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Whole body, long-axis rotational training improves lower extremity neuromuscular control during single leg lateral drop landing and stabilization ☆

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ABSTRACT

Background: Poor neuromuscular control during sports activities is associated with non-contact lower extremity injuries. This study evaluated the efficacy of progressive resistance, whole body, long-axis rotational training to improve lower extremity neuromuscular control during a single leg lateral drop landing and stabilization.

Methods: Thirty-six healthy subjects were randomly assigned to either Training or Control groups. Electromyographic, ground reaction force, and kinematic data were collected from three pre-test, post-test trials. Independent sample t-tests with Bonferroni corrections for multiple comparisons were used to compare group mean change differences ($P \leq 0.05/21 \leq 0.0023$).

Findings: Training group gluteus maximus and gluteus medius neuromuscular efficiency improved 35.7% and 31.7%, respectively. Training group composite vertical–anteroposterior–mediolateral ground reaction force stabilization timing occurred 1.35 s earlier. Training group knee flexion angle at landing increased by 3.5°. Training group time period between the initial two peak frontal plane knee displacements following landing increased by 0.17 s. Training group peak hip and knee flexion velocity were 21.2°/s and 20.1°/s slower, respectively. Time period between the initial two peak frontal plane knee displacements following landing and peak hip flexion velocity mean change differences displayed a strong relationship in the Training group ($r^2 = 0.77$, $P = 0.0001$) suggesting improved dynamic frontal plane knee control as peak hip flexion velocity decreased.

Interpretation: This study identified electromyographic, kinematic, and ground reaction force evidence that device training improved lower extremity neuromuscular control during single leg lateral drop landing and stabilization. Further studies with other populations are indicated.

1. Introduction

Lower extremity injuries sustained during sports activities can lead to long-term, and/or permanent health reductions (Hootman et al., 2007). Many knee injuries, including anterior cruciate ligament (ACL) rupture occur from non-contact mechanisms, such as landing from a jump or performing a sudden directional change (Yu et al., 2002). The loading response that occurs as the foot impacts the ground during single leg landings creates a chain reaction through multiple joint linkages (Joseph et al., 2008; Yu et al., 2002).

Poor lower extremity neuromuscular control during sports movements may create potentially injurious alignment and excessive knee joint forces (Pollard et al., 2010). Maintaining lower extremity neuromuscular control depends on both cortically programmed, and reflex-supplied mechanisms (Lephart et al., 2000; Williams et al., 2001). Lower extremity neuromuscular control represents unconscious efferent responses to afferent signals that facilitate dynamic lower extremity joint stability (Lephart et al., 2000). Movements such as a single leg lateral drop landing and stabilization (SLDLS) challenge lower extremity neuromuscular control by altering muscle force and length feedback during sudden jump landing deceleration (Fitzgerald et al., 2001; Williams et al., 2001). Effective eccentric muscle function is essential during jump landings to oppose potentially injurious alignment and extreme joint load states, particularly at the knee (Gerber et al., 2009; LaStayo et al., 2003, 2008).

Previous studies have reported that lower extremity neuromuscular control in healthy men during jump performance depended on enhanced muscle activation efficiency, and effective regulation of lower extremity angular joint displacement and velocity (Bosco et al., 1982; Bosco and Viitasalo, 1982). In a study of 13 subjects at a mean 3.3 years following unilateral ACL reconstruction, reduced lower extremity neuromuscular control was indicated by a greater time needed to achieve postural stabilization following a single leg step down from a 19 cm tall step at the surgical lower extremity compared to the non-surgical lower extremity (Colby et al., 1999). Increased knee injury risk has also been related to shallow hip and knee flexion angles at initial jump landing (Hewett et al., 2006a; Pollard et al.,

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2010). Increased peak hip and knee flexion among subjects with long-term ACL deficiency has been reported to be kinematic compensations to increase lower extremity neuromuscular control during single leg hop for distance performance (Gauffin and Tropp, 1992). The ability to reduce hip and knee flexion velocity during jump landings has also been related to improved lower extremity neuromuscular control and decreased knee injury risk among athletically active individuals (Hewett et al., 2006a).

Since having sufficient lower extremity neuromuscular control is vital to lower extremity injury prevention, training programs try to facilitate its development using movements that simulate sports function, particularly targeting the hip and knee joints (Hewett et al., 2006b; Imwalle et al., 2009; Myer et al., 2008). A training device that develops effective integrated trunk and lower extremity neuromuscular control during simulated sport movements might be a useful supplement to existing lower extremity injury prevention programs (Imwalle et al., 2009; Myer et al., 2008; Pollard et al., 2010). The Ground Force 360 device (Center of Rotational Exercise Inc., Clearwater, FL, USA) was designed to develop integrated trunk-lower extremity neuromuscular control using progressive resistance, whole body, long-axis rotational training during simulated sports movements (Fig. 1). During upright, weightbearing function, trunk and lower extremity movements, load transfer, and muscle power are synchronously coupled (Gracovetsky and Iacono, 1987; van Wingerden et al., 1993; Vleeming et al., 1995). Gluteus maximus and hamstring neuromuscular activation in particular is highly integrated with axial trunk rotation (van Wingerden et al., 1993; Vleeming et al., 1995). Injury prevention studies have identified direct relationships between increased knee injury risk and trunk neuromuscular control deficits (Zazulak et al., 2007a,b). As carefully performed movement patterns become more automatic through repetitious practice they also become more neuromuscularly and biomechanically efficient (Wu et al., 2008). The primary reason for this improved efficiency is enhanced neuromuscular connectivity (Green and Wilson, 2000; Wu et al., 2008). The close integration between trunk and lower extremity movements that occurs with Ground Force 360 device training may provide a useful, non-impact method for improving the lower extremity neuromuscular control needed to enhance dynamic knee stability during movements such as a SLDLS.

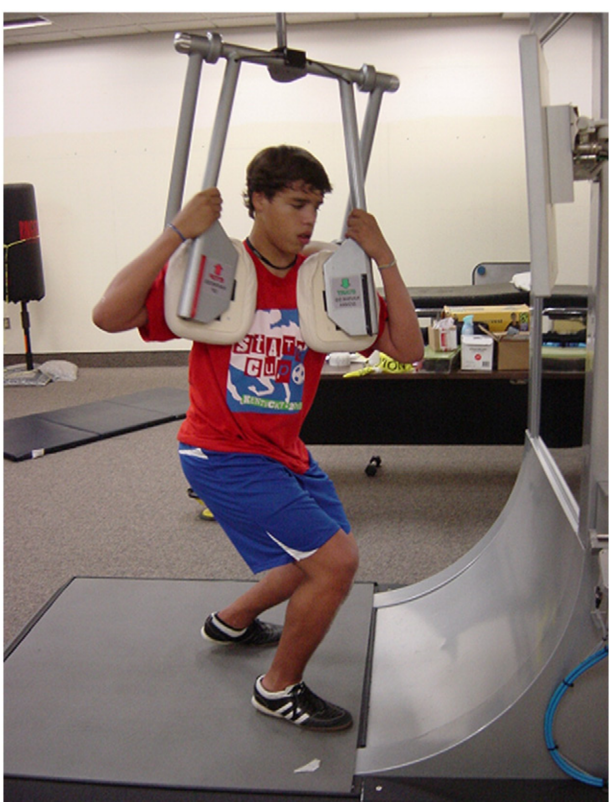

Fig. 1. Integrated trunk and lower extremity training in the Ground Force 360 device (Center of Rotational Exercise, Inc., Clearwater, FL, USA).

Table 1
Subject demographics: mean (standard deviation) [minimum, and maximum].

	Training group ($n=18$)	Control group ($n=18$)
Age (yrs)	22.3 (2.3)	25.4 (6.9)
Height (cm)	173.6 (10.5)	177.7 (8.5)
Pre-test subject weight (kg)	70.0 (9.4)	75.7 (12.1)
Post-test subject weight (kg)	70.8 (10)	74.2 (10)
IKDC physical activity scale level (median)	3 [2–4]	3 [2–4]
Exercise program or sports activity participation	9 of 18 (50%) subjects regularly participated in recreational running or weight training, 9 of 18 (50%) regularly participated in soccer, basketball, volleyball, tennis, flag football, or swimming.	16 of 18 (88.9%) subjects regularly participated in recreational running, 10 of 18 (55.6%) regularly participated in weight training, 6 of 18 (33.3%) regularly participated in basketball, soccer, flag football or tennis, and 5 of 18 (27.8%) regularly participated in recreational cycling.

The purpose of this study, which represents part of a larger project (Nyland et al., 2010), was to evaluate the efficacy of progressive resistance, whole body, long-axis rotational training for improving the lower extremity neuromuscular control of healthy subjects during SLDLS performance. The SLDLS movement was selected because it requires both sagittal and frontal plane lower extremity motion control, and primarily eccentric neuromuscular activation. The study hypothesis was that the Training group would display greater mean change differences compared to the Control group, suggesting improved lower extremity neuromuscular control. A secondary hypothesis was that the Training group would display a stronger relationship between peak hip flexion velocity reductions and the time period between the initial two peak frontal plane knee angular displacements following landing, also suggesting improved lower extremity neuromuscular control.

2. Methods

2.1. Experimental design

This was a prospective, randomized controlled study using a pre-test, post-test design with statistical comparison of mean change differences between data collection sessions. The time period between pre-test and post-test data collection was 4.0 ± 0.5 weeks (range = 3.5 to 5 weeks) for both groups.

2.2. Subject recruitment and group assignment

The Institutional Review Boards of the University of Louisville and Norton Healthcare, Louisville, KY approved this study. To be considered for study inclusion subjects had to be between 18 and 50 years of age, be participating in an exercise program or sports activity at least twice weekly, be without low back injury history or current low back pain, be without current lower extremity injury, and have no history of lower extremity surgery other than partial menisectomy (*and be at least 2 years post-surgery*).

Written informed consent was obtained from each subject. Forty-six potential subjects responded to campus flyer advertisements. Ten potential subjects were rejected from study participation because of previous knee ligament reconstruction, low back injury history, the desire to increase existing exercise program or sports activity volume during the study period, or because of an inability to comply with the study time commitment. Using a random numbers table with block randomization for gender, subjects were assigned to a Training or Control group. The International Knee Documentation Committee (IKDC) Physical Activity Scale (1 = highly competitive sports person, 2 = well-trained and frequently sporting, 3 = sporting sometimes, 4 = non-sporting) was used to determine subject perceived activity level (Table 1). Subjects continued regular exercise program or sport activities during the study period without increasing intensity, frequency, or volume. Female subjects were required to provide a negative pregnancy test at study initiation. Based on allocated time requirements, Training group subjects were reimbursed $120 for study participation, and Control group subjects were reimbursed $20.

2.3. Data collection

To provide a comprehensive profile of lower extremity neuromuscular control during SLDLS performance, EMG, lower extremity kinematic, and ground reaction force data were synchronously collected. Prior to each data collection and training session subjects performed a 10 min stationary cycling warm-up at a subjectively comfortable intensity. This was followed by 5 min of static stretching with subjects selecting stretches that they regularly performed prior to exercise program or sports activities. Subjects were then instructed in SLDLS performance from a 15.2 cm tall step using their preferred stance lower extremity. The preferred stance lower extremity was operationally defined as the lower extremity that subjects preferred to use for stance when kicking a ball. Subjects were instructed to flex their contralateral lower extremity knee approximately 45º to raise the foot off the step, assuming a single leg stance position on the step with the preferred stance lower extremity. Following this, they were instructed to jump laterally from the step down to the force plate using only the preferred stance lower extremity, performing a soft, controlled single leg landing with a flexed knee, and attempting to achieve and maintain stability as quickly as possible. Subjects performed 3–4 practice trials prior to data collection. After SLDLS practice subjects stood motionless on the force plate on their preferred stance lower extremity and bodyweight was determined. Following this, subjects performed 3 SLDLS trials.

Surface electrode sites at the preferred stance lower extremity were cleansed with isopropyl alcohol and shaved. Figure eight shaped Ag/AgCl bipolar adhesive electrodes (4 cm × 2.2 cm) with two circular conductive areas (each 1 cm diameter) and a 2 cm inter-electrode distance (Dual electrode #272, Noraxon, Scottsdale, AZ) were applied to the skin in parallel to the mid-muscle belly of gluteus maximus, gluteus medius, vastus medialis, rectus femoris, vastus lateralis, medial hamstrings, biceps femoris, and the medial head of gastrocnemius (SENIAM, 2010). A reference electrode was applied over the anterior superior iliac spine of the preferred stance lower extremity. Electrode sites were demarcated with an oil-based skin marker to enable consistent pre-test, post-test placement. Electromyographic (EMG) data were collected using an eight channel cable system (MyoSystem 1200, Noraxon, Scottsdale, AZ) with a 10–500 Hz bandwidth, >0 MΩ differential input impedance, a common mode rejection ratio of 100 db @ 50/60 Hz, and a 1000 Hz data sampling rate. Following instruction in appropriate muscle activation during manual muscle testing, mean maximal volitional isometric contraction (MVIC) EMG amplitudes (μV) were determined for each muscle or muscle group with an approximately 2 s time to peak activation, 6 s peak activation hold time, and 2 s gradual relaxation time using standard manual muscle test techniques (Kendall et al., 2005).

Two cm diameter retro-reflective markers were applied via adhesive discs to the skin approximately overlying the third lumbar spinous process, the greater trochanter (over cycling type shorts), the lateral femoral epicondyle, over the shoe approximately 2 cm distal to the lateral malleolus protuberance, and at the fifth metatarsal head of the preferred stance lower extremity. Markers enabled two-dimensional sagittal and frontal plane kinematic data collection with a 60 Hz sampling rate, using two video cameras (Sony DCR-HC30, Tokyo, Japan). One video camera was positioned perpendicular to the sagittal plane calibration space (0.9 m wide by 1.4 m tall) and one was positioned perpendicular to the frontal plane calibration space (0.9 m wide by 1.4 m tall). The cameras enabled two-dimensional frontal and sagittal plane kinematic data collection (Simi Motion 2D, Unterschleissheim, Germany). Hip angle was defined as the angle formed by the markers positioned over the third lumbar spinous process (low back), greater trochanter (hip), and lateral femoral epicondyle (knee). This angle has also been referred to as the trunk flexion angle representing composite movement between the hip and trunk (Blackburn and Padua, 2009). Sagittal and frontal plane knee angles were defined as the angle formed by markers positioned over the greater trochanter (hip), lateral femoral epicondyle (knee) and immediately distal to the lateral malleolus (ankle). Ankle angle was defined as the angle formed by markers positioned over the lateral femoral epicondyle (knee), immediately distal to the lateral malleolus (ankle) and over the head of the fifth metatarsal (foot). Good consistency has been reported between two- and three-dimensional kinematic analyses of dynamic frontal plane knee angles during side jump movements (McLean et al., 2005). Ground reaction force data were collected using a force plate (Model 9286AA, Kistler, Winterthur, Switzerland) and a 1000 Hz sampling rate.

2.4. Training program

Training group subjects participated in nine, approximately 20 min exercise sessions using the Ground Force 360 device (approximately 2 sessions/week). The computerized training device used compressed air to provide concentric or concentric-to-eccentric progressive resistance. The device harness provided up to 15.2 cm of side-to-side excursion, and 280º one-way, long-axis rotation. The open device frame provided an unobstructed view for monitoring user performance and a mirror in front of the user provided visual performance feedback. Training group subjects performed 7 exercise sets/session (Table 2). During exercise performance subjects were instructed to assume an athletic ready position of slight trunk, hip, and knee flexion, and ankle dorsiflexion. From this position they were instructed to maintain tightened abdominal muscles as they rotated their hips and trunk in unison as quickly as possible, back and forth through the available device range of motion during whole body long-axis rotation concentric activation, and with controlled deceleration during eccentric activation. Long-axis device rotation was set at 60 ± 10º one-way rotation (120 ± 20º total rotation) based on subject comfort. Foot position was adjusted between exercise sets from standard athletic ready position placement (at or slightly greater than shoulder-width apart) to diagonal placement (stride position with the left foot forward for concentric left rotation and with the right foot forward for concentric right rotation) to modify frontal and transverse plane lower extremity alignment and better facilitate hip abductor–adductor and internal–external rotator neuromuscular contributions (Neumann, 2010). The Borg Rating of Perceived Exertion Scale was used to monitor and control for subject perceived exercise intensity (Borg et al., 1987). The duration of each exercise set was timed and subjects received between set rest periods based on a 3:1 rest-to-work ratio.

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Table 2
Training group Ground Force 360 device regimen: mean (standard deviation). Rating of perceived exertion scale range [6 = no exertion, 20 = maximal exertion] (Borg et al., 1987). 13 somewhat hard, 15 hard or heavy.

Session #	Set #	Mode	Subjective intensity	Rating of perceived exertion	Resistance (kg/cm^2)	Repetitions	Foot placement
1–5	1	Two-way concentric rotation	Low	13.1 (1.8)	2.64 (0.84)	20	Standard
	2	Two-way concentric rotation	Moderate	13.9 (2)	3.24 (0.84)	10	Standard
	3	Concentric left rotation–Eccentric right rotation	Moderate-to-high	14.2 (1.7)	4.27 (1.27)	10	Standard
	4	Concentric right rotation–Eccentric left rotation	Moderate-to-high	14.2 (1.8)	4.27 (1.27)	10	Standard
	5	Concentric left rotation–Eccentric right rotation	Moderate	13.6 (1.7)	3.61 (1.05)	10	Diagonal
	6	Concentric right rotation–Eccentric left rotation	Moderate	13.8 (2)	3.60 (1.05)	10	Diagonal
	7	Two-way concentric rotation	Moderate-to-low	13.4 (2)	2.38 (0.63)	20	Standard
6-9	The same subjective intensity, resistance progressions, and foot placements were used. The exercise mode for the fifth and sixth exercise sets changed to one-way concentric left and right rotation, respectively. The repetition goal changed to Set 1. = 15 repetitions, Sets 2–6 = 8 repetitions, and Set 7. = 15 repetitions. This was a planned study modification to maintain subject cognitive focus.						

2.5. Data analysis

Full wave rectification, 60 Hz notch filtering, and 50 ms root mean square smoothing was applied to the EMG signals determined during the 6 s peak manual muscle testing activation period and for SLDLS trials between initial landing vertical ground reaction force production and return to single leg stance bodyweight. All EMG signal smoothing and analysis was performed using MyoResearch software version 2.10 (Noraxon, Scottsdale, AZ).

Mean SLDLS trial EMG signal amplitudes were then standardized to % MVIC (determined during the 6 s peak activation period while manual muscle testing). Standardized mean SLDLS trial EMG signal amplitudes were then divided by peak vertical ground reaction force observed during landing (N) (*standardized to subject bodyweight* (*N*)). This provided a valid and reliable measurement of lower extremity neuromuscular efficiency (Bosco et al., 1982; Bosco and Viitasalo, 1982; Cannon et al., 2001). The mean of these "unit-less" trial values was then determined for the pre-test and post-test data collections and were expressed as mean percent change.

Frontal plane knee alignment during SLDLS was standardized to alignment during relaxed single leg stance (Training group = 4.7 ± 3° knee valgus, Control group = 5.3 ± 3° knee valgus) such that increased relative knee valgus during landing is indicated by an increasing positive value, and increased relative knee varus is indicated by an increasing negative value. To further delineate lower extremity neuromuscular control, the initial two peak frontal plane knee angular displacements that occurred after landing and the time period between those peaks were analyzed. Smaller frontal plane knee angular displacement peak magnitudes and a longer time period between peaks suggests better frontal plane lower extremity neuromuscular control (Hewett et al., 2006a; Kernozek et al., 2005). Composite vertical–anteroposterior–mediolateral ground reaction force stabilization timing represented the sum of the time between landing onset and when single leg stance bodyweight values were consistently re-established for vertical (±20 N), anteroposterior (±5 N) and mediolateral (±5 N) ground reaction forces divided by three.

2.6. Statistical analysis

Independent sample t-tests were used to determine pre-test, post-test mean change differences (Dimitrov and Rumrill Jr., 2003). Dependent variables considered indicative of improved lower extremity neuromuscular control during SLDLS were: increased lower extremity neuromuscular efficiency as defined by a reduced standardized mean SLDLS trial EMG amplitude/peak vertical ground reaction force ratio (Bosco et al., 1982; Bosco and Viitasalo, 1982; Cannon et al., 2001), earlier single leg postural stabilization timing (Colby et al., 1999), decreased peak hip and knee flexion velocities during landing (Hewett et al., 2006a), increased hip and knee flexion at landing (Gauffin and Tropp, 1992; Olsen et al., 2004; Hewett et al., 2006a; Pollard et al., 2010), and an increased time period between the initial two peak frontal plane knee displacements, suggesting improved frontal plane knee control (Hewett et al., 2006a; Kernozek et al., 2005).

A pilot study of four subjects (2 men, 2 women) that met study inclusion criteria was performed to determine preliminary measurement reliability. Intraclass correlation coefficients (ICC) and 95% confidence intervals (95% CI) were calculated to describe the mean pre-test, post-test measurement reliability obtained without intervention and with 4 weeks between sessions. The $ICC_{3,1}$ formula was selected, since only one tester evaluated the subject population and compared mean measurements (Shrout and Fleiss, 1979). Moderate to high reliability was observed for gluteus maximus (0.97, 95% CI = 0.68–0.99), gluteus medius (0.94, 95% CI = 0.71–0.97), vastus medialis (0.97, 95% CI = 0.60–0.99), rectus femoris (0.96, 95% CI = 0.60–0.99), vastus lateralis (0.97, 95% CI = 0.81–0.99), medial hamstrings (0.95, 95% CI = 0.60–0.99), biceps femoris (0.89, 95% CI = 0.72–0.97), and medial gastrocnemius (0.97, 95% CI = 0.82–0.99) standardized EMG measurements during SLDLS performance. Moderate to high reliability was observed for hip (0.94, 95% CI = 0.60–0.99; 0.98, 95% CI = 0.70–0.99), knee (0.98, 95% CI = 0.88–0.99; 0.95, 95% CI = 0.84–0.99), and ankle (0.93, 95% CI = 0.63–0.99; 0.93, 95% CI = 0.70–0.99) initial and peak sagittal plane angular displacement magnitudes, respectively, and for peak hip (0.96, 95% CI = 0.78–0.99), knee (0.98, 95% CI = 0.88–0.99), and ankle (0.88, 95% CI = 0.71–0.99) angular velocities. Moderate to high reliability was observed for the first (0.94, 95% CI = 0.61–0.99) and second (0.92, 95% CI = 0.70–0.99) peak frontal plane knee displacement following landing and for the time period between them (0.97, 95% CI = 0.69–0.99). Moderate to high reliability was also observed for peak vertical (0.98, 95% CI = 0.80–0.99; 0.94, 95% CI = 0.61–0.99), anteroposterior (0.98, 95% CI = 0.85–0.99; 0.99, 95% CI = 0.88–0.99) and mediolateral (0.97, 95% CI = 0.82–0.99; 0.96, 95% CI = 0.87–0.98) ground reaction force magnitude and timing, respectively, and for composite vertical, mediolateral, and anteroposterior ground reaction force stabilization timing (0.97, 95% CI = 0.89–0.99).

From these reliability data minimal detectable change (MDC) values were calculated for all dependent variables using the following formula: MDC = $\text{t-score}_{\text{level of confidence}}$ x mean pre-test measurement standard deviation x $\sqrt{(2[1\text{-}ICC_{3,1}])}$(Haley and Fragala-Pinkham, 2006). Values for standardized EMG measurements were 0.11, 0.12, 0.14, 0.13, 0.06, 0.10, 0.11, and 0.13 for gluteus maximus, gluteus medius, vastus lateralis, rectus femoris, vastus medialis, medial hamstrings, biceps femoris and medial gastrocnemius, respectively. Values for sagittal plane lower extremity kinematic displacement measurements of hip flexion (4.4°, 2.6°), knee flexion (3.1°, 3.2°) and ankle dorsiflexion (4.4°, 4.2°) at initial landing and at peak displacement, respectively were also determined. Values for peak hip, knee, and ankle velocity were 16.2°/s, 18°/s, and 38°/s, respectively. Values for peak frontal plane lower extremity kinematic displacement measurements were 2.7°, 3.4°, and 0.15 s for initial peak

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frontal plane knee displacement, second peak frontal plane knee displacement, and for the time period between the initial two peak frontal plane knee displacements, respectively. Values for peak vertical, anteroposterior, and mediolateral ground reaction force magnitude measurements were 41.2 N, 85.2 N, and 38.9 N, respectively. Values for peak vertical, anteroposterior, and mediolateral ground reaction force timing measurements were 0.014 s, 0.06 s, and 0.019 s, respectively. The MDC value for composite vertical–anteroposterior–mediolateral stabilization timing was 0.12 s. An alpha level of $P \leq 0.05$ with Bonferroni corrections for multiple comparisons ($0.05/21 < 0.0023$) was selected to indicate statistical significance. Coefficient of determination (r^2) analysis was used to better delineate the relationship between the mean change differences for peak hip flexion velocity and the time period between the initial two peak frontal plane knee displacements following landing. All statistical analysis was performed using SPSS version 11.0 software (SPSS, Chicago, IL).

3. Results

3.1. Surface electromyography

Training group gluteus maximus and gluteus medius neuromuscular efficiency mean change differences improved 35.7% and 31.7% *(both exceeding MDC values)*, respectively while the Control group decreased 14.6% and 16.0%, respectively. No other muscle or muscle group displayed statistically significant group mean change differences (Table 3).

3.2. Composite ground reaction force timing

Training group composite vertical–anteroposterior–mediolateral ground reaction force stabilization timing mean change difference occurred 1.35 s earlier (pre-test = 4.32 ± 1.4 s, post-test = 2.97 ± 1.1 s), *exceeding the MDC value*, while the Control group mean change difference displayed a slight delay 0.24 s (pre-test = 3.76 ± 1.1 s, post-test = 4.00 ± 1.6 s) ($t = 4.08$, $P < 0.0001$).

3.3. Kinematics

Training group knee flexion angle at landing mean change difference revealed 3.5° greater flexion *(exceeding the MDC value)*, while the Control group knee flexion angle at landing mean change difference decreased 0.5°. The time period between the initial two peak frontal plane knee displacements during landing mean change difference was increased 0.17 s in the Training group *(exceeding the MDC value)*, but occurred 0.20 s earlier in the Control group. Training group peak hip flexion velocity mean change difference was 21.2°/s slower (*exceeding the MDC value*), while the Control group mean change difference was 12.5°/s faster. Training group peak knee flexion velocity mean change difference was 20.1°/s slower (*exceeding the MDC value*), while the Control group was 4.6°/s faster (Table 4).

4. Discussion

Although injury prevention training programs have helped reduce the incidence of non-contact lower extremity injuries that occur during sports, they still occur far too often (Hootman et al., 2007; McLean, 2008). A deficiency in some training programs is failure to adequately consider the influence of integrated trunk-lower extremity function on composite lower extremity neuromuscular control during maneuvers such as jump landings (Imwalle et al., 2009; McLean, 2008; Zazulak et al., 2007a,b). Therefore injury prevention programs that combine weightbearing, integrated trunk and lower extremity neuromuscular control training through functionally relevant sports movement simulations, with a bias toward eccentric activation may be more effective (Coventry et al., 2006; LaStayo et al., 2003).

Earlier single leg postural stabilization timing, decreased postural sway displacement, and reduced medial gastrocnemius, peroneus longus, and tibialis anterior EMG amplitudes have been reported following 4 weeks of bi-weekly maximum effort leg press sessions suggesting improved lower extremity neuromuscular control (Bruhn et al., 2004). Following an 8 week eccentric cycling ergometry program LaStayo et al. (2008) reported that subjects could decelerate the same pedal resistance at a decreased vastus lateralis EMG amplitude suggesting a reduced neural drive requirement to withstand high knee loads. Increased lower extremity neuromuscular efficiency during vertical jumping has also been characterized by decreased leg extensor standardized EMG amplitude/vertical ground reaction force magnitude ratios (Bosco et al., 1982; Bosco and Viitasalo, 1982). Each of these studies used healthy subjects of similar age to the subjects that participated in our study.

Excessive or poorly controlled knee valgus during single leg landings, in conjunction with a shallow knee flexion angle, and excessive quadriceps femoris activation, increases ACL injury risk (Ford et al., 2003; Hewett et al., 2005; Kernozek et al., 2005; Shimokochi and Shultz, 2008). Large magnitude or poorly controlled frontal plane knee movements contribute more to ACL injury risk than isolated transverse or sagittal plane events alone (Pollard et al., 2010). Developing effective composite trunk and lower extremity neuromuscular control can help counter-balance excessive and/or poorly controlled lower extremity alignment, movement, and knee joint loads during single leg jump landings (Blackburn and Padua, 2009; Myer et al., 2008).

The greater mean change differences observed in the Training group for increased knee flexion angle at landing, decreased peak hip flexion velocity, decreased peak knee flexion velocity, and a longer

Table 3
Standardized mean EMG amplitude/peak vertical ground reaction force results: mean (standard deviation).

	Training group			Control group			
	Pre-test	Post-test	Mean % change	Pre-test	Post-test	Mean % change	
Gluteus maximus	0.56 (0.2)	0.36 (0.1)	-35.7	0.48 (0.2)	0.55 (0.2)	+14.6	$t = -4.0$, $P < 0.0001$[a]
Gluteus medius	0.63 (0.2)	0.43 (0.1)	-31.7	0.50 (0.2)	0.58 (0.2)	+16.0	$t = -4.0$, $P < 0.0001$[a]
Vastus lateralis	0.52 (0.3)	0.49 (0.2)	-5.8	0.60 (0.3)	0.58 (0.2)	-3.3	NS
Rectus femoris	0.65 (0.3)	0.66 (0.3)	+1.5	0.61 (0.3)	0.57 (0.3)	-6.6	NS
Vastus medialis	0.71 (0.2)	0.74 (0.3)	+4.2	0.85 (0.3)	0.79 (0.3)	-7.1	NS
Medial hamstrings	0.34 (0.2)	0.38 (0.3)	+11.8	0.43 (0.2)	0.36 (0.4)	-16.3	NS
Biceps femoris	0.33 (0.1)	0.30 (0.1)	-9.1	0.35 (0.3)	0.41 (0.2)	+17.1	NS
Gastrocnemius	0.48 (0.2)	0.55 (0.3)	+14.6	0.52 (0.2)	0.45 (0.1)	-13.5	NS

[a] $P \leq 0.0023$.

Table 4
Kinematic variable results: mean (standard deviation).

	Training group			Control group			
	Pre-test	Post-test	Mean change	Pre-test	Post-test	Mean change	
Hip flexion at landing (º)	25.6 (12)	26.6 (8)	+1	26.1 (8)	27.1 (9)	+1	NS
Peak hip flexion (º)	42.1 (16)	44.6 (14)	+2.5	41.6 (12)	43.4 (14)	+1.8	NS
Peak hip velocity (º/s)	−126.1 (25)	−104.9 (31)	−21.2	−130.5 (59)	−143.0 (57)	+12.5	$t = 3.4$, $P = 0.002$[a]
Knee flexion at landing (º)	27.8 (4)	31.3 (4)	+3.5	29.1 (5)	28.6 (9.1)	−0.5	$t = 3.3$, $P = 0.002$[a]
Peak knee flexion (º)	60.6 (10)	60.5 (8)	−0.1	54.9 (10)	55.7 (10)	+0.8	NS
First peak frontal plane knee displacement (º)	−0.5 (6)	−2.3 (5)	−1.8	−0.5 (5)	−3.3 (6)	−2.8	NS
Second peak frontal plane knee displacement (º)	−1.3 (9)	3.3 (9)	4.6	2.0 (7)	3.4 (9)	1.4	NS
Time between 1st–2nd peak frontal plane knee displacements (s)	0.60 (0.24)	0.77 (0.29)	+0.17	0.74 (0.33)	0.54 (0.33)	−0.20	$t = 3.5$, $P = 0.001$[a]
Peak knee velocity (º/s)	209.8 (73)	189.7 (57)	−20.1	221.1 (39)	225.7 (36)	+4.6	$t = -3.3$, $P = 0.002$[a]
Ankle dorsiflexion at landing (º)	10.9 (7)	7.3 (5)	−3.6	10.8 (7)	7.1 (7)	−3.7	NS
Peak ankle dorsiflexion (º)	25.6 (7)	26.1 (5)	+0.5	22.1 (4)	25.1 (3)	+3.0	NS
Peak ankle dorsiflexion velocity (º/s)	351.5 (73)	319.2 (51)	−32.3	337.1 (77)	315.7 (69)	−21.4	NS

[a] $P < 0.0023$.

time period between the initial two peak frontal plane knee displacements after landing suggest improved lower extremity neuromuscular control. Performing a jump landing in greater knee flexion is known to improve lower extremity shock absorption and decrease ACL strain (Gauffin and Tropp, 1992; Olsen et al., 2004; Pollard et al., 2010). Although peak frontal plane knee angular displacement values did not display group differences, coefficient of determination analysis revealed a strong direct relationship between mean change differences for the time period between the initial two peak frontal plane knee displacements and peak hip flexion velocity reductions in the Training group ($r^2 = 0.77$, $P = 0.0001$)(Fig. 2). This suggests that 77% of the variability in the time period between the initial two peak frontal plane knee displacements after landing for the group that trained on the Ground Force 360 device could be explained by peak hip flexion velocity reductions. Only the Training group displayed significant mean change differences for these variables, suggesting improved lower extremity neuromuscular control following device training. This supports the relationship between hip neuromuscular control and dynamic frontal plane knee stability.

Training group mean change differences also indicated earlier composite vertical–anteroposterior–mediolateral ground reaction

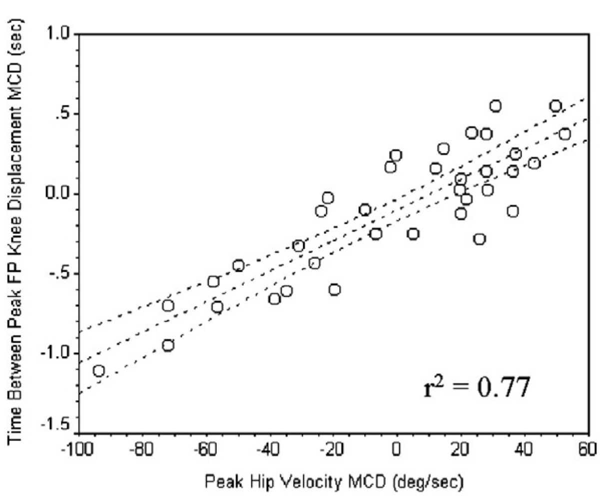

Fig. 2. Coefficient of determination fit line with 95% confidence interval revealed a strong, direct relationship ($r^2 = 0.77$, $P = 0.0001$) between the mean change differences (MCD) for peak hip flexion velocity and the time period between the initial two peak frontal plane (FP) knee displacements following landing. Only the Training group displayed significant mean change differences for these variables, suggesting improved lower extremity neuromuscular control.

force stabilization timing, also suggesting improved lower extremity neuromuscular control and dynamic stability. Delayed stabilization timing is more frequently observed among individuals with impaired dynamic knee stability such as following ACL injury (Colby et al., 1999). Comparable peak ground reaction force magnitudes and onset timing between groups suggests similar SLDLS technique and intensity (Table 5).

Training group mean change differences also indicated improved gluteus maximus and gluteus medius neuromuscular efficiency with reduced standardized EMG amplitude/vertical ground reaction force ratios. This finding in combination with comparable MVIC values observed during pre- and post-test data collection sessions suggests that Training group subjects were better able to effectively withstand sudden single lower extremity loads with a concomitant reduced neural drive requirement for these muscles. Similar findings have been reported by others for lower leg muscles following high intensity leg pressing (Bruhn et al., 2004), the vastus lateralis following eccentric cycling ergometry (LaStayo et al., 2008), and the leg extensors during vertical jumping (Bosco et al., 1982; Bosco and Viitasalo, 1982).

4.1. Study limitations

Study limitations include a lack of three-dimensional kinematic and inverse dynamic kinetic analyses. This addition would have better delineated specific hip, knee, and ankle segmental kinematic and kinetic contributions to SLDLS performance. Subtle differences in surface electrode placement, movement, and subcutaneous tissue thickness changes between the pre-test and post-test data collection sessions may have also influenced standardized EMG amplitude values. Specifically, slightly altered muscle length in association with hip and knee joint angle and velocity changes between data collection sessions may also have influenced standardized EMG amplitude values. However, given the magnitude of the mean percent change difference that was observed for gluteus maximus and gluteus medius neuromuscular efficiency we believe that training induced central drive changes was the primary stimulus.

Also, given the relatively short training period, study results only represent early training adaptations. The initial 3 weeks of resistance training programs primarily produce neural system effects of improved recruitment responsiveness and efficiency (Moritani and deVries, 1979). The findings of this mean 4 week duration study did not provide information regarding possible long-term training benefits. Lastly, the primary investigator oversaw all aspects of device adjustments and settings, subject technique, and rest period

Table 5
Ground reaction force (GRF) magnitude and onset timing results: mean (standard deviation).

	Training group			Control group		
	Pre-test	Post-test	Mean change	Pre-test	Post-test	Mean change
Peak vertical GRF (N/kg)	20.5 (2.9)	20.5 (2.6)	0.0	19.5 (2.7)	19.4 (2.6)	−0.1
Peak vertical GRF timing (ms)	82 (10)	81 (10)	−1.0	79 (10)	78 (20)	−1.0
Peak anteroposterior GRF (N/kg)	−0.32 (3.2)	0.28 (3.1)	+0.60	−0.68 (3.0)	0.18 (3.1)	+0.86
Peak anteroposterior GRF timing (ms)	47 (30)	40 (30)	−7.0	59 (40)	43 (30)	−16.0
Peak mediolateral GRF (N/kg)	−2.77 (0.6)	−2.78 (0.6)	+0.01	−2.76 (0.7)	−2.71 (0.7)	−0.05
Peak mediolateral GRF timing (ms)	84 (30)	83 (30)	−1.0	89 (30)	84 (40)	−5.0

monitoring. Differences may exist when subjects independently adjust settings and select different movements and training modes.

5. Conclusions

Ground Force 360 device training improved the lower extremity neuromuscular control of healthy subjects during SLDLS performance. These findings are particularly meaningful because no training session included any jump or jump landing tasks or their associated impact loads and increased lower extremity injury risks. For these reasons Ground Force 360 device training may also be a useful supplement to rehabilitation programs following hip, knee or ankle surgery when jumping cannot be safely performed because of the potentially adverse effects on tissue healing and remodeling. Further study is needed to evaluate device training among other subject groups that might benefit from having improved lower extremity neuromuscular control such as athletically active adolescents (Myer et al., 2008), and patients that desire to safely return to sports with jumping components following lower extremity surgery such as ACL reconstruction (Gerber et al., 2009).

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