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Muscle Synergies in People With Chronic Ankle Instability During Anticipated and Unanticipated Landing-Cutting Tasks

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Context: Although neuromuscular deficits in people with chronic ankle instability (CAI) have been identified, previous researchers have mostly investigated the activation of multiple muscles in isolation. Investigating muscle synergies in people with CAI would provide information about the coordination and control of neuromuscular activation strategies and could supply important information for understanding and rehabilitating neuromuscular deficits in this population.

Objective: To assess and compare muscle synergies using nonnegative matrix factorization in people with CAI and healthy control individuals as they performed different landing-cutting tasks.

Design: Cross-sectional study.

Setting: Laboratory.

Patients or Other Participants: A total of 11 people with CAI (5 men, 6 women; age = 22 ± 3 years, height = 1.68 ± 0.11 m, mass = 69.0 ± 19.1 kg) and 11 people without CAI serving as a healthy control group (5 men, 6 women; age = 23 ± 4 years, height = 1.74 ± 0.11 m, mass = 66.8 ± 15.5 kg) participated.

Main Outcome Measure(s): Muscle synergies were extracted from electromyography of the lateral gastrocnemius, medial gastrocnemius, fibularis longus, soleus, and tibialis anterior (TA) muscles during anticipated and unanticipated landing-cutting tasks. The number of synergies, activation coefficients, and muscle-specific weighting coefficients were compared between groups and across tasks.

Results: The number of muscle synergies was the same for each group and task. The CAI group exhibited greater TA weighting coefficients in synergy 1 than the control group ($P = .02$). In addition, both groups demonstrated greater fibularis longus ($P = .03$) weighting coefficients in synergy 2 during the unanticipated landing-cutting task than the anticipated landing-cutting task.

Conclusions: These results suggest that, although both groups used neuromuscular control strategies of similar complexity or dimensionality to perform the landing-cutting tasks, the CAI group displayed different muscle-specific weightings characterized by greater emphasis on TA function in synergy 1, which may reflect an effort to increase joint stability to compensate for ankle instability.

Key Words: central nervous system, neuromuscular activation, electromyography

Key Points

- People with chronic ankle instability (CAI) and healthy control individuals used neuromuscular control strategies of similar complexity or dimensionality during dynamic tasks.
- The CAI group emphasized synergy-specific tibialis anterior function during the early- and late-stance phases of unanticipated and anticipated landing-cutting tasks compared with the healthy group.
- Both groups emphasized synergy-specific fibularis longus function during unanticipated versus anticipated landing-cutting tasks.

A sprain of the ligaments on the lateral side of the ankle joint is one of the most common musculoskeletal injuries. Up to 70% of people who sprain their ankles develop chronic ankle instability (CAI) and experience lingering mechanical and functional deficits.[1] Although researchers[2] have evaluated and developed rehabilitation strategies for people with ankle sprains and CAI, the economic burden of these conditions in the United States amounts to 4 to 6 billion dollars in annual health care charges. Furthermore, people with CAI face higher risks of developing more serious clinical sequelae, such as ankle osteoarthritis.[3,4]

Investigators have assessed neuromuscular function in those with CAI and identified numerous deficits,[5] such as less muscle activity in the tibialis anterior (TA), medial gastrocnemius (MG), fibularis longus (FL), and gluteus medius[6]; delayed activation of the FL[7]; and longer activation (ie, duration) of the FL.[8] Despite these characterizations of deficits in neuromuscular function, the authors of conventional electromyography (EMG) studies typically analyze and interpret the activation of each muscle independently. However, the central nervous system (CNS) is believed to produce and coordinate movement by reducing the dimensionality of the activation of many muscles into synergies.[9] Therefore, aside from very basic cocontraction analyses, conventional analyses do not fully capture the way the CNS controls muscular activation during dynamic tasks. Given that fast and dynamic movements are controlled by coordinating multiple muscles together into muscle synergies,[10] it may be beneficial to use a research framework that examines neuromuscular activation patterns in people with CAI during such movements via muscle synergy analysis.

One way to study muscle synergies is via nonnegative matrix factorization (NMF).[9,11,12] Nonnegative matrix factorization analysis of muscle synergies has been used to identify deficits in neuromuscular activation in patients with neurologic conditions, such as stroke or cerebral palsy.[13,14] This analysis provides meaningful insights about CNS control, such as the complexity of an individual's strategy to control the activations of multiple muscles during movement, which is reflected in the number of muscle synergies present during a particular task.[15] Specifically, people who exhibit a smaller number of muscle synergies appear to use a less complex control strategy, as demonstrated by patients with neurologic conditions who appear to control the activation of multiple muscles with fewer or merged versions of the muscle synergies found in healthy people.[15] For example, patients who have had a stroke use fewer muscle synergies to control the lower extremity muscles during walking than healthy control individuals.[15] In addition, fewer muscle synergies also appear to be associated with movement deficits.[15] Similarly, Ambrosini et al[16] reported that patients poststroke increased the number of synergies after rehabilitation (cycling with functional electrical stimulation). Because CAI is also often considered to affect centrally mediated neuromuscular control in a similar way as other neurologic conditions,[17–19] investigating muscle synergies in people with CAI may provide unique insight into their neuromuscular control strategies during dynamic tasks. Therefore, the purpose of our study was to assess and compare muscle synergies using NMF in people with and those without CAI during landing-cutting tasks. We hypothesized that people with CAI would (1) use fewer (ie, less complex) muscle synergies, (2) exhibit different muscle-specific functional roles (ie, weightings) in muscle synergies, and (3) display task-specific differences in muscle synergies.

METHODS

Participants

A total of 11 people with CAI (CAI group; 5 men, 6 women; age = 22 ± 3 years, height = 1.68 ± 0.11 m, mass = 69.0 ± 19.1 kg) and 11 people without CAI serving as a healthy control group (CON group; 5 men, 6 women; age = 23 ± 4 years, height = 1.74 ± 0.11 m, mass = 66.8 ± 15.5 kg) were recruited to participate in the study. Initial screening and inclusion in the CAI group was based on a questionnaire (Modified Ankle Instability Instrument) that established the history and severity of previous ankle sprains.[20,21] Recruits were excluded if they had a history of bilateral ankle sprains, lower extremity fractures or knee injuries, or both. In addition, the Foot and Ankle Disability Index (FADI; CAI group: 90.3 ± 0.4; CON group: 100 ± 0.0) and FADI-Sport (CAI group: 88.6 ± 9.1; CON group: 100 ± 0.0) questionnaires were used to assess ankle joint function,[22] and Tegner scores (CAI group: 5.3 ± 1.2; CON group: 5.3 ± 1.0) were used to quantify the physical activity levels of participants. All individuals provided written informed consent, and the study was approved by the Human Research Protection Program of the University of Michigan.

Procedures

We recorded ground reaction forces (GRFs) using a force platform (AMTI) and muscle activations using a desktop surface EMG system (model Bagnoli; Delsys). Muscle EMG and GRF data were collected at sampling frequencies of 1200 Hz. Five single-differential EMG electrodes (model DE-2.1; Delsys) were attached to the soleus (SL), MG, lateral gastrocnemius (LG), TA, and FL muscles. The bars on the electrodes were 99.9% silver, 10 mm in length, and 1 mm in width and were separated by 10 mm. The skin was cleaned using alcohol swabs, and EMG sensors were placed according to standard recommendations.[23] To limit cross-talk, we determined the final location of the EMG sensors by asking participants to perform isolated muscle contractions. After a brief warm-up, they were instructed to execute up to 5 trials each of an anticipated landing-cutting task (Ant) and an unanticipated landing-cutting task (Unant). For both tasks, they performed a forward jump over a 15-cm box and onto a force plate. The distance between the initial position and the force plate was normalized to each person's *leg length*, which was defined as the distance between the anterior-superior iliac spine and the medial malleolus of the same limb. Individuals were instructed to land on their involved limb and perform a 90° cutaway from their landing limb as quickly as possible after landing on the force plate. For the CON group, the *involved limb* was defined as the dominant limb, which was the limb they used to kick a ball. During Ant, the direction of the cut was given to the participant before each jump. During Unant, the direction of the cut was indicated via an electronic signal displayed on a computer screen, which was positioned at waist height in front of the force plate. The signal was triggered when the person broke a light beam that was projected from a light gate, which was positioned halfway between the initial start position and the force plate. *Successful trials* were defined as those that were correctly performed according to the instructions and cutting direction and in which the foot was placed entirely on the force plate. Data from 1 participant in the CON group were excluded because of problems with the GRF data. In addition, 2 people in the CAI group were unable to perform Unant. Based on these exclusions, we analyzed a total of 123 trials (ie, 22 participants × 2 tasks × 3 trials − 9 bad trials).

Data Processing

Data were analyzed from the stance phase of each task. The stance phase of the landing-cutting tasks was based on GRF thresholds of 10 N for both touchdown and takeoff. The EMG data were low-pass filtered using a cutoff frequency of 450 Hz and high-pass filtered using a cutoff frequency of 20 Hz. The filtered EMG data were rectified and smoothed using a low-pass filter at a cutoff frequency of 10 Hz. The smoothed activation data for each muscle were normalized to the maximum activation observed during all trials and time normalized to 101 data points such that 0% represented touchdown and 100% represented takeoff (Figure 1).[24] Each muscle's activation was further divided by its SD to obtain the unit variance so that NMF could extract equally weighted muscle synergies.[25] The smoothed and normalized muscle activation data from each participant and each task were organized into a 5 × 303 input matrix (ie, 5 rows for all 5 muscles and 303 columns for 3 trials of 101 data points). The NMF algorithm decomposed the 5 × 303 input matrix (E) into the synergy vector (W) and activation coefficient (C):

$$E = \sum_{i=1}^{N_{\mathrm{synergy}}} W_i C_i + \varepsilon, \tag{1}$$

where N_{synergy} is the number of muscle synergies and ε is the residual error.[25]

The NMF algorithm was then applied iteratively with an increasing number of muscle synergies (eg, in the case of 1 synergy, the algorithm would extract 1 set of synergy vectors and 1 activation coefficient). The NMF algorithm and output with 3 synergies, 3 sets of synergy vectors, and 3 activation coefficients are depicted in Figure 2. The total variance accounted for ($\mathrm{VAF_{Total}}$; Equation 2) and variance accounted for by each muscle ($\mathrm{VAF_{Each}}$; Equation 3) were calculated iteratively with continuously larger numbers of synergies until the following criteria were met: (1) $\mathrm{VAF_{Total}} \geq 90\%$ and (2) $\mathrm{VAF_{Each}} \geq 75\%$.[25,26] In most datasets, 3 synergies were the appropriate number of muscle synergies for the criteria. In addition, the VAF values indicated the extent to which the EMG profile of all muscles ($\mathrm{VAF_{Total}}$) and each individual muscle ($\mathrm{VAF_{Each}}$) could be explained by the respective number of synergies. Muscle synergies were sorted based on timing of the peak activation coefficient and cosine similarity of synergy vectors[27]:

$$\mathrm{VAF_{Total}} = 1 - \frac{\sum_{i=1}^{p}\sum_{j=1}^{n}\left(\varepsilon_{i,j}\right)^2}{\sum_{i=1}^{p}\sum_{j=1}^{n}\left(E_{i,j}\right)^2}, \tag{2}$$

where p is the number of muscles and n is the number of time points, and

$$\mathrm{VAF_{Each_m}} = 1 - \frac{\sum_{j=1}^{n}\left(\varepsilon_{m,j}\right)^2}{\sum_{j=1}^{n}\left(E_{m,j}\right)^2}, \tag{3}$$

where m is each muscle.

Statistical Analysis

The independent variables were group (CAI and CON) and task (Ant and Unant). The dependent variables were the number of muscle synergies, the $\mathrm{VAF_{Total}}$ of each synergy, and the muscle-specific weightings from each of the extracted synergies. In addition, we used the cosine-similarity values and zero-lag cross-correlation coefficients to assess whether muscle synergy data from each group and

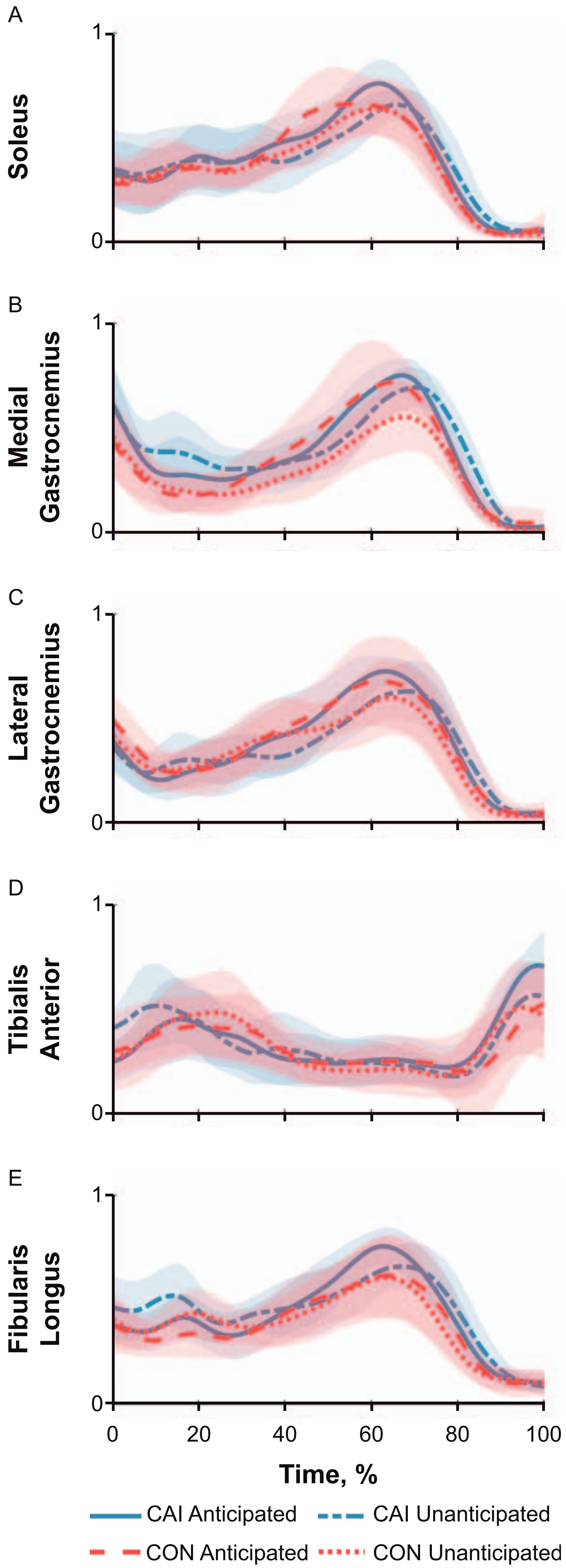

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Figure 1. Normalized muscle activity (mean ± SD) in people with chronic ankle instability (CAI) and healthy control individuals (CON) during anticipated and unanticipated landing-cutting tasks.

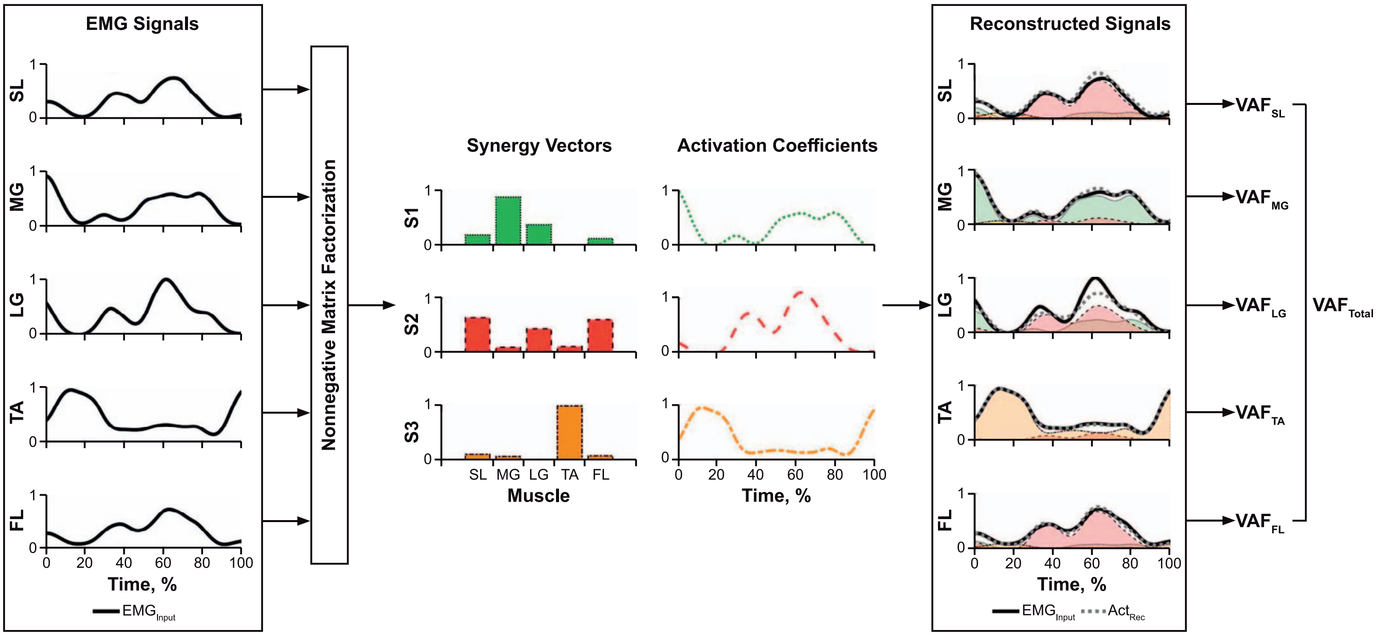

Figure 2. Nonnegative matrix factorization workflow. Abbreviations: EMG, electromyography; FL, fibularis longus; LG, lateral gastrocnemius; MG, medial gastrocnemius; S1, synergy 1; S2, synergy 2; S3, synergy 3; SL, soleus; TA, tibialis anterior; VAF, variance accounted for.

task were similar (ie, CON Ant versus Unant, CAI Ant versus Unant, CON versus CAI Ant, CON versus CAI Unant) and thus appropriate for statistical comparison. Specifically, the zero-lag cross-correlation coefficients were used to evaluate the similarity of each synergy's activation coefficient and ensure that the activation pattern of each synergy (ie, time-variant components) captured similar bursts of muscle activity. Similarly, we used the cosine-similarity values to determine the similarity of each synergy vector (ie, weighting coefficients) and ensure that the muscles that were active in each synergy (ie, time-invariant components) were the same. Cross-correlation and cosine-similarity values >0.80 were considered to exhibit high similarity.[27] Normality of all dependent variables was checked using the Jarque-Bera test.[28] The Wilcoxon rank sum test was conducted to compare the number of synergies between the CAI and CON groups separately for each task. Separate 2 × 2 analyses of variance were calculated to compare the VAF$_{\text{Total}}$ and muscle-specific weightings for each of the extracted synergies between groups (CAI and CON) and across tasks (Ant and Unant). We used MATLAB (version 2019a; The MathWorks, Inc) and set the α level at .05.

RESULTS

Dimensionality of Muscle Synergies

Based on the VAF results, we identified 2 to 4 synergies as the appropriate number to represent the EMG data of each trial. In most cases, 3 synergies were sufficient to reconstruct the EMG data (Figure 3), and the average VAF$_{\text{Total}}$ for 3 synergies was approximately 93%. We observed no difference in the number of synergies expressed by the CAI and CON groups during either task (CON Unant: 3.1 ± 0.6; CAI Unant: 2.8 ± 0.4; CON Ant: 3.1 ± 0.6; CAI Ant: 3.1 ± 0.5; Figure 4). In addition, no interaction or main effects were found for VAF$_{\text{Total}}$ with 1 synergy (CON Unant: 46.6% ± 15.9%; CAI Unant: 51.0% ± 6.9%; CON Ant: 55.4% ± 13.7%; CAI Ant: 51.2% ± 7.6%), 2 synergies (CON Unant: 81.2% ± 9.3%; CAI Unant: 83.5% ± 4.6%; CON Ant: 84.6% ± 7.1%; CAI Ant: 85.4% ± 5.1%), 3 synergies (CON Unant: 92.0% ± 3.6%; CAI Unant: 93.3% ± 3.2%; CON Ant: 94.0% ± 2.1%; CAI Ant: 94.1% ± 2.7%), or 4 synergies (CON Unant: 97.4% ± 1.3%; CAI Unant: 97.8% ± 1.3%; CON Ant: 97.8% ± 0.9%; CAI Ant: 97.9% ± 1.2%; Figure 5).

Similarity of Muscle Synergies

The cosine-similarity values for the synergy vectors of synergies 1 and 2 were >0.80 for all group and task comparisons (Table 1). In contrast, the cosine-similarity values for the synergy vectors of synergy 3 were <0.80 for 3 of 4 group comparisons (Table 1). Specifically, the cosine-similarity value for the synergy vectors of the comparison between CON and CAI for Unant was 0.85. The zero-lag cross-correlation coefficients of all activation coefficients from all respective group and task comparisons were >0.80 (Table 2).

Functional Interpretation of Muscle Synergies

Muscles were considered part of a specific synergy (ie, activated or recruited by that synergy) if their weighting coefficients were >0.3.[29] The activation coefficient of synergy 1 reflected muscle activation during the early (approximately 10% to 20%)- and late (approximately 80% to 100%)-stance phases of the landing-cutting tasks. The muscle-specific weightings in the synergy vector of synergy 1 reflected mainly TA activation. Therefore, it seems that synergy 1 functions to control the ankle angle at ground contact and during late stance. The activation coefficient of synergy 2 captured muscle activation during the middle-

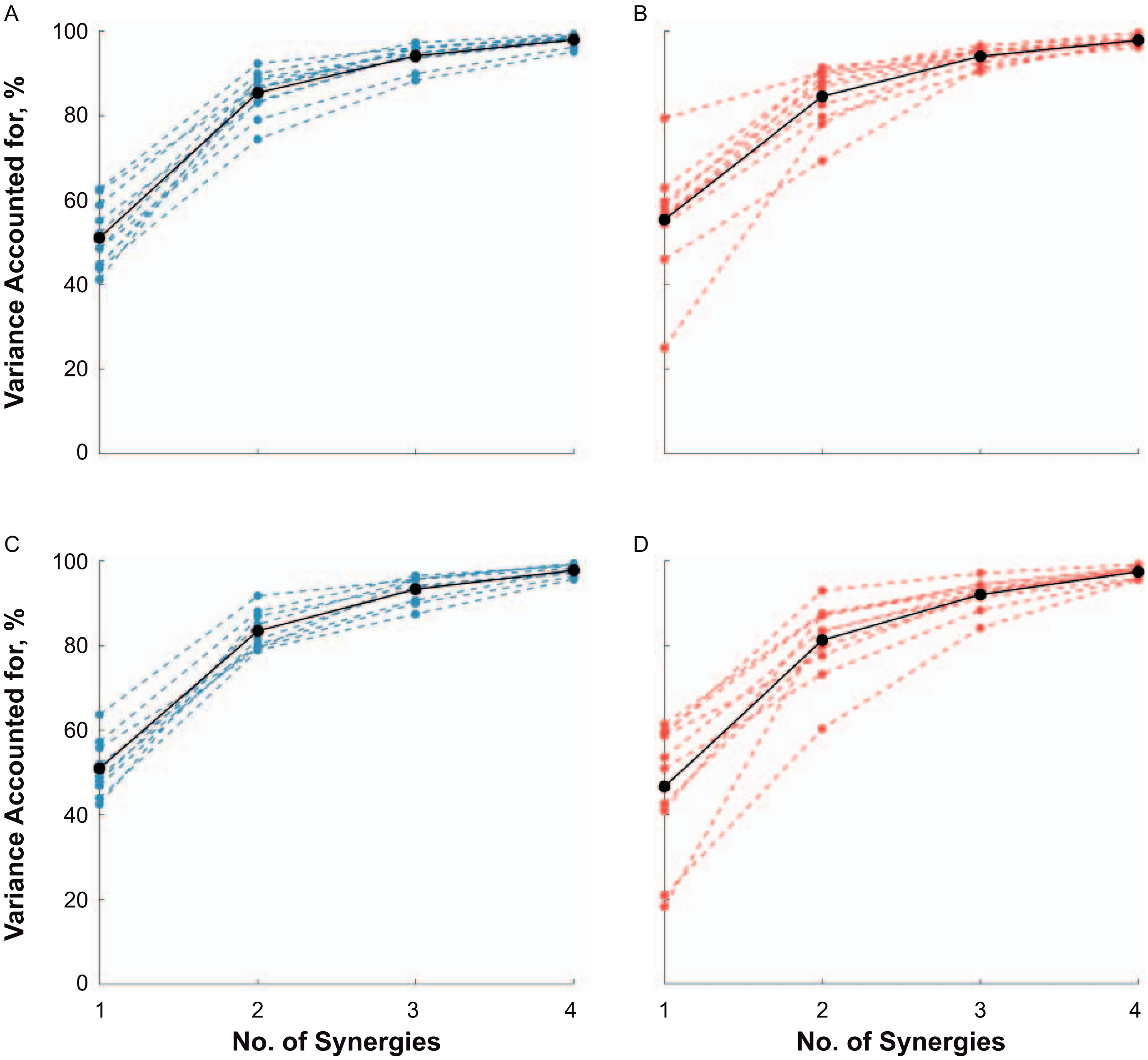

Figure 3. Variance accounted for the synergies for, A, the chronic ankle instability group, B, the healthy control group during the anticipated landing-cutting task, C, the chronic ankle instability group, and D, the healthy control group during the unanticipated landing-cutting task. The black line indicates the average.

stance phase (approximately 30% to 80%) of the landing-cutting tasks, and the muscle-specific weightings in the synergy vector associated with this synergy indicated that it reflected primarily FL activation. Given that the FL acts mainly to evert the ankle in the frontal plane, the function of synergy 2 thus seems to be related to the transition from forward to lateral motion during the midportion of the landing-cutting tasks. Similarly, the activation coefficient of synergy 3 captured muscle activation in the middle-stance phase of the landing-cutting tasks. Unlike synergy 2, however, the muscle-specific weightings in the vector for synergy 3 were associated with activation of the SL, MG, and LG. Hence, synergy 3 seems to play a role in propulsion and helps accelerate the body toward the new cutting direction.

Group and Task Differences in Muscle Synergies

We observed no group × task interactions for any of the individual muscle weightings for any of the synergy vectors and synergies. However, main effects existed for group in the weightings for the SL and TA in the synergy vector of synergy 1 (Figures 6 and 7). Specifically, the task-averaged weightings of the TA were larger ($P = .02$) in the CAI than the CON group, whereas the task-averaged weightings of the SL were smaller ($P = .03$) in the CAI than the CON group. We also noted main effects for task in the individual muscle weightings of the MG and FL in the synergy vector of synergy 2 (Figures 6 and 7). The group-averaged weightings of MG were larger ($P = .03$) during Ant than Unant, and the group-averaged weightings of FL were smaller ($P = .03$) during Ant than Unant.

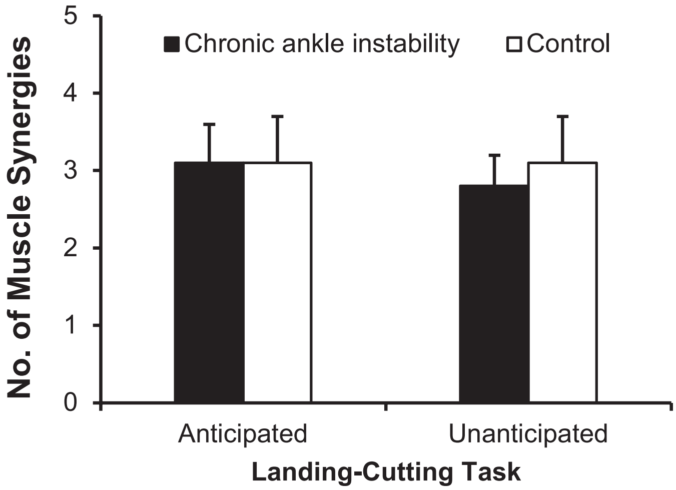

Figure 4. The number of muscle synergies for each group during each landing-cutting task.

DISCUSSION

The purpose of our study was to use NMF and extract muscle synergies to investigate and compare neuromuscular control strategies in people with and those without CAI during landing-cutting tasks. To our knowledge, we are the first to investigate muscle synergies during dynamic tasks in individuals with CAI. In addition, our novel findings provide a reference for the future evaluation of muscle synergy in people with CAI. The results showed no difference in the dimensionality of muscle synergies between the CON and CAI groups during Ant and Unant. Whereas the first 2 muscle synergies were similar for both groups and tasks, a third synergy accounted for individual differences in both groups and tasks. The CAI group exhibited greater TA weightings and smaller SL weightings in synergy 1 than the CON group. Both groups displayed smaller MG weightings and greater FL weightings in synergy 2 during Unant than Ant. Together, these results partially supported our hypothesis that people with CAI would demonstrate different weightings in specific muscle synergies, but this difference did not depend on the task. Conversely, the results did not support our hypothesis that people with CAI would use a less complex neuromuscular control strategy than the CON group. Therefore, our findings suggest that those with CAI used different neuromuscular control strategies during dynamic tasks. Future researchers who focus on muscle synergy analysis may provide novel insights about neuromuscular deficits in people with CAI.

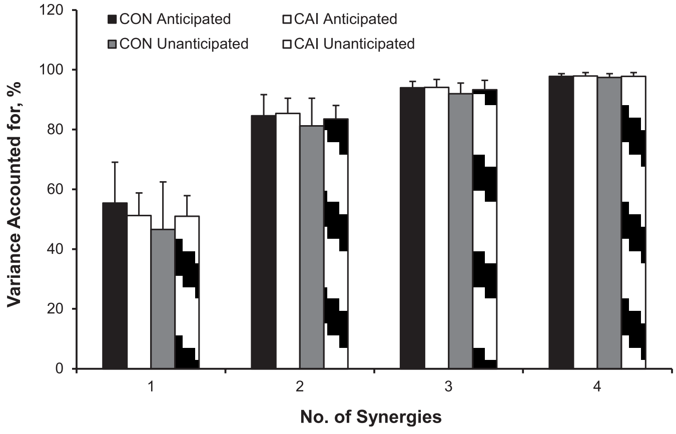

Figure 5. Variance (mean ± SD) accounted for the synergies for each group during landing-cutting tasks. Abbreviations: CAI, chronic ankle instability group; CON, healthy control group.

Table 1. Cosine-Similarity Coefficients for Each Synergy Vector Comparison (Mean ± SD)

	Synergy		
Comparison	1	2	3
CON Ant versus Unant	0.98 ± 0.02	0.82 ± 0.13	0.78 ± 0.17
CAI Ant versus Unant	0.99 ± 0.01	0.83 ± 0.12	0.78 ± 0.20
CON versus CAI Ant	0.98 ± 0.02	0.81 ± 0.12	0.72 ± 0.22

Abbreviations: Ant, anticipated landing-cutting task; CAI, chronic ankle instability group; CON, healthy control group; Unant, unanticipated landing-cutting task.

Dimensionality of Muscle Synergies

The dimensionality of muscle synergies did not differ between the CAI and CON groups. This finding did not agree with our hypothesis that people with CAI would exhibit fewer synergies or that each muscle synergy would exhibit a greater $\text{VAF}_{\text{Total}}$. These hypotheses were based on the work of researchers[15] who suggested that people with certain neurologic conditions (eg, stroke, cerebral palsy) exhibit fewer muscle synergies and simpler neuromuscular control strategies. Fewer muscle synergies, as observed in people with neurologic conditions, may be due to greater cocontractions and result in less efficient movement.[15,30] However, given that people in the CAI group exhibited the same number of muscle synergies during both landing-cutting tasks, it appears that they used a similar synergy-based neuromuscular control strategy during landing-cutting tasks regardless of the associated cognitive load.

Different Weightings of Muscle Synergies

Analysis of the individual muscle weightings in each synergy reflected both group and task main effects. Specifically, in synergy 1, the weighting of the TA muscle was greater for the CAI than the CON group, which indicated that people with CAI recruited the TA muscle to a greater extent than people in the CON group. Given that the activation coefficient of synergy 1 captured muscle activity during the early phase of stance during the landing-cutting tasks, individuals with CAI may emphasize sagittal-plane positioning of the ankle around touchdown to a greater extent, which is clinically important because ankle

Table 2. Zero-Lag Cross-Correlation Coefficients for Each Activation Coefficient Comparison (Mean ± SD)

	Synergy		
Comparison	1	2	3
CON Ant versus Unant	0.85 ± 0.11	0.88 ± 0.10	0.88 ± 0.08
CAI Ant versus Unant	0.87 ± 0.07	0.91 ± 0.06	0.89 ± 0.06
CON versus CAI Ant	0.84 ± 0.10	0.90 ± 0.09	0.89 ± 0.07
CON versus CAI Unant	0.86 ± 0.08	0.90 ± 0.06	0.90 ± 0.06

Abbreviations: Ant, anticipated landing-cutting task; CAI, chronic ankle instability group; CON, healthy control group; Unant, unanticipated landing-cutting task.

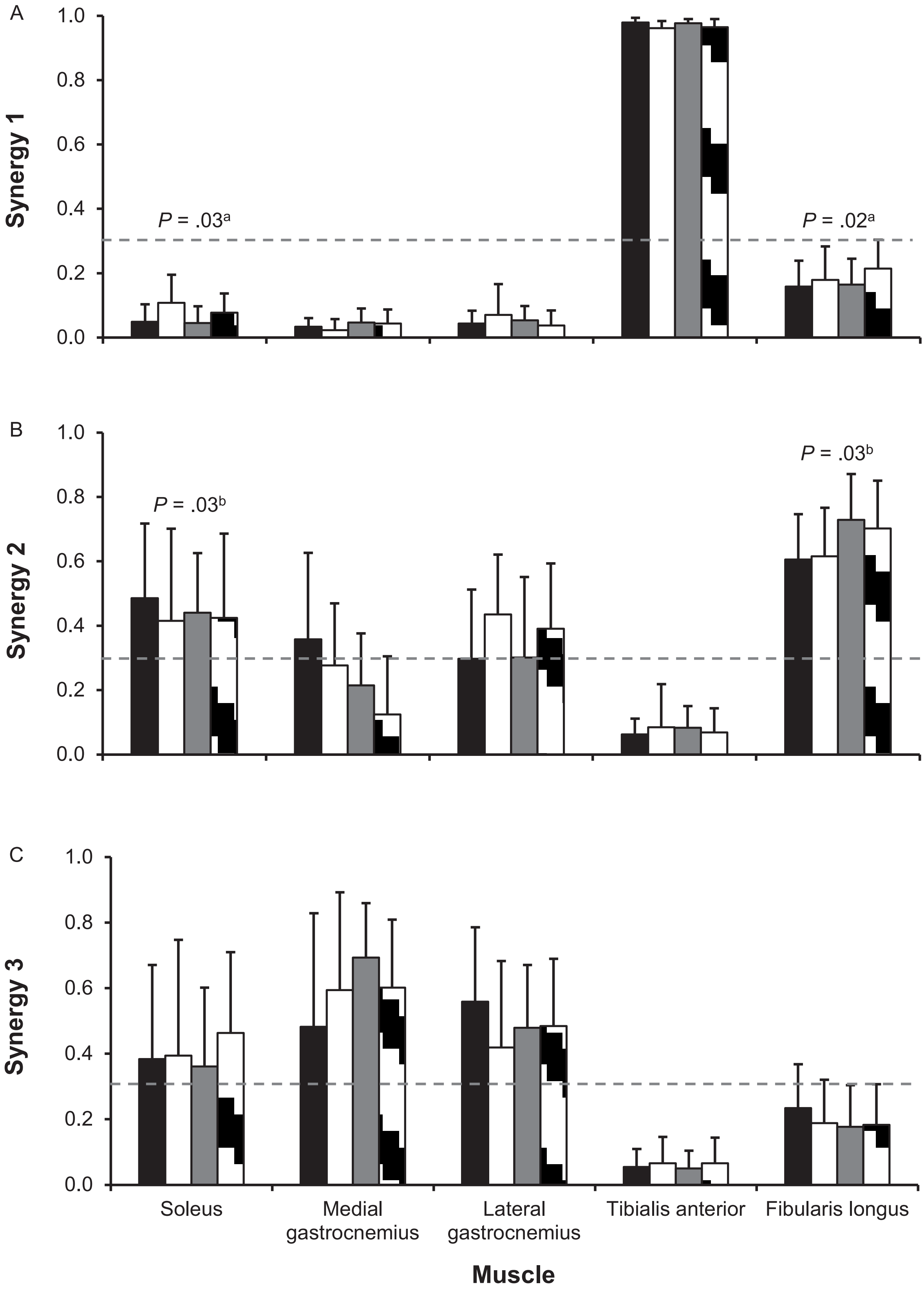

Figure 6. Muscle synergy vectors (and muscle-specific weightings) extracted from each group during anticipated and unanticipated landing-cutting tasks. A, Synergy 1. B, Synergy 2. C, Synergy 3. [a] Main effect for group. [b] Main effect for task. Abbreviations: CAI, chronic ankle instability group; CON, healthy control group.

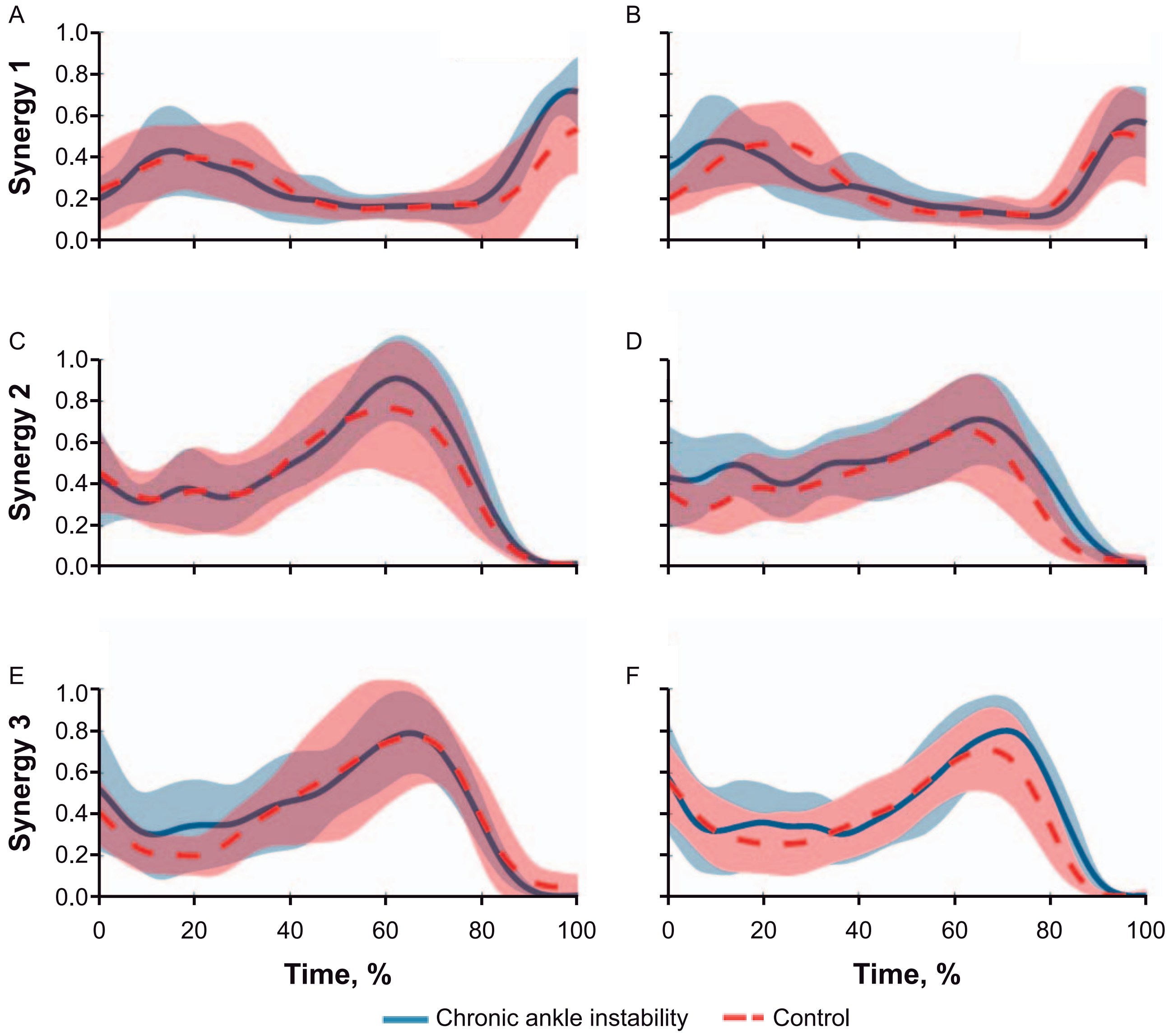

Figure 7. Activation coefficients extracted from each group during each task. A, Synergy 1 anticipated landing-cutting task. B, Synergy 1 unanticipated landing-cutting task. C, Synergy 2 anticipated landing-cutting task. D, Synergy 2 unanticipated landing-cutting task. E, Synergy 3 anticipated landing-cutting task. F, Synergy 3 unanticipated landing-cutting task.

positioning affects foot-ground clearance and mitigates the risk of unanticipated contact.[31,32] In addition, computer simulations have suggested that greater ankle dorsiflexion at the instant of foot contact is associated with a smaller external moment arm of GRFs about the subtalar joint, which could lessen the risk of subsequent ankle sprains.[33] Moreover, a dorsiflexed position increases stability of the ankle joint because the surfaces of the ankle joint become more congruent as dorsiflexion increases. People with CAI are thought to compensate for their lack of stability by dorsiflexing the ankle joint to achieve a more close-packed and stable position.[6] Lastly, greater emphasis on TA activation during the early stance phase may also enhance coactivation and ankle-joint stability. For example, using a musculoskeletal model, DeMers et al[34] showed that increased cocontraction of the ankle invertors and evertors might improve ankle joint stability and decrease ankle injury risk. However, the interpretation related to the greater weighting of the TA by the CAI group could be deemed less protective with respect to the control of frontal-plane motion because the TA also inverts the foot at the ankle. This interpretation therefore provides an alternative explanation of the results and may indicate that the CAI group also exhibited altered frontal-plane control. Thus, it may be necessary to investigate ankle joint kinematics and kinetics in future studies to support or refute this interpretation. Although group differences existed in the weighting of the SL muscle for synergy 1, the magnitudes of this weighting were very small (ie, well below the 0.3 threshold) and hence may not reflect clinically or functionally important differences in muscle coordination between the CON and CAI groups.[29]

In addition to the group difference between CON and CAI in individual muscle weightings for synergy 1, we found a task difference between Ant and Unant for synergy 2. Specifically, both groups displayed greater weighting of the FL muscle and less weighting of the MG muscle during Unant than Ant. Based on the timing of muscle activity, the activation coefficient for synergy 2 suggests that people used this synergy to control muscle activation during the middle phase of stance during landing-cutting tasks. The increase in FL weighting, therefore, may reflect greater emphasis on transitioning from forward to lateral motion via greater activation of frontal-plane muscles. Furthermore, greater FL weighting may also suggest an attempt to maintain balance and ankle joint stability in the frontal plane when the landing-cutting task is performed with more uncertainty and without knowing the direction of movement.[35] Interestingly, the increase in FL weighting was accompanied by a decrease in MG weighting; although based on the thresholds, the changes in MG weighting in synergy 2 indicate that the MG muscle would be considered active during Ant but not during Unant. Because the MG muscle is an important contributor to propulsive forces during landing-cutting tasks,[36] this change in neuromuscular control may mean that both groups emphasized frontal-plane stability over propulsion during Unant.

Limitations and Future Directions

Our study had several limitations. First, the sample recruited was relatively small (only 22 participants), which may have affected the results, and the procedures may warrant replication in future studies with a larger sample. Second, we recorded EMG from only 5 lower limb muscles. Using a small number of muscles in the NMF methods may have led to an overestimation of the VAF by the muscle synergies, which may in turn have resulted in the extraction of fewer muscle synergies and subsequently affected the interpretation of their dimensionality.[37] However, the minimum number of muscles for appropriate implementation of the NMF methods is considered 4.[37] Moreover, the 5 muscles in our study captured the major kinesiologic functions of the ankle joint (ie, plantar flexion, dorsiflexion, inversion, and eversion) and thus likely still adequately represented neuromuscular strategies of ankle motions from a muscle synergy-based control perspective. Considering that researchers[38] have reported that people with CAI exhibited functional deficits in the proximal muscles, it would be valuable to include these muscles (eg, gluteus maximus) in future studies to better characterize and more comprehensively understand the functional role of muscle synergies in people with CAI. Third, the muscles that were investigated are located relatively close to each other, which may have led to cross-talk and affected the EMG signal. Using indwelling electrodes in future research may help reduce the potential for cross-talk via more direct measurement of a muscle's activation. Fourth, the muscle activation magnitude was normalized based on the maximum value observed across all trials and tasks. Whereas the use of maximal voluntary isometric contractions represents another normalization method, the maximum observed value is commonly used in NMF research[29,39] and may be acceptable in our study because dynamic landing and cutting tasks impose high task demands that may actually elicit near-maximum muscle activation levels.[40] Similarly, normalizing the EMG to values obtained from a quasistatic position (eg, a standing trial) may be another option that is less sensitive to group differences in activation than values obtained during dynamic movements. Fifth, we analyzed only landing-cutting tasks in a laboratory setting. Given that ankle sprains also occur during other movements (eg, walking) and in other environments (eg, uneven or inclined surfaces), assessing muscle synergies across a variety of tasks and conditions may provide additional information that could hold important clinical implications.

CONCLUSIONS

Across the various Ant and Unant, people with CAI used global neuromuscular control strategies that were like those of the healthy control group. However, regardless of the landing-cutting task, individuals with CAI exhibited slight differences in how they recruited their ankle-dorsiflexor muscles. Specifically, people with CAI relied on greater TA weighting in the synergy that controlled muscle activation during the early- and late-stance phases of landing-cutting tasks. These findings suggest that, although those with CAI displayed similar complexities of muscle synergy during dynamic tasks, they also used neuromuscular control strategies in a muscle-specific manner that is consistent with increasing joint stability and mitigating the risks of reinjury. Still, recurrent ankle sprains are multifactorial, and it is difficult to explain the high rates of injury based solely on isolated findings from our study.

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