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Effect of Limiting Ankle-Dorsiflexion Range of Motion on Lower Extremity Kinematics and Muscle-Activation Patterns During a Squat

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Context: Limitations in gastrocnemius/soleus flexibility that restrict ankle dorsiflexion during dynamic tasks have been reported in individuals with patellofemoral pain (PFP) and are theorized to play a role in its development. ***Objective:*** To determine the effect of restricted ankle-dorsiflexion range of motion (ROM) on lower extremity kinematics and muscle activity (EMG) during a squat. The authors hypothesized that restricted ankle-dorsiflexion ROM would alter knee kinematics and lower extremity EMG during a squat. ***Design:*** Cross-sectional. ***Participants:*** 30 healthy, recreationally active individuals without a history of lower extremity injury. ***Interventions:*** Each participant performed 7 trials of a double-leg squat under 2 conditions: a no-wedge condition (NW) with the foot flat on the floor and a wedge condition (W) with a 12° forefoot angle to simulate reduced plantar-flexor flexibility. ***Main Outcome Measures:*** 3-dimensional hip and knee kinematics, medial knee displacement (MKD), and ankle-dorsiflexion angle. EMG of vastus medialis oblique (VMO), vastus lateralis (VL), lateral gastrocnemius (LG), and soleus (SOL). One-way repeated-measures ANOVAs were performed to determine differences between the W and NW conditions. ***Results:*** Compared with the NW condition, the wedge produced decreased peak knee flexion ($P < .001$, effect size [ES] = 0.81) and knee-flexion excursion ($P < .001$, ES = 0.82) while producing increased peak ankle dorsiflexion ($P = .006$, ES = 0.31), ankle-dorsiflexion excursion ($P < .001$, ES = 0.31), peak knee-valgus angle ($P = .02$, ES = 0.21), and MKD ($P < .001$, ES = 2.92). During the W condition, VL ($P = 0.002$, ES = 0.33) and VMO ($P = .049$, ES = 0.20) activity decreased while soleus activity increased ($P = .03$, ES = 0.64) compared with the NW condition. No changes were seen in hip kinematics ($P > .05$). ***Conclusions:*** Altering ankle-dorsiflexion starting position during a double-leg squat resulted in increased knee valgus and MKD, as well as decreased quadriceps activation and increased soleus activation. These changes are similar to those seen in people with PFP.

Keywords: knee, patellofemoral pain, medial knee displacement, dynamic knee valgus

Injuries in physically active populations most often occur in the lower extremity, with up to 42% of these injuries occurring at the knee.[1] Chronic knee pain such as patellofemoral pain syndrome (PFP) is one of the most common forms of knee overuse injury.[1–4] Due to the high prevalence of this condition among physically active individuals, it is important to understand factors that may predispose an individual to the development of PFP.

Frontal-plane knee malalignment during dynamic activities is theorized to be a risk factor for PFP.[5–9] Specifically, increased frontal-plane knee motion during dynamic activities may play a role in the development of PFP due to alterations in joint loading at the patellofemoral joint and increased stress on the periarticular structures of the patella.[5,7] Excessive frontal-plane knee motion has been theorized to be caused by a variety of factors during functional tasks, including hip-muscle weakness.[9] However, the relationship between hip-muscle strength, activation level, and frontal-plane knee motion has been equivocal.[10–12] Another factor that may influence frontal-plane alignment is the foot and ankle complex.[13] Dynamic frontal-plane motion has been described as a combination of joint motion including the ankle, hip, and knee.[14] However, little evidence exists that links sagittal-plane ankle motion and frontal-plane knee motion.[13]

Limited ankle-dorsiflexion range of motion (ROM) due to gastrocnemius[15] and soleus[16] tightness is known to exist in individuals with PFP. The risk of developing PFP as a result of limited ankle dorsiflexion has been attributed to a series of biomechanical compensations in response

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to the limited ankle ROM.[16] Specifically, decreased dorsiflexion during weight-bearing tasks limits the ability to lower the body's center of mass, encouraging increased subtalar-joint pronation and tibial internal rotation to gain additional motion. Increased tibial internal rotation requires a concomitant increase in femoral internal rotation and has been linked with a knee valgus position.[17] These alterations in frontal- and transverse-plane motion at the hip and knee are likewise theorized to lead to the development of PFP.[9]

Limited ankle-dorsiflexion ROM has also been shown to influence clinical measures of knee motion. Bell et al[13] grouped individuals based on performance during a double-leg squat. They compared lower extremity strength and ROM between unimpaired subjects who kept their knees over their toes during a squat and subjects with medial knee displacement (MKD) in which the patella passed medial to the great toe during the squat. They found that individuals with MKD displayed limited ankle-dorsiflexion ROM compared with a group that did not show evidence of MKD. By using a heel lift to eliminate ankle-dorsiflexion restrictions, subjects reduced MKD. To better understand the role of gastrocnemius- and soleus-muscle flexibility as potential risk factors for PFP, it is important to understand the influence of ankle-dorsiflexion ROM on lower extremity kinematics and muscle activation. Restricting ankle-dorsiflexion ROM may produce effects in lower extremity kinematics and muscle-activation patterns during a squat task comparable to those commonly observed in individuals with PFP. Therefore, the purpose of this investigation was to determine the effect of reduced dorsiflexion ROM on 3-dimensional kinematics at the hip and knee, ankle-dorsiflexion angle, quadriceps activation, gastrocnemius activation, and soleus activation during the descent phase of a squat task. We hypothesized that restricting ankle-dorsiflexion ROM using a wedge under the foot would result in hip and knee kinematic changes and muscle-activation changes for the surrounding knee musculature.

Methods

Participants

Thirty subjects (15 men, 15 women; height 173.5 ± 12.1 cm; weight 72.0 ± 16.4 kg) completed the test protocol. Subjects were healthy and ranged in age from 18 to 30 years. They were required to be physically active, which was defined as engaging in 30 minutes of physical activity a day for a minimum of 3 days/wk. Subjects were excluded if they reported lower extremity injury in either leg within the past 3 months that had required them to miss physical activity for at least 1 day or required physician referral, if they had had lower extremity surgery within the past year, or if they had current knee pain at time of testing. Before participation, all subjects read and signed an informed-consent form approved by the university's institutional review board.

Procedures

Subjects reported to a research laboratory for a single testing session lasting approximately 1 hour. They were required to wear standard running shoes, spandex shorts, and a spandex T-shirt or sports bra. Data from the subject's dominant leg, which was defined as the leg used to kick a ball a maximal distance, were used for all data analyses. Before data collection, subjects warmed up on a stationary cycle ergometer (Schwinn Airdyne Upright Bike, Nautilus, Inc) for 5 minutes at a self-selected pace.

Surface electromyography (EMG; Delsys Bagnoli-8, Boston, MA) was used to record muscle activity of the quadriceps (vastus medialis oblique [VMO] and vastus lateralis [VL]), lateral gastrocnemius (LG), and soleus (SOL) using Ag-AgCl single differential surface electrodes (Delsys Inc, Boston, MA) with a sampling rate of 1200 Hz. Unit specifications for the EMG system included a common-mode rejection ratio of 92 dB and amplifier gain of 1000. The skin was prepared before electrode placement using standard procedures including shaving and cleansing with isopropyl alcohol. The electrode for the VL was placed over the VL, approximately 10 cm superior and 7 cm lateral to the superior border of the patella oriented at 10° to the vertical.[18] For the VMO the electrode was placed approximately 4 cm superior and 3 cm medial to the superomedial border of the patella oriented at a 55° angle.[18] The electrode for the LG was placed over the bulge of the lateral head of the gastrocnemius.[19] Electrode placement for the SOL was just medial to the Achilles tendon, inferior to the midpoint of the lower leg.[19] A reference electrode was placed over the tibial tuberosity of the test limb. All electrode placements were reinforced with prewrap and athletic tape and checked for crosstalk with manual muscle testing before data collection.

Each subject was fitted with reflective markers to enable recording kinematic data of the lower extremity during the squatting tasks. The movement of the reflective markers was captured by 7 infrared video cameras (Vicon Motion Systems, Centennial, CO) at a sampling rate of 120 Hz. All data were collected using Vicon Nexus Software (version 1.1, Vicon Motion Systems). Reflective markers were attached to subject's L5–S1 space and bilaterally on the following landmarks: anterosuperior iliac spine, greater trochanter, medial and lateral femoral epicondyles, midthigh, midshank, medial and lateral malleoli, head of the fifth metatarsal, head of the first metatarsal, calcaneus, and acromion process. All markers were placed over spandex clothing (trochanter) or footwear (metatarsal and calcaneal markers) if they could not be applied directly to the skin at a given landmark. A Helen Hayes marker set was used with the addition of markers placed over the greater trochanter. Before trial-data collection, a static calibration trial was recorded for each subject with the subject standing with feet shoulder width apart and both arms abducted to 90°. Medial markers were removed from both legs after static calibration. A right-hand global reference system was defined in which the *x*-axis was positive in the anterior direction, the *y*-axis was positive to the left of each subject, and the *z*-axis was positive in the superior direction. Hip-joint angles were calculated with Euler-angle conventions of flexion (–)/extension (+) about the *y*-axis, abduction (–)/adduction (+) about the *x*-axis, and internal (+)/external (–) rotation about the *z*-axis. Knee-joint angles were calculated with Euler-angle conventions of flexion (+)/extension (–) about the *y*-axis, varus (+)/valgus (–) about the *x*-axis, and internal (+)/external (–) rotation about the *z*-axis. Ankle-joint angles were calculated with Euler-angle conventions of plantar flexion (+)/dorsiflexion (–) about the *y*-axis; the other 2 axes were not related to the purpose of this study.

Subjects performed 7 double-leg squats under 2 separate counterbalanced conditions: a no-wedge (NW) condition with the foot flat on the floor (Figure 1) and a wedge (W) condition with a 12° forefoot angle (Figure 2). The wedge ran along the full length of the foot and was designed to place the subject's ankle in approximately 12° of dorsiflexion before performing the double-leg squat task. It was used to put the subject's ankle in a starting position that was closer to the end range of dorsiflexion motion, which would mimic a restriction in plantar-flexor muscle flexibility. Thus, during the W condition the subjects would reach their end range of dorsiflexion sooner, thereby minimizing the amount of motion available during the double-leg squat. A 12° wedge was determined through pilot testing to cause a change in overall lower body kinematics but not restrict a subject's ability to perform the overhead squat task. The subjects began the squat task from a standardized start position with feet shoulder width apart, toes facing straight ahead, arms overhead, and heels on the floor. They were instructed to perform a double-leg squat "as if they were sitting in a chair" to a comfortable depth. Each subject was allowed a maximum of 5 practice repetitions and was provided 2 minutes of rest between the practice and test trials. Each individual was asked to stand with feet shoulder width apart without the wedge in place, and this procedure was mimicked for the W condition to standardize foot position between conditions.

Three trials of maximal voluntary isometric contractions (MVICs) were performed for the VMO and VL, SOL, and LG after performing all test trials. For VMO and VL, the subject sat in a dynamometer chair with knees and hips flexed to 90° and was instructed to extend the knee forcefully into an immovable strap for 5 seconds. SOL MVIC was collected with the subject in a quadruped position with knees and hips flexed to 90° and an immovable strap around the heads of the metatarsals of the test limb. LG MVIC was collected with the subject lying prone with the test limb off the end of a table and an immovable strap across the metatarsal heads. For both the SOL and LG MVICs, the subjects plantar-flexed against the strap with maximal effort. Each trial lasted 5 seconds, and the middle 3 seconds of each were averaged and used for normalization.

All data were imported into Motion Monitor Software (version 7.72, Innovative Sports Training, Inc, Chicago IL) and reduced using a customized MatLab (Mathworks, Natick MA) program. EMG data were corrected for direct-current bias, band-pass filtered (zero phase lag, fourth order, Butterworth) with cutoff frequencies of 20 and 350 Hz and notched filtered (59.5–60.5 Hz). EMG data were smoothed using a 20-millisecond root-mean-square sliding-window function. Kinematic data were filtered using a fourth-order low-pass Butterworth filter with a 12-Hz cutoff frequency.[20,21] Squat depth was operationally defined as peak knee-flexion angle. Average EMG amplitude and peak kinematic values were calculated during the descending phase of each squat, which was found using knee-flexion angle and defined as the onset of knee flexion through peak knee-flexion angle. Kinematic-excursion values were calculated by subtracting the minimum value from the peak value for each variable. MKD was calculated by subtracting the starting knee-center value immediately

Figure 1 — Start position of the distal segments for the no-wedge condition.

Figure 2 — Start position of the distal segments for the wedge condition.

before the beginning of each squat from the maximum value during the squat (displacement).

Statistical Analyses

All data analyses were performed using SPSS version 17.0 (SPSS, Inc, Chicago, IL). Separate repeated-measures ANOVAs were run for each dependent variable, and the within-subject factor was wedge condition (2 levels: W, NW). A priori alpha level was set at $P \leq .05$.

Results

One male subject was removed from data analysis due to equipment malfunctions during data collection, leaving 29 subjects for data analysis. Each variable was examined for skewness and kurtosis using cutoff values of –2 to 2 for skewness and –3 to 3 for kurtosis. All variables were in this range except MKD and SOL EMG. Square-root transformations were performed on these variables and rechecked to verify that they met skewness and kurtosis cutoff values. Transformed values were used for the statistical tests.

To verify that the wedge increased ankle dorsiflexion, we compared the average starting values between conditions. With the wedge in place, ankle dorsiflexion increased by 9.5° compared with the NW condition (NW 0.7° ± 2.8°, W –8.8° ± 3.0°; $F_{28} = 687.0$, $P < .001$, effect size = 3.2). Table 1 presents peak kinematic variables during each condition. With the wedge in place, knee flexion decreased while knee valgus and dorsiflexion increased. Table 2 presents excursions between conditions. With the wedge in place, participants had less knee flexion and ankle dorsiflexion but greater MKD (NW 0.0007 ± 0.002 m, W 0.027 ± 0.014 m; $F_{28} = 31.79$, $P < .01$, effect size = 2.92) excursion than in the NW condition. Finally, we observed significant changes in muscle activation (Table 3). With the wedge in place, SOL activity increased while VL and VMO activity decreased. No difference was observed in the LG between conditions.

Table 1 Peak Joint Angles (°) During the Descending Phase of the Squat for Each Condition, Mean ± SD (95% Confidence Interval)

	No wedge	Wedge	P	F_{28}	Effect size
Knee					
flexion (+)/extension (–)	96.7 ± 17.3 (89.1, 102.3)	81.1 ± 19.2 (73.8, 88.4)	<.01	46.77	0.81
varus (+)/valgus (–)	–3.4 ± 3.2 (–4.6, –2.2)	–4.1 ± 3.3 (–5.3, –2.8)	.02	6.45	0.21
internal rotation (+)/external rotation (–)	10.0 ± 8.1 (7.0, 13.1)	9.4 ± 8.0 (6.3, 12.4)	.64	0.23	0.07
Ankle plantar flexion (+)/dorsiflexion (–)	–28.0 ± 6.4 (–30.5, –25.6)	–30.0 ± 6.3 (–32.4, –27.6)	<.01	8.71	0.31
Hip					
extension (+)/flexion (–)	–62.2 ± 13.6 (–67.5, –57.1)	–63.1 ± 14.0 (–68.5, –57.83)	.45	0.59	0.06
adduction (+)/abduction (–)	–3.4 ± 3.4 (–4.7, –2.1)	–2.5 ± 3.3 (–4.1, –0.82)	.12	2.55	0.26
internal rotation (+)/external rotation (–)	–6.0 ± 3.9 (–7.6, –4.5)	–5.2 ± 3.4 (–6.6, –3.9)	.09	3.01	0.20

Table 2 Displacement Values (°) for Each Condition, Mean ± SD (95% Confidence Interval)

	No wedge	Wedge	P	F_{28}	Effect size
Knee					
sagittal plane	100.2 ± 16.8 (93.6, 106.9)	85.1 ± 18.5 (77.9, 92.6)	<.01	41.8	0.82
frontal plane	17.7 ± 11.3 (13.4, 22.0)	16.9 ± 11.2 (12.6, 21.1)	.15	2.23	0.07
rotation	13.8 ± 9.1 (10.4, 17.3)	12.6 ± 9.2 (9.1, 16.1)	.39	0.78	0.13
Ankle sagittal plane	28.8 ± 6.0 (26.5, 31.1)	21.3 ± 5.9 (19.0, 23.5)	<.01	110.17	1.25
Hip					
sagittal plane	70.7 ± 14.7 (65.2, 76.3)	71.9 ± 14.5 (66.4, 77.5)	.29	1.16	0.08
frontal plane	7.1 ± 3.7 (5.7, 8.5)	8.6 ± 4.8 (6.8, 10.4)	.09	3.04	0.31
rotation	21.6 ± 11.1 (17.3, 25.8)	21.3 ± 11.6 (16.9, 25.7)	.70	0.15	0.03
Body center of mass x (m)	0.05 ± 0.03 (0.05, 0.07)	0.07 ± 0.03 (0.06, 0.08)	<.01	13.5	0.71

Range of motion was calculated by subtracting the minimum and maximum values during the descending phase of the squat. Center of mass in the x direction corresponds to anteroposterior displacement.

Table 3 Average EMG (%MVIC) During the Descending Phase of the Squat for Each Condition, Mean ± SD (95% Confidence Interval)

	No wedge	Wedge	*P*	F_{28}	Effect size
Vastus lateralis	62.4 ± 21.8 (53.9, 70.8)	55.1 ± 18.6 (47.9, 62.3)	<.01	12.23	0.33
Vastus medialis oblique	66.3 ± 26.1 (56.2, 76.4)	61.2 ± 21.1 (53.0, 69.4)	.03	5.63	0.20
Soleus	21.8 ± 2.6 (16.3, 27.9)	23.4 ± 2.4 (18.0, 29.7)	.03	5.46	0.64
Lateral gastrocnemius	18.5 ± 11.2 (14.2, 22.9)	19.8 ± 12.4 (15.0, 24.6)	.21	1.68	0.10

A secondary analysis was performed to describe changes in the body's center of mass (COM). COM position was calculated with Motion Monitor software using anthropometric data.[22] Body-COM displacement was calculated in the anteroposterior (x) direction relative to the global reference system. The wedge resulted in increased posterior COM displacement (Table 2).

Discussion

The purpose of this investigation was to determine the effect of simulated gastrocnemius/soleus tightness, which would limit dorsiflexion ROM, on lower extremity kinematics and muscle activation. Based on our results, decreased sagittal-plane motion at the ankle leads to the following kinematic changes: decreased knee flexion (peak and excursion), increased knee valgus (peak), increased MKD, and reduced dorsiflexion (excursion). Restricting sagittal-plane ankle motion also resulted in increased SOL activation and decreased quadriceps activation during the descent portion of the squat. We hypothesized that changes would occur in muscle activation and knee kinematics to compensate for the ankle joint's decreased ROM. Our results support this hypothesis and provide further evidence that restrictions in ankle motion (in this case due to the placement of a wedge under the foot) alter kinematics and muscle activation at the knee joint.

For this study, we chose to alter the starting position of the ankle using a wedge that extended the entire length of the foot. This altered the ankle position at the beginning of the squat by placing the foot in dorsiflexion. As a result, participants achieved greater peak dorsiflexion and experienced less dorsiflexion excursion. We chose this method because it restricted ankle motion by lengthening the LG and SOL and forcing the talus into the mortise. There may be other explanations for kinematic changes caused by the wedge, such as alterations to the COM. However, given that this is the first study involving this combination of variables, we felt that this was an appropriate approach. Other interventions such as ankle bracing or manual blocking of the shin could be used in future examinations to determine if they produce similar changes in kinematics and muscle activity.

In this investigation, the peak knee-valgus angle increased by approximately 1° when sagittal-plane ankle ROM was limited with the wedge. While the increase in knee-valgus angle during the W condition was small in absolute magnitude (1° increase), this represented an overall increase in knee-valgus angle of 18% during the W condition compared with the NW condition. Knee-valgus angle has been shown to predict noncontact anterior cruciate ligament injury[23] and is theorized to cause PFP.[9] Other research has shown that placing a wedge under the heel to increase the starting plantar-flexion angle and allow for greater ankle-dorsiflexion motion during a squat caused a visual decrease in MKD compared with performing a squat without the wedge.[13] The combined findings from that investigation and our current study highlight the potential influence of ankle-dorsiflexion ROM on knee-valgus motion during a squatting task.

Limiting the amount of available ankle-dorsiflexion ROM during the W condition resulted in a concomitant decrease in peak knee-flexion angle and ROM. Peak knee-flexion angle decreased by approximately 15° during the W condition compared with the NW condition. This large change in knee-flexion angle represented a 16% decrease in knee flexion, which is similar to the percentage increase in knee-valgus angle. We hypothesize that limited ankle-dorsiflexion ROM during the W condition resulted in an inability to achieve full knee flexion (15% decrease), which resulted in a compensatory increase in knee-valgus angle (18% increase) as the individual attempted to lower the body's COM during the squat motion. We did not observe any changes in hip motion (peak and excursion) during the W condition, which suggests that squat kinematic alterations during the W condition are most readily apparent at the knee (decreased knee flexion and increased knee valgus). It makes intuitive sense that decreased ankle motion would influence knee kinematics more than hip kinematics. Rotation of the hip and knee did not change between conditions. The wedge may have resulted in greater changes in these variables if we had recruited a population with significant MKD or valgus alignment.[13]

The compensatory changes associated with limiting ankle-dorsiflexion motion may have clinical relevance as decreased knee flexion, increased knee valgus, and decreased dorsiflexion have been implicated as body postures associated with increased risk of PFPS.[7,16,24,25] Increased knee valgus has been implicated as a risk factor in PFPS due to the forces that subsequently occur at the lateral aspect of the patellofemoral joint. Pathologic groups have shown knee-valgus angulations of just 2° more than those of symptom-free samples.[7] Because individuals may perform a squatlike activity hundreds of times per day, a small change in knee kinematics has the potential to contribute to the development of PFP. An increase in knee valgus is often associated with tightness of the iliotibial band and a lateral tracking of the patella in the patellofemoral joint.[5,8,26] The resulting malalignment of the patella can cause an increase in joint reaction force over a smaller area of the lateral surface of the femoral trochlea and an increase in tensile force on the medial stabilizing structures.[27] This alteration in contact stresses is theorized to lead to the development of PFPS.[9]

Decreased knee-flexion angle has also been implicated in knee injuries.[28] Boling et al[28] identified decreased knee flexion during a jump-landing task as a prospective risk factor of PFP. In addition, women who landed with less knee flexion had greater knee-valgus angles and knee-adductor moments than women who landed with increased hip and knee flexion at landing.[29] These factors can increase frontal-plane motion and negatively affect patellofemoral-joint mechanics.[9] Finally, decreased knee-flexion angles have been observed during dynamic tasks in individuals with PFPS by several researchers.[24,30] Our data demonstrate that the wedge was associated with decreased knee-flexion angle and decreased quadriceps activation. It is important to note that our subject sample was free of PFP. The most likely explanation is that the decreased quadriceps activation is directly linked to knee-flexion angle. As the body's COM is lowered during the squat, increased levels of knee flexion and quadriceps activation are required. When the participants were on the wedge they could not increase the knee-flexion angle any farther, so greater quadriceps activation was not required.

Limiting ankle-dorsiflexion ROM during the squatting task resulted in decreased activity of the quadriceps musculature and increased activity of the SOL during the descent phase of the squat. These changes are likely due to the changes observed in knee-flexion and ankle-dorsiflexion kinematics during the W condition. As previously indicated there was a significant decrease in peak knee-flexion angle and ROM during the W condition, which most likely accounts for the decrease in quadriceps (VMO and VL) muscle activity. Quadriceps activation is necessary during the descent phase of the squat to control knee-flexion motion and prevent the knee from collapsing in the sagittal plane. After the amount of knee-flexion motion was restricted during the W condition, less quadriceps activity was required. SOL activation was significantly increased during the descent phase of the squat task. We believe that greater SOL activation was required during the W condition to control the larger ankle-dorsiflexion angle as the SOL acts to eccentrically resist ankle-dorsiflexion motion. LG activity remained unchanged, possibly due to its diarthrodial nature.

We also observed greater MKD during the W condition, which indicates that the knee center was collapsing medially during the squat. MKD is important because it is a measure that represents how a clinician might assess dynamic alignment during a squat. We quantified MKD by tracking the knee center via motion analysis; however, MKD has been identified as a faulty movement pattern and is theorized to increase the risk of knee injury.[13,31] One major difference between previous investigations and our study was that we did not seek participants with visually identifiable MKD. If we had included a greater number of individuals with poor lower extremity alignment, it is possible that squat performance on the wedge would have exacerbated MKD, given the previously shown propensity for tight and weak ankle musculature. In addition, the relationship between knee valgus and MKD is not clear and needs further investigation.

This study had several limitations. Squat depth and cadence were not controlled because we wanted to observe natural kinematic compensations imposed by the wedge. Squat depth and cadence most likely had a greater influence on EMG, so future research should focus on controlling for these factors. In addition, due to the low effect sizes for the EMG data, we caution readers on the clinical significance of the muscle-activation results. We did not evaluate symptomatic individuals for this study, so future research should determine if these findings hold true in a symptomatic population. Another limitation is that markers were placed over shoes, which may have affected our kinematic data at the ankle.

Conclusions

Clinically, these findings may suggest that the natural compensation for LG and SOL tightness is decreased sagittal-plane motion and increased frontal-plane knee motion. Over time, this may lead to other imbalances throughout the kinetic chain, making the individual more susceptible to overuse or acute knee injuries such as PFP. While most research has assessed static malalignment at the foot and muscle imbalances at the hip with respect to PFPS, few have considered the role of limited ankle dorsiflexion. This study suggests that, for at least some individuals, restrictions in available ankle ROM may cause changes in sagittal- and frontal-plane kinematics at the knee joint. Findings from this study may differ from those of other research to date because of the novel nature of the task performed. Further research is needed to better illuminate how restrictions in available ankle ROM can lead to overuse injury at the knee.

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