

## Assessment of neuromuscular response characteristics at the knee following a functional perturbation

By: Sandra J. Shultz, David H. Perrin , J. Milton Adams , Brent L. Arnold , Bruce M. Gansneder, Kevin P. Granata

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### **Abstract:**

Previous research indicates that both the extent and timing of muscular activation at the knee can be influenced by muscle activity state, joint angle, weight-bearing status and trunk position. However, little research to date has evaluated protective neuromuscular response times and activation patterns to an imposed perturbation with the knee joint in a functional, weight-bearing stance. Hence, we designed a lower extremity perturbation device to produce a sudden, forward and either internal or external rotation moment of the trunk and femur relative to the weight-bearing tibia. Surface electromyography (EMG) recorded long latency reflex times of the medial and lateral quadriceps, hamstring and gastrocnemius muscles in 64 intercollegiate lacrosse and soccer players in response to both internal and external rotation perturbation. We found the gastrocnemius fired significantly faster than the hamstring, which in turn fired significantly faster than the quadriceps. There was also a significant difference in activation times of the medial and lateral hamstring not found for the quadriceps or gastrocnemius muscles. Our findings confirmed that reactive neuromuscular responses following this functional perturbation differ markedly from those previously reported using seated, partial weight-bearing and/or uni-planar models under relaxed conditions. **Keywords:** Electromyography; Neuromuscular control; Anterior cruciate ligament; Long latency reflex; Reaction time

### **Article:**

#### **INTRODUCTION**

Multiple studies have indicated that the speed at which muscle force can be generated may be a more important determinant in providing joint stabilization and potential injury prevention than traditional muscular strength [1– 5]. In studies evaluating the effects of training and fatigue on muscular response times, it is apparent that timing can be significantly improved with proper training and significantly delayed when fatigued [5,6]. Therefore, timely activation of the musculature in response to joint perturbations appears to be a relevant issue when considering the system's ability to stiffen the joint and prevent excessive joint deformation and ligament strain. Hence, considerable research has examined muscular response characteristics and activation patterns while subjecting the knee to sudden perturbations or stress [1,3,7–15]. The majority of these studies have measured response characteristics under resting conditions while applying joint perturbations in an unloaded or partially loaded lower extremity which does not

mimic most ligament injury mechanisms [1,3,7,12,14–16]. However, there is sufficient evidence to suggest that muscular response characteristics under sudden loading differ considerably depending on the activity state of the muscle [17–21], joint and trunk position [22–24], movement velocity [11], weight-bearing status [24–26] and prior training [1,6].

Reflexive activation following a perturbation with the muscle(s) actively contracting appears to differ considerably from a muscle at rest. When a contracting muscle is unexpectedly stretched, the afferent activity can vary substantially from trial to trial and can result in a weak or absent monosynaptic response and a more prominent long loop response [27]. Conversely, when a contracting muscle is suddenly unloaded, a period of reflexive silencing in the electromyographic signal is often observed [20,28]. These response variations have been demonstrated by Marsden [20] following sudden loading and unloading of a constantly contracting muscle.

Research also indicates that weight-bearing status and joint angle may impact neuromuscular activation patterns and joint stability at the knee. Hsieh and Walker [25] and Markolf et al. [26] used cadaver specimens to assess the effects of various joint compressive loads on joint stiffness and laxity. Both studies demonstrated significant increases in stiffness and corresponding decreases in laxity with increasing levels of compressive loads at various flexion angles and directions of applied force. Other cadaveric studies have demonstrated that the angle of knee flexion will also influence knee stability, as the hamstrings are ineffective in limiting anterior and tibial rotary translation at knee flexion angles less than 15–30° [22,23].

In vivo, Wilk et al. [24] demonstrated that muscle forces and the degree of co-contraction differed considerably between open and closed kinetic chain exercise. They compared tibiofemoral joint forces and electromyographic activity (normalized to per cent maximal voluntary isometric contraction) of the quadriceps, hamstrings and gastrocnemius during open-chain knee extension and closed-chain leg press and squat exercises. Compressive forces were significantly greater in the squat compared to leg press which was in turn significantly greater than the leg extension exercise. Additionally, they found that hamstring co-contraction was influenced by trunk position relative to the knee joint with greater activation noted when the trunk was placed directly over the knee in the squat as compared to a more posterior displacement in the leg press. Hamstring co- activity and peak activity also varied considerably, depending on knee flexion angle. These findings indicate that weight-bearing status (axial loading), knee flexion angle as well as the relative position of the trunk to the knee can significantly impact joint stability by altering joint compressive forces and co-activation patterns [24].

From these findings, it is reasonable to question whether invoking a perturbation in a seated, moderate weight-bearing [15] or open-chain [3,14] limb position with the trunk positioned posterior to the knee joint will result in the same muscular activation patterns as a more functional, full weight-bearing position. Reflexive activation patterns in response to sudden loading of the knee in a functional weight-bearing application have yet to be studied in a controlled manner and may prove useful in understanding the neuromuscular control strategies that contribute to joint stiffening and stabilization during sport activity. Therefore, our purpose was to design a lower extremity perturbation device to assess neuromuscular response

characteristics and activation patterns of the knee musculature following an unanticipated perturbation in a single leg weight-bearing stance under active muscular conditions.

## METHODS

### *Subjects*

Sixty-four (32 female, 32 male) intercollegiate athletes (age=20.0±1.2 years, height= 172.5±9.1 cm, weight=72.4±13.1 kg) participated in the study. At the time of data collection, each athlete: (1) was actively participating in their sport; (2) had no history of surgery for knee ligament injury in the dominant limb; (3) had no other history of injury to the dominant extremity in the last 6 months that might influence neuromuscular response characteristics at the knee; (4) was free of symptomatic foot, knee and hip malalignments that would detract from the subject's ability to perform a single leg squat; and (5) was otherwise healthy with no current complaints of pain. Prior to participation in the study, all subjects read and sign an informed consent form approved by the University's Institutional Review Board for the Protection of Human Subjects.

### *Instrumentation*

An eight-channel Myosystem 2000 EMG (Noraxon, Scottsdale, AZ) was used to record temporal activation patterns immediately following lower extremity perturbation. Unit specifications included an amplifier gain of 1 mV/V, a frequency bandwidth of 16–500 Hz, CMRR 114 dB, input resistance from 20 MΩ to 1 GΩ, and a sampling rate of 1000 Hz. Bipolar Ag–AgCl surface electrodes measuring 10 mm in diameter with a centerto-center distance of 2.0 cm were placed over quadriceps, hamstring and gastrocnemius muscles. To acquire, store and analyse the electromyography (EMG) data, we used Data Pac III Version 1.1 Lab Application Systems software (Run Technologies, Laguna Hills, CA).

To produce the perturbation, we designed a lower extremity perturbation device (LEPD) to produce a sudden, unanticipated forward and either internal rotation (IR) or external rotation (ER) moment of the trunk and femur relative to the weight-bearing tibia (Fig. 1). Subjects were restrained by two horizontal kevlar cables measuring 4 ft in length, one attached at each hip via a Viper Waist belt (Speed City, Portland, OR). The release end of each cable was attached to a height-adjustable, wall-mounted quick-release trigger mechanism. Each release mechanism consisted of an archery bow release, an activating trigger and an impulse switch. Upon trigger release, a contact switch was simultaneously depressed, sending a voltage signal to the computer via a BNC connection to trigger the recording of EMG activity and accurately mark the time of stimulus. One trigger produced a 9 V signal and the other a 3.4 V signal to allow us to differentiate between ER and IR events.

The subject's position was standardized while attached to the LEPD using an electrogoniometer (Penny

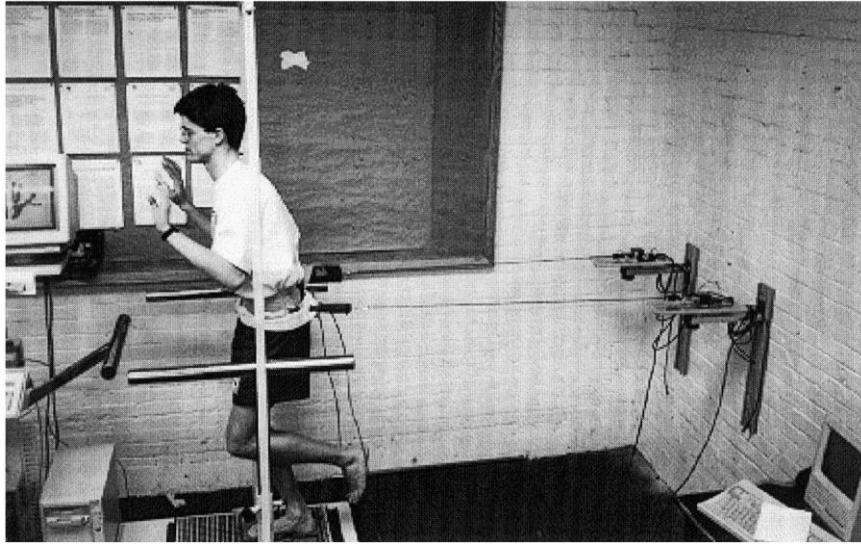


Fig. 1. Lower extremity perturbation device (University of Virginia, Charlottesville, VA).

and Giles, Santa Monica, CA) to confirm knee flexion angle and the Chattecx Balance System (Chattanooga Group, Inc., Hixson, TN) force platform to consistently position the trunk so that the subject's center of pressure was directed through the middle of the foot.

### **Reliability and validity of the LEPD**

To assess validity and response consistency of the LEPD, the Chattecx Balance System was used to provide an objective measure of postural response to the imposed perturbation. To determine validity, we compared maximum postural sway measures prior to and following the release of cable tension. Individual subject trials (10 trials each) were averaged and used for data analysis. A repeated measures ANOVA confirmed that maximum sway distance was significantly increased following perturbation [ $F(1,9)=1277.30$ ,  $P<0.0001$ ].

To establish reliability or consistency of the subject's response following perturbation, day-to-day variability in postural sway measures were assessed. Using Formula (2,k) as described by Shrout and Fleiss [29], intraclass correlation coefficients (ICCs) were calculated for maximal medial/lateral and anterior/posterior sway for IR and ER perturbation: IR, 0.81 horizontal sway and 0.84 vertical sway; ER, 0.69 horizontal sway and 0.71 vertical sway.

Because of the dynamic nature of postural response to an imposed functional perturbation, some variation in postural responses between trials was expected. This in part explains our rationale for using an average of multiple trials for data analysis in the research protocol. While the reliability coefficients were not exceptionally high, they adequately support the premise that the LEPD creates a forward and rotational perturbation in a reasonably consistent manner. This was further supported qualitatively by subjects consistently reporting a forward and rotational sensation and by the tester's visual observation of the Chattecx Balance System display screen which tracks the movement of the subject's center of pressure.

### ***Procedures***

The subject's skin was prepped and surface electrodes were placed in a parallel arrangement over the midline of the vastus lateralis and vastus medialis (midway between the motor point and distal tendon), medial and lateral hamstrings (mid-belly), and medial and lateral gastrocnemius (mid-belly) muscles. All electrode placements were confirmed with manual muscle testing and checked for cross-talk. An electrogoniometer was positioned over the lateral aspect of the knee joint using a line from the greater trochanter to lateral femoral condyle, and a line from the fibular head to lateral malleolus to align the stationary segments along the long axis of the lateral thigh and lower leg. The electrodes and electrogoniometer were further secured with an elastic bandage to prevent cable tensioning and movement artifact. Finally, the waist harness was attached to fit snugly at the level of the anterior superior iliac spine (ASIS) and the height of the trigger mechanism was adjusted to a position level with the ASIS with the subject in the flexed knee test position.

The position of each subject prior to perturbation was standardized using the following procedures to achieve a consistent knee angle and trunk position relative to the knee joint. The subject stood on the Chattecx Balance System platform with the foot plate sized to allow approximately 1/2-inch clearance beyond the toe and heel. With cables attached and height properly adjusted at the wall, subjects were instructed to stand upright with back straight, arms folded across the chest and to lean into the cables with their hips. From this position, the subject was asked to slowly bend their knee until they achieved a knee flexion angle of  $\approx 35^\circ$  as indicated by the electrogoniometer. Using the Chattecx's visual training target which displayed center of pressure relative to the foot, the footplates were adjusted (anterior/posterior) on the platform until the cursor was located in the middle of the target (bullseye) once the test position had been achieved. Subjects were allowed to practice the movement sequence until they were able to consistently reproduce this position.

Prior to testing, subjects were provided a minimum of three practice trials for both IR and ER perturbation to become acquainted and comfortable with the task. Once the subject was properly positioned, either the left or right cable was released at an unanticipated, random time interval of 1–10 s to produce either IR or ER perturbation. Subjects were instructed to look straight ahead and react to the perturbation by trying to maintain their single leg balance following cable release. Each subject completed 20 trials using the dominant leg. Trials were equally split between IR and ER with order randomized to avoid anticipatory responses. Subjects rested 30 s between trials and supported their weight on their non-dominant leg during rest periods to avoid fatigue.

### *Signal processing and analysis*

From time of stimulus, EMG activity was recorded for 100 ms prior to and 500 ms following cable release using a trigger sweep acquisition mode. The acquired raw signal was digitally processed, using a centered (symmetric) root mean square (RMS) algorithm, with a 10 ms time constant. Two threshold event buffers with differentiating upper peak filters identified and separated the IR and ER events via the differential voltage recorded from their respective release triggers. Individual events for IR and ER were visually inspected and selected based on the following criteria: (1) long latency reflex identified within 150 ms following cable release; (2) baseline muscle activity sufficiently quiet and stable to insure an acceptable signal-to-noise ratio; (3) readable signal obtained from all six muscle sites and (4) signal free of movement artifact to

allow clear interpretation of the signal. If a trial failed to meet any of the above criteria, the event was deleted and excluded from further analysis. Using the first five trials that met the selection criteria, the signal was averaged for each muscle and served as the representative signal to determine muscle onset times.

Muscle response time (MRT) was defined as the time delay between the initiation of the perturbation and the onset of EMG activity at the long latency response level, within the first 30–150 ms following cable release. Because of the inconsistency of the monosynaptic response [20,27] and the difficulty of separating voluntary responses from long latency responses under active muscular conditions, we evaluated only long latency responses in this study. Although the time frame in which this response is thought to occur varies considerably in the literature [15,20,27,30], this time frame is in agreement with literature specifically evaluating long latency reflexes under active/reactive muscular conditions [20,30] and was confirmed during pilot testing of our research model.

The 100 ms pre-trigger reference interval was used to determine average baseline activity for each muscle just prior to cable release. Reflex onset time was then defined as the time point when myoelectric activity first exceeded two standard deviations of the average baseline activity in the hamstrings and gastrocnemius, and one standard deviation in the quadriceps. In order to be considered a reflex event, myoelectric activity had to exceed the given threshold level for a minimum of 10 ms. We used a 1 SD threshold for the quadriceps because of its higher baseline activity in the pre-perturbation testing position compared to the gastrocnemius and hamstring muscles. Given the quadriceps higher baseline activity, a 2 SD threshold resulted in artificially longer delays when compared to the other muscles. Hence, we found the sensitivity of a 1 SD threshold for the quadriceps to produce more reliable event markings, and to be more comparable to the 2 SD threshold of the hamstrings and gastrocnemius. The improved sensitivity in determining muscle onset was easily confirmed visually due to a reflexive silencing in the quadriceps that consistently preceded reflex activation. (Although this reflexive inhibition in the quadriceps would technically be considered the first reflex event, our interest was in the time to reflex activation resulting in muscle stiffening.) Fig. 2 illustrates a typical averaged signal obtained, with stimulus trigger and onset times marked for each muscle using the described methods.

Based on the data acquisition and processing just described, muscle response times for the six muscle sites and two release conditions (IR and ER) were compiled in spreadsheet format and loaded into SPSS Statistical Software Package (Allegiant Technologies, Inc.) for analysis. Two separate one-way repeated measures ANOVAs with two within variables (muscle group, muscle side) were used to compare long latency reflex response times and activation patterns by muscle groups (quadriceps, hamstring and gastrocnemius) and muscle side (medial vs. lateral) for both external rotation and internal rotation perturbation conditions. Tukey's post hoc analysis further evaluated any significant main effects and interactions. All analyses were performed at an alpha level of  $P < 0.05$ .

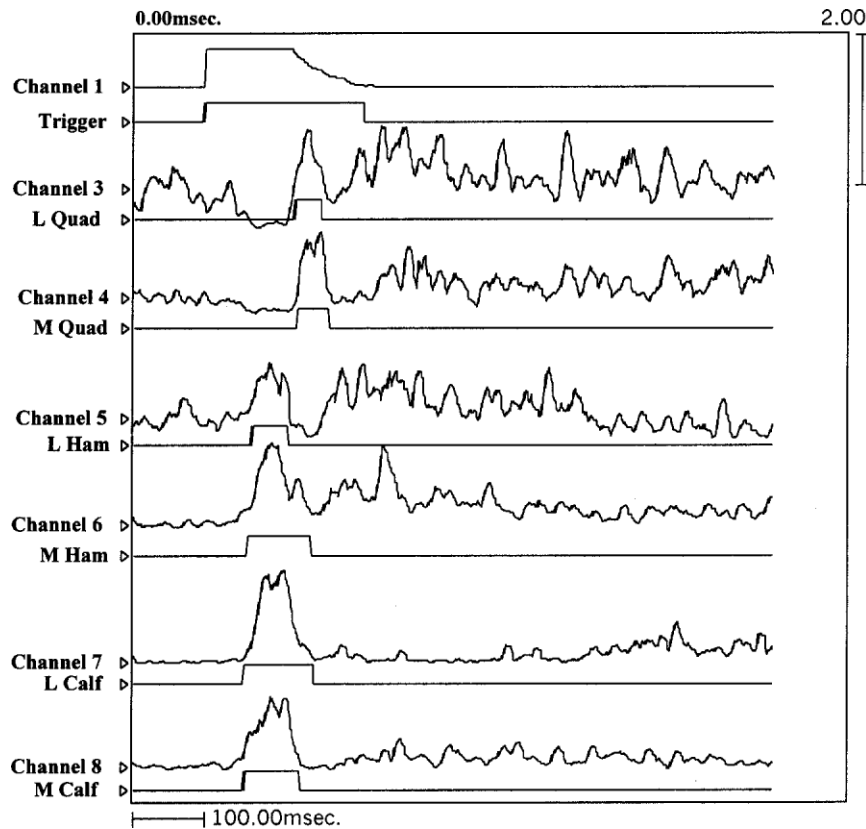


Fig. 2. Averaged signal obtained from a single subject, illustrating time of trigger onset and time of reflex onset for each muscle.

## RESULTS

### *Measurement reliability and error variance*

To assess the reliability and error variance of the acquired data, ICCs and standard error of measurements (SEMs) were computed for (1) consistency of subject performance across time (early vs. late trials), and (2) intratester reliability for trial selection criteria and computer assisted determination of muscle onset time. All data were collected by the primary investigator.

### **Performance consistency across time**

In order to assess the consistency of subject performance across repeated trials and to determine whether a training effect was occurring in later trials, the first two acceptable trials were compared to the last two acceptable trials for each muscle under both IR and ER perturbation conditions. Twelve subjects were chosen in a systematic random fashion by selecting every fifth subject (N=12). ICCs and SEMs were calculated using ICC formula 2,k [29] and are listed in Table 1. We found fair to good performance consistency between early and late trials for most muscles, with ICCs ranging from 0.64 to 0.83. However, performance consistency of the lateral hamstring and medial calf (IR), and lateral and medial calf (ER) were quite poor, ranging from 0.12 to 0.31.



**Table 1**  
Intraclass correlation (ICC) and standard error of measurement (SEM) to determine trial-to-trial reliability between the average onset times of the first two and last two acceptable trials ( $N=12$ )

	L Quad	M Quad	L Ham	M Ham	L Calf	M Calf
External rotation						
ICC	0.70	0.83	0.64	0.68	0.31	0.18
SEM	9.19	6.17	8.89	5.80	6.79	8.12
Internal rotation						
ICC	0.76	0.83	0.12	0.70	0.80	0.26
SEM	7.24	7.24	17.86	5.36	5.04	10.18

### Trial selection and determination of muscle onset time

One month following the completion of data analysis, data interpretation methods were repeated on the same 12 subjects to assess intratester reliability of trial selection and computer assisted determination of muscle onset times. ICCs and SEMs using formula 2,k [29] are displayed in Table 2. ICCs ranged from 0.85 to 0.98, indicating that the investigator was able to consistently select acceptable trials and obtain similar results with repeat analysis.

**Table 2**  
Intra-tester reliability estimates for trial selection criteria and computer assisted determination of muscle onset times (using first five acceptable trials) ( $N=12$ )

	L Quad	M Quad	L Ham	M Ham	L Calf	M Calf
External rotation						
ICC	0.98	0.98	0.95	0.97	0.87	0.89
SEM	2.17	1.84	2.95	1.38	3.34	2.46
Internal rotation						
ICC	0.92	0.98	0.98	0.94	0.93	0.85
SEM	4.42	1.95	3.55	2.06	1.66	3.26

### Statistical results

Table 3 list the means, standard deviations and standard errors for muscle reaction time of the medial and lateral quadriceps, hamstrings and gastrocnemius muscles for both external and internal rotation perturbation.

### Muscle reaction time

There was a main effect for muscle group for both ER [ $F^{(2,126)}=141.45$ ,  $P<0.0001$ ] and IR [ $F_{(2,126)}=137.23$ ,  $P<0.0001$ ]. Tukey's post hoc analysis indicated the gastrocnemius muscle group responded significantly faster than the hamstring group, which in turn responded faster than the quadriceps muscle group (Fig. 3). There was also a significant interaction between muscle group and muscle side for both ER [ $F_{(2,126)}=13.38$ ,  $P<0.0001$ ] and IR [ $F_{(2,126)}=27.07$ ,  $P<0.0001$ ]. Plotting the interaction (Fig. 4) and Tukey's post hoc analysis revealed a



significant difference between the medial and lateral hamstring while no significant difference was found between the medial and lateral gastrocnemius or medial and lateral quadriceps.

**Table 3**  
Mean, standard deviation (SD) and standard error (SE) of muscle reaction times (msec) for internal and external rotation perturbation (N=64 total)

	Quadriceps		Hamstrings		Gastrocnemius	
	Lateral	Medial	Lateral	Medial	Lateral	Medial
<b>Internal rotation</b>						
Mean	94.27	91.00	76.86	58.41	56.09	53.23
SD	18.57	17.81	22.39	11.05	9.39	8.82
SE	2.32	2.23	2.80	1.38	1.17	1.10
<b>External rotation</b>						
Mean	95.34	93.03	70.41	60.05	57.69	55.84
SD	20.86	21.24	15.64	9.14	9.24	8.18
SE	2.61	2.66	1.96	1.14	1.15	1.02

## DISCUSSION

This study establishes the use of a lower extremity perturbation model to provide a valid and reliable assessment of muscular timing and activation patterns following a functional, weight-bearing perturbation. Our findings confirmed that muscular responses following a functional, rotation perturbation differ considerably from those previously reported using non/partial weight-bearing or uniplanar perturbations under resting muscle conditions.

### *Measurement reliability and error variance*

Given the many intrinsic and extrinsic factors that are known to influence the EMG signal [31–34], the inherent variability in EMG data is a concern and must be considered in the overall analysis of muscular response characteristics. Therefore, in order to quantify the expected variations for this research model, reliability estimates and SEMs were computed for subject performance consistency and intratester interpretation reliability.

### **Subject performance consistency**

In order to obtain the low ICC values for the medial gastrocnemius (IR and ER), lateral gastrocnemius (ER) and lateral hamstring (IR), there needed to be an appreciable variation in response times between trials or in subjects by trials. Analysis of the sources of variance for the lateral gastrocnemius (ER) and medial gastrocnemius (ER and IR) indicate that low ICCs were due primarily to a large ratio of error variance (random subject variability within trials), not trial variance (systematic change across time) compared to the between subject variance. Therefore, with the possible exception of the lateral hamstring, there was no apparent learning or training effect over time that could affect the validity of the trial selection criteria. Only the lateral hamstring (IR) demonstrated both a high trial variance and a high error variance. Consistent with this finding, mean and standard deviation values for lateral hamstring muscle reaction time (IR) decreased by =10 ms and =15 ms respectively in later trials compared to earlier trials. Therefore, it appears that the lateral hamstring responded quicker and more consistently with repeated testing. Why this response variation occurred only in the lateral hamstring (IR) is unclear and should be addressed in future studies.

A companion to the intraclass correlation is the standard error of measurement (SEM) which provides a unit value of measurement precision or in this case, subject performance consistency. With the exception of the lateral hamstring in response to internal rotation perturbation, SEM values ranged from 5 to 10 ms. Taking a 95% confidence interval around these values, individual muscle reaction times varied by  $\pm 10$ – $20$  msec of their “true” muscle reaction time. Clearly, these values indi-

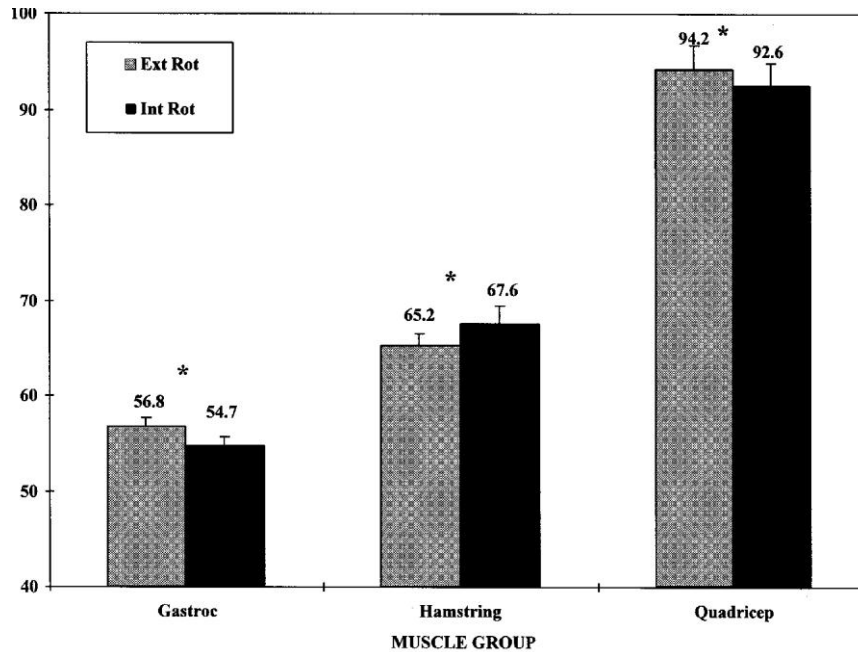


Fig. 3. Comparison of muscle group response times for both internal and external rotation perturbation. Values are means and error bars indicate SE. Asterisk indicates significance relative to other muscle groups (Tukeys,  $P < 0.05$ ).

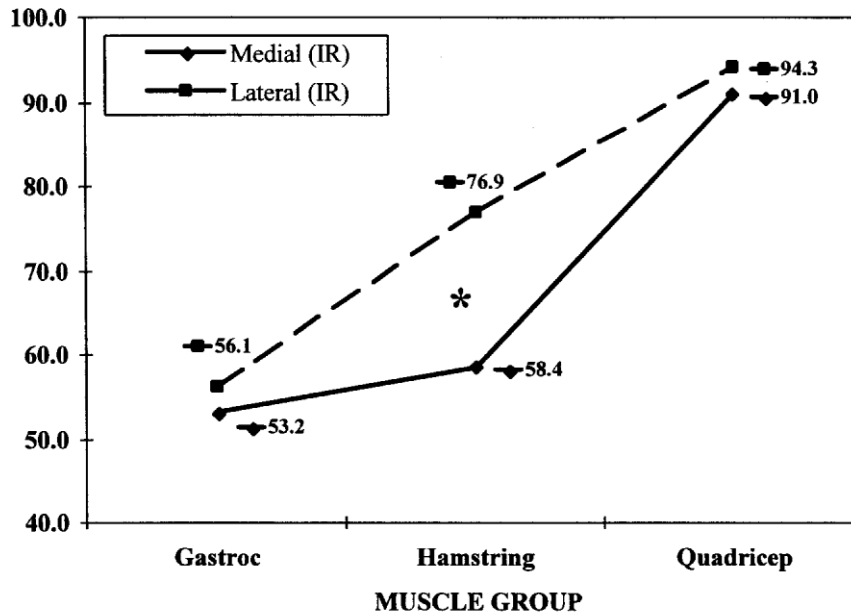


Fig. 4. Interaction of muscle group and muscle side, internal rotation perturbation. Values represent means. Asterisk indicates a significance difference between the medial and lateral hamstrings not found for quadriceps or gastrocnemius.

cate that subjects vary considerably in their neuromuscular responses on individual trials.

Because of the dynamic nature of the measured functional response to perturbation, low ICCs and high SEM values were not surprising. Human performance is variable, and it is unlikely that muscular responses would be identical across all trials when correcting for dynamic postural disturbances. This response variation was also demonstrated in postural sway measures during reliability and validity testing of the perturbation device. This supports the primary rationale for determining muscle onset based on a composite or average of five trials rather than a single, individual trial. Thus, by taking an average of five trials, a more general representation of the subjects performance, muscle response characteristics can be obtained.

### *Measurement of long latency responses*

Consistent with previous research [20,27] and our pilot data, the long latency response was the first to occur, with monosynaptic responses occurring infrequently in individual muscles and/or perturbation trials. There did not appear to be any consistency or trend as to when a monosynaptic response occurred in a particular subject, muscle or trial. However, monosynaptic responses were observed more frequently in the quadriceps than either the hamstrings or gastrocnemius muscles. Because of their infrequent occurrence, the presence of this short latency response was usually lost once multiple trials were averaged. However, on occasion, a reflex event was identified during the first 20–30 ms of the averaged signal, and was disregarded in the analysis because of its time occurrence. The occasional presence of these monosynaptic events helped to confirm that the long latency reflex was in fact the response under investigation.

### *Comparison of muscle group latency times*

Actual long latency responses in this study averaged 55.7 ms for the gastrocnemius, 66.3 ms for the hamstring and 93.4 ms for the quadriceps. While these values were within our expected time frame, we did note considerable differences when compared with previous literature evaluating long latency responses. These variations may be explained by the differences in research/perturbation models.

Wojtys and colleagues [5,6,12,15], reported substantially longer average values (130–150 ms) for all three muscles. In the perturbation method first described by Wojtys and Huston [15], the knee joint was isolated and the tibia was manually displaced anteriorly with muscles in a relaxed state prior to the perturbation. Therefore, it is likely that different mechanisms may have been responsible for the reflexive activation when compared to the current study. With the muscles in a relaxed state, other joint structures, such as capsule and ligament, may have provided the proprioceptive feedback to initiate the muscular response. Since ligaments are thought to provide proprioceptive feedback only at high loads and at end ranges [14], reflexive responses generated from ligaments are likely to represent longer delays than those generated by a muscle stretch reflex, the more likely mechanism of activation here. Other instrumentation methods, such as delays between the initiation of the stimulus and joint loading, EMG and data interpretation methods may also account for the discrepancies in latency times.

While Chan et al. [30] demonstrated values more consistent with our study, they reported faster times for the quadriceps (67.5 ms) and slower times for the gastrocnemius (108.3 ms) muscles. These discrepancies may also be explained by the manner in which the perturbation occurred to

elicit the response. In Chan et al. [30], the perturbation occurred in an open-chain condition, evaluating gastrocnemius responses with isolated ankle joint displacement, and quadriceps responses with isolated knee joint displacement. From these data, Chan et al. theorized that due to the greater distance of the gastrocnemius from the spinal cord and cortical centers, long latency times would be slower in the gastrocnemius than in the hamstring or quadriceps muscles. Conversely, the current study evaluated reflex responses of all three muscles simultaneously during a lower extremity perturbation in a closed-chain, weight-bearing stance. Given the forward and rotational nature of the perturbation, it is likely that a stretch reflex (or other proprioceptive stimulus) occurred first in the gastrocnemius and hamstring muscles, initiating an earlier response than was found in the quadriceps. A similar “distal to proximal” activation in a closed-chain environment has also been demonstrated by Nashner [35] with postural perturbations. However, while this explains the longer latency of the quadriceps relative to the gastrocnemius, it does not adequately explain the absolute shorter latencies we observed in the gastrocnemius. Shorter absolute latencies may be due to different subject populations (athlete vs. sedentary and spinal cord transected), activity state prior to perturbation, or variations in the data acquisition, processing or muscle onset detection methods employed. The later is difficult to determine, since Chan et al. did not report EMG instrumentation and interpretation methods in sufficient detail to allow an adequate comparison.

The increased latency of the quadriceps can be further explained by the presence of a “silent period” or reflexive silencing of muscle activity that we consistently observed in the quadriceps prior to reflexive muscular activation. This silent period was previously demonstrated by Marsden [20] with sudden unloading of an actively contracting thumb, generating a silencing of EMG activity at 50 ms and lasting an additional 50 ms before activity returned. It is likely that a similar “unloading” phenomenon occurred in the quadriceps, considering the forward perturbation with cable release while the quadriceps were actively contracting. Given the additional delay created by this inhibitory reflex, long latency reflex activation times averaging 93.4 ms would appear to be reasonable and in close agreement with Marsden’s [20] findings.

#### *Comparison of medial and lateral muscle latency times*

Because of the rotational nature of the imposed perturbation and evidence that medial and lateral aspects of the quadriceps and hamstrings can be preferentially activated with variations in limb alignment and direction of joint loading [9,36,37], myoelectric reflexes were recorded for both the medial and lateral aspects of each muscle. While we found a significant difference in the medial and lateral response times of the hamstring for both internal and external rotation, there was no difference in medial and lateral response times of the quadriceps or gastrocnemius muscles.

One potential explanation for the medial vs. lateral firing time differences in the hamstrings not found in the quadriceps or gastrocnemius muscles is their functional grouping. For the quadriceps and gastrocnemius muscle groups, there is a single common distal insertion point through the patellar and achilles tendons respectively. However, the medial and lateral hamstrings diverge at the knee and insert on the medial and lateral aspects of the posterior tibia respectively. As a result, the medial and lateral hamstrings function independently to either medial or laterally rotate the tibia relative to the femur [38]. Conversely, lateral rotation at the knee is not a unilateral function of the quadriceps or gastrocnemius muscles [38,39]. Therefore,

based on function alone, it would seem reasonable that the medial and lateral hamstrings should be capable of independent activation, especially in light of the rotational component of the imposed perturbation. However, if this were the only explanation, one would expect the lateral hamstring to respond quicker with rotation in one direction and the medial hamstring to respond quicker with rotation in the opposite direction. This was clearly not the case as the lateral hamstring was significantly slower for both internal and external rotation perturbation conditions.

An alternative explanation for these findings may be the differential innervation of the medial and lateral hamstrings. While the medial and lateral aspects of the quadriceps and gastrocnemius muscles are innervated by the same nerve (femoral and tibial respectively), the hamstrings are innervated by both the tibial (semitendinosus, semimembranosus and long head of biceps femoris) and common fibular (short head of biceps femoris) nerves [38]. Since a muscle stretch reflex is typically thought to initiate a non-specific activation of the stretched muscle and its synergists [27,40,41], muscles that are innervated by the same spinal nerve would be likely to be activated simultaneously. However, if the tibial and common fibular nerves were not similarly activated with the imposed perturbation, and the recording area of the biceps femoris was innervated by the common peroneal, this may result in different latency times from the medial hamstring. To further support this common nerve response theory, long latency reflex times of the medial hamstring, medial gastrocnemius and lateral gastrocnemius, each known to be innervated by the tibial nerve, were not found to be significantly different.

### *Recruitment order*

For both internal and external rotation, the same recruitment order was observed with the gastrocnemius muscle group firing first, followed by the hamstrings and then the quadriceps muscle groups. This activation pattern differs markedly from that of Wojtys and Huston [15] with anterior tibial translation. In the dominant extremity of healthy subjects, they reported a preferential recruitment order of initial activation of the hamstrings (45% of the time), followed by the quadriceps, then activation of the gastrocnemius at the intermediate response level. Unfortunately, the authors did not indicate how this recruitment order was determined and whether the time delay between each muscle recruited was in fact significantly different. When comparing mean differences in their intermediate latency times for each muscle ( $G=142.8$ ,  $H=141.3$ ,  $Q=143.1$ ) [15], the values do not appear to be significantly different. Furthermore, there was no difference in the medial and lateral responses of the hamstring ( $MH=141.1$ ,  $LH=141.4$ ). Therefore, it appears from their values that the anterior tibial translation perturbation may have elicited more of a “co-contraction” response of all muscles simultaneously to stiffen the knee rather than a true preferential recruitment order. However, this cannot be concluded definitively based on the information presented by the authors.

Conversely, our study demonstrated a clear preferential activation pattern, most likely due to the closed-chain environment in which the perturbation occurred. As mentioned previously, this activation order is similar to that of a postural response found by Nashner [35,42]. He theorized that during stance, the functionally related muscles of the leg will activate in a fixed pattern such that compensatory motions are simultaneously generated at the ankle, knee and hip [35]. In other words the gastrocnemius and hamstring would be activated together and the tibialis anterior and quadriceps would be activated together. He also demonstrated a distal to proximal firing pattern,

with the distal muscles preceding the proximal muscles by approximate 10–15 ms [35]. Although in Nashner's study the perturbation was created at the ankle by moving the foot platform rather than a cable release at the hip, it seems reasonable that a similar co-activation of functional groups would occur to maintain stance equilibrium. This, in fact, was the case for the medial and lateral gastrocnemius and medial hamstring, but not the lateral hamstring.

Other than the possible explanations previously given, it is unclear why the lateral hamstring would have greater variability and demonstrate significantly longer delays compared to the medial and lateral gastrocnemius and medial hamstring. Given the role of the hamstring in stabilizing the knee and preventing anterior and rotary tibial translation [1,13,22,23,43,44], it would seem that delays in hamstring firing may be detrimental to adequate knee stiffening and joint protection. However, with the perturbation model used in our study, it is reasonable that a protective delay or inhibition in lateral hamstring activation may occur with internal rotation of the femur on the fixed tibia (i.e. internal rotation perturbation condition). If the lateral hamstring were activated first, this would act to externally rotate the tibia, which could potentially increase, rather than relieve the rotatory stress placed on the anterior cruciate ligament. While this explanation seems logical, one would not expect the same protective response with external rotation of the femur on the tibia (external rotation perturbation condition). Although not as pronounced (10.4 ms vs. 18.5 ms) this delay was in fact present with external rotation as well. Further study is needed to adequately explain this delay.

## CONCLUSION

Little research to date has evaluated protective neuromuscular response and recruitment patterns with the knee joint in a functional environment. However, previous research indicates that numerous factors such as muscle activity state, knee and hip flexion angles, and trunk position can significantly influence both the extent and ability of the hamstring muscles to stabilize the knee [22–24]. Our results confirm that neuromuscular activation patterns in response to rotational joint perturbations in a functional, weight-bearing stance are quite different from those previously reported using non/partial weight-bearing or uniplanar perturbation models. Therefore, to fully appreciate the neuromuscular contribution to joint stability and control during functional activities when knee injuries are likely to occur, using similar functional testing environments may prove invaluable.

Furthermore, evaluating response characteristics in a weight-bearing stance provides an avenue by which to determine the effects of limb alignment (e.g. Q-angle, knee recurvatum, subtalar pronation), body position and joint flexion angles (knee and hip) on joint stress and neuromuscular control of stability. Whether any of these factors adversely affect lower extremity limb alignment sufficiently to diminish the ability of the neuromuscular system to adequately respond and stiffen the joint may be an important finding in our understanding of risk factors associated with the anterior cruciate ligament and other lower extremity injuries. Future research should therefore manipulate and evaluate the effect of these variables on muscle response times and activation patterns. It is our belief that the functional research model presented here provides a well-controlled environment in which to provide this evaluation.

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