Journal of Athletic Training 2015;50(7):688–696
doi: 10.4085/1062-6050-50.2.02

www.natajournals.org

Muscle-Activation Onset Times With Shoes and Foot Orthoses in Participants With Chronic Ankle Instability

Bart Dingenen, MSc, PT*; Louis Peeraer, PhD, PT*†; Kevin Deschamps, PhD, Pod*; Steffen Fieuws, PhD‡; Luc Janssens, Eng*§; Filip Staes, PhD, PT*

*Musculoskeletal Rehabilitation Research Group, Department of Rehabilitation Sciences; ‡Interuniversity Institute for Biostatistics and Statistical Bioinformatics; and §Department of Electrical Engineering, Faculty of Engineering Technology Services, KU Leuven, Belgium; †Thomas More Kempen University College, Mobilab, Geel, Belgium

Context: Participants with chronic ankle instability (CAI) use an altered neuromuscular strategy to shift weight from double-legged to single-legged stance. Shoes and foot orthoses may influence these muscle-activation patterns.

Objective: To evaluate the influence of shoes and foot orthoses on onset times of lower extremity muscle activity in participants with CAI during the transition from double-legged to single-legged stance.

Design: Cross-sectional study.

Setting: Musculoskeletal laboratory.

Patients or Other Participants: A total of 15 people (9 men, 6 women; age = 21.8 ± 3.0 years, height = 177.7 ± 9.6 cm, mass = 72.0 ± 14.6 kg) who had CAI and wore foot orthoses were recruited.

Intervention(s): A transition task from double-legged to single-legged stance was performed with eyes open and with eyes closed. Both limbs were tested in 4 experimental conditions: (1) barefoot (BF), (2) shoes only, (3) shoes with standard foot orthoses, and (4) shoes with custom foot orthoses (SCFO).

Main Outcome Measure(s): The onset of activity of 9 lower extremity muscles was recorded using surface electromyography and a single force plate.

Results: Based on a full-factorial (condition, region, limb, vision) linear model for repeated measures, we found a condition effect ($F_{3,91.8} = 9.39$, $P < .001$). Differences among experimental conditions did not depend on limb or vision condition. Based on a 2-way (condition, muscle) linear model within each region (ankle, knee, hip), earlier muscle-activation onset times were observed in the SCFO than in the BF condition for the peroneus longus ($P < .001$), tibialis anterior ($P = .003$), vastus medialis obliquus ($P = .04$), and vastus lateralis ($P = .005$). Furthermore, the peroneus longus was activated earlier in the shoes-only ($P = .02$) and shoes-with-standard-foot-orthoses ($P = .03$) conditions than in the BF condition. No differences were observed for the hip muscles.

Conclusions: Earlier onset of muscle activity was most apparent in the SCFO condition for ankle and knee muscles but not for hip muscles during the transition from double-legged to single-legged stance. These findings might help clinicians understand how shoes and foot orthoses can influence neuromuscular control in participants with CAI.

Key Words: footwear, insoles, ankle sprains, neuromuscular system, electromyography

Key Points

- Shoes and foot orthoses accelerated muscle-activation onset times of the ankle and knee but not the hip in participants with chronic ankle instability.
- Earlier muscle-activation onset times were most prominent in the shoes-with-custom-foot-orthoses condition.
- At the ankle, the muscle-activation onset time of the peroneus longus was earlier in the shoes-only, shoes-with-standard-foot-orthoses, and shoes-with-custom-foot-orthoses conditions than in the barefoot condition, and the muscle-activation onset time of the tibialis anterior was earlier in the shoes-with-custom-foot-orthoses condition than in the barefoot condition.
- At the knee, the muscle-activation onset times of the vastus medialis obliquus and vastus lateralis were earlier in the shoes-with-custom-foot-orthoses condition than in the barefoot condition.
- The results may help clinicians understand how shoes and foot orthoses can influence neuromuscular control of the lower extremity in participants with chronic ankle instability.

Lateral ankle sprains are estimated to account for approximately 15% of all sport injuries.[1] Even more concerning than the initial ankle sprain is the large proportion of patients with residual symptoms and recurrent ankle sprains for months to years after the initial injury.[2] The occurrence of repetitive ankle sprains and the feeling of the ankle "giving way" with slight or no perturbation has been defined as *chronic ankle instability* (CAI).[3]

The transition task from double-legged to single-legged stance during barefoot (BF) conditions has been shown to discriminate between uninjured participants and participants with CAI. Researchers have reported that muscle-activation onset times typically were delayed[4,5] and postural stability was impaired[6] in participants with CAI, indicating the use of another strategy to shift weight from double-legged to single-legged stance. However, it is unclear whether findings from BF tests represent typical daily situations when shoes, and for some persons foot orthoses, are often worn.

The human foot is the first point of contact between the body and a supporting surface. The cutaneous mechanoreceptors on the planar surface of the foot are an important source of sensory information,[7] which is considered essential for achieving and maintaining functional joint stability.[8] Shoes and foot orthoses act as an interface between the body and a supporting surface and can influence the sensory feedback from these mechanoreceptors by increasing the contact area between the foot and the supporting surface.[7,9] Furthermore, the small kinematic alterations of the rear foot and tibia that have been described with the use of shoes and foot orthoses[10] may put the ankle joint in a more neutral position, thereby improving the capacity of the ankle mechanoreceptors to provide more accurate proprioceptive input toward the central nervous system.[11] Changing the sensory input to these mechanisms consequently would change the motor output.[7]

Evidence is increasing that shoes and foot orthoses can influence lower extremity muscle activation.[10,12–14] Dingenen et al[14] were the first investigators to measure the influence of shoes and foot orthoses on muscle-activation onset times of the entire lower extremity in uninjured participants during the transition from double-legged to single-legged stance. Their results showed that shoes and foot orthoses can accelerate muscle-activation onset times of the peroneus longus. No differences were reported in more proximal muscles. Recently, researchers have suggested that future investigators should be focused on the influence of shoes and foot orthoses on neuromuscular control, especially in participants with injuries, such as CAI,[10,13,14] to increase our understanding of how positive clinical outcomes from the use of shoes and foot orthoses can be achieved.[11] Altering or improving proprioceptive information and muscle-activation patterns in participants with CAI would be clinically beneficial, given that their proprioceptive and neuromuscular deficits have been described.[15]

To our knowledge, no investigators have focused on the influence of shoes and foot orthoses on muscle-activation onset times of the entire lower extremity in participants with CAI during the transition from double-legged to single-legged stance. Therefore, the purpose of our study was to evaluate the influence of shoes and foot orthoses on muscle-activation onset times during the transition from double-legged to single-legged stance in participants with CAI. Based on the proposed effects of shoes and foot orthoses on lower extremity neuromuscular control, we hypothesized that shoes and foot orthoses would accelerate muscle-activation onset times compared with a BF condition.

METHODS

Participants

We selected 15 participants (9 men, 6 women) with CAI from a population of university students of the Faculty of Kinesiology and Rehabilitation Sciences of KU Leuven (Table 1). A self-report questionnaire was used to determine whether volunteers met the criteria to participate in the study.[5] We included men and women who were between the ages of 18 and 45 years; had worn foot orthoses for at least 6 weeks; had a history of at least 2 lateral ankle sprains of the same ankle in the 2 years before the study; and reported a subjective feeling of *giving way*, defined as "the regular occurrence of uncontrolled and unpredictable episodes of excessive inversion of the rear foot, which do not result in acute lateral ankle sprain,"[16] or a *feeling of ankle-joint instability*, defined as "the situation whereby during activities of daily living and sporting activities the subject feels that the ankle joint is unstable, usually associated with a fear of sustaining an acute ligament sprain."[16] Exclusion criteria were a history of surgery to the musculoskeletal structures of either the lower extremity or back, a history of fracture in either lower extremity, or an acute injury to the musculoskeletal structures of other joints of the lower extremity in the 3 months before the study that affected joint integrity and function and that resulted in at least 1 interrupted day of desired physical activity.[17] We also excluded volunteers with the following conditions: Parkinson disease, multiple sclerosis, cerebrovascular accident, peripheral neuropathy, circulatory disorder, or serious joint disorders (eg, rheumatoid arthritis, osteoarthritis).[14]

Table 1. Characteristics of Participants and Foot Orthoses

Characteristic	Mean ± SD
Age, y	21.8 ± 3.0
Height, cm	177.7 ± 9.6
Mass, kg	72.0 ± 14.6
Foot length, cm	25.7 ± 1.8
Navicular drop of more affected limb, mm	4.9 ± 3.3
Navicular drop of less affected limb, mm	5.5 ± 2.7
Correction of navicular drop of more affected limb with standard foot orthoses, mm	3.7 ± 3.0
Correction of navicular drop of more affected limb with custom foot orthoses, mm	3.3 ± 3.1
Correction of navicular drop of less affected limb with standard foot orthoses, mm	4.4 ± 2.3
Correction of navicular drop of less affected limb with custom foot orthoses, mm	3.9 ± 2.3
Hardness of standard foot orthoses, Shore A	60.0 ± 0.0
Hardness of custom foot orthoses, Shore A	40.0 ± 14.1
Comfort of standard foot orthoses, visual analog scale score, mm	65.6 ± 10.6
Comfort of custom foot orthoses, visual analog scale score, mm	77.8 ± 13.2

Thirteen participants were right-limb dominant, and 2 participants were left-limb dominant. Eight participants reported bilateral CAI. Three of these 8 participants could define 1 limb as being more unstable than the other. In participants with bilateral CAI, the self-reported more unstable limb was considered to be the *more affected limb*. The *dominant limb*, defined as the preferred limb to kick a ball, was identified as the *more affected limb* when the more unstable limb could not be identified. Participants had worn the custom foot orthoses for 35.7 ± 21.3 months. Ten participants always used custom foot orthoses, 4 participants used them only during sports activities, and 1 participant reported using them sometimes. Eleven participants started using foot orthoses due to foot or ankle problems; 1 participant, due to lower back pain; and 3 participants, due to a combination of lower extremity problems.

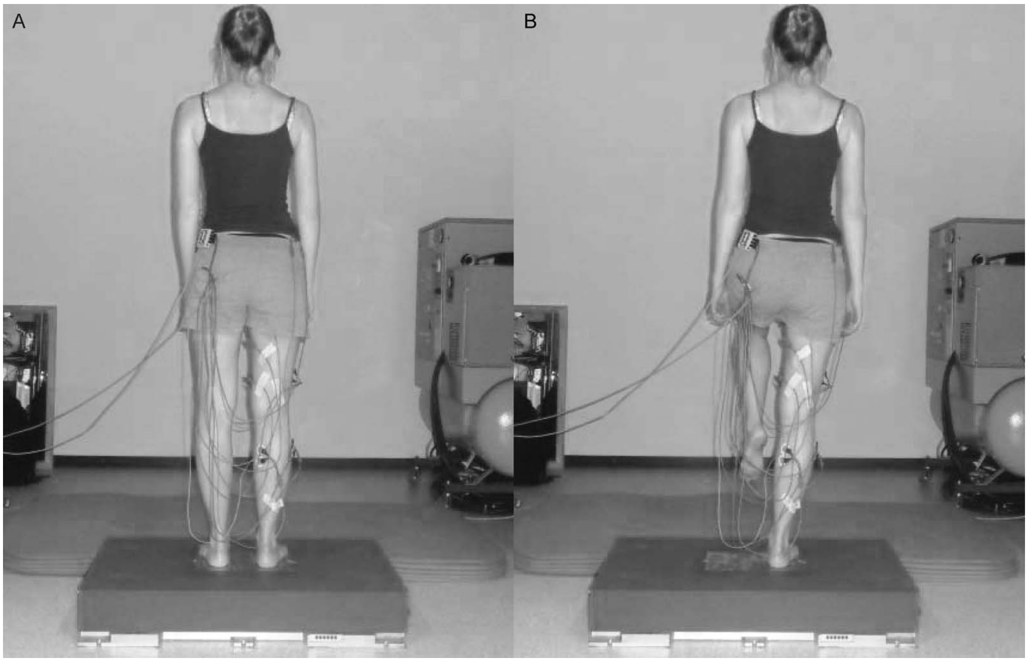

Figure 1. Experimental set-up. Surface electromyography and force-plate data were measured during the transition from A, double-legged stance to B, single-legged stance.

All participants provided written informed consent, and the study was approved by the Commissie Medische Ethiek van de Universitaire Ziekenhuizen KU Leuven.

Data Collection and Procedures

Data collection and procedures were identical to those in the study of Dingenen et al.[14] The transition task from double-legged to single-legged stance that the participants performed is illustrated in Figure 1. This task has been used to study muscle-activation onset times after injury, including CAI,[4–6] and to investigate the influence of shoes and foot orthoses on these onset times in uninjured participants.[14]

Data Analysis

The data analysis was identical to that in the study of Dingenen et al.[14] To avoid errors, we compared the muscle-activation onset time determined by the algorithm with the muscle-activation onset time identified visually.[18] In most cases, we did not need to change this automatically determined muscle-activation onset time. However, in some cases, such as an increase in muscle activity not related to the transitional movement (possibly an artifact) or in muscles where baseline activity is increased during double-legged stance, the algorithm may place the muscle-activation onset time too early or too late compared with the visual judgment, in which the muscle-activation onset time is determined based on the earliest visual rise in electromyography (EMG) activity beyond the steady state during double-legged stance.[18]

Statistical Analysis

We used a *t* test to compare the hardness and navicular-drop correction between the standard and custom foot orthoses and the navicular drop between limbs. Comfort scores of standard and custom foot orthoses were compared with the Wilcoxon signed rank test. Muscles were grouped according to their regions: ankle (gastrocnemius, peroneus longus, tibialis anterior), knee (vastus medialis obliquus, vastus lateralis), and hip (adductor longus, tensor fasciae latae, gluteus medius, gluteus maximus). The differences in muscle-activation onset times as a function of condition (4 levels), region (3 levels), limb (2 levels), and vision (2 levels) were evaluated using a linear model for repeated measures. Within each region, a 2-way linear model for repeated measures was used to evaluate the differences in muscle-activation onset times as a function of condition (4 levels) and muscle (ankle: 3 levels; knee: 2 levels; hip: 4 levels). A 3-way linear model for repeated measures was used to evaluate the interactions with the factors of limb (more affected, less affected) and vision (eyes open, eyes closed). To evaluate the difference among conditions within each muscle, a post hoc analysis was conducted. In all models, we relaxed the strict assumption of the classic repeated-measures analysis of variance, using a larger number of variables to describe the covariance matrix.[19] The model *F* tests were based on the Kenward-Roger adjusted degrees-of-freedom solution, an approach specifically proposed for small-sample settings. We used Tukey adjustments for multiple comparisons within each model. In the analysis of the post hoc results for each muscle, these adjustments were made only for the 6 pairwise comparisons among the experimental conditions. The α level was set at .05. All analyses were performed using SAS System for Windows (version 9.2; SAS Institute Inc, Cary, NC).

RESULTS

Participants and Foot Orthoses Characteristics

The navicular drop was not different between the more affected and less affected limbs ($t_{14} = -0.968$, $P = .35$). The

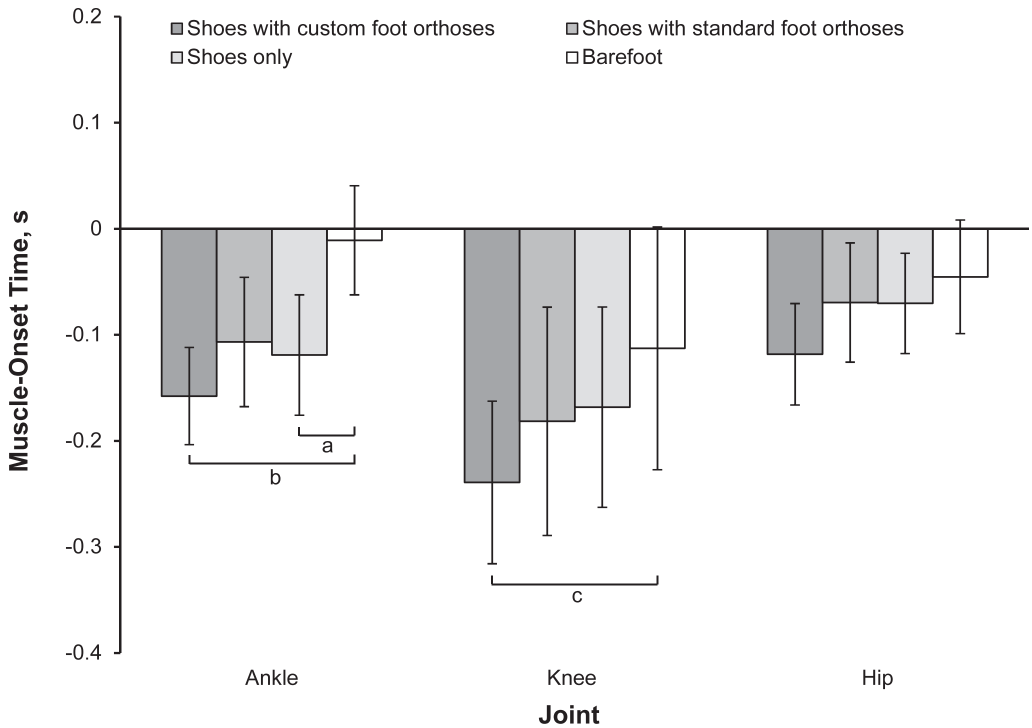

Figure 2. Muscle-activation onset times (means and 95% confidence intervals) in 3 regions (ankle, knee, hip) for 4 conditions of the more affected limb with eyes open. [a] Indicates difference ($P < .05$). [b] Indicates difference ($P < .001$). [c] Indicates difference ($P < .01$).

navicular-drop correction was not different between the custom and standard foot orthoses for the more affected ($t_{13} = -0.548$, $P = .60$) and less affected ($t_{13} = -0.902$, $P = .38$) limbs. Comfort scores were higher ($U = -49$, $P = .003$), and the hardness was lower ($t_{14} = -5.477$, $P < .001$) in the custom than in the standard foot orthoses. The satisfaction rate of the custom foot orthoses was 4.3 ± 0.6.

Muscle Activity

Based on the analysis of the 4 factors (condition, region, limb, and vision) combined, we observed a difference among conditions ($F_{3,91.8} = 9.39$, $P < .001$) and among regions ($F_{2,98} = 4.32$, $P = .02$) but did not observe an interaction between region and condition ($F_{6,120} = 0.87$, $P = .52$). We noted no interactions between condition and limb ($F_{3,87.3} = 0.18$, $P = .91$) or condition and vision ($F_{3,87.3} = 0.44$, $P = .72$). Within each region, no evidence suggested that the condition-muscle analyses depended on limb or vision ($P > .05$). Furthermore, we did not observe an effect of limb ($F_{1,68.3} = 0.22$, $P = .64$) or vision ($F_{1,66.2} = 1.23$, $P = .27$). Therefore, and to simplify data reporting, we present only the results of the 2-way condition-muscle analysis within each region for the more affected limb with eyes open (Figures 2 and 3).[14]

Within the ankle region, a difference was noted among conditions ($F_{3,26.1} = 9.53$, $P < .001$) and among muscles ($F_{2,28.3} = 7.26$, $P = .003$). Furthermore, we observed an interaction ($F_{6,33.2} = 4.20$, $P = .003$) between condition and muscle. Irrespective of the muscle, muscle-activation onset times in the shoes-only (SO; $P = .02$) and shoes-with-custom-foot-orthoses (SCFO; $P < .001$) conditions were earlier than in the BF condition (Figure 2). The muscle-activation onset times of the peroneus longus ($F_{3,25.2} = 10.27$, $P < .001$) and tibialis anterior ($F_{3,25} = 5.51$, $P = .005$) were different among conditions, but the onset times of gastrocnemius activity ($F_{3,26.6} = 1.15$, $P = .35$) were not. The onset times of peroneus longus activity were earlier in the SO ($P = .02$), shoes-with-standard-foot-orthoses (SSFO; $P = .03$), and SCFO ($P < .001$) conditions than in the BF condition. The onset times of tibialis anterior activity were earlier in the SCFO than in the BF condition ($P = .003$; Figure 3).

In the knee region, we noted a difference among conditions ($F_{3,14.9} = 6.66$, $P = .005$) but not among muscles ($F_{1,14.3} = 0.90$, $P = .36$). We did not observe an interaction between condition and muscle ($F_{3,14.5} = 1.32$, $P = .31$). Irrespective of the muscle, muscle-activation onset times were earlier in the SCFO than in the BF condition ($P = .007$; Figure 2). The onset times of vastus medialis obliquus ($F_{3,15.5} = 3.83$, $P = .03$) and vastus lateralis ($F_{3,14.5} = 8.39$, $P = .002$) activity were different among conditions. Earlier onset times were noted for the vastus medialis obliquus ($P = .04$) and vastus lateralis ($P = .005$) activity in the SCFO than in the BF condition (Figure 3).

Within the hip region, we observed a difference among muscles ($F_{3,28.8} = 41.18$, $P < .001$) but not among conditions ($F_{3,46.5} = 1.46$, $P = .24$). We noted an interaction between condition and muscle ($F_{9,77.1} = 2.29$, $P = .03$). The muscle-activation onset times for all hip muscles were not different among conditions ($P > .05$; Figure 3).

The results of the post hoc analyses (differences among all conditions for all muscles) are shown in Table 2. The mean differences and associated 95% confidence intervals among all conditions for all muscles are presented in Table 3.

DISCUSSION

We are the first investigators to evaluate the influence of shoes and foot orthoses on lower extremity muscle-

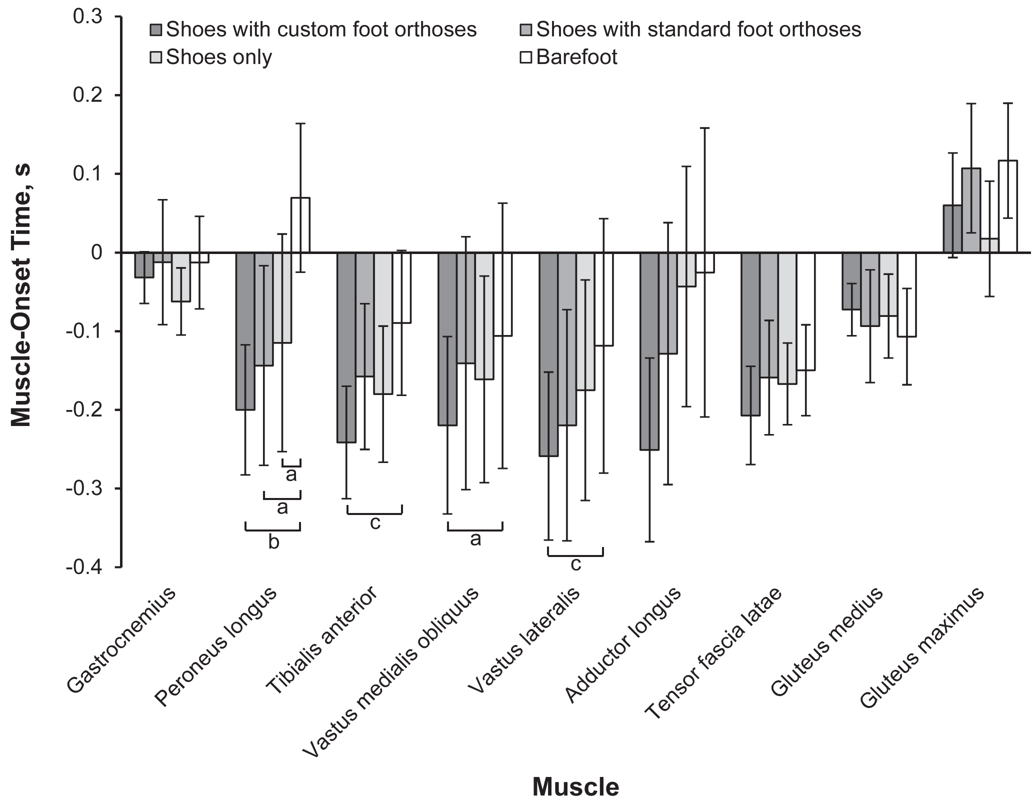

Figure 3. Muscle-activation onset times (means and 95% confidence intervals) of the gastrocnemius, peroneus longus, tibialis anterior, vastus medialis obliquus, vastus lateralis, adductor longus, tensor fascia latae, gluteus medius, and gluteus maximus of the more affected leg with eyes open. [a] Indicates difference ($P < .05$). [b] Indicates difference ($P < .001$). [c] Indicates difference ($P < .01$).

activation onset times during the transition from double-legged to single-legged stance in participants with CAI. Earlier muscle-activation onset times were most prominent in the SCFO condition and were observed not only around the ankle but also around the knee.

At the ankle, the onset times of peroneus longus activity were earlier in the SO, SSFO, and SCFO conditions than in the BF condition, whereas the onset time of tibialis anterior activity was earlier in the SCFO than in the BF condition. Dingenen et al[14] also reported decreased latencies of muscle-activation onset for peroneus longus activity in the SO and SCFO conditions compared with the BF condition in uninjured participants but no alterations in onset times of tibialis anterior activity. Baur et al[20] reported higher peroneal preactivity but no effect on the timing of peroneus longus activation after an 8-week foot-orthoses intervention in participants with running-related overuse injuries. Earlier activation of the peroneus longus and tibialis anterior in participants with CAI could be clinically relevant, given that the magnitude of the differences among conditions that we reported exceeded the standard errors of differences between repeated measurements (70 milliseconds for the peroneus longus and 90 milliseconds for the tibialis anterior) noted during the transition from double-legged to single-legged stance in participants with CAI.[5] During functional activities, the dynamic restraints of the lower extremity joints need to react very quickly to overcome the external forces created by the movement. However, some evidence exists that reflexive mechanisms alone may not act quickly enough to prevent lateral ankle sprains.[21] Authors[22,23] investigating the injury mechanisms of ankle-inversion sprains have shown that these injuries occurred very quickly after foot contact. The time to reach maximum inversion during these injury mechanisms was 80 to 170 milliseconds,[22,23] whereas the first peroneal EMG activity could be observed at approximately 54 milliseconds during laboratory trapdoor tests in healthy participants.[21] Hoch and McKeon[24] reported that these peroneal reaction times occurred even later in participants with CAI. Furthermore, the electromechanical delay, which is approximately 72 milliseconds, has to be added to this 54-millisecond reaction time.[21] Therefore, Konradsen et al[21] suggested that active joint protection cannot be expected until approximately 126 milliseconds after heel strike. This implies that preparatory muscle-activation mechanisms are essential to compensate for these delays and to provide ankle stability during functional tasks. Accelerating the timing of the dynamic-restraint mechanism may lead to a system that is better prepared to react properly to load acceptance.[25]

A loss of anticipatory muscle activity impairs the ability to control the reactive forces caused by the voluntary movement and has been associated with an increased reinjury risk.[26] This phenomenon has been demonstrated in a variety of musculoskeletal pathologic conditions, including CAI.[26] Van Deun et al[4] demonstrated that participants with CAI had an increased latency of lower extremity muscle activation compared with uninjured participants during a transition task performed BF. In our study, only the muscle activity of the peroneus longus and gluteus maximus was initiated after the transition from double-legged to single-legged stance in the BF condition. The

Table 2. **Post Hoc Analysis (*P* Values) of the Experimental Conditions for All Muscles in the More Affected Limb With Eyes Open**

	Conditions					
Muscle	Barefoot Versus Shoes Only	Barefoot Versus Shoes With Standard Foot Orthoses	Barefoot Versus Shoes With Custom Foot Orthoses	Shoes Only Versus Shoes With Standard Foot Orthoses	Shoes Only Versus Shoes With Custom Foot Orthoses	Shoes With Standard Foot Orthoses Versus Shoes With Custom Foot Orthoses
Gluteus maximus	.08	.99	.55	.15	.69	.73
Gluteus medius	.87	.98	.74	.98	>.99	.93
Tensor fasciae latae	.95	.99	.30	>.99	.61	.46
Adductor longus	>.99	.96	.20	.86	.09	.40
Vastus lateralis	.29	.09	.005[a]	.90	.06	.90
Vastus medialis obliquus	.41	.86	.04[a]	.89	.38	.39
Tibialis anterior	.28	.66	.003[a]	.97	.45	.28
Peroneus longus	.02[a]	.03[a]	<.001[a]	.97	.33	.74
Gastrocnemius	.38	>.99	.86	.48	.64	.93

[a] Indicates difference ($P < .05$).

Table 3. **Mean Differences (Associated 95% Confidence Intervals) Among Conditions for All Muscles of the More Affected Limb With Eyes Open**

	Conditions					
Muscle	Barefoot Versus Shoes Only	Barefoot Versus Shoes With Standard Foot Orthoses	Barefoot Versus Shoes With Custom Foot Orthoses	Shoes Only Versus Shoes With Standard Foot Orthoses	Shoes Only Versus Shoes With Custom Foot Orthoses	Shoes With Standard Foot Orthoses Versus Shoes With Custom Foot Orthoses
Gluteus maximus	0.099 (−0.011, 0.210)	0.010 (−0.107, 0.127)	0.057 (−0.049, 0.162)	−0.090 (−0.205, 0.026)	−0.043 (−0.146, 0.061)	0.047 (−0.064, 0.158)
Gluteus medius	−0.026 (−0.111, 0.059)	−0.013 (−0.112, 0.085)	−0.034 (−0.107, 0.038)	0.013 (−0.080, 0.106)	−0.008 (−0.074, 0.057)	−0.021 (−0.104, 0.061)
Tensor fasciae latae	0.017 (−0.064, 0.099)	0.009 (−0.088, 0.106)	0.057 (−0.031, 0.146)	−0.008 (−0.101, 0.085)	0.040 (−0.045, 0.125)	0.048 (−0.052, 0.148)
Adductor longus	0.018 (−0.232, 0.268)	0.103 (−0.159, 0.365)	0.225 (0.005, 0.446)	0.085 (−0.152, 0.322)	0.208 (0.006, 0.409)	0.122 (−0.089, 0.333)
Vastus lateralis	0.056 (−0.167, 0.280)	0.101 (−0.127, 0.329)	0.14 (−0.060, 0.340)	0.045 (−0.168, 0.257)	0.084 (−0.100, 0.268)	0.039 (−0.151, 0.229)
Vastus medialis obliquus	0.055 (−0.166, 0.276)	0.035 (−0.211, 0.281)	0.114 (−0.092, 0.320)	−0.021 (−0.237, 0.195)	0.058 (−0.123, 0.239)	0.079 (−0.124, 0.282)
Tibialis anterior	0.091 (−0.041, 0.223)	0.068 (−0.068, 0.205)	0.152 (0.030, 0.274)	−0.022 (−0.155, 0.110)	0.061 (−0.056, 0.179)	0.084 (−0.039, 0.206)
Peroneus longus	0.184 (0.009, 0.359)	0.213 (0.048, 0.379)	0.270 (0.138, 0.401)	0.029 (−0.167, 0.225)	0.085 (−0.083, 0.254)	0.056 (−0.102, 0.215)
Gastrocnemius	0.050 (−0.026, 0.125)	−0.000 (−0.104, 0.103)	0.019 (−0.050, 0.088)	−0.050 (−0.142, 0.043)	−0.030 (−0.087, 0.026)	0.019 (−0.068, 0.107)

onset times of activity for all other muscles did not typically occur after the transition. This can be explained by the definition of *Mx onset* provided by Dingenen et al[14] and the fact that movement speed was not standardized in the studies of Van Deun et al.[4,5] Therefore, one should be cautious when interpreting the magnitude of the differences among conditions we reported based on these studies.[4,5] Furthermore, muscle-activation times are known to be task dependent.[27] Therefore, comparing our results with those reported in other studies in which different tasks (eg, quicker trapdoor tasks) were used is difficult.

In agreement with previous studies,[4,14] our participants' muscle-activation onset times were not influenced by the vision condition. Furthermore, we observed no differences among conditions for the gastrocnemius, thereby supporting the literature.[14,28] The biarticular course of the gastrocnemius and the fact that this muscle is unlikely to have a substantial function as an invertor or evertor of the rear foot during the transition task may provide reasonable explanations for these observations.

In contrast with previous studies,[14,28] we noted differences in muscle-activation onset times among conditions in the knee muscles. Decreased latencies of muscle-activation onset of the vastus medialis obliquus and vastus lateralis were observed in the SCFO condition compared with the BF condition. In accordance with the peroneus longus and tibialis anterior, the magnitude of these differences exceeded the standard errors of differences between repeated measurements (90 milliseconds for the vastus medialis obliquus and 120 milliseconds for the vastus lateralis) that Van Deun et al[5] reported during the transition from double-legged to single-legged stance in participants with CAI. Dingenen et al[14] described a trend toward an earlier activation of these muscles in uninjured participants among these conditions but did not observe differences. The more homogeneous group in the current study may explain why differences among conditions at the knee were more apparent. Most of the other researchers who investigated the influence of foot orthoses on knee-muscle activity have focused solely on the level of activation and not on muscle-activation onset times.[29,30] However, we do not know whether an increase or decrease in muscle activity is beneficial or detrimental in relation to injury.[12]

Optimizing neuromuscular control of the hip muscles is considered crucial for lower extremity function. Researchers[29,31] have indicated that hip-muscle activity also may adapt to the use of foot orthoses. However, in accordance with the results of the uninjured participants,[14] no differences among conditions were observed in muscle-activation onset times of the hip muscles. The interaction among conditions and muscles can be explained by the larger differences among conditions for the adductor longus than for the other hip muscles.

Whereas the exact physiologic pathways describing how the reported changes in muscle-activation onset times may occur have not been defined, our observations confirmed the assumption that shoes and foot orthoses may have important neuromuscular effects. In the conditions where shoes and foot orthoses were used, participants were more likely to activate the ankle and knee muscles earlier, whereas the hip muscles were unaffected. A possible explanation for this neuromuscular bottom-up effect of shoes and foot orthoses is the sensory reweighting theory. Carver et al[32] presumed that persons rely on the sensory inputs that provide the most functionally reliable information. The sensory information coming from the foot and ankle may become more reliable when shoes and foot orthoses are used, because the cutaneous mechanoreceptors on the plantar surface of the foot may be stimulated more through the increased contact area between the foot and supporting surface.[9] Improving the sensory input coming from the foot and ankle toward the central nervous system might be clinically beneficial, given that proprioceptive deficits have been reported in participants with CAI.[15] A modulation of the afferent input by these mechanisms toward the central nervous system subsequently may lead to an adaptation of the efferent output.[7] However, we cannot exclude that biomechanical effects also contributed to the aforementioned results even though researchers[10] recently questioned the movement-control function of foot orthoses. In most studies focusing on the biomechanical influence of foot orthoses, investigators[10] have measured small and nonsystematic changes of the rear foot and tibia. Nester et al[33] concluded that the kinematic effects of foot orthoses were most obvious at the rear-foot complex, whereas the more proximal joints (knee, hip, pelvis) were generally unaffected during walking. Smaller kinematic changes in the proximal regions, therefore, may cause smaller changes in muscle activation, but even subtle changes in lower extremity alignment achieved by foot orthoses may facilitate proprioceptive mechanisms and muscle activity.[11] Therefore, the functioning of shoes and foot orthoses may be mediated by an interplay of biomechanical and neuromuscular effects.[14,20] However, this theoretical framework is only speculative and far from conclusive.

Researchers[28,30] have suggested that appropriate footwear may be necessary to maximize the potential benefits of foot orthoses. Our results showed that shoes can also decrease muscle-activation latencies but to a lesser extent than the conditions for which the custom foot orthoses are also worn. Dingenen et al[14] reported similar results. This indicates that the observed findings in the SCFO condition may result from an interplay between the shoes and the foot orthoses. We find it interesting that only peroneus longus muscle activity was initiated earlier during the SSFO than in the BF condition, whereas peroneus longus, tibialis anterior, vastus medialis obliquus, and vastus lateralis muscle activities were initiated earlier in the SCFO than in the BF condition despite a similar navicular-drop correction for both types of foot orthoses. Our participants had not adapted to the SSFO condition. This may provide a reasonable explanation for these differences, because researchers[20,34] have suggested that a wearing period of at least 4 weeks is needed to see full adaptations to foot orthoses. Furthermore, the comfort scores for the SSFO condition were lower. This difference can be considered clinically relevant[35] and may be important because orthotic comfort has been related to muscle activation.[36] The lower hardness level of the custom foot orthoses may further explain this observation, given that softer foot orthoses generally are considered to be more comfortable.[20]

Clinical Relevance

Rehabilitation programs for CAI vary but typically involve therapeutic modalities to improve dynamic ankle stability. One possible way to achieve this goal is to improve the timing of the dynamic-restraint mechanism with specific neuromuscular exercises.[26] Our results support the observations of Dingenen et al[14] and imply that shoes and foot orthoses can also influence neuromuscular function in participants with CAI. In future studies, investigators should evaluate the possible additional value or interaction between exercise therapy and the use of shoes and foot orthoses in participants with CAI. Moreover, they should determine whether these changes in muscle function are associated with positive clinical outcomes.

Limitations

A few limitations need to be addressed. The material and design of the custom foot orthoses, duration of use of custom foot orthoses, and reasons for using custom foot orthoses varied. One could suggest that more specific criteria for the custom foot orthoses would elicit differences more clearly. Using surface EMG, we could not measure the activity of deeper muscles, such as the tibialis posterior. This muscle acts as a dynamic stabilizer of the rear foot and medial longitudinal arch, but given its deep location within the posterior compartment of the lower leg, the measurement of tibialis posterior muscle activity was not possible using our methods without cross-talk from various superficial muscles.[37] More invasive intramuscular electrodes in conjunction with ultrasound imaging to visualize the target zone are better able to measure tibialis posterior muscle activity.[37] Eight of the 15 participants reported bilateral CAI. Caution, therefore, is warranted when interpreting the differences between limbs. However, the main purpose of our study was to investigate differences in muscle-activation onset times among conditions rather than differences between limbs. The height of the medial arch support of the standard foot orthoses was based on the correction of half of the navicular-drop excursion.[14] Nevertheless, the final navicular-drop correction of the standard foot orthoses was slightly more than our goal of 50% of navicular-drop correction, with a difference of 1.3 mm for the more affected limb and 1.7 mm for the less affected limb. The corrections of the navicular drop with standard foot orthoses were 3.7 mm for the more affected limb and 4.4 mm for the less affected limb (Table 1), whereas 50% of the navicular drop would have been 2.5 mm (4.9/2) for the more affected limb and 2.8 mm (5.5/2) for the less affected limb. The correction, therefore, was 1.2 mm (range, 3.7 to 2.5 mm) and 1.6 mm (range, 4.4 to 2.8 mm) more than our goal of 50% for the more affected and less affected limbs, respectively. However, considering the reported measurement error of the navicular-drop measurement (range, 1.47 to 3.66 mm for intratester comparisons),[38,39] we did not consider this difference clinically relevant. Furthermore, these standard foot orthoses also had a neutral rear-foot posting, which could have influenced the position of the rear foot and, subsequently, the navicular height. The cumulative effect of the measurement error and the rear-foot posting could have contributed to the slightly increased correction of the navicular drop. Using this method instead of more sophisticated technologies to create or select the standard foot orthoses may be a limitation of the study. However, considering the standard foot orthoses as "standard" after further customization would be difficult. Furthermore, researchers[29,40] have shown that the degree of posting may not influence EMG variables. The same neutral running shoes were used for all participants. Material properties of the shoes, therefore, were identical across participants, but the disadvantage of this choice was that participants lacked habituation to these shoes. We could not evaluate the possible biomechanical alterations of shoes or foot orthoses because we did not measure kinematics. However, we were measuring the influence on muscle-activation onset times. Given the cross-sectional nature of this study, we do not know how these neuromuscular changes can develop over time. Therefore, researchers should evaluate the short- and long-term neuromuscular adaptations of shoes and foot orthoses in future studies.

CONCLUSIONS

We observed that shoes and foot orthoses accelerated muscle-activation onset times of the ankle and knee but not the hip in participants with CAI. These results may help clinicians understand how shoes and foot orthoses can influence neuromuscular control of the lower extremity in participants with CAI. Given that shoes and foot orthoses are often used during daily activities, one should be careful when generalizing the findings reported from studies in which only BF tests are conducted.

REFERENCES

1. Hootman JM, Dick R, Agel J. Epidemiology of collegiate injuries for 15 sports: summary and recommendations for injury prevention initiatives. *J Athl Train*. 2007;42(2):311–319.
2. van Rijn RM, van Os AG, Bernsen RM, Luijsterburg PA, Koes BW, Bierma-Zeinstra SM. What is the clinical course of acute ankle sprains? A systematic literature review. *Am J Med*. 2008;121(4):324–331.
3. Hertel J. Functional anatomy, pathomechanics, and pathophysiology of lateral ankle instability. *J Athl Train*. 2002;37(4):364–375.
4. Van Deun S, Staes FF, Stappaerts KH, Janssens L, Levin O, Peers KK. Relationship of chronic ankle instability to muscle activation patterns during the transition from double-leg to single-leg stance. *Am J Sports Med*. 2007;35(2):274–281.
5. Van Deun S, Stappaerts K, Levin O, Janssens L, Staes F. Stability of measurement outcomes for voluntary task performance in participants with chronic ankle instability and healthy participants. *J Athl Train*. 2011;46(4):366–375.
6. Dingenen B, Staes FF, Janssens L. A new method to analyze postural stability during a transition task from double-leg stance to single-leg stance. *J Biomech*. 2013;46(13):2213–2219.
7. Nurse MA, Nigg BM. The effect of changes in foot sensation on plantar pressure and muscle activity. *Clin Biomech (Bristol, Avon)*. 2001;16(9):719–727.
8. Riemann BL, Lephart SM. The sensorimotor system: part II. The role of proprioception in motor control and functional joint stability. *J Athl Train*. 2002;37(1):80–84.
9. Sesma AR, Mattacola CG, Uhl TL, Nitz AJ, McKeon PO. Effect of foot orthotics on single- and double-limb dynamic balance tasks in patients with chronic ankle instability. *Foot Ankle Spec*. 2008;1(6):330–337.
10. Mills K, Blanch P, Chapman AR, McPoil TG, Vicenzino B. Foot orthoses and gait: a systematic review and meta-analysis of literature pertaining to potential mechanisms. *Br J Sports Med*. 2010;44(14):1035–1046.
11. Richie DH Jr. Effects of foot orthoses on patients with chronic ankle instability. *J Am Podiatr Med Assoc*. 2007;97(1):19–30.

12. Murley GS, Landorf KB, Menz HB, Bird AR. Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: a systematic review. *Gait Posture*. 2009;29(2): 172–187.
13. Scott LA, Murley GS, Wickham JB. The influence of footwear on the electromyographic activity of selected lower limb muscles during walking. *J Electromyogr Kinesiol*. 2012;22(6):1010–1016.
14. Dingenen B, Peeraer L, Deschamps K, Fieuws S, Janssens L, Staes F. The influence of shoes and foot orthoses on lower extremity muscle activation onset times in healthy subjects during the transition from double-leg stance to single-leg stance. *J Sports Med Phys Fitness*. 2015;55(1–2):16–24.
15. Hertel J. Sensorimotor deficits with ankle sprains and chronic ankle instability. *Clin Sports Med*. 2008;27(3):353–370.
16. Delahunt E, Coughlan GF, Caulfield B, Nightingale EJ, Lin CW, Hiller CE. Inclusion criteria when investigating insufficiencies in chronic ankle instability. *Med Sci Sports Exerc*. 2010;42(11):2106–2121.
17. Gribble PA, Delahunt E, Bleakley C, et al. Selection criteria for patients with chronic ankle instability in controlled research: a position statement of the International Ankle Consortium. *J Orthop Sports Phys Ther*. 2013;43(8):585–591.
18. Hodges PW, Bui BH. A comparison of computer-based methods for the determination of onset of muscle contraction using electromyography. *Electroencephalogr Clin Neurophysiol*. 1996;101(6):511–519.
19. Verbeke G, Molenberghs G. *Linear Mixed Models in Practice: An SAS-Oriented Approach*. New York, NY: Springer-Verlag; 1997.
20. Baur H, Hirschmuller A, Muller S, Mayer F. Neuromuscular activity of the peroneal muscle after foot orthoses therapy in runners. *Med Sci Sports Exerc*. 2011;43(8):1500–1506.
21. Konradsen L, Voigt M, Hojsgaard C. Ankle inversion injuries: the role of the dynamic defense mechanism. *Am J Sports Med*. 1997; 25(1):54–58.
22. Mok KM, Fong DT, Krosshaug T, et al. Kinematics analysis of ankle inversion ligamentous sprain injuries in sports: 2 cases during the 2008 Beijing Olympics. *Am J Sports Med*. 2011;39(7):1548–1552.
23. Fong DT, Ha SC, Mok KM, Chan CW, Chan KM. Kinematics analysis of ankle inversion ligamentous sprain injuries in sports: five cases from televised tennis competitions. *Am J Sports Med*. 2012; 40(11):2627–2632.
24. Hoch MC, McKeon PO. Peroneal reaction time after ankle sprain: a systematic review and meta-analysis. *Med Sci Sports Exerc*. 2014; 46(3):546–556.
25. Santello M. Review of motor control mechanisms underlying impact absorption from falls. *Gait Posture*. 2005;21(1):85–94.
26. Crow J, Pizzari T, Buttifant D. Muscle onset can be improved by therapeutic exercise: a systematic review. *Phys Ther Sport*. 2011; 12(4):199–209.
27. Duysens J, Levin O. Ankle sprains: getting off on the wrong foot. *Exerc Sport Sci Rev*. 2010;38(3):143–149.
28. Rose HM, Shultz SJ, Arnold BL, Gansneder BM, Perrin DH. Acute orthotic intervention does not affect muscular response times and activation patterns at the knee. *J Athl Train*. 2002;37(2):133–140.
29. Hertel J, Sloss BR, Earl JE. Effect of foot orthotics on quadriceps and gluteus medius electromyographic activity during selected exercises. *Arch Phys Med Rehabil*. 2005;86(1):26–30.
30. Nawoczenski DA, Ludewig PM. Electromyographic effects of foot orthotics on selected lower extremity muscles during running. *Arch Phys Med Rehabil*. 1999;80(5):540–544.
31. Bird AR, Bendrups AP, Payne CB. The effect of foot wedging on electromyographic activity in the erector spinae and gluteus medius muscles during walking. *Gait Posture*. 2003;18(2):81–91.
32. Carver S, Kiemel T, Jeka JJ. Modeling the dynamics of sensory reweighting. *Biol Cybern*. 2006;95(2):123–134.
33. Nester CJ, van der Linden ML, Bowker P. Effect of foot orthoses on the kinematics and kinetics of normal walking gait. *Gait Posture*. 2003;17(2):180–187.
34. Hirschmuller A, Baur H, Muller S, Helwig P, Dickhuth HH, Mayer F. Clinical effectiveness of customised sport shoe orthoses for overuse injuries in runners: a randomised controlled study. *Br J Sports Med*. 2011;45(12):959–965.
35. Mills K, Blanch P, Vicenzino B. Identifying clinically meaningful tools for measuring comfort perception of footwear. *Med Sci Sports Exerc*. 2010;42(10):1966–1971.
36. Mundermann A, Nigg BM, Humble RN, Stefanyshyn DJ. Orthotic comfort is related to kinematics, kinetics, and EMG in recreational runners. *Med Sci Sports Exerc*. 2003;35(10):1710–1719.
37. Semple R, Murley GS, Woodburn J, Turner DE. Tibialis posterior in health and disease: a review of structure and function with specific reference to electromyographic studies. *J Foot Ankle Res*. 2009;2:24.
38. Sell KE, Verity TM, Worrell TW, Pease BJ, Wigglesworth J. Two measurement techniques for assessing subtalar joint position: a reliability study. *J Orthop Sports Phys Ther*. 1994;19(3):162–167.
39. Vinicombe A, Raspovic A, Menz HB. Reliability of navicular displacement measurement as a clinical indicator of foot posture. *J Am Podiatr Med Assoc*. 2001;91(5):262–268.
40. Murley GS, Bird AR. The effect of three levels of foot orthotic wedging on the surface electromyographic activity of selected lower limb muscles during gait. *Clin Biomech (Bristol, Avon)*. 2006;21(10): 1074–1080.

Address correspondence to Bart Dingenen, MSc, PT, Department of Rehabilitation Sciences, KU Leuven, Tervuursevest 101 b1501, Leuven, 3001, Belgium. Address e-mail to bart.dingenen@faber.kuleuven.be.