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Original research

Effects of ankle destabilization devices and rehabilitation on gait biomechanics in chronic ankle instability patients: A randomized controlled trial

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ABSTRACT

Patients with chronic ankle instability (CAI) have altered gait patterns, which are characterized by increased inversion positioning during gait. Ankle destabilization devices increase peroneus longus muscle activation during gait, which may increase eversion.

Objective: To determine whether incorporating destabilization devices into a 4-week impairment-based rehabilitation program has beneficial effects on gait biomechanics and surface electromyography (sEMG) compared to impairment-based rehabilitation without destabilization devices in CAI patients.

Design: Randomized controlled trial.

Setting: Laboratory.

Participants: Twenty-six CAI patients.

Outcome measures: Patients completed baseline gait trials and were randomized into no device or device groups. Groups completed 4-weeks of rehabilitation with or without devices, and then completed post-intervention gait trials. Lower extremity sagittal and frontal plane kinematics and kinetics and sEMG activity were measured.

Results: The device group increased dorsiflexion during mid-late stance and had lower normalized sEMG amplitude for the peroneus longus during early stance and mid-swing after rehabilitation. The no device group had less peroneus brevis sEMG activity during early stance after rehabilitation.

Conclusion: Incorporating destabilization devices in a 4-week rehabilitation program was an effective method of improving dorsiflexion during the stance phase of gait. However, impairment-based rehabilitation, regardless of instability tool, was not effective at improving frontal plane motion.

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1. Introduction

Lateral ankle sprains are the most common musculoskeletal injury to occur in people who participate in athletics (Hootman, Dick, & Agel, 2007; Nelson, Collins, Yard, Fields, & Comstock, 2007) or recreational activity (Waterman, Owens, Davey, Zacchilli, & Belmont, 2010). Recurrence rates have been estimated to be as high as 70% after an isolated ankle sprain (McKay, Goldie, Payne, & Oakes, 2001), with approximately 30% percent of patients continuing to have symptoms lasting greater than 1 year after their initial sprain (van Rijn, van Os, Bernsen, Luijsterburg, Koes, & Bierma-Zeinstra, 2008). Symptoms of repetitive ankle sprains and episodes of instability that occurs greater than one year following initial sprain has been termed chronic ankle instability (CAI) (Delahunt, Coughlan, Caulfield, Nightingale, Lin, & Hiller, 2010). Many characteristics of CAI include impaired proprioception (Forkin, Koczur, Battle, & Newton, 1996; Garn & Newton, 1988; Konradsen, Ravn, & Sorensen, 1993; Lentell, Baas, Lopez, McGuire, Sarrels, & Snyder, 1995; Willems, Witvrouw, Verstuyft, Vaes, & De

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Clercq, 2002), decreased neuromuscular control (Konradsen, Olesen, & Hansen, 1998; McKeon & Hertel, 2008; McKeon, Ingersoll, Kerrigan, Saliba, Bennett, & Hertel, 2008; Olmsted, Carcia, Hertel, & Shultz, 2002; Perrin, Bene, Perrin, & Durupt, 1997), decreased range of motion (ROM) (Denegar, Hertel, & Fonseca, 2002; Drewes, McKeon, Kerrigan, & Hertel, 2009; Wilson, Gieck, Gansneder, Perrin, Saliba, & McCue, 1998), decreased strength (Konradsen et al., 1998; Munn, Beard, Refshauge, & Lee, 2003; Willems et al., 2002), and altered gait (Brown, 2011; Delahunt, Monaghan, & Caulfield, 2006a, b; Drewes, McKeon, Paolini, et al., 2009; Monaghan, Delahunt, & Caulfield, 2006).

With regards to gait, CAI patients show greater ankle inversion positioning during late swing through early stance phase and spend a longer time on the lateral aspect of the foot during the stance phase, which may predispose them to recurrent ankle sprains (Delahunt et al., 2006a; Drewes, Lee, McKeon, Paolini, Kerrigan, & Hertel, 2008; Drewes, McKeon, Paolini, et al., 2009). Furthermore, individuals demonstrate decreased foot clearance and dorsiflexion (Brown, 2011). It has been hypothesized that these alterations could be a result of mechanical changes of the lateral ankle after injury (Gupta, Paul, & Sandhu, 2008; P. O. McKeon, McKeon, Mattacola, & Lattermann, 2011), a change in pre-programmed motor response (Konradsen et al., 1998; Monaghan et al., 2006), a decrease in the ability to detect this pathological position (Konradsen et al., 1998), or a combination of both mechanical and sensoriomotor deficits. Ankle destabilization devices have been developed to improve the faulty motor response and detection of increased inversion associated with CAI by implementing a feed-forward mechanism in response to an anticipated bout of instability (Forestier & Toschi, 2005).

Ankle destabilization devices possess an articulator that forces the patient into plantar flexion, inversion, and internal rotation in a controlled manner during movement. These devices are unique, compared to other instability tools, because they can be worn during functional exercises. A study on two specific ankle destabilization devices, the Myolux Athletik and Myolux II (Cevres Santé, Le Bourget-du-Lac, France) (Fig. 1) found a pronounced increase in the peroneus longus surface electromyography (sEMG) amplitude prior to initial contact while the device was worn during treadmill walking in CAI patients (Donovan, Hart, & Hertel, 2015). The increased activation of the peroneus longus prior to initial contact may provide more stability to the ankle joint by pulling the ankle out of the inverted position in late swing. As the devices caused alterations prior to initial contact, it shows their ability to cause a feed-forward response mechanism that may be learned over time if implemented into a rehabilitation program for CAI. Therefore, the purpose of this study was to determine whether incorporating ankle destabilization devices in a 4-week impairment-based rehabilitation program improves ankle, knee, and hip kinematics, kinetics, vGRF and lower leg sEMG measures when compared to a no device group.

2. Methods

2.1. Study design

A single-blinded randomized controlled trial comparing 4 weeks of supervised impairment-based progressive rehabilitation with (device group) and without ankle destabilization (no device group) on frontal and sagittal plane ankle, knee, and hip kinematics and kinetics and sEMG activity (anterior tibialis, peroneus brevis, peroneus longus, and medial gastrocnemius) in young adults with CAI was completed. All procedures were approved by the Institution's investigational review board prior to enrolling patients. This study was part of a larger study; therefore, the same patients and impairment-based rehabilitation program were used as in a previously published paper (Donovan et al., 2016).



Fig. 1. Myolux Athletik (top) and Myolux II (bottom).

2.2. Patients

Twenty-six young adults (age = 21.34 ± 2.9, sex = (M = 7,F = 19), height = 168.96 ± 10.6 cm, weight = 70.73 ± 13.7 kg) with CAI were recruited over 1 year period from a University setting and surrounding community to participate in this study (Table 1). Patients were classified as having CAI if they had a history of more than one ankle sprain with the initial sprain occurring greater than one year ago, current self-reported functional deficits due to ankle symptoms that was qualified by a score of <85% on the Foot and Ankle Ability Measure (FAAM) Sport scale and a ≥ 11 on the Identification of Functional Instability scale (IdFAI) (Gribble et al., 2013; Simon, Donahue, & Docherty, 2012) (Carcia, Martin, & Drouin, 2008). Patients were excluded if they had an ankle fracture, ankle surgery, an ankle sprain within the past 6 weeks, or any other current lower extremity pathology (Gribble et al., 2013). We used a cut-off score of an 85% versus the commonly reported 80% for the FAAM-Sport because we believe self-reported deficits of 15% are clinically relevant and is a large enough deficit where treatment could cause large improvements.

Table 1
Patient baseline demographics and ankle sprain history.

	No device (n = 13)		Device (n = 13)	
	Mean	95% confidence interval	Mean	95% confidence interval
Age (years)	21.5	1.6	21.3	1.8
Height (cm)	169.1	5.8	168.8	3.7
Mass (kg)	75.3	7.4	66.1	7.0
Number of sprains	3.1	0.8	6.2	2.9
Last sprain (months)	24.5	12.2	10.3	5.3
First sprain (years)	5.6	1.9	7.9	2.8
Baseline foot and ankle ability measure-activities of daily living %	87.7	4.3	85.8	3.9
Baseline foot and ankle ability measure-sport %	65.9	9.9	67.1	7.3
Identification of functional ankle instability	22.9	0.9	23.2	2.8

*There were no differences between groups at baseline.

3. Instruments

3.1. Ankle destabilization devices

The Myolux Athletik and Myolux II (Cevres Santé, Le Bourget-du-Lac, France) were the destabilization devices used in the device group. The Myolux Athletik consists of a half boot with an articulator beneath the calcaneus that allows for approximately 45 ° of combined inversion, internal rotation and plantar flexion verses the Myolux II, which is a full length sandal that allows for about 30 ° of combined motion. Both devices were used in the device group of this study (Fig. 1). During the rehabilitation program, patients progressed from using the Myolux II to using the Myolux Athletik to make the exercise more challenging.

3.2. Motion capture system

Three-dimensional joint kinematics of the ankle, knee, and hip were measured using the *TrackSTAR* electromagnetic motion analysis system (Ascension Technologies, Inc., Burlington,

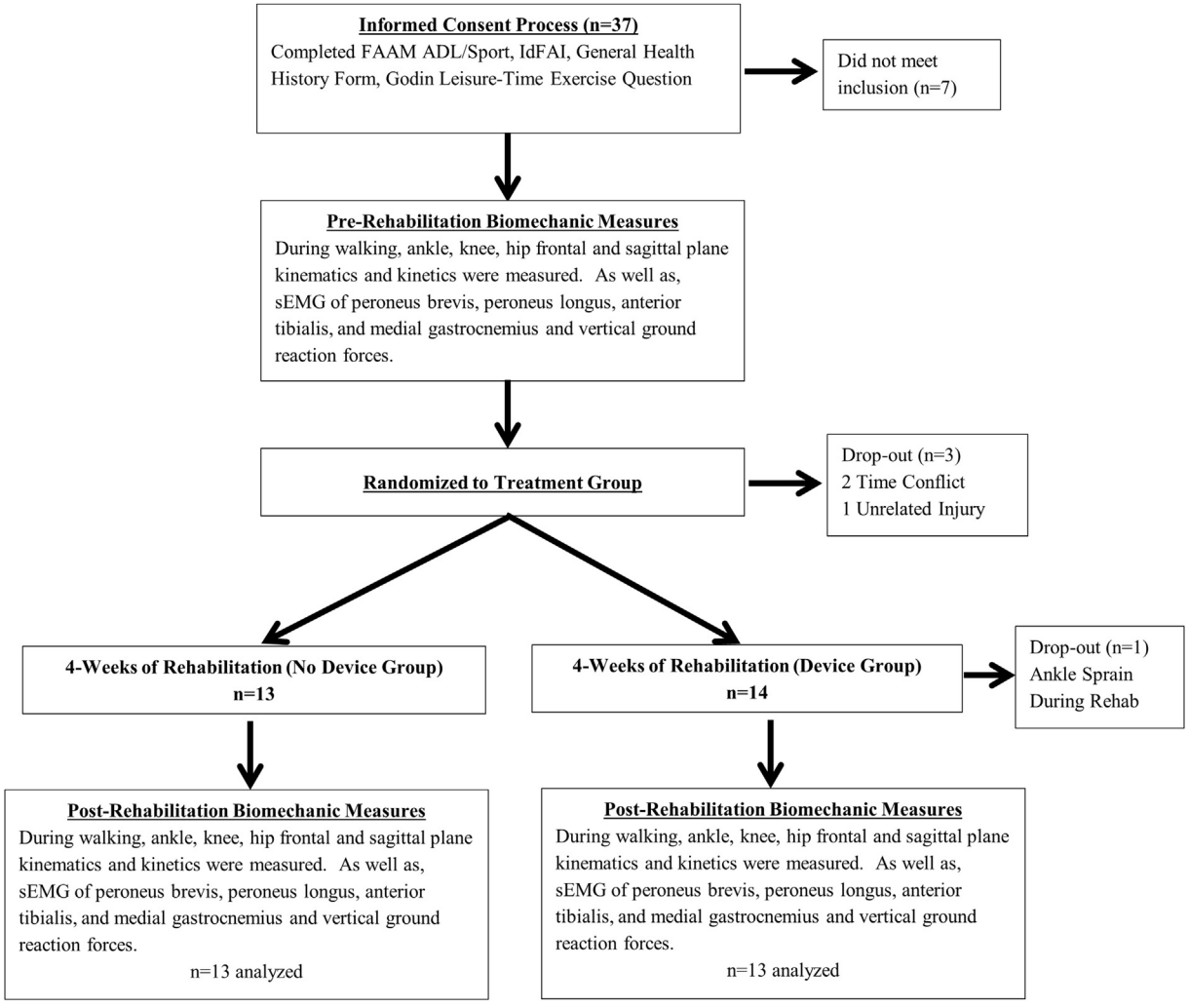


Fig. 2. A CONSORT diagram that outlines the flow of patients through the study.

Vermont) controlled by MotionMonitor software (Version 8, Innovative Sports Training, Inc., Chicago, Illinois) at a sampling rate of 144 Hz. A non-conductive forceplate (Bertec Corporation, Columbus, Ohio) with a sampling rate of 1440 Hz was embedded into a walk-way and synchronized with the electromagnetic tracking device and used to collect ground reaction forces and for determination of initial contact and terminal stance during walking trials.

3.3. Surface electromyography

Surface EMG was measured using rectangular DE 2.1 differential EMG sensors (Delsys, Boston, MA). The sensors consisted of two 1 mm parallel bars separated by about 1 cm and were placed over the middle portion of the muscle belly parallel to muscle fiber orientation. The signal was amplified with a gain of 1000 and digitized with a 4 channel acquisition system (Bagnoli EMG system, Delsys, Boston, MA) at 1000 Hz. Input impedance was $>10^{15}\Omega/0.2$ pF with a signal to noise ratio of 1.2uV. Data was collected and processed using MotionMonitor software (Innovative Sports Training, Inc., Chicago, Illinois). Data was filtered using a 10–500 band-pass filter and smoothed using a 50-sample moving window root mean square (RMS) algorithm as recommended by Konrad (2005). Data for each muscle was normalized to the corresponding muscle during quiet standing.

3.4. Participant set-up

Surface EMG electrodes were placed over the anterior tibialis, peroneus brevis, peroneus longus and medial gastrocnemius after the skin had been properly cleaned (Donovan et al., 2015). Electrode placement was visually analyzed for cross-talk by individually manual muscle testing each muscle. Next, 10 electromagnetic sensors (4 bilateral and 2 central) were placed on the participant's (posterior calcaneus, dorsal aspect of the first metatarsal, lateral mid-shank, lateral mid-thigh, the base of the sacrum, and the 12th vertebrae of the thoracic spine) using double-sided tape and secured with Leuokotape® and elastic wraps to minimize movement. With regards to the calcaneus sensors, holes were cut from the shoes to ensure these sensors could be placed directly on the skin (Chinn, Dicharry, & Hertel, 2013). An 11th moveable sensor was attached to a stylus and used for digitization of each joint. As previously reported by Brown (2011), digitization of the segments and joints were completed by pointing to the proximal and distal longitudinal and horizontal landmarks using the stylus. Specific landmarks were: right/left ASIS, cervical spine vertebrae 7, thoracic spine vertebrae 12, lumbar spine vertebrae 5, right/left medial knee joint line, right/left lateral knee joint line, right/left medial malleolus, right/left lateral malleolus, and right/left 2nd toe.

3.5. Procedures

Each patient was fitted in a neutral shoe (Brooks Defyance, Seattle, WA) and were set-up with the sEMG electrodes and the electromagnetic tracking sensors as previously described. Next, patients were instructed to walk across the 6 m walk-way at their normal pace. The investigator altered their starting place to ensure their involved limb landed on the forceplate for each trial. Once the participant felt that they were walking normally and were consistently stepping on the forceplate each trial, they completed 15 separate walking trials across the walk-way. After the 15 trials were recorded; patients were instructed to return to the lab 2 days later

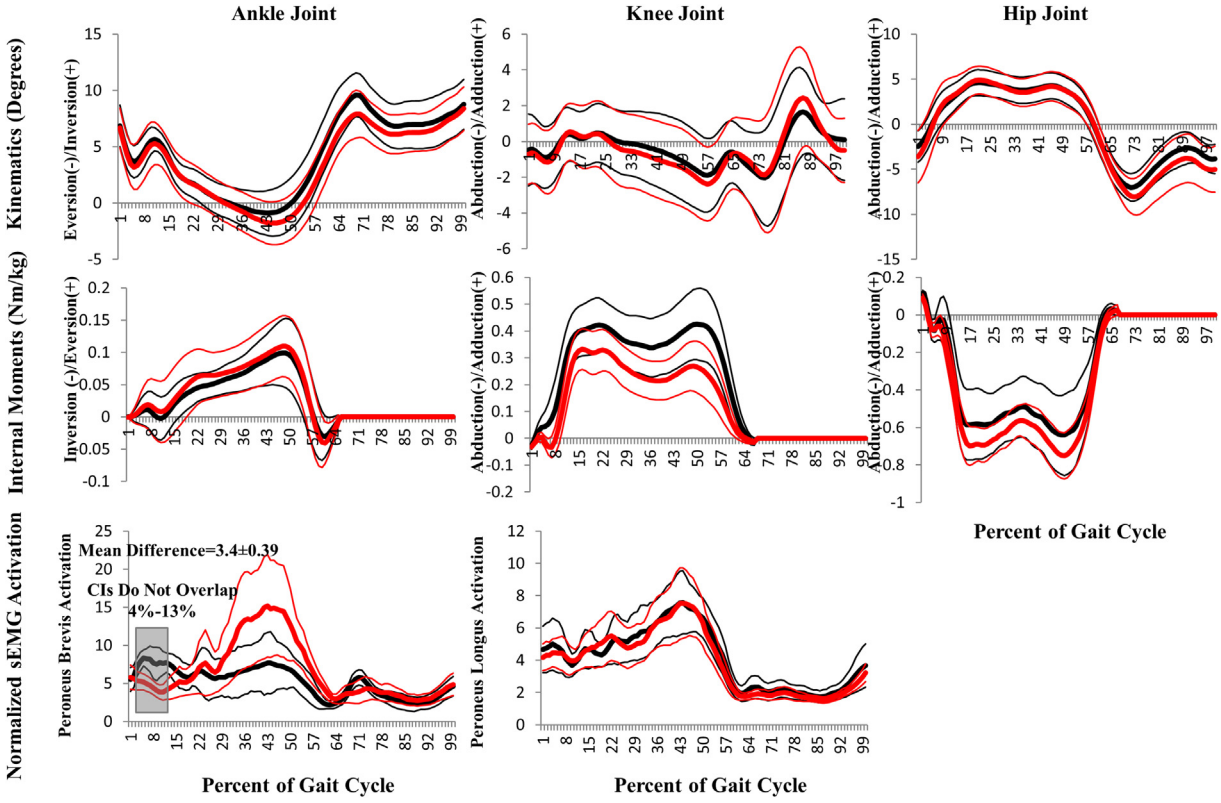


Fig. 3. No device group means with 90% confidence intervals (CIs) for frontal plane ankle, knee, and hip kinematics, internal moments and sEMG activation during the gait cycle (1%–100%) pre and post rehabilitation. Stance phase represents approximately 0–68% of the gait cycle. Black represents pre-rehabilitation and red represents post rehabilitation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

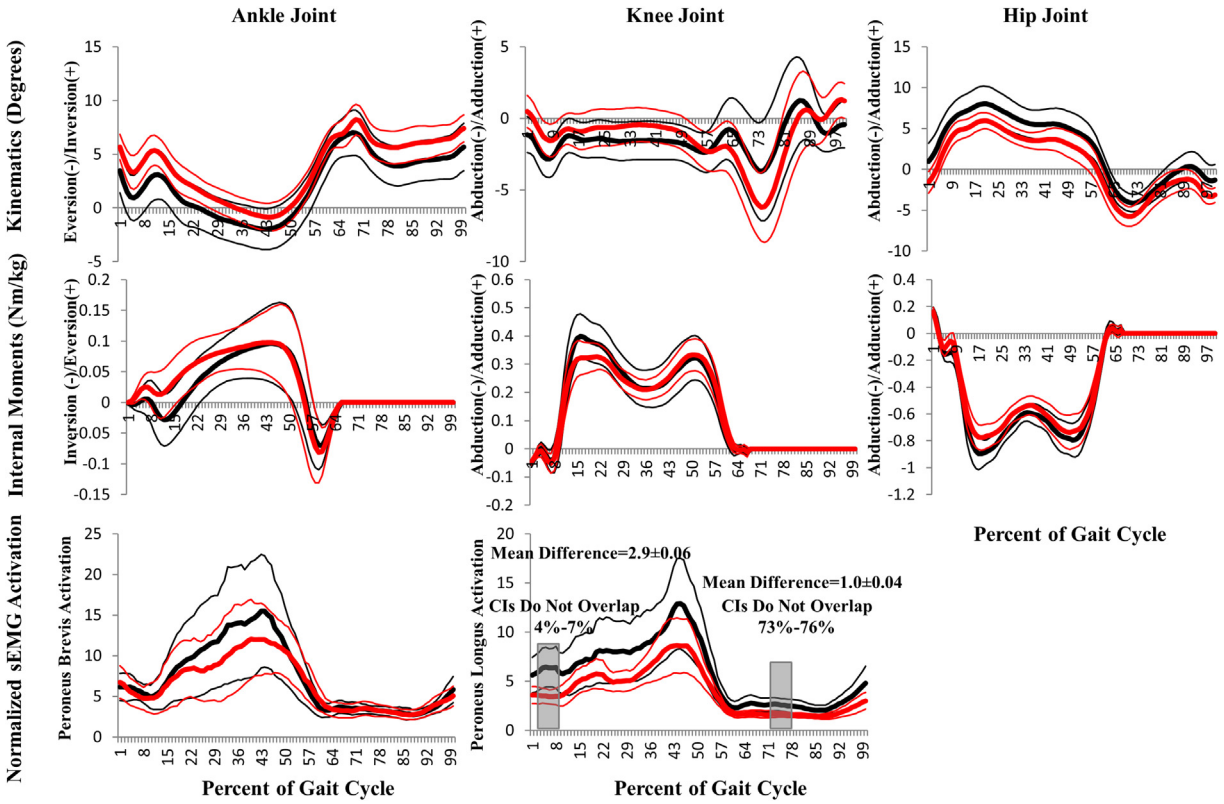


Fig. 4. Device group means with 90% confidence intervals (CIs) for frontal plane ankle, knee, and hip kinematics, internal moments and sEMG activation during the gait cycle (1%–100%) pre- and post-rehabilitation. Stance phase represents approximately 0–68% of the gait cycle. Black represents prerehabilitation and red represents post-rehabilitation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

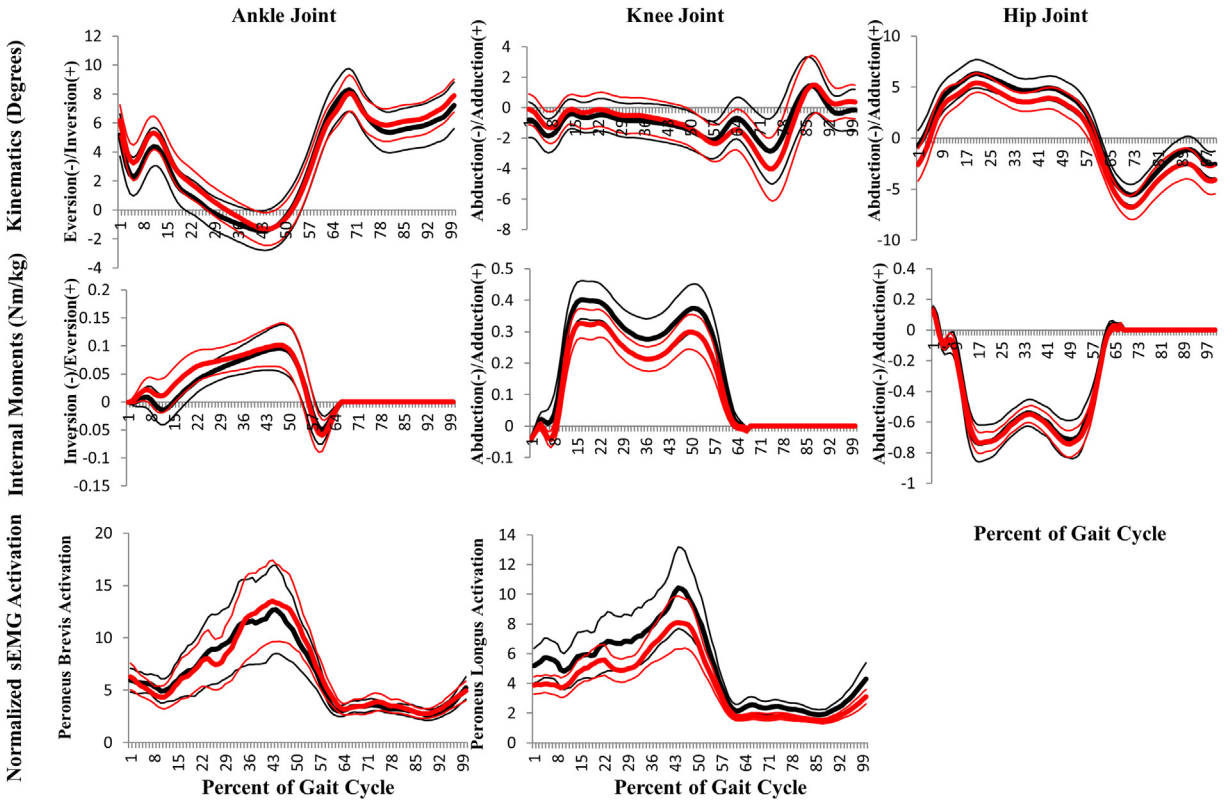


Fig. 5. Pooled group means with 90% confidence intervals (CIs) for frontal plane ankle, knee, and hip kinematics, internal moments and sEMG activation during the gait cycle (1%–100%) pre- and post-rehabilitation. Stance phase represents approximately 0–68% of the gait cycle. Black represents pre-rehabilitation and red represents post-rehabilitation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

for their first rehabilitation session. At this time, they were randomly assigned to either the no device or device group via concealed envelope allocation. The randomization schedule was constructed by one of the investigators utilizing a computerized random number generator prior to commencement of patient recruitment. This investigator was not involved in the screening or testing procedures of any participants. Patients completed 12 supervised rehabilitation sessions over 4-weeks. After they completed the program, they returned to the lab 2–7 days later and repeated the gait trials as previously described. The investigator who completed data collection was blinded to group assignment throughout data collection and data reduction. A CONSORT diagram outlines the flow of patients (Fig. 2).

3.6. Rehabilitation program

All rehabilitation was supervised and progressed by a Certified Athletic Trainer (ATC). On average, all sessions took the same amount of time to complete for both groups (No device: 65.2 min; Device: 66.3 min). The full rehabilitation program can be found as an appendix to a previous study (Donovan et al., 2016). The rehabilitation program was created based on recommendations of a novel rehabilitation paradigm for treatment of CAI patients (Donovan & Hertel, 2012). The rehabilitation program for both groups consisted of exercises that address deficits in range of motion (ROM), strength, balance, and functional activity. The ATC used clinical judgment for initial intensity of each exercise and progressed them to the next exercise after they could successfully meet the pre-established progression criteria (Donovan et al., 2016). The no device group incorporated traditional unstable surfaces such as foam pads and discs during the balance and functional exercises, while the device group incorporated both ankle destabilization devices during the balance and functional exercises. In addition, the device group used the ankle destabilization devices during treadmill walking, while the no device group completed shod treadmill walking. The treadmill walking portion of the rehabilitation program was matched for time between each group (Donovan et al., 2016).

3.7. Data reduction

The kinematic data were filtered with a low-pass 4th-order, Butterworth filter at a cut-off frequency of 14.5 Hz (Boling, Padua, Marshall, Guskiewicz, Pyne, & Beutler, 2009). Euler rotation method (Y, X, Z) or (flexion/extension, adduction/abduction, internal/external rotation) was used to calculate ankle, knee, and hip joint rotations. Vertical ground reaction force (vGRF) (N) was normalized to each participant's body mass (kg) and internal joint moments (N*m/kg) were normalized to the participants height and mass (Protopapadaki, Drechsler, Cramp, Coutts, & Scott, 2007).

Using the MotionMonitor software, the 15 stride cycles of the involved limbs were re-sampled to 100 frames so that each frame represents one percent of the stride cycle. One complete stride cycle was the time between one initial contact until the next initial contact of the same foot occurred. The first initial contact event was measured by using the forceplate. Since we only had one embedded forceplate, the second initial contact event of the same foot was measured by using a MotionMonitor software algorithm to calculate vertical foot velocity. This technique has been widely accepted as accurate and reliable to detect gait events (Jasiewicz et al., 2006;

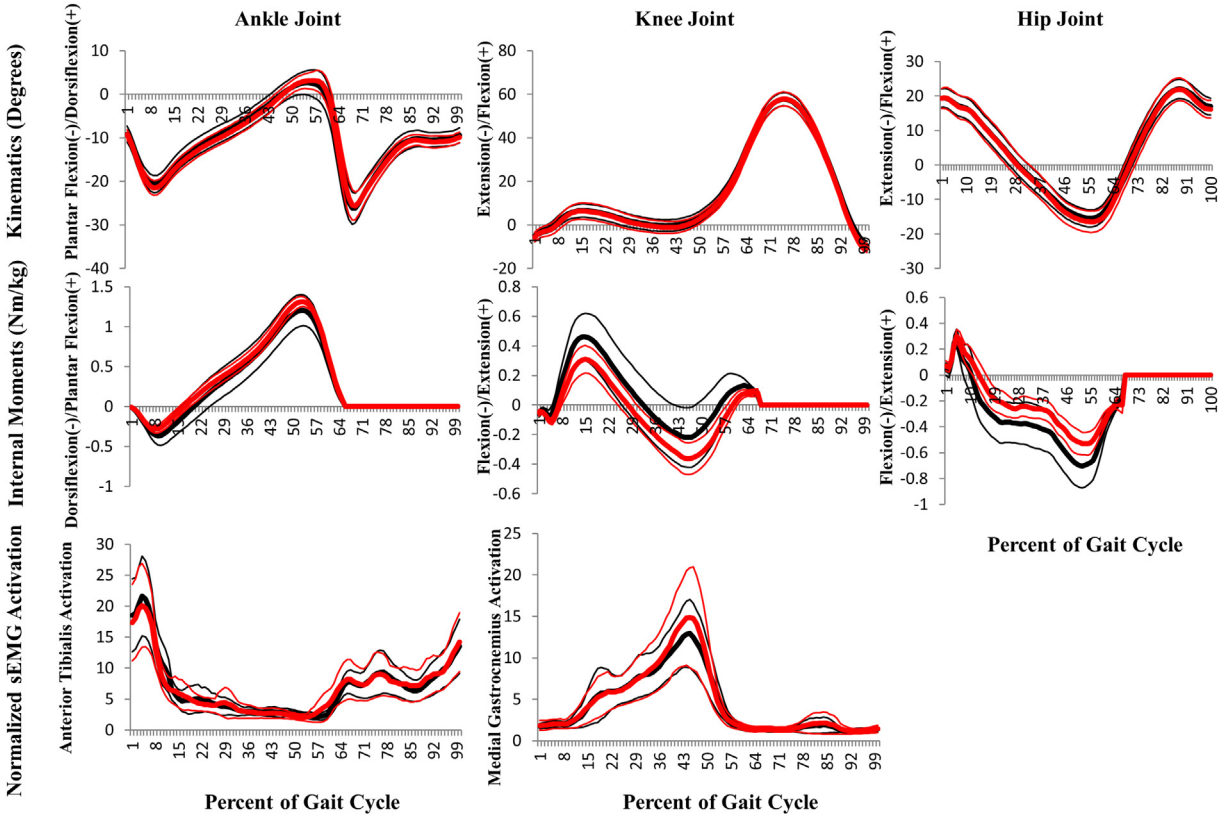


Fig. 6. No device group means with 90% confidence intervals (CIs) for sagittal plane ankle, knee, and hip kinematics, internal moments and sEMG activation during the gait cycle (1%–100%) pre- and post-rehabilitation. Stance phase represents approximately 0–68% of the gait cycle. Black represents pre-rehabilitation and red represents post-rehabilitation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

O'Connor, Thorpe, O'Malley, & Vaughan, 2007). In addition to using this technique, we visually inspected each trial to ensure the computer generated initial contact event, represented the same time the animation appeared to heel strike. This was completed for all ankle, knee, and hip kinematics and kinetics and for the normalized sEMG amplitude of the anterior tibialis, peroneus brevis, peroneus longus, and medial gastrocnemius. This data reduction technique has been previously reported using other software programs (Chinn et al., 2013; McKeon et al., 2009).

3.8. Statistical analysis

For the dependent variables ankle, knee, and hip frontal and sagittal kinematics, kinetics and vGRF and for sEMG activity of the anterior tibialis, peroneus brevis, peroneus longus, and medial gastrocnemius, group means and associated 90% confidence intervals (CIs) were calculated across all 100 points of the gait cycle (Chinn et al., 2013). A sequence of times series CI analyses were performed across the entire gait cycle to determine any increments where the CIs did not overlap. This was completed comparing pre-rehabilitation to post-rehabilitation within the device group and pre-rehabilitation to post-rehabilitation within the no device group. In the event that there were no differences with-in each group following rehabilitation, we chose to pool the no device and device groups into one group and complete a pre-rehabilitation to post-rehabilitation comparison for the combined group to assess for the overall effectiveness of an impairment-based rehabilitation program on lower limb biomechanics. If CIs did not overlap for at least 3 consecutive time increments, those increments in the gait cycle were considered statistically significant (Drewes, McKeon, Kerrigan, et al., 2009; Drewes, McKeon, Paolini, et al., 2009; McKeon et al., 2009).

4. Results

All pre-post rehabilitation frontal and sagittal plane kinematics and kinetics for the ankle, knee, hip and sEMG can be found in Figs. 3–8. The device group had increased dorsiflexion motion during mid-late stance ((Mean difference ± SD; % of cycle): 5.4 ± 1.6°; 45–64%) and had lower normalized sEMG amplitude for the peroneus longus during early stance (2.9 ± 0.06; 4–7%) and mid-swing (1.0 ± 0.04; 73–76%) after rehabilitation. The no device group had less peroneus brevis sEMG activity during early stance (3.4 ± 0.39; 4–13%) after rehabilitation. There were no other changes in ankle, knee, hip kinematics and kinetics for either group following rehabilitation. In addition, there were no differences in vGRF after rehabilitation for any group (Fig. 9).

5. Discussion

Our primary findings were that incorporating ankle destabilization devices into an impairment-based rehabilitation program did not improve ankle inversion positioning during gait. Furthermore, we found that when we combined both groups, there were no improvements in ankle inversion kinetics or kinematics during gait after rehabilitation. However, we did find that ankle destabilization devices caused an increase in dorsiflexion during the mid to late stance phase of gait. In addition, we also found a decrease in peroneus brevis sEMG activity during early stance in the no device group and a decrease in peroneus longus sEMG activity during early

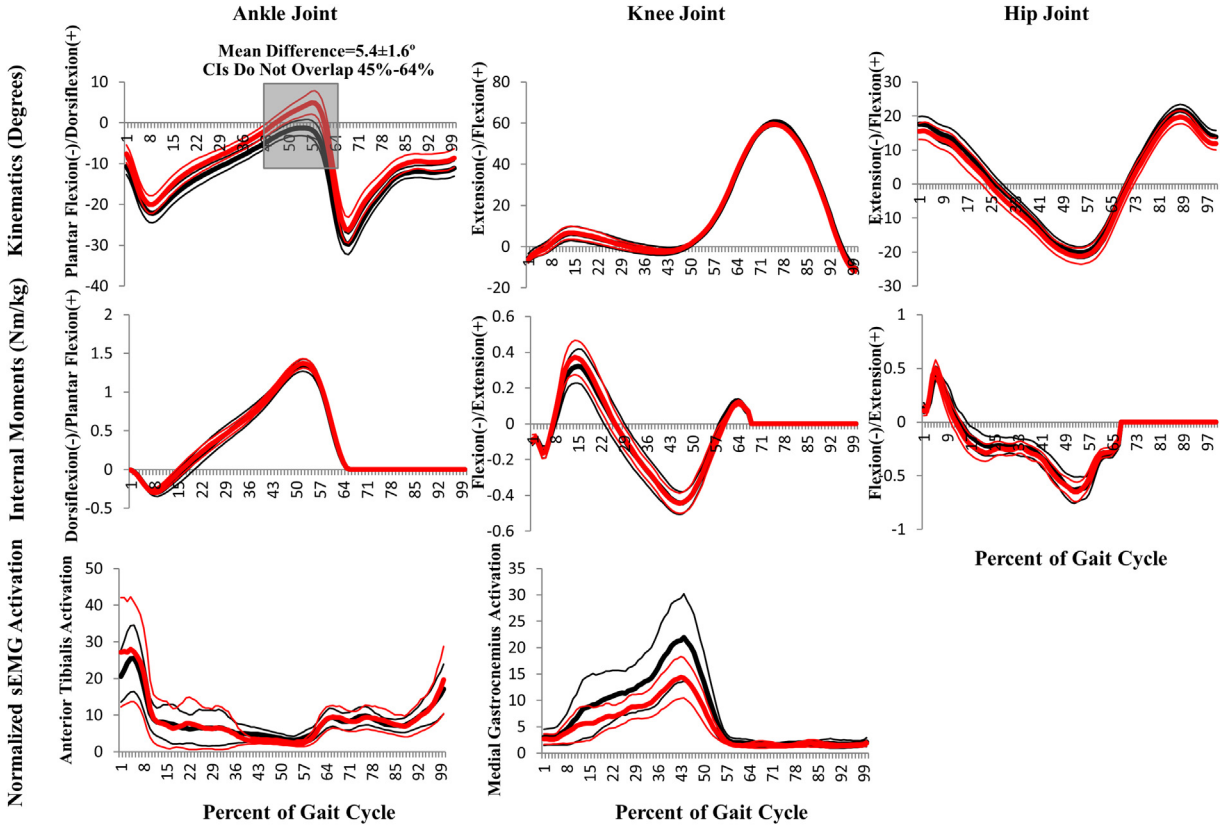


Fig. 7. Device group means with 90% confidence intervals (CIs) for sagittal plane ankle, knee, and hip kinematics, internal moments and sEMG activation during the gait cycle (1%–100%) pre- and post-rehabilitation. Stance phase represents approximately 0–68% of the gait cycle. Black represents pre-rehabilitation and red represents post-rehabilitation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

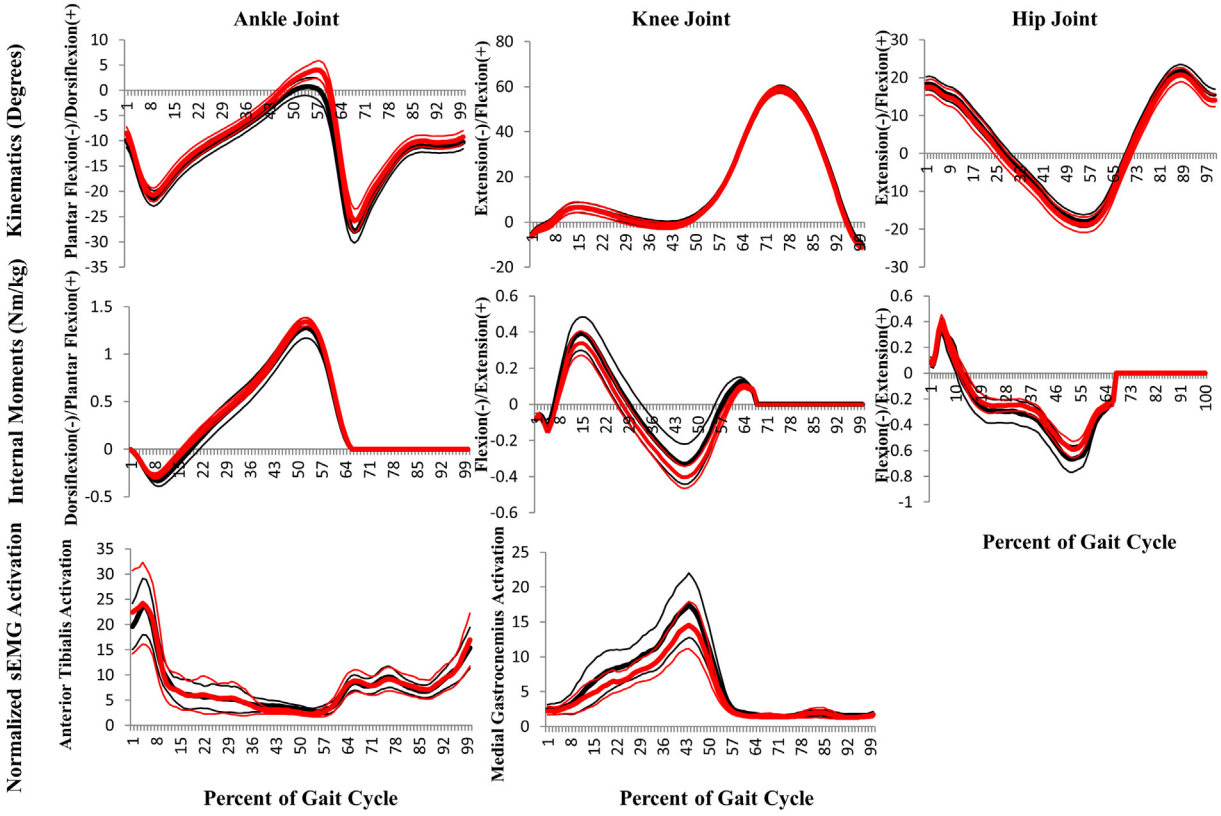


Fig. 8. Pooled group means with 90% confidence intervals (CIs) for sagittal plane ankle, knee, and hip kinematics, internal moments and sEMG activation during the gait cycle (1%–100%) pre- and post-rehabilitation. Stance phase represents approximately 0–68% of the gait cycle. Black represents pre-rehabilitation and red represents post-rehabilitation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

stance and mid-swing in the device group after rehabilitation. Finally, after rehabilitation, we found no changes in proximal joint kinematics or kinetics for the device, no device, or pooled groups.

5.1. Ankle frontal plane kinematics and kinetics

We found no alterations after rehabilitation in inversion-eversion motion or inversion-eversion moments for the device, no device or pooled groups. These findings are of interest because following rehabilitation, these patients had improvements in self-reported function (FAAM-Sport Pre = 66.5%, Post = 86.3%), dorsiflexion ROM, ankle strength, and balance (Donovan et al., 2016). The improvements in self-reported function, dorsiflexion ROM, ankle strength and balance were similar in the device group, the no device group, and when the groups were pooled (Donovan et al., 2016). This suggests that improvements in these other clinical measures associated with CAI are not associated with altered ankle frontal plane kinematics and kinetics. These results are consistent with McKeon et al. (2009)., who examined the effects of a balance training protocol on ankle gait measures during treadmill walking and a case-report by O'Driscoll, Kerin, & Delahunt (2011)., who examined the effects of a 6-week dynamic neuromuscular training program on ankle joint function. Neither study (McKeon et al., 2009; O'Driscoll et al., 2011) found changes in ankle inversion motion throughout the gait cycle; however, both studies reported large improvements in the patients' self-reported function. In addition, the McKeon et al. (2009). study found improved shank/rearfoot coupling stability measured by a reduction in shank/rearfoot coupling variability using a continuous relative phase analysis, and the O'Driscoll et al. (2011) study found decreased plantar flexion during a jump landing task and decrease vGRF during gait.

It appears that in order to decrease ankle inversion positioning throughout gait in CAI patients, clinicians must incorporate specific gait training modalities in conjunction with impairment-based rehabilitation. Gait training using visual feed-back from a 3-dimensional motion capture system, has been shown to effectively reduce the knee adduction moment in healthy knee varus aligned individuals over 8 treatment sessions (Barrios, Crossley, & Davis, 2010). This study (Barrios et al., 2010) shows that it is possible to alter frontal plane movement patterns; however, to our knowledge there has not been a gait training technique at the ankle to cause lasting reduction in inversion motion during walking. Specific to the ankle destabilization devices, it appears the feed-forward mechanism that has previously been reported (Donovan et al., 2015) may not have been great enough to improve the foot position or cause lasting changes. Even though our impairment-based rehabilitation program improved FAAM-Sport score to a mean of 86%, which would no longer qualify these patients of enrolling in this study based on recent recommendations (Gribble et al., 2013), we recognize that there is still a 14% deficit from the maximum score on the self-reported function scale.

5.2. Ankle sagittal plane kinematics and kinetics

We found approximately a 6° increase in peak dorsiflexion in the device group after rehabilitation during the mid-to-late stance phases of gait. However, we did not find differences in sEMG activity of the anterior tibialis or medial gastrocnemius or differences in dorsiflexion/planter flexion moments that correspond to this improvement in dorsiflexion motion. Furthermore, we found that both groups had equal improvements in standing straight knee/standing bent knee dorsiflexion after rehabilitation (Donovan et al.,

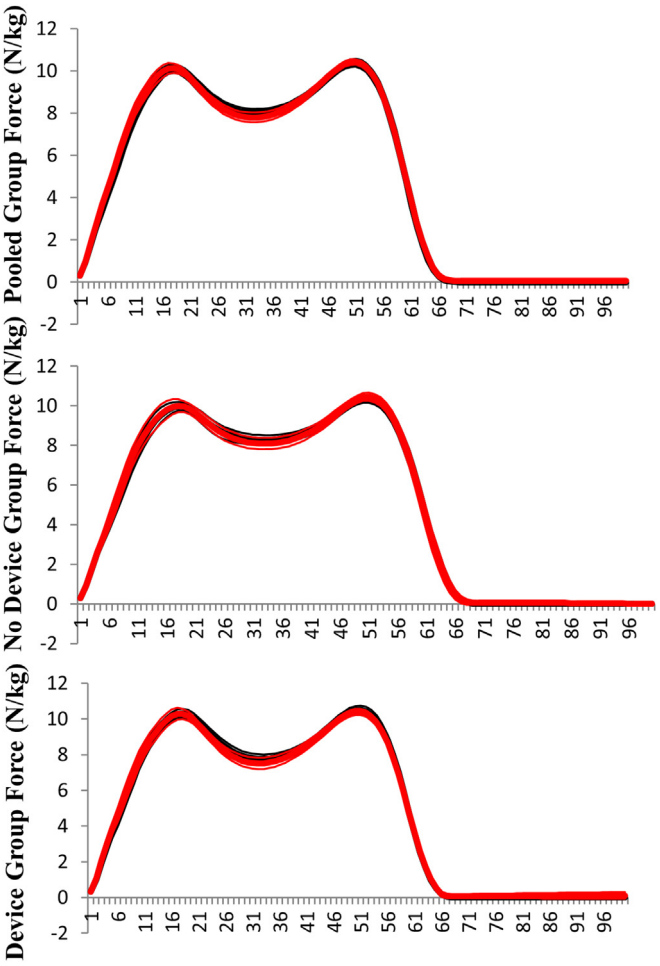


Fig. 9. Vertical ground reaction force (VGRF) normalized to mass means with 90% confidence intervals (CIs) during the gait cycle (1%–100%) pre- and post-rehabilitation. Stance phase represents approximately 0–68% of the gait cycle. Black represents pre-rehabilitation and red represents post-rehabilitation. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of this article.)

2016). We believe the increase in dorsiflexion may be a result of the design of the devices with regards to the amount of plantar flexion it forces individuals to be in during ground contact phases of walking or other functional tasks. These devices may have implemented a feed-forward response to the amount of plantar flexion they were undergoing during the stance phase of gait. Even though there was not a statistically significant reduction in medial gastrocnemius activity during mid-late stance after rehabilitation, there was about a 10% reduction in mean sEMG activity after rehabilitation during the phase where the improvements of dorsiflexion were found. Improvement in dorsiflexion during the stance phase of gait, may cause an increase in joint stability since the boney congruency of the joint is increased with more dorsiflexion (Hertel, 2002).

5.3. Surface EMG activity of the peroneus brevis and peroneus longus

We found a reduction in peroneus brevis activation during early stance in the no device group after rehabilitation and a reduction in peroneus longus activation during early stance and during mid-swing in the device group after rehabilitation. These findings refute our original hypothesis that both unstable surfaces and destabilization devices would increase muscle activity during gait and that the ankle destabilization devices would cause an isolated increase in peroneus brevis and peroneus longus sEMG activation. Even though these sEMG activation differences were statistically significant, they were not large enough to cause alterations in kinematics or kinetics during walking, but they do suggest that rehabilitation may cause a preservation strategy during walking where individuals can complete the same task with less muscle activity. In a study completed by Feger, Donovan, Hart and Hertel (2015), they found that the peroneus longus was activated for approximately 15% longer of the gait cycle than healthy individuals. The authors concluded that the increased peroneal activation may cause a fatiguing response of the muscle, which decreases its ability to adequately protect against sudden perturbations. Based on our study design, we cannot conclude that the reduction in sEMG activity of ankle evertors is advantageous, but similar sEMG activation patterns have been shown during the application of ankle bracing in CAI patients (Barlow, Donovan, Hart, & Hertel, 2015), which are devices that are successful at decreasing lateral ankle sprains (McGuine, Brooks, & Hetzel, 2011; McGuine, Hetzel, Wilson, & Brooks, 2012). Furthermore, it has been shown that healthy individuals do not activate their evertor musculature until mid-stance (Feger, Donovan, Hart, & Hertel, 2015).

5.4. Limitations

Although impairment-based rehabilitation did not change frontal plane gait kinematics or kinetics, we did not include any joint coupling variability measures (McKeon et al., 2009; Plisky, Rauh, Kaminski, & Underwood, 2006). These analyses may provide more insight on subtle gait alterations after rehabilitation. In addition, we did not monitor walking speed during the trials, but relied on a self-selected pace for each patient. Theoretically, walking speed may have been altered following the intervention; however, given relatively no changes in joint moments or vGRF, we do not believe any changes in walking speed would have been large enough to be clinically meaningful. Also, all gait measures were collected in a neutral shoe and not the destabilization devices and post rehabilitation data collection took place 2–7 days following rehabilitation. It is possible that the feed-forward mechanism of the devices diminished prior to post-testing. However, we believe that this is not clinically relevant since we would want gait alterations to last longer than 2–7 days following a 4-week intervention. Furthermore, this study was limited to one follow-up data collection session. Therefore, it remains unclear of how successful this rehabilitation program is at reducing recurrent ankle sprains.

6. Conclusion

Despite causing improvements in self-reported function, a four week progressive rehabilitation program that incorporates ankle destabilization devices or unstable surfaces during balance and functional exercises did not alter frontal plane kinematics or kinetics of the ankle, knee and hip joint in CAI patients. Ankle destabilization devices caused an increase in dorsiflexion ROM during mid to late stance during gait after rehabilitation, which may contribute to an increase in ankle joint stability. Rehabilitation programs to treat CAI should encompass exercises to improve all deficits associated with CAI and include specific gait training exercises to decrease excessive inversion ROM throughout the gait cycle.

Conflict of interest

None declared.

Ethical approval

The University of Virginia's Institution Review Board for Health Sciences (IRB-HSR-16992) has approved this project.

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