms of absolute error) in the absence of vision suggesting that vision (for the individuals who could do the task with eyes closed) does not influence task accuracy differently during this functional weight-bearing task in competent BVL individuals. However, without vision, VM and VL activity were increased during extension phases of the SLS task for both groups. This was also the case on foam, with both flexion and extension phases impacted similarly. MH activity was increased on foam for flexion and extension of the nonperturbed events and during flexion only for the perturbed events. It was noticed that the mean activity of the VL in the anticipatory phases was greater in the BVL group.

Summary

Task dependent muscle activation of the knee is characterized by a complex interaction of the many systems. The ability to move efficiently and respond effectively to perturbations is an important factor contributing to the performance of functional activities. The long latency response is increased with training in healthy, young adults. It is also reduced in individuals with bilateral vestibular loss. This is the first study to examine the effects of BVL on performance accuracy as well as anticipatory, reflex, and volitional muscle activation of the knee during a dynamic weight-bearing task.

The main purpose of the present study was to examine and compare motor performance and strategies used during the controlled single leg squat task with training over 2 days. Also, we hoped to compare long latency responses of the quadriceps and hamstrings in response to unexpected perturbations of this task in healthy individuals and those with BVL. The first study provided information about the ability to improve performance accuracy with perturbations based on the feedback available. It also showed concomitant changes in the LLR of quadriceps muscles with learning. In the second study, it was found that competent subjects with BVL show similar performance accuracy as healthy individuals during the SLS, with the exception of endpoint error. Muscle strategies are slightly different and vary on firm and foam surfaces. A significant finding was that the LLR is reduced in this group in response to unexpected perturbations, especially when visual feedback is absent. Rehabilitation and/or time living with bilateral vestibular deficiency can lead to a reorganization of the central nervous system, which may partly explain the alterations in neuromuscular control. More research is needed to determine the relationship between the long latency response and fall risk and if different training dosages with perturbations affect these in both healthy and vestibular-deficient populations.

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