

Alteration of ankle kinematics and muscle activity during heel contact when walking with external loading

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Abstract

Purpose The risk of recurrent ankle sprains could be reduced if the ankle joint is positioned in a more closed pack position (eversion and dorsiflexion) with enhanced co-activation of Tibialis Anterior (TA) and Peroneus Longus (PL) around heel contact in walking. We examined whether such alteration can be induced by augmenting ankle position error in the inversion and plantarflexion directions.

Methods Fifteen young, healthy subjects participated in this study. They walked on a treadmill while a light weight was affixed to the dorsal–lateral side of the test foot to create afferent error signals in the inversion and plantarflexion directions. We recorded and analyzed subjects' ankle kinematics and electromyographic (EMG) activity of TA and PL before (baseline), during (adaptation), and after (post adaptation) walking with the weight. Our analyses focused on 30 ms before and after heel contact where ankle sprains are most likely to occur.

Results Subjects increased ankle eversion before and after heel contact during the adaptation period. This increase carried over to the post adaptation period following removal of the weight. The weight also induced an increase in the TA activity before heel contact during late adaptation, although this increase did not carry over to the post adaptation period. No significant changes were observed in ankle dorsiflexion, PL activity, and muscle co-activation.

Conclusion Our error-driven approach is feasible to reduce ankle inversion around heel contact in walking, and may have clinical implication on intervention of recurrent ankle sprains.

Keywords Ankle · Gait · Motor learning · Error augmentation · Sensorimotor adaptation

Abbreviations

TA	Tibialis anterior
PL	Peroneus longus
EMG	Electromyography
LMM	Linear mixed model
CNS	Central nervous system
CCI	Co-contraction index
QTM	Qualisys track manager
IEMG	Integrated electromyography

Introduction

Ankle sprains are one of the most common injuries in sports and recreation (Ferran and Maffulli 2006; Garrick 1977). The most common type of ankle sprain is lateral or inversion injuries where the foot tilts inwards excessively, damaging the lateral structures of the ankle complex (Hertel 2002). Many individuals who experience an ankle sprain

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eventually develop chronic ankle instability (Peters et al. 1991; Smith and Reischl 1986), which is characterized by lingering pain, swelling, and most problematically, recurrent ankle sprains (Gutierrez et al. 2009). A potential factor that could contribute to recurrent ankle sprains is walking with excessive ankle inversion (Delahunt et al. 2006; Monaghan et al. 2006). This position increases the tendency for the ankle to further move into inversion after heel contact, and thus is a position more susceptible to inversion injuries (Hertel 2002; Konradsen and Voigt 2002).

Stabilization of the ankle joint around heel contact in walking could reduce the risk of recurrent ankle sprains. This could be achieved by two motor strategies. The first strategy is to place the ankle joint in a more everted and dorsiflexed position (i.e., closed pack position) around heel contact. The second strategy is to increase co-activation of Tibialis Anterior (TA) and Peroneus Longus (PL), as suggested by previous investigation on ankle sprain ‘copers’ (Dundas et al. 2014). The term *coper* refers to individuals who successfully cope with the damage caused by the initial ankle sprain and do not develop chronic ankle instability. These individuals increased co-activation of TA and PL as a strategy to stabilize the ankle joint during challenging activities such as jump landing and curb walking (Gutierrez et al. 2012).

Ankle kinematics and muscle activity during walking could be modulated by error signals generated by controlled force perturbation. A previous study used robot-generated ankle plantar flexion torque to induce increases in dorsiflexion and TA activity during early swing (Blanchette et al. 2011). While the perturbation initially caused the ankle to move toward plantar flexion (error augmentation), subjects quickly reduced this error by increasing ankle dorsiflexion and TA activity (error correction). Following removal of the resistance, the increased ankle dorsiflexion and TA activity during the early swing phase persisted (aftereffects). The aftereffects lasted shortly for approximately 20 cycles before the amplitude reduced to 37 % of the initial value. This approach has been tested in patients post stroke to improve drop foot and increase ground clearance during walking (Blanchette et al. 2014).

The error detection and correction processes induced by force perturbation appear to involve both feedback and feedforward control mechanisms (Gordon et al. 2010; Lam et al. 2006). The force perturbation stimulates the ankle proprioceptors and mechanical receptors to generate error feedback, and the central nervous system (CNS) modifies motor outputs in response to the feedback to minimize movement error. The CNS also modifies motor outputs proactively to minimize movement error in anticipation to the upcoming perturbation. Such feedforward control is evident by lingering aftereffects observed *after* the perturbation and error feedback are removed (Blanchette and Bouyer 2009; Emken and Reinkensmeyer 2005).

In this study, we asked if a similar error-driven approach could be used to increase ankle eversion and dorsiflexion as well as co-activation of TA and PL around heel contact in walking. To address this question, we used a sandbag to create external loading to deviate the ankle joint away from its regular state during walking. We hypothesized that (a) external loading pushes the ankle toward inversion and plantar flexion would induce increased ankle eversion and dorsiflexion as well as co-activation of TA and PL before and after heel contact during walking; (b) these alterations would persist, in the short term, after the perturbation is removed.

Methods

Subjects

A total of 15 subjects (7 females and 8 males) participated in this study. They were young and healthy (mean age = 22.6 ± 2.1 years; mean height = 175.1 ± 10 cm; mean weight = 73.5 ± 13.8 kg) without medical conditions affecting the leg joints and walking. In this study, we used each subject’s dominant leg as the test leg, as determined by the leg used to kick a ball (Hoffman and Payne 1995). The right leg was the dominant leg for all subjects. Before participation, subjects were required to sign a consent form approved by the Northeastern University Human Subject Research Protection office. This study has been performed in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki.

Instrumentation

A 6-camera motion analysis system (Oqus 300, Qualisys AB, Gothenburg, Sweden) was used to capture subjects’ ankle joint kinematics of the test leg during treadmill walking. Qualisys track manager (QTM) software was used to track and record the kinematic data.

To create an ankle model, reflective markers were placed on the major bony landmarks of the shank and foot of the test leg, including medial and lateral femoral epicondyles, medial and lateral malleoli, the second toe, and the fifth metatarsal head. A rigid cluster of four markers was placed on the calf muscle as tracking markers for the shank. In addition, two separate markers were placed on the heel and the lateral side of the calcaneus. These two markers, along with the marker on the second toe, were used as the tracking markers for the foot. After the model was created, the markers on the medial femoral condyle, medial malleolus, and fifth metatarsal head were removed for walking trials. The foot models were created using Visual 3D software (C-Motion Inc, Rockville, MD).

Two electromyographic (EMG) electrodes (Trigno Wireless, Delsys Inc, Natick, MA) were placed on the TA



Fig. 1 A 1-lb weight was placed on the dorsal–lateral side of the foot as a perturbation. The weight was affixed on the foot using an elastic bandage (not shown)

and PL of the test leg to record the muscle activity. EMG electrodes were placed on the skin over each muscle belly aligned in parallel with the muscle fibers and in locations detailed by Perotto et al. (2011). Prior to application of EMG electrodes, the skin was cleaned with alcohol wipes and hairs were removed using a razor. EMG signals were sampled at a frequency of 2000 Hz.

Data collection procedures

Subjects participated in a single data collection session in which they were required to walk on a treadmill. The session was divided into baseline, adaptation, and post adaptation periods. During the baseline period, subjects walked on a treadmill without any perturbation for 1 min. During the adaptation period, subjects walked on the treadmill for 5 min, while a 1-lb sandbag was placed to the dorsal–lateral side of the test foot, specifically on the 3rd, 4th, and 5th metatarsal bones, to create an error signal in the inversion and plantarflexion directions (Fig. 1). The sandbag was firmly affixed to the foot using an elastic bandage. Based on our pilot tests conducted prior to this current study, a 1-lb weight was sufficient for individuals to sense the perturbation but was light enough for subjects to overcome without strenuous effort, which minimized the risk of ankle injuries during the walking trial. During the post-adaptation period, the weight was removed while subjects continued walking on the treadmill for another minute. The walking speed was set at each subject's self-selected comfortable speed, which was determined prior to data collection. Subjects' comfortable speeds ranged from 1.8 to 2 mph.

Data analysis

The kinematic outcome variables included the mean ankle position in the frontal and sagittal planes over the last 30 ms of the swing phase (pre-landing phase) and over the first 30 ms of the stance phase (post-landing phase). The kinematic data were analyzed using Visual 3D. The marker data were smoothed using a low-pass filter (2nd-order Butterworth filter) with a cutoff frequency of 6 Hz. The filter was run in both forward and reverse directions to remove phase lag. The ankle was modeled as a three-degree-of-freedom joint with the center midway between the markers on medial and lateral malleoli. An Euler angle approach was used to calculate the ankle angle in each plane. The sequence of rotation was XYZ, where *X* is dorsiflexion–plantarflexion, *Y* is inversion–eversion, and *Z* is abduction–adduction. Heel contact was determined as the time frame when the heel marker changes from moving in the forward direction to the backward direction. This method was developed by Zeni et al. (2008) and was shown to have excellent concurrent validity against the force plate method.

The outcome variables for muscle activity included: (a) the integrated EMG (IEMG) of TA during the pre- and post- landing phases; (b) the IEMG of PL during the pre- and post-landing phases; (c) the co-contraction index (CCI) of TA and PL during the pre- and post- landing phases. The EMG data were analyzed using customized software written in LabVIEW language (National Instruments, Austin, TX). The signals were band-pass filtered (2nd-order Butterworth filter, with cutoff frequencies of 10 Hz and 500 Hz), rectified, and smoothed with a low-pass filter (2nd-order Butterworth filter, with a cutoff frequency of 6 Hz) before integration. The filter was run in both forward and reverse directions to remove phase lag.

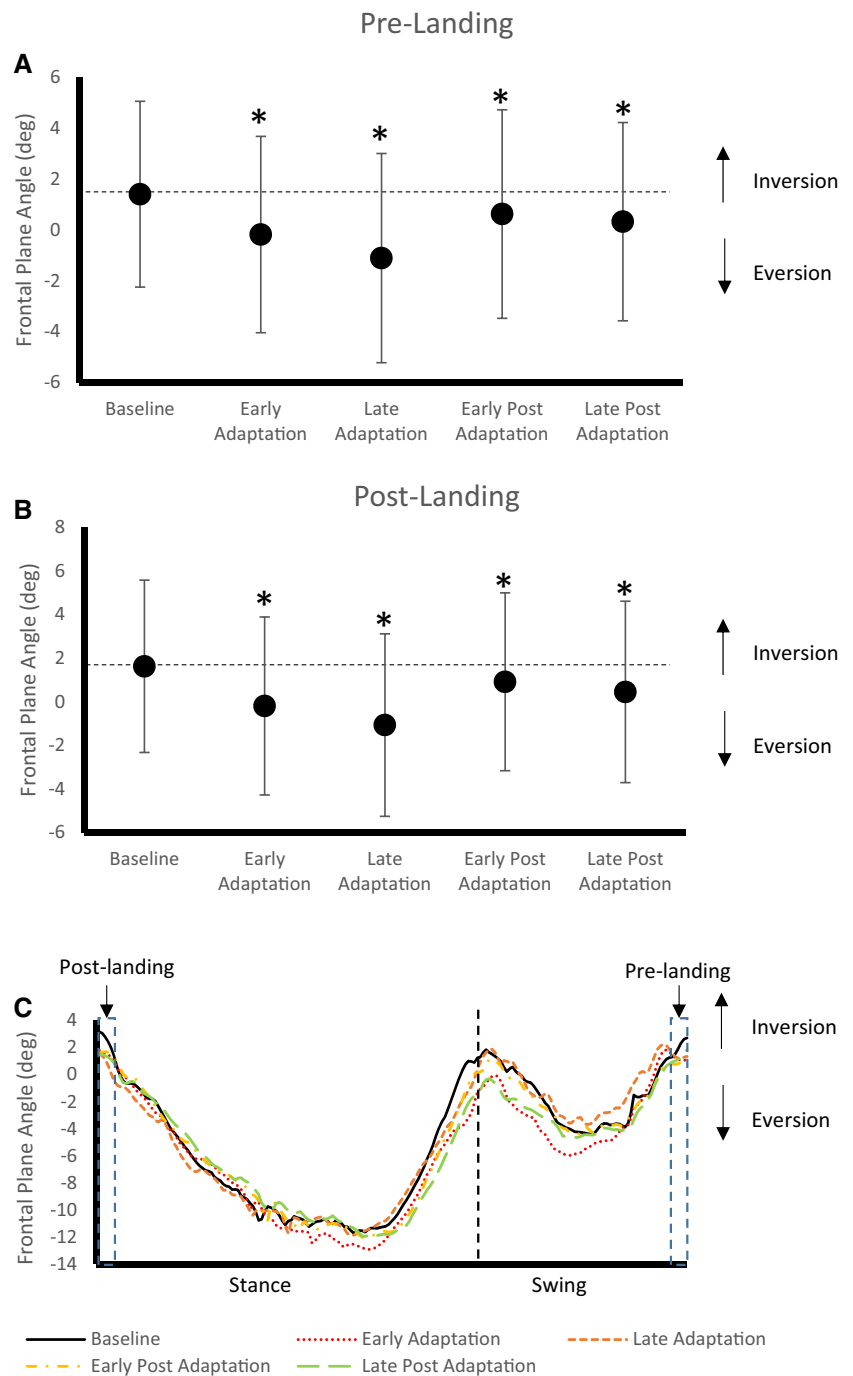
The CCI were calculated using the following algorithm: $\frac{IEMGs}{IEMGI} \times (IEMGs + IEMGI)$, where IEMGs were the IEMG of the less active muscle and IEMGI was the IEMG of the more active muscle (Rudolph et al. 2001).

Each outcome variable was averaged across the last 5 cycles of the baseline period (baseline), the first 5 cycles of adaptation period (early adaptation), the last 5 cycles of adaptation period (late adaptation), the first 5 cycles of post adaptation period (early post adaptation), and the 16th–20th cycles of the post adaptation period (late post adaptation). We selected the 16th–20th cycles for analysis based on a previous study showing that aftereffects on ankle kinematics and EMG activity can persist for approximately 20 cycles (Blanchette et al. 2011).

Statistical analysis

We used linear mixed models (LMM) to compare each outcome measure across baseline, early adaptation, and

Fig. 2 **a** The mean ankle inversion–eversion angle in the pre-landing phase across subjects in each test period. **b** The mean ankle inversion–eversion angle in the post-landing phase across subjects in each test period. **c** The mean ankle kinematic trajectory in the frontal plane from a representative subject (case 14) across test periods. In **a, b**, error bars represent standard deviation. The Asterisk value was significantly different from baseline ($p < 0.05$)



late adaptation values. The correlation between repeated observations was modeled as “unstructured” to allow the observed data to dictate the correlations. When significance was detected, post hoc tests with Bonferroni correction were conducted. All statistical analyses were carried out using IBM SPSS version 21 (Armonk, NY: IBM Corp.), with two-sided tests and a type-1 error rate of 0.05.

Results

Ankle kinematics in the frontal plane

Figure 2a shows the mean ankle inversion–eversion angle in the pre-landing phase across subjects. On average, subjects’ ankle position was slightly inverted during the baseline period ($1.4 \pm 3.6^\circ$). After the weight is applied,

the ankle position became more everted during the early adaptation period ($-0.2 \pm 3.9^\circ$) compared to the baseline, and moved further into eversion during the late adaptation period ($-1.1 \pm 4.1^\circ$). Following removal of the weight, the ankle moved toward inversion, but remained more everted than the baseline during the early ($0.6 \pm 4.1^\circ$) and late ($0.32 \pm 3.9^\circ$) post adaptation periods. The LMM detected significant differences across the periods examined ($p < 0.01$). Post hoc tests indicated that the baseline value was significantly different from the early adaptation value ($p < 0.01$), the late adaptation value ($p < 0.01$), the early post adaptation value ($p = 0.03$), and the late post adaptation value ($p < 0.01$).

Figure 2b shows the mean ankle inversion–eversion angle in the post-landing phase across subjects. The trend of ankle position change over the periods examined was similar to that observed in the pre-landing phase ($p < 0.01$, LMM). Post hoc tests indicated that the baseline value ($1.6 \pm 4^\circ$) was significantly different from the early adaptation value ($-0.2 \pm 4.1^\circ$, $p < 0.01$), the late adaptation value ($-1.1 \pm 4.2^\circ$, $p < 0.01$), the early post adaptation value ($0.9 \pm 4.1^\circ$, $p = 0.03$), and the late post adaptation value ($0.44 \pm 4.2^\circ$, $p < 0.01$).

Figure 2c shows the mean ankle kinematic trajectories in the frontal plane during each period examined from a representative subject (case 14). This subject demonstrated an increase in ankle eversion from the baseline in both pre- and post-landing phases during adaptation and post-adaptation periods, which was consistent with the results shown in Fig. 2a, b.

Ankle kinematics in the sagittal plane

Figure 3a shows the mean ankle dorsiflexion–plantarflexion angle in the pre-landing phase across subjects. On average, subjects' ankle position was in slight plantarflexion during baseline ($-0.7 \pm 4.2^\circ$). After the weight was applied, the ankle moved further toward plantarflexion during the early adaptation period ($-1.7 \pm 3.1^\circ$). While the ankle moved back toward the baseline during the late adaptation period ($-1.1 \pm 4^\circ$), the position was still in more plantarflexion compared to the baseline. Following removal of the weight, subjects' ankle position remained in more plantarflexion compared to the baseline during both early ($-1.2 \pm 3.8^\circ$) and late ($-1 \pm 4.1^\circ$) post adaptation periods. However, the LMM suggested that none of the aforementioned deviations from the baseline was significant ($p = 0.25$).

Figure 3b shows the mean ankle dorsiflexion–plantarflexion angle during the post-landing phase across subjects in the baseline ($-0.6 \pm 4.3^\circ$), early adaptation ($-1.4 \pm 3.2^\circ$), late adaptation ($-0.8 \pm 4.3^\circ$), early post adaptation ($-1.1 \pm 4.3^\circ$), and late post adaptation periods ($-1 \pm 4.5^\circ$). The trend of ankle position change over these

periods was similar to that observed in the pre-landing phase, and the LMM also did not detect significance in this trend ($p = 0.33$).

Figure 3c shows the mean ankle kinematic trajectories in the sagittal plane during each period examined from the same representative subject (case 14). This subject did not demonstrate an obvious change in ankle dorsiflexion–plantarflexion angles from the baseline in both pre- and post-landing phases during adaptation and post-adaptation periods, which was consistent with the results shown in Fig. 3a, b.

TA EMG activity

The EMG analyses for both TA and PL were based on 14 subjects; one subject (case 6) was excluded due to technical issues with recording. Figure 4a shows the mean TA IEMG data across subjects in the pre-landing phase (Fig. 4a). On average, the TA activity increased from the baseline ($36.2 \pm 15.9 \text{ uV} \times \text{s}$) during the early adaptation period ($38.5 \pm 15.4 \text{ uV} \times \text{s}$), and further increased during the late adaptation period ($41.2 \pm 14.5 \text{ uV} \times \text{s}$). Following removal of the weight, however, the TA activity dropped below the baseline during both early ($33.9 \pm 14.5 \text{ uV} \times \text{s}$) and late ($34.3 \pm 16 \text{ uV} \times \text{s}$) post adaptation periods. The LMM detected significance in TA IEMG across the periods examined ($p < 0.01$). Post hoc tests indicated that only the late adaptation value was significantly different from the baseline value ($p = 0.01$, Fig. 4a).

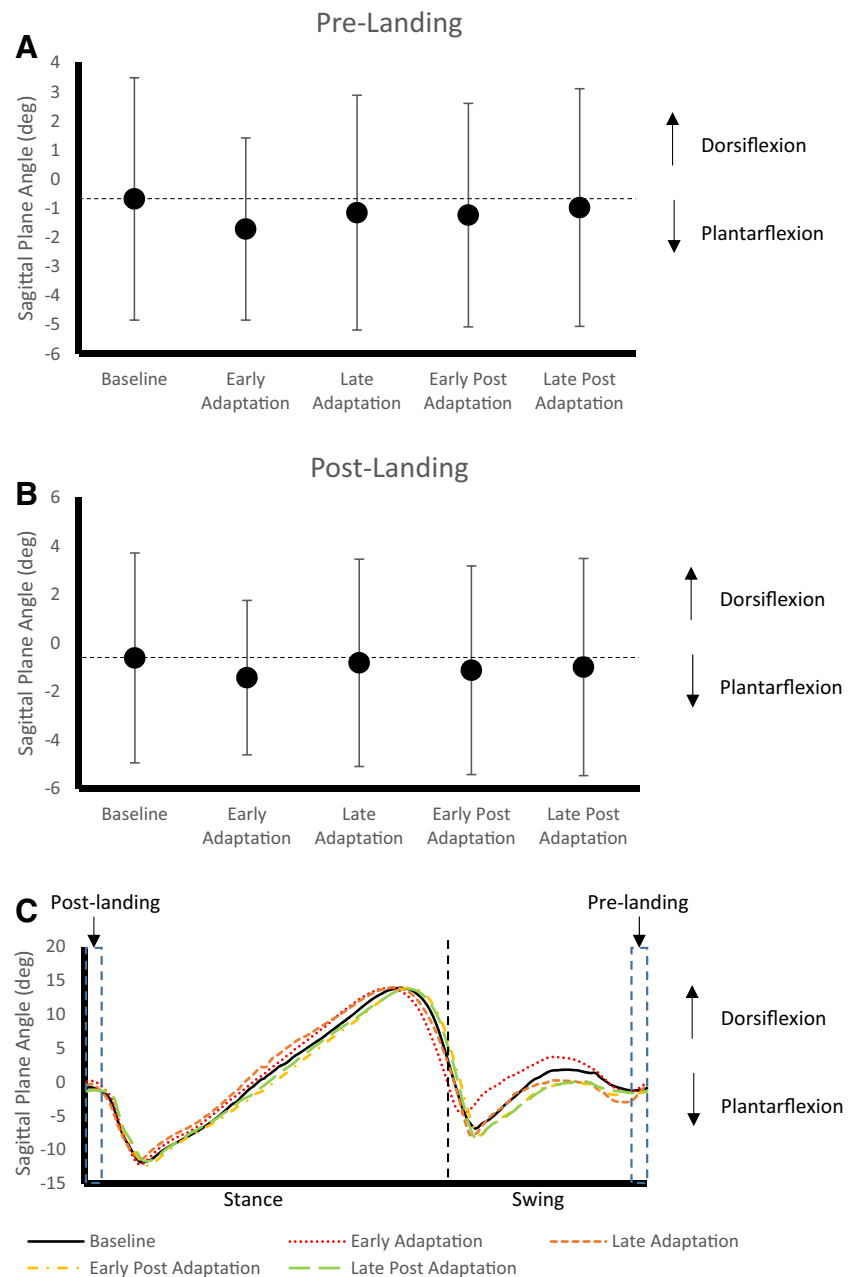
Figure 4b shows the mean TA IEMG data across subjects in the post-landing phase. Here, the TA activity was higher than the baseline ($40.7 \pm 17.6 \text{ uV} \times \text{s}$) during the late adaptation period ($45.5 \pm 20.5 \text{ uV} \times \text{s}$), but was close to the baseline during the early adaptation period ($39.8 \pm 14.7 \text{ uV} \times \text{s}$) and early ($37.6 \pm 16.5 \text{ uV} \times \text{s}$) and late post adaptation periods ($40.4 \pm 19.8 \text{ uV} \times \text{s}$). While the LMM detected significance in the TA IEMG in the post-landing phase across the periods examined ($p = 0.01$), post hoc tests indicated that none of the adaptation or post adaptation values was significantly different from the baseline ($p > 0.2$).

Figure 4c shows the mean TA EMG pattern during each period examined from the same representative subject (case 14). This subject showed an increase in TA activity in the pre-landing phase from the baseline, especially during the adaptation period. In the post-landing phase, the differences in the TA activity between baseline and adaptation/post adaptation periods were not as obvious as those observed in the pre-landing phase.

PL EMG activity

Figure 5a shows the mean PL IEMG in the pre-landing phase across subjects in the baseline ($9.1 \pm 3.1 \text{ uV} \times \text{s}$), early adaptation ($9.1 \pm 3.1 \text{ uV} \times \text{s}$), late adaptation

Fig. 3 **a** The mean ankle dorsiflexion–plantarflexion angle in the pre-landing phase across subjects in each test period. **b** The mean ankle dorsiflexion–plantarflexion angle in the post-landing phase across subjects in each test period. **c** The mean ankle kinematic trajectory in the sagittal plane from a representative subject (case 14) across test periods. In **a, b**, error bars represent standard deviation



($10.3 \pm 3.4 \text{ uV} \times \text{s}$), early post adaptation ($8.3 \pm 2.9 \text{ uV} \times \text{s}$), and late post adaptation periods ($8.4 \pm 2.7 \text{ uV} \times \text{s}$). Figure 5b shows the mean PL IEMG in the post-landing phase across subjects in the baseline ($9.6 \pm 3.2 \text{ uV} \times \text{s}$), early adaptation ($9.2 \pm 2.9 \text{ uV} \times \text{s}$), late adaptation ($11.1 \pm 3.7 \text{ uV} \times \text{s}$), early post adaptation ($9.2 \pm 3 \text{ uV} \times \text{s}$), and late post adaptation periods ($9.8 \pm 3.2 \text{ uV} \times \text{s}$). In both pre- and post-landing phases, the PL activity was higher than the baseline during the late adaptation period, but was close to the baseline during the rest of the adaptation and post adaptation periods. The LMM detected significance in PL IEMG for both pre- ($p = 0.05$) and post-landing phases ($p < 0.01$). However, post hoc tests indicated that none of

the adaptation or post adaptation values was significantly different from the baseline ($p > 0.1$).

Figure 5c shows the mean PL EMG pattern during each period examined from the same representative subject (case 14). No obvious differences were observed in the PL activity between baseline and adaptation/post adaptation periods in either pre- or post-landing phase.

Co-activation of TA and PL

Figure 6a shows the mean CCI between TA and PL in the pre-landing phase across subjects in the baseline (11.9 ± 4.3), early adaptation (11.7 ± 4.7), late adaptation

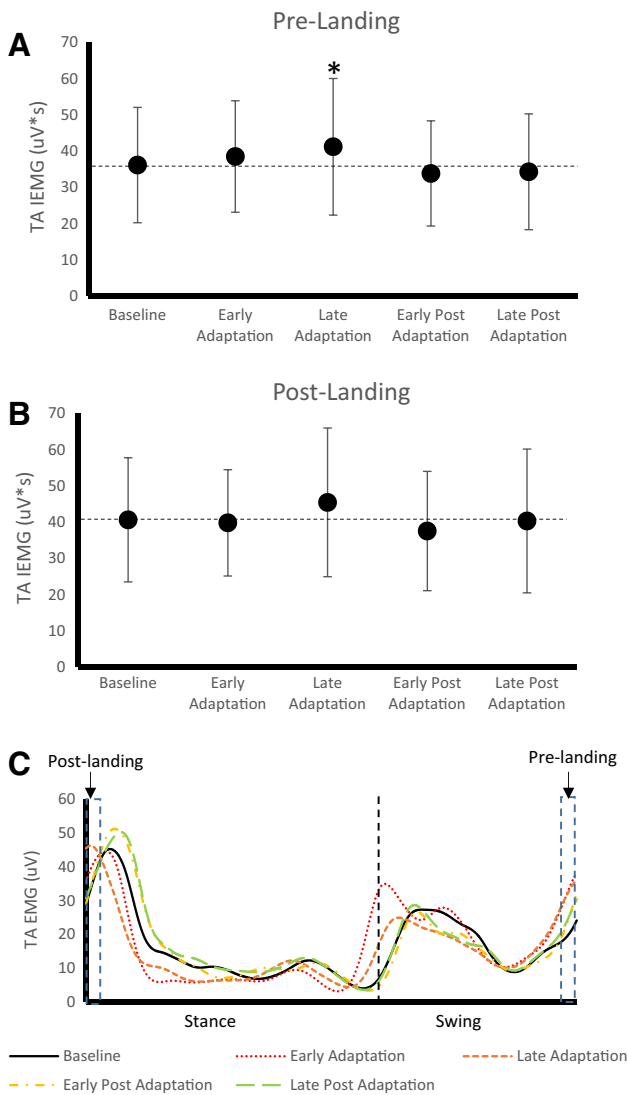


Fig. 4 **a** The mean TA IEMG in the pre-landing phase across subjects in each test period. **b** The mean TA IEMG in the post-landing phase across subjects in each test period. **c** The mean TA EMG pattern from a representative subject (case 14) across test periods. Error bars represent standard deviation. The Asterisk value was significantly different from baseline ($p < 0.05$)

(13.5 ± 4.6), early post adaption (10.7 ± 4.2), and late post adaptation periods (10.8 ± 3.8). Figure 6b shows the mean CCI in the post-landing phase across subjects in the baseline (11.3 ± 4.6), early adaptation (11.7 ± 4.1), late adaptation (14.4 ± 4.8), early post adaptation (11.9 ± 4.1), and late post adaptation periods (12.7 ± 4.4). In both pre- and post-landing phases, the CCI was higher than the baseline during the late adaptation period, but was close to the baseline during the early adaptation period and early and late post adaptation periods. The LMM detected significance in the CCI for only the post-landing phase ($p < 0.01$), but

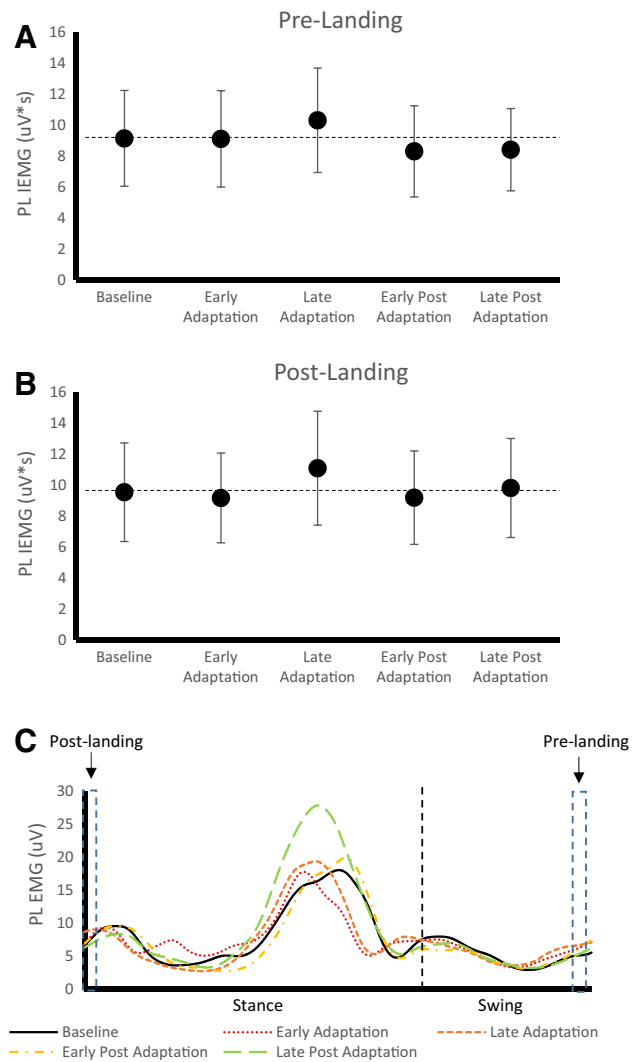


Fig. 5 **a** The mean PL IEMG in the pre-landing phase across subjects in each test period. **b** The mean PL IEMG in the post-landing phase across subjects in each test period. **c** The mean PL EMG pattern from case 14 across test periods. In **a**, **b**, error bars represent standard deviation

not the pre-landing phase ($p = 0.07$). However, post hoc tests indicated that none of the adaptation or post adaptation values was significantly different from the baseline in the post-landing phase ($p > 0.17$).

Discussion

Healthy individuals immediately counteracted the 1-lb weight placed on the dorsal-lateral side of the foot by increasing the ankle eversion angle in the pre- and post-landing phases. This increase persisted even after the

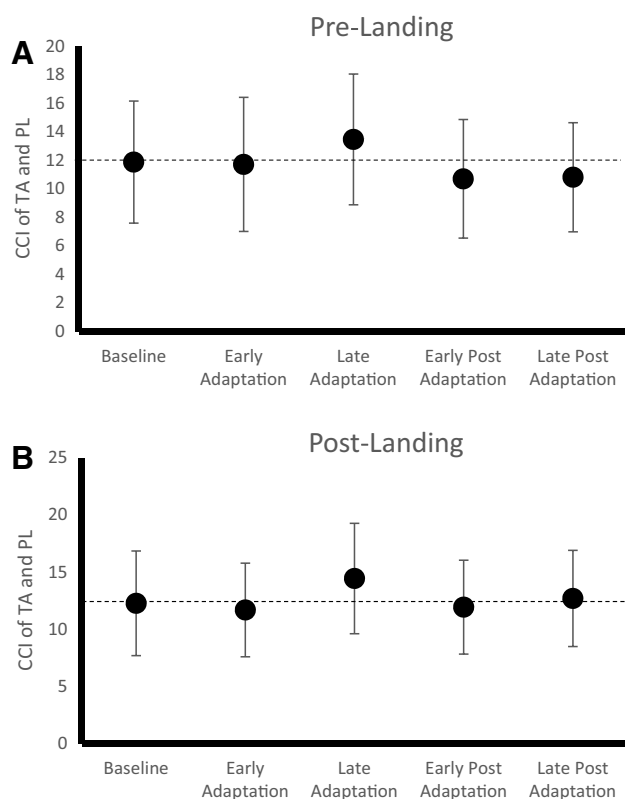


Fig. 6 **a** The mean co-contraction index (CCI) of TA and PL across subjects in each test period. **b** The CCI of TA and PL across subjects in each test period. Error bars represent standard deviation

weight was removed during the post adaptation period. The increase appeared to be small (though significant) in magnitude, which was expected, following just one short session of adaptation. The increased ankle eversion was a result of error augmentation, detection, and correction, which largely involved feedback control. However, feedforward mechanism was also involved in this process, as demonstrated by the finding that increased ankle eversion persisted following the removal of the weight (i.e., aftereffect). Without the weight, the aftereffect could not be a reaction to error feedback, but rather, it represented a change in feedforward commands in anticipation of the perturbation in upcoming gait cycles (Blanchette and Bouyer 2009; Emken and Reinkensmeyer 2005).

The weight was not able to increase ankle dorsiflexion in the pre- and post-landing phases during either adaptation or post-adaptation periods. Individuals seem to avoid performing heel contact with increased dorsiflexion, as this change may shift the contact spot more proximally to the body, creating an unnatural contact pattern that could interrupt the heel rocker and dynamics of gait. A previous study used a robotic device to generate perturbation in the plantarflexion direction during swing phase of gait, and induced an increase in dorsiflexion during the early swing phase

(Blanchette et al. 2011). Although this previous study did not examine kinematic changes around heel contact, it showed average ankle trajectories across subjects in the sagittal plane over a gait cycle before, during, and after perturbation. Based on the trajectories, the dorsiflexion angle was also not affected by the perturbation around heel contact. While this was consistent with our results, we cannot rule out the possibility that the lack of change observed in Blanchette et al. (2011) was because the force perturbation was provided only during early swing but not around heel contact.

The weight did not affect co-activation of TA and PL before and after heel contact during gait. The PL activity did not deviate away from the baseline during and after walking with the weight, despite the increase in ankle eversion. This may be because there are other ankle muscles including Peroneus Brevis and Tertius that could contribute to eversion. We did observe an increase in the TA activity in the pre-landing phase during the late adaptation period, but the increase did not carry over to the post adaptation period. A previous study found that increased TA activity in early swing induced by force perturbation could persist following removal of the perturbation (Blanchette et al. 2011). Their findings and ours together suggested that the ability of increased TA activity to carry over to the post adaptation period may be phase dependent (i.e., the increase in early swing can carry over, but that in pre-landing cannot). Phase-dependent carryover may suggest different control mechanisms for the muscle. The control of TA may involve more feedback mechanism around heel contact, and therefore the EMG activity returned to the baseline right after the external loading was removed. In contrast, the control may involve more feedforward mechanism during early swing, and therefore the increased EMG activity was able to persist without presence of the external loading. Phase-dependent carryover was also observed in Blanchette and Bouyer (2009) where they examined how people adapt Hamstrings activity to an elastic force field during walking. The increased Hamstrings activity around toe off was able to carry over to post adaptation period, but that from mid to late swing could not.

The results of this study may have clinical implication in ankle rehabilitation. Individuals with chronic ankle instability tend to walk with an overly inverted foot around heel contact (Delahunt et al. 2006; Monaghan et al. 2006). This could be a reason why they recurrently sprain the ankle during daily walking activities. Thus, correcting abnormal ankle positioning during walking should be a rehabilitation goal for ankle instability. However, existing rehabilitation programs for ankle instability do not show positive effects on correcting abnormal positioning during walking (Coughlan and Caulfield 2007; McKeon et al. 2009). An issue with these programs is that they are often based on

static balance training (e.g., single leg stance), and therefore the training effect may not transfer effectively to a dynamic situation such as walking.

The error-driven approach tested in this current study was performed in walking, and was able to induce an increased ankle eversion around heel contact. Thus, it may have the potential to help individuals with chronic ankle instability correct abnormal ankle positioning during walking and in turn, reduce the risk of recurrent ankle sprains. The increase in eversion found in this study appeared to be small (Fig. 2a, b) and probably did not have clinical significance. This may be because the adaptation period was short (5 min) or the parameters of the external loading (e.g., magnitude, timing, and application point) were not optimal. Despite the small magnitude, the results provided initial evidence to support that the error-driven approach could guide individuals to adapt toward eversion. In addition, the error-driven approach modulates the ankle kinematics through both feedback and feedforward pathways, as discussed previously. This seems to be especially suitable for individuals with chronic ankle instability as they appear to have deficits in both feedback (Konradsen and Ravn 1990) and feedforward controls (Caulfield et al. 2004). However, before the approach can be applied to clinical practice, many questions need to be answered. For example, can the approach induce similar adaptation in individuals with chronic ankle instability? Can long term, repeated adaptation lead to a greater and longer lasting aftereffect that can be used in everyday walking? These questions are outside the scope of the current study, and will be addressed in our future research.

A major limitation of this study was the use of a sandbag to perturb the ankle. The force created by the sandbag was constant throughout the whole gait cycle, and we were not able to target the perturbation around heel contact. This is an alternative explanation why we did not find PL EMG changes around heel contact and why the increased TA EMG around heel contact did not carry over to the post adaptation period. In Blanchette et al. (2011), a robotic device was used to target the force perturbation during early swing, and they were able to induce an aftereffect of increased TA activity in the target time frame.

In addition, we were not able to ensure that the external torque created by the weight was constant across subjects. While the same amount of weight was placed on the 3rd, 4th, and 5th metatarsal bones for all subjects, the resulting external torques may have been somewhat different from one subject to another. For subjects with a smaller foot, we had to move the sandbag closer to the ankle joint, which shortened the moment arm and therefore the resulting torque. Also, the location of weight placement may have been suboptimal for generating sufficient plantarflexion torque, which may explain why we did not find significant

changes in dorsiflexion. The aforementioned limitations can be addressed by using a robotic device to produce force perturbation rather than using a sandbag. To continue this line of research, we have started developing a robotic device that allows us to better control the magnitude, direction, and timing of force perturbation (Murphy et al. 2014).

The ankle eversion changes detected in this study were small in magnitude despite the significance. The internal validity of the study would be weakened if the changes were smaller than measurement error. We made every effort to minimize potential measurement error. For example, the same researcher placed the markers on all subjects and processes all marker data to avoid inter-rater errors. After each testing period, the researcher re-inspected the markers to ensure that they stayed in the same locations to minimize errors in within-subject comparisons.

Based on QTM outputs, the residuals for measured marker positions were mostly below 1 mm and never exceeded 3 mm in all collected walking trials. However, it was unknown how much these minimal residuals affected the results of joint angle calculation, which was a limitation of the study. Using a similar three-degree-of-freedom ankle model and an older version of the Qualisys motion capture system (ProReflex), a previous study examined the reliability of ankle kinematic measures for a treadmill walking task (Collins et al. 2009). The results showed that the measures had high within-session reliability in both sagittal and frontal planes (coefficient of multiple correlation >0.93). While these results provided evidence to support that the measurement error may be low in our study, we could not conclude this until our lab performs our own reliability tests. This study used a one group, repeated measures design to test the hypotheses. This followed how previous studies examined adaptation and after-effects in healthy individuals (e.g., Blanchette and Bouyer 2009; Emken and Reinkensmeyer 2005). However, having a control group adapting to a negligible weight would have strengthened the internal validity of the study.

Conclusion

We tested an error-driven approach to induce increases in ankle eversion, ankle dorsiflexion, and co-activation of TA and PL around heel contact in walking. Error signals were generated in the inversion and plantarflexion directions by placing a light weight to the dorsal–lateral side of the foot. The weight induced an increase in ankle eversion before and after heel contact, and the increase persisted even after the weight was removed. We concluded that the approach is feasible for reducing ankle inversion during walking in healthy persons, and may have the potential to correct abnormal positioning in individuals with chronic ankle instability.

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Conflict of interest We, the authors of this manuscript, affirm that we have no financial and personal relationships with other people or organizations that could inappropriately influence our work.

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