# Effect of Prophylactic Knee Bracing on Anterior Cruciate Ligament Agonist and Antagonist Muscle Forces During Perturbed Walking

Raneem Haddara,* MESc, Vahidreza Jafari Harandi,* PhD, and Peter Vee Sin Lee,*† PhD

*Investigation performed at the University of Melbourne, Melbourne, Victoria, Australia*

**Background:** Anterior cruciate ligament (ACL) injuries most commonly occur after a perturbation. Prophylactic knee braces (PKBs) are off-the-shelf braces designed to prevent and reduce the severity of knee injuries during sports, yet their effectiveness has been debated.

**Purpose:** To identify differences in ACL agonist and antagonist muscle forces, during braced and unbraced conditions, while walking with the application of unexpected perturbations.

**Study Design:** Controlled laboratory study.

**Methods:** A total of 20 recreational athletes were perturbed during walking at a speed of 1.1 m/s, and motion analysis data were used to create patient-specific musculoskeletal models. Static optimization was performed to calculate the lower-limb muscle forces. Statistical parametric mapping was used to compare muscle forces between the braced and unbraced conditions during the stance phase of the perturbed cycle.

**Results:** The brace reduced muscle forces in the quadriceps (QUADS), gastrocnemius (GAS), and soleus (SOL) but not in the hamstrings. The peak QUADS muscle force was significantly lower with the brace versus without at 49% to 60% of the stance phase ($28.9 \pm 12.98$ vs $14.8 \pm 5.06$ N/kg, respectively; $P < .001$) and again at 99% of the stance phase ($1.7 \pm 0.4$ vs $3.6 \pm 0.13$ N/kg, respectively; $P = .049$). The SOL muscle force peak was significantly lower with the brace versus without at 25% of the stance phase ($1.9 \pm 1.7$ vs $4.6 \pm 3.4$ N/kg, respectively; $P = .031$) and at 39% of the stance phase ($1.9 \pm 1.4$ vs $5.3 \pm 5.6$ N/kg, respectively; $P = .007$). In the GAS, there were no significant differences between conditions throughout the whole stance phase except between 97% and 100%, where the braced condition portrayed a smaller peak force ($0.23 \pm 0.13$ vs $1.4 \pm 1.1$ N/kg for unbraced condition; $P = .024$).

**Conclusion:** These findings suggested that PKBs that restrict knee hyperextension and knee valgus/varus motion can alter neuromuscular patterns, which result in a reduction of QUADS force.

**Clinical Relevance:** Understanding the way PKBs alter muscle function and knee mechanics can provide invaluable information that will help in making decisions about their use. Further studies should investigate different types of braces and perturbations to evaluate the effectiveness of PKBs.

**Keywords:** anterior cruciate ligament (ACL); perturbations; musculoskeletal modeling; injury; muscle force; prophylactic knee braces

Prophylactic knee braces (PKBs) are off-the-shelf knee braces designed to prevent and reduce the severity of knee injuries, such as anterior cruciate ligament (ACL) tears, sustained during sporting activity.[15] Rehabilitation from ACL injuries is very expensive, and these injuries often result in pain, predisposition to osteoarthritis and other morbidities, quadriceps (QUADS) weakness and other neuromuscular deficiencies, and other long-term effects that present as a major rehabilitation challenge.[40] As a result, researchers have focused on injury prevention, such as neuromuscular training programs, that aim to strengthen and alter dangerous neuromuscular coordination patterns to prevent injuries.[26] The QUADS[11,13,35,39,43] and gastrocnemius (GAS)[1,13,19,42] are known as ACL antagonists and have been shown to contribute to ACL tears through anterior tibial translation (ATT) in cadavers, at low flexion angles. On the other hand, the hamstrings (HAMS) and soleus (SOL) are ACL agonists that provide counterbalancing forces against ATT by the QUADS and GAS.[13,19,35,42,43] Nonetheless, not all programs are effective

The Orthopaedic Journal of Sports Medicine, 9(2), 2325967120981641
DOI: 10.1177/2325967120981641

in reducing ACL injury rates,[48] with causes including lack of compliance,[46] specificity toward a particular skill level,[29] and total exposure time.[57]

PKBs are meant to combat the factors that contribute to the failure of training programs as well as targeting less modifiable risk factors such as the person's anatomy. PKBs were initially designed to protect the knee from contact injuries, such as a lateral impact to the knee joints, without compromising normal function. However, researchers have also sought to study how such an externally applied device can affect and alter muscle function and knee mechanics; they have found differences in kinematics,[7,59] kinetics,[7,59] and even muscle forces[16,23] during various movements. Despite the popularity and widespread use of these braces, research in this area has shown conflicting results.[14,23,36,50,54] The majority of the motion studies[7,59] has examined changes in kinematics and kinetics with PKBs. Two studies[16,23] have examined the effect of bracing on muscle forces using in vivo motion analysis as well as a combined in vivo/in vitro method. Moreover, two other studies[7,50] have focused on using surface electromyography (EMG) to find differences in muscle activity due to bracing during a variety of dynamic activities. Ewing et al[16] found that the HAMS and vasti (vastus lateralis, vastus medialis, vastus intermedius) muscles produced significantly greater flexion and extension torques, respectively, and greater peak muscle forces with bracing during an anticipated double-leg drop landing. Similarly, Hangalur et al[23] also examined the effects of a PKB during drop landing and found a reduction in muscle forces and ACL strain. Osternig and Robertson[50] reported significant reductions during running in the EMG activity of 6 lower-limb muscles crossing the knee and ankle with prophylactic bracing compared with control condition. Likewise, Branch et al[6] evaluated 2 types of custom-fitted braces on ACL-deficient and healthy patients and reported a decrease in QUADS and HAMS activity during the stance phase of side-step cutting. However, much is left unknown about the effect of PKB on muscle forces during other movements, specifically during unplanned, unanticipated movements.

Current literature in motion analysis evaluates knee braces during anticipated and planned motions, which may not provide us with a full understanding of their effectiveness. Over 96% of ACL injuries occur during sudden, unexpected, dynamic movements. These injuries most commonly occur during awkward dynamic body movements, landing, and perturbations.[5,21,31,33,49] Perturbations make the athlete unbalanced or lose control, which can ultimately lead to injury. As a result, examining the effect of a prophylactic brace on muscle forces in relation to ACL injury during unplanned perturbation may provide a better understanding of the effectiveness of the braces in a more realistic scenario. In turn, this will provide further insight into the efficacy of knee bracing and potential brace designs that can prevent injury. Currently, PKBs are not widely used as preventive interventions due to an absence of direct evidence of their effectiveness. The dynamic evaluation of the effectiveness of PKBs is a complex problem influenced by many variables; the braces can mechanically alter and restrict joint kinematics, or they can modify the neuromuscular activity of the lower limb.

As the effect of prophylactic bracing on neuromuscular coordination patterns is still not fully understood, we opted to examine how a PKB can perform during unanticipated movements. The purpose of this study was to examine the effect of PKBs during an unanticipated perturbation during which the wearer cannot plan for the specific movement. This will provide insight on how braces that restrict knee hyperextension and knee valgus/varus affect neuromuscular coordination and muscle forces during unanticipated perturbations. Although ACL injuries do not usually happen during walking, potentially injurious movement patterns during a disturbance to natural balance while walking could provide insight on what may be reproduced, on a higher scale, during high-impact and high-speed athletic tasks that can effectively tear the ACL.[28] Furthermore, perturbation training, at low speeds, is used for nonoperative rehabilitation after ACL injury and postoperatively after reconstruction, on the assumption that control at low speeds will transfer to higher speed activities.[24] Moreover, perturbed walking has been proven to reveal dynamic stabilization strategies; perturbations have commonly been used as models of injury mechanism to understand the neuromuscular and biomechanical responses to potentially injurious actions.[17,18,38,45,47,52,53,55] It was hypothesized that bracing would decrease ACL antagonist muscle forces (QUADS and GAS) and increase ACL agonist muscle forces (HAMS and SOL).

## METHODS

### Participant Recruitment and Preparation

Between July 2018 and November 2019, twenty healthy participants (10 men and 10 women) were recruited for this study. Participation criteria included individuals between the ages of 18 and 30 years who are regular participants in sports that included jumping, cutting, and/or pivoting >50 hours per year. Exclusion criteria included a history of or current chronic injury such as a fracture, tendon tear,

†Address correspondence to Peter Vee Sin Lee, PhD, Department of Biomedical Engineering, University of Melbourne, Victoria, 3010, Australia (email: pvlee@unimelb.edu.au).

*Department of Biomedical Engineering, University of Melbourne, Melbourne, Victoria, Australia.

Final revision submitted August 9, 2020; accepted August 24, 2020.

One or more of the authors has declared the following potential conflict of interest or source of funding: This work was mainly supported by the Melbourne International Research Scholarship and in part by The Commonwealth of Australia (represented by the Defence Science and Technology Group) through a Defence Science Partnerships Agreement. Additionally, POD Orthotic Pty provided the knee braces used in this study. AOSSM checks author disclosures against the Open Payments Database (OPD). AOSSM has not conducted an independent investigation on the OPD and disclaims any liability or responsibility relating thereto.

Ethical approval for this study was obtained from the University of Melbourne (ID No. 1034932.9).

or ligament tear to either lower extremity; no history of vestibular dysfunction; and no recent history (past 6 months) of acute lower-limb injury that necessitated medical attention. All participants signed an informed consent form approved by the relevant human research ethics committees. Each participant was fitted with a commercial PKB (K8 2.0; POD Active Pty) on both legs. The knee brace consisted of upper and lower forged carbon composite frames connected by medial and lateral hinges that included synthetic ligaments. These ligaments were inspired by the properties of native knee ligaments, providing progressive and multidirectional motion control. Proper fit was determined by measuring the width of the participant's knees using the calipers and instructions provided from the manufacturer for a comfortably tight fit. A total of 45 retroreflective markers (0.014-m diameter) were mounted on each participant's trunk, pelvis, thigh, shank, and feet using a custom marker set based on the study by Dorn et al[12] (with minor variations). Instead of a sacral cluster, 2 posterior superior iliac spine markers were placed. Calibration markers were affixed to the participant's medial and lateral femoral condyles and medial and lateral malleoli to define joint centers. The medial and lateral epicondyle markers were removed during the trials due to placement of the braces. Surface EMG was placed on the skin over the muscle bellies of the rectus femoris, vastus medialis, vastus lateralis, biceps femoris, semitendinosus, medial GAS, SOL, and tibialis anterior.

## Instrumentation

The trials were completed using a computer-assisted rehabilitation environment (Motekforce Link). This system features a semi-immersive virtual reality screen, 10-camera (MX T20 S) motion capture system (Vicon Inc), and 1 × 2–m dual-belt treadmill instrumented with 2 force plates (Motekforce Link). EMG data were sampled at 2400 Hz using a wireless EMG system (Avanti Trigno; Delsys).

## Perturbation Protocol

The patients walked at a speed of 1.1 m/s for 5 minutes before any perturbations were administered. The perturbation was a deceleration perturbation in which the belt under the dominant leg suddenly decelerated to a speed of 0 m/s with a maximum acceleration of 2.9 m/s$^2$ about 85 milliseconds after toe off.[22] This ultimately applies the perturbation at heel strike, causing the patient to experience sudden deceleration and temporarily lose balance. Three successful perturbations were recorded with a minimum of a 15-second washout period between them without the PKB (control) and another 3 with the PKB.

## Data Preprocessing

The raw (labeled) kinematic marker, ground-reaction force (GRF), and muscle EMG data from each trial were exported from VICON Nexus (Version 2.8.2; VICON, Oxford Metrics). Marker trajectories and GRF data were smoothed with 10-Hz fourth-order Butterworth filters.[32] Raw EMG data were also high-pass filtered using a fourth-order, zero-lag recursive Butterworth filter with a cutoff frequency of 20 Hz, full-wave rectified, and then low-pass filtered with a cutoff frequency of 15 Hz.[16] The resulting linear envelopes were normalized to each muscle's maximum EMG amplitude reached during stance phase.

## Rigid-Body Modeling

Rigid-body modeling was performed using OpenSim, an open-source musculoskeletal modeling software, by following a standard pipeline.[10] A 10-segment, 25 degrees-of-freedom (DOF) generic model (Gait 2392) was scaled to each patient's mass and anthropometry using the static trial data. The hip was modeled as a ball-and-socket joint with 3 DOF, while the knee was modeled as a translating hinge joint with 3 DOF (flexion-extension, rotations, and valgus-varus). The lower-limb joints were actuated by 92 musculotendon units, where each unit was represented by a Hill-type muscle in series with an elastic tendon.[56] Inverse kinematics computed the joint angles at each time instant by matching the model markers with the experimental marker trajectories such that the sum of the squared error distances between the 2 marker sets was minimized.[37] Residual reduction analysis was used to create simulations that were dynamically consistent with the experimentally recorded GRFs.[10,44] Next, static optimization solved for the individual musculotendinous forces by decomposing the net joint moments in an optimization problem that minimized the sum of the squares of all muscle activations.[9] The muscle force results from static optimization were validated against EMG; data were qualitatively consistent with the computationally estimated muscle activations (Figure 1). Each walking cycle was normalized to 100% starting from heel strike to toe-off.

## Statistical Analysis

The mean of three successful trials per condition for each patient was calculated, and the dominant leg was used for analysis. Muscles were combined into functional groups by calculating the sum of the contributions from each muscle within the group: QUADS (rectus femoris, vastus medialis, vastus intermedius, and vastus lateralis), HAMS (biceps femoris long head, biceps femoris short head, semimembranosus, and semitendinosus), and GAS (medial and lateral compartments of GAS). Statistical parametric mapping (SPM) was used to statistically compare muscle forces between men and women.[20] Specifically, an SPM paired $t$ test was used to compare the QUADS, HAMS, GAS, and SOL muscle forces during braced and unbraced conditions ($\alpha = .05$). All SPM analyses were implemented using the open-source spm1d code (Version M0.1; www.spm1d.org) in Matlab (R2019a, 9.6.0.3427; The Mathworks).

# RESULTS

The study participants consisted of 10 healthy men (age, 22.2 ± 3.50 years; height, 1.76 ± 0.07 m; mass, 71.3 ± 11.6

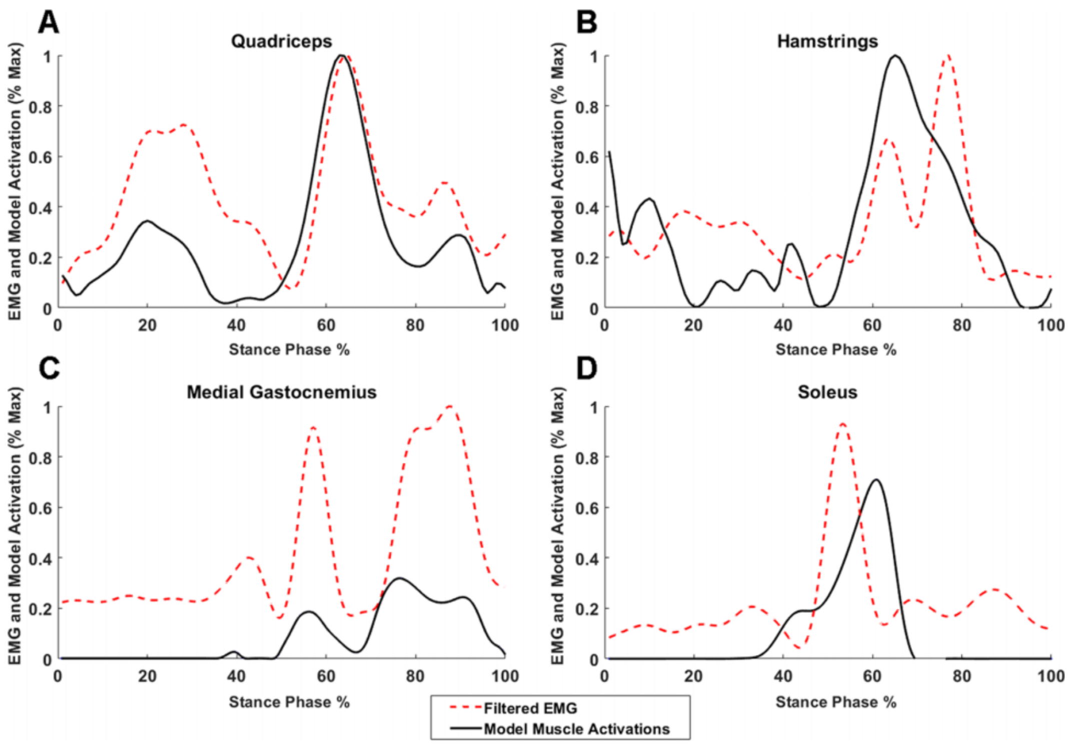


**Figure 1.** Comparison between filtered experimental EMG (red dashed line) and simulated muscle activations (solid black line) of the (A) quadriceps (rectus femoris, vastus medialis, and vastus lateralis), (B) hamstrings (biceps femoris and semitendinosus), (C) medial gastrocnemius, and (D) soleus for a representative patient. EMG and model muscle activations were normalized to their maximum values during stance phase. EMG, electromyography.

kg) and 10 healthy women (age: 22.5 ± 3.24 years; height, 1.64 ± 0.08 m; mass, 57.1 ± 7.02 kg). Compared with the unbraced condition, the brace reduced muscle forces in the QUADS, GAS, and SOL. The peak QUADS muscle force was significantly lower with versus without the brace at 49% to 60% of the stance phase (28.9 ± 12.98 vs 14.8 ± 5.06 N/kg, respectively; $P < .001$) and at 99% of the stance phase (1.7 ± 0.4 vs 3.6 ± 0.13 N/kg; $P = .049$) (Figure 2). There were no significant differences between conditions in the HAMS (Figure 3). The SOL muscle forces were significantly lower for the braced condition at 25% of the stance phase (1.9 ± 1.7 N/kg [braced] vs 4.6 ± 3.4 N/kg [unbraced]; $P = .031$) and at 39% (1.9 ± 1.4 N/kg [braced] vs 5.3 ± 5.6 N/kg [unbraced]; $P = .007$) (Figure 4). In the GAS, the only significant difference between conditions was at 97% to 100% of the stance phase, during which the braced condition portrayed a smaller GAS peak force (0.23 ± 0.13 vs 1.4 ± 1.1 N/kg for unbraced; $P = .024$) (Figure 5). There were no significant differences in knee joint angles between braced and unbraced conditions (Figure 6).

## DISCUSSION

The brace significantly reduced QUADS peak forces from 28.9 to 14.8 N/kg at $P < .001$, which is in line with our hypothesis. Moreover, we hypothesized an increase in HAMS force with the PKB. However, our study showed no significant differences between braced and unbraced conditions during walking with perturbations. Looking at the lower leg muscles, the GAS's force—which is an ACL agonist—was reduced in the last 3% of the gait cycle from 1.4 to 0.23 N/kg with bracing. This does not fulfill our hypothesis, as a reduction occurred in the terminal stance phase, whereas a reduction in the peak force was desired. Conversely, we hypothesized a force increase in the SOL—an ACL agonist; however, the brace had no effect on the highest peak force. Instead, the brace slightly delayed the SOL force production, causing lower SOL forces with the brace to be observed at 25% and 39% of the stance phase.

The reduction in QUADS peak forces from 28.9 to 14.8 N/kg at $P < .001$ agreed with our hypothesis. This is a beneficial change, as the QUADS can strain the ACL when they contract at low flexion angles and have been shown to induce ACL tears in cadavers.[11,13,35,39,43] Cadaveric and in vivo studies[11,13,30,35,39] have shown that QUADS contraction has an effect on ACL strain between 0° to 45° of flexion. The smaller the knee flexion, the greater the ACL strain with the same QUADS force. This was evident in a study by Beynnon et al[4], in which they performed a static in vivo study where they applied 100 N of anterior shear load at 30° of knee flexion; 2 of 7 braces provided strain-shielding effects. However, when this was compounded with isometric QUADS contraction, none of the braces were able to provide protection to the ACL. Similarly, Branch et al[6] also conducted an in vivo study that applied

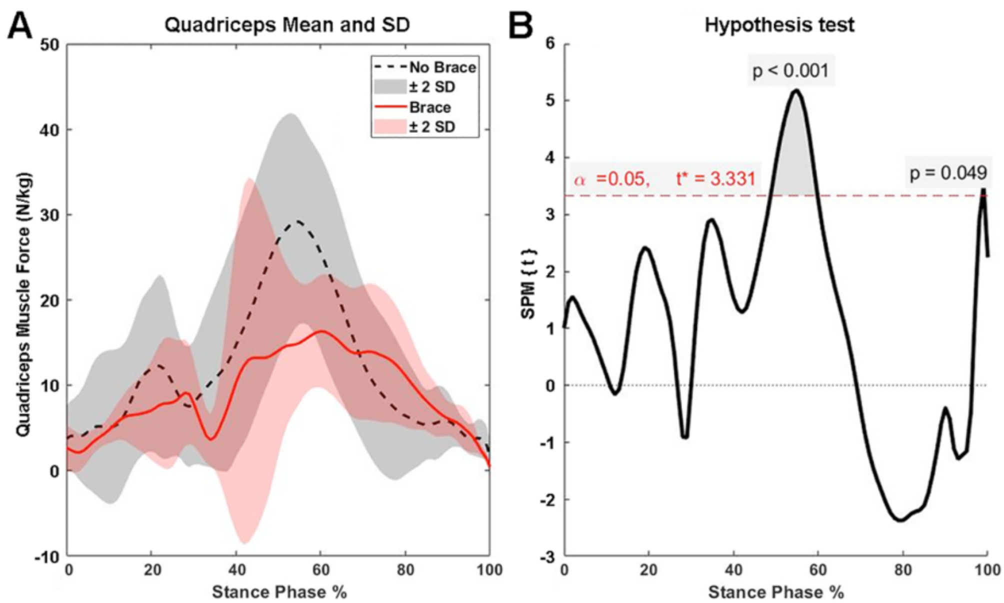


**Figure 2.** (A) Mean quadriceps muscle forces with no brace (dashed black line) ± 2 SD (gray band) and with a prophylactic knee brace (solid red line) ± 2 SD (red band) during perturbed walking. (B) The paired *t* test statistic SPM {t} (solid black line). The critical threshold of 3.331 (red dashed line) was exceeded at time = 49% to 60% and at 99% of the stance phase with a suprathreshold cluster probability value of $P < .001$ and $P = .049$, respectively, indicating a significantly lower muscle force peak with the brace than without. SPM, statistical parametric mapping.

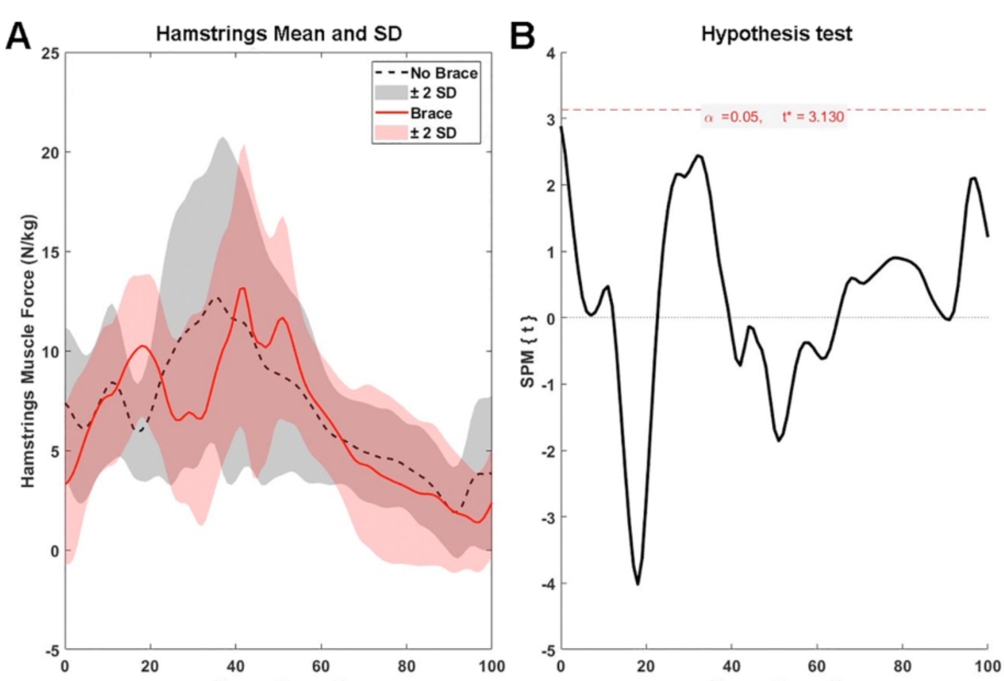


**Figure 3.** (A) Mean hamstrings muscle forces with no brace (black dashed line) ± 2 SD (gray band) and with a prophylactic knee brace (solid red line) ± 2 SD (red band) during perturbed walking. (B) The paired *t* test statistic SPM {t} (solid black line). The critical threshold of 3.130 (red dashed line) was not exceeded at any instance, indicating that there are no significant differences between braced and unbraced conditions for hamstrings force. SPM, statistical parametric mapping.

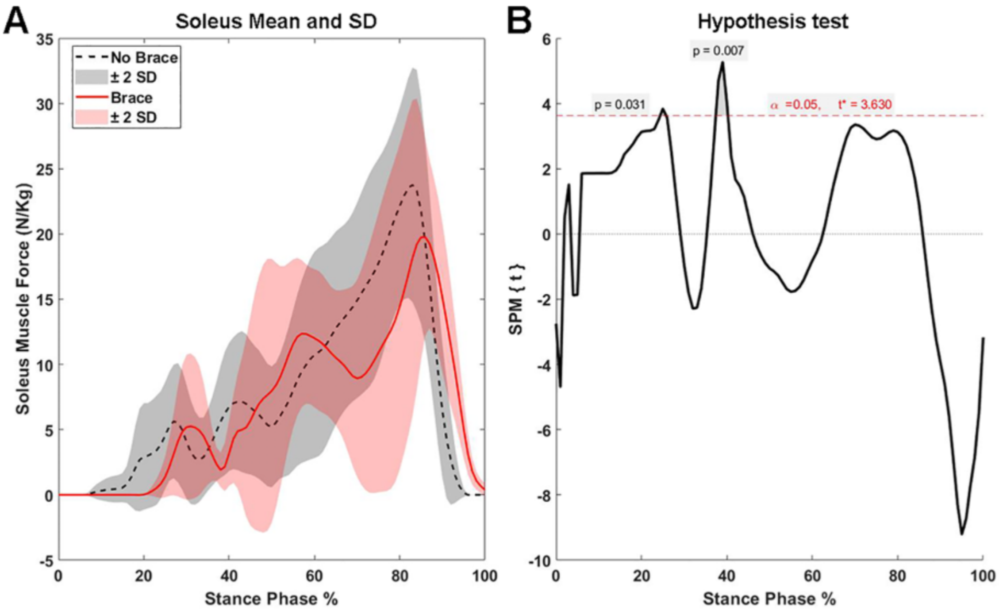


**Figure 4.** (A) Mean soleus muscle forces with no brace (black dashed line) ± 2 SD (gray band) and with a prophylactic knee brace (solid red line) ± 2 SD (red band) during perturbed walking. (B) The paired *t* test statistic SPM {t} (solid black line). The critical threshold of 3.630 (red dashed line) was exceeded at time = 25% and at 39% of the stance phase with suprathreshold cluster probability value of *P* = .031 and *P* = .007, respectively. SPM, statistical parametric mapping.

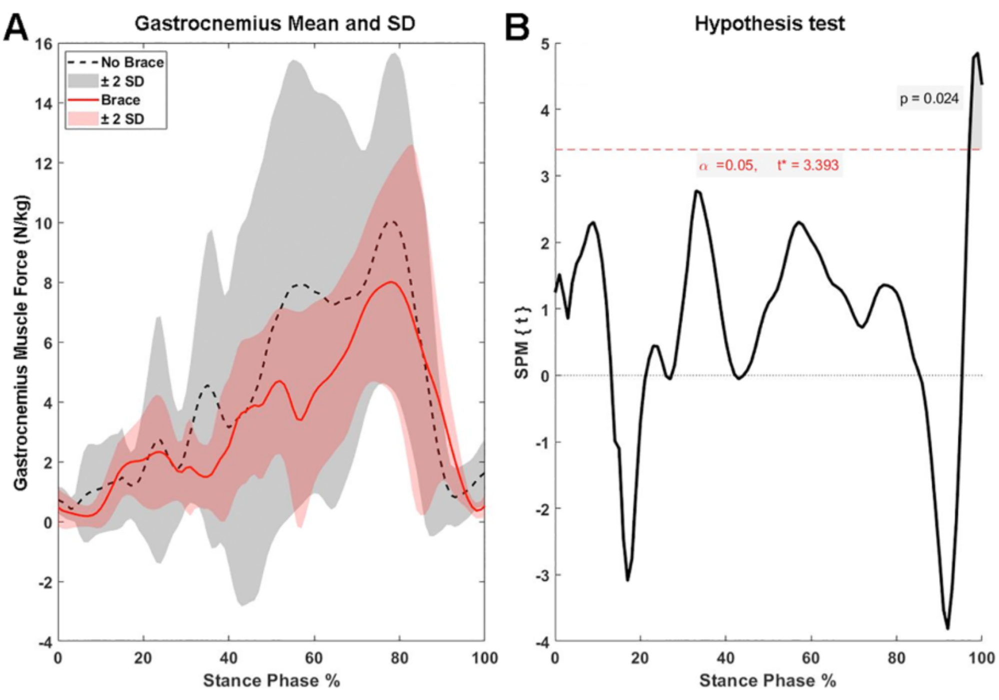


**Figure 5.** (A) Mean gastrocnemius muscle forces with no brace (black dashed line) ± 2 SD (gray band) and with a prophylactic knee brace (solid red line) ± 2 SD (red band) during perturbed walking. (B) The paired *t* test statistic SPM {t} (solid black line). The critical threshold of 3.393 (red dashed line) was exceeded at time = 97% to 100% of the stance phase with a suprathreshold cluster probability value of *P* = .024, indicating a significantly lower muscle force peak with the brace than without. SPM, statistical parametric mapping.

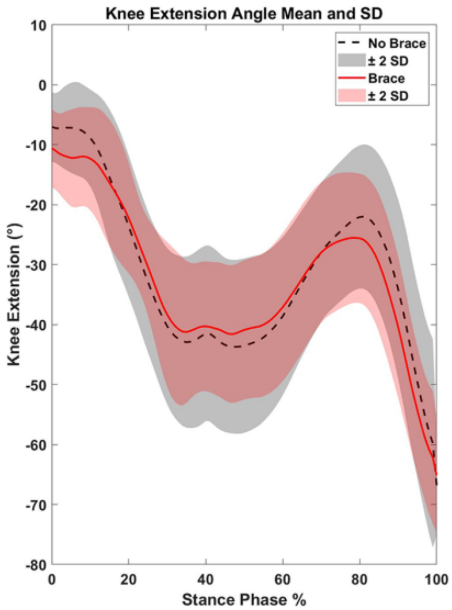


**Figure 6.** Mean knee extension angle (positive) and flexion (negative) with no brace (dashed black line) ± 2 SD (gray band) and with brace (solid red line) ± 2 SD (red band) during perturbed walking.

anterior shear load with active QUADS force and showed similar results. These studies[4,6,23] show that PKBs do not reduce ACL injury by solely providing mechanical stability; changes in neuromuscular coordination and native firing patters are also required to reduce ACL strain.

Looking at the QUADS peak in our study, the peak occurred between 30° and 40° of knee flexion angle (Figure 6) with the brace, which falls within the "low flexion" angle range of 0° to 45°. This decrease in QUADS force from our results is in agreement with the study by Hangalur et al,[23] where the brace reduced peak QUADS force from 5653 to 4666 N during a drop-landing motion. Similarly, our findings are in line with Osternig and Robertson[50] and Branch et al,[6] who both reported significant reductions in QUADS EMG while wearing a knee brace compared with an unbraced condition during running and side-step cutting, respectively. On the other hand, Ewing et al[16] illustrated the opposite findings during drop landing. In that study, the QUADS showed an increase in force in the braced condition compared with the unbraced condition. Although Ewing et al[16] studied a similar brace made by the same company, opposite results were observed in relation to QUADS forces. This is likely the result of the difference in movement or the absence of the anticipation factor in our study compared with their planned movement study.

The HAMS are ACL agonists that can decrease ACL strain.[13,19,35,43] Therefore, we hypothesized an increase in HAMS force with the PKB. However, our study showed no significant differences between braced and unbraced conditions during walking with perturbations. On the other hand, a few studies examining muscle forces[23] and EMG[6,50] reported a reduction in HAMS muscle force and activity. Ewing et al,[16] however, illustrated an increase in HAMS forces during the braced condition in landing. This disparity in results could be attributed to the variations in movements (ie, running,[50] side-step cutting,[6] drop-landing,[16,23] and walking with perturbation), which may evoke different responses. Nonetheless, the studies conducted by Hangalur et al[23] and Ewing et al[16] were drop-landing and showed opposite results. This suggests that all PKBs may not work equally, instead altering the muscle activations and forces in a similar pattern.

The GAS force, which is an ACL agonist, was reduced from 1.4 to 0.23 N/kg with bracing in the final 3% of the gait cycle. Although a reduction in the GAS force is desired as it increases ACL strain by inducing ATT,[1,13,19,42] the reduction happened at the end of the stance phase and not during the large peak GAS force between 60% and 80% of the stance phase. This does not fulfill our hypothesis, as a reduction in the peak force was desired. Correspondingly, the study by Ewing et al[16] did not find any GAS force differences during the drop landings. Yet, the study by Osternig and Robertson[50] depicted a significant reduction in GAS EMG activity during running. Again, this might be credited to the differences in the PKB brace design.

Conversely, the SOL is an ACL agonist; it induces posterior tibial translation at all flexion angles, with the greatest translations happening at 50° of flexion.[13] However, the brace had no effect on the highest peak force. Instead, the brace slightly delayed the SOL force production, causing lower SOL forces with the brace to be observed at 25% and 39% of the stance phase. Similarly, neither Ewing et al[16] nor Osternig and Robertson[50] found any differences in SOL muscle forces and EMG during landing and running, respectively.

Our results do not fully align with the literature describing examined muscle forces during planned landing.[16,23] This is likely the result of differences in movement (that is, drop landing versus perturbation) or because of differences in the biomechanics between planned and unplanned movements, as shown by several studies.[3,8,34,51,58] These differences suggest that bracing might not consistently provide favorable muscle force patterns during a wide range of movements. Currently, PKBs are not widely used as preventive interventions due to an absence of direct evidence of their effectiveness. The dynamic evaluation of the effectiveness of PKBs is a multifaceted challenge affected by many variables; the braces can mechanically change and restrict joint angles, or they can modify the neuromuscular activity of the lower limb.[14] The braces used in this study are meant to restrict hyperextension of the knee using mechanical stoppers, reduce knee valgus/varus, and add overall joint stability using the synthetic ligaments. Nonetheless, because of the complexity of the braces' mechanisms of action, it is difficult to precisely compare our results with other prophylactic brace studies. Other studies have used different braces, conducted different motions, and had patients perform planned and anticipated movements.[3,8,34,51,58]

Since the effect of prophylactic bracing on neuromuscular coordination patterns is still not fully understood, we opted to examine how a PKB can perform during unanticipated movements. The purpose of this study was to examine the effect of a specific kind of PKB during an unanticipated perturbation, where the wearer cannot anticipate the specific movement. This provided us with insight on how braces that restrict knee hyperextension and knee valgus/varus affect neuromuscular coordination and muscle forces during unanticipated perturbations. Considering the disparity of epidemiological research on PKBs, determining the extent to which different kinds of PKBs alter lower-limb function during a variety of movements will help us in understanding the efficacy of preventive bracing with similar properties.

### Limitations of the Study and Future Work

Musculoskeletal modeling is a great tool to predict in vivo muscle forces without invasive methods; nonetheless, it has some limitations. First, the musculoskeletal models were scaled based on a generic model. Using real imaging data from the subjects, such as magnetic resonance imaging, would create a more accurate model. Second, we assumed that the inverse dynamics joint moments were satisfied by muscle forces alone and therefore did not include the effect of the brace, which may also exert a moment about the knee joint, directly into the model.[16] Unlike functional knee braces, which are specifically designed to off-load certain compartments of the tibiofemoral joint, the exact mechanism of prophylactic knee bracing is unknown. Therefore, nonmuscular contributions were assumed to be negligible compared with the moments generated by the muscles. Third, static optimization methods do not include activation dynamics and have been critiqued for their failure to predict cocontraction. Despite this, static optimization in musculoskeletal modeling is a powerful technique to quantify how muscles function and has been used and validated by many researchers.[1,2,16,25,41] Additionally, the perturbations were introduced during walking, which is unlikely to cause injury. Nevertheless, perturbed walking has been proven to reveal dynamic stabilization strategies; perturbations have commonly been used as models of injury mechanism to understand the neuromuscular and biomechanical responses to potentially injurious actions.[17,18,45,47,52,53,55] Many studies[17,18,22,45,47,52,53,55] have performed perturbations during walking and were able to observe differences. For example, Hurd et al[28] studied the kinematic differences in normal and perturbed walking between male and female athletes. They concluded that female athletes exhibited characteristics during normal and perturbed walking, which may increase the risk of an ACL injury, and the repetition of these harmful patterns during provocative athletic movements may lead to injury. Furthermore, potentially injurious movement patterns during a disturbance to natural balance while walking could provide insight on what may be reproduced, on a higher scale, during high-impact and high-speed athletic tasks that can effectively tear the ACL.[28] Moreover, the ACL ligament strain was not directly measured; therefore, it is difficult to accurately discern the impact of bracing on the ACL. Instead, we can infer what might happen to the ACL based on surrounding muscle forces. Finally, the perturbations may have introduced an inertial artifact in the sagittal plane moment component of the GRF. Although the compensatory mechanism developed by Hnat and van den Bogert[27] has shown great success in in high speeds (96.75%), the model was able to accurately predict only the pitch moment at the speed of 1.2 m/s by 72.06%. Given that our study was conducted at 1.1 m/s, which is at a slower speed than in the study by Hnat and van den Bogert, we assumed the artifact would be smaller and even more inaccurately compensated for using an inertial compensation method. Therefore, we opted to not compensate for the error, as the predicted output underestimates the pitch moment. Despite these limitations of musculoskeletal modeling, it remains a powerful tool in understanding biomechanics without invasive procedures. Future work can build upon these findings, such as examining the effects of other forms of perturbations and unplanned movements using PKBs.

## CONCLUSION

This study showed that PKBs that restrict knee hyperextension and knee valgus/varus altered neuromuscular patterns, thereby resulting in a reduction of QUADS force. Further research may include examining the effects of other types of perturbations and unplanned movements using PKBs. Understanding the way PKBs alter muscle function and knee mechanics can provide invaluable information that will help in decision-making regarding their use.

## ACKNOWLEDGMENT

The authors thank POD Orthotic for providing the knee braces used in this study. The authors also acknowledge Cameron Patrick from the Statistical Consulting Centre at the University of Melbourne for his assistance with the statistical analysis and data interpretation.

## REFERENCES

1. Adouni M, Shirazi-Adl A, Marouane H. Role of gastrocnemius activation in knee joint biomechanics: gastrocnemius acts as an ACL antagonist. *Comput Methods Biomech Biomed Engin*. 2016;19(4):376-385.
2. Anderson FC, Pandy MG. Static and dynamic optimization solutions for gait are practically equivalent. *J Biomech*. 2001;34(2):153-161.
3. Besier TF, Lloyd DG, Ackland TR, Cochrane JL. Anticipatory effects on knee joint loading during running and cutting maneuvers. *Med Sci Sports Exerc*. 2001;33(7):1176-1181.
4. Beynnon BD, Pope MH, Wertheimer CM, Johnson RJ, Fleming BC, Howe CE. The effect of functional knee braces on anterior cruciate ligament strain, in vivo. *J Bone Joint Surg Am*. 1992;74:1298-1312.
5. Boden BP, Torg JS, Knowles SB, Hewett TE. Video analysis of anterior cruciate ligament injury abnormalities in hip and ankle kinematics. *Am J Sports Med*. 2009;37(2):252-259.
6. Branch T, Hunter R, Reynolds P. Controlling anterior tibial displacement under static load: a comparison of two braces. *Orthopedics*. 1988;11(9):1249-1252.
7. Branch TP, Hunter R, Donath MMM, et al. Dynamic EMG analysis of anterior cruciate deficient legs with and without bracing during cutting. *Am J Sports Med*. 1989;17(1):35-41.

8. Brown TN, Palmieri-Smith RM, McLean SG. Sex and limb differences in hip and knee kinematics and kinetics during anticipated and unanticipated jump landings: implications for anterior cruciate ligament injury. *Br J Sports Med*. 2009;43(13):1049-1056.
9. Crowninshield RD, Brand RA. A physiologically based criterion of muscle force prediction in locomotion. *J Biomech*. 1981;14(11): 793-801.
10. Delp SL, Anderson FC, Arnold AS, et al. OpenSim: open source to create and analyze dynamic simulations of movement. *IEEE Trans Biomed Eng*. 2007;54(11):1940-1950.
11. DeMorat G, Weinhold P, Blackbur T, Chudik S, Garrett W. Aggressive quadriceps loading can induce noncontact anterior cruciate ligament injury. *Am J Sports Med*. 2004;32(2):477-483.
12. Dorn TW, Schache AG, Pandy MG. Muscular strategy shift in human running: dependence of running speed on hip and ankle muscle performance. *J Exp Biol*. 2012;215(11):1944-1956.
13. Elias JJ, Faust AF, Chu YH, Chao EY, Cosgarea AJ. The soleus muscle acts as an agonist for the anterior cruciate ligament: an in vitro experimental study. *Am J Sports Med*. 2003;31(2):241-246.
14. Erickson AR, Yasuda K, Beynnon B, Johnson R, Pope M. An in vitro dynamic evaluation of prophylactic knee braces during lateral impact loading. *Am J Sports Med*. 1993;21(1):26-35.
15. Ewing KA, Begg RK, Galea MP, Lee PVS. Effects of prophylactic knee bracing on lower limb kinematics, kinetics, and energetics during double-leg drop landing at 2 heights. *Am J Sports Med*. 2016;44(7): 1753-1761.
16. Ewing KA, Fernandez JW, Begg RK, Galea MP, Lee PVS. Prophylactic knee bracing alters lower-limb muscle forces during a double-leg drop landing. *J Biomech*. 2016;49(14):3347-3354.
17. Ferber R, Osternig LR, Woollacott MH, Wasielewski NJ, Lee JH. Gait perturbation response in chronic anterior cruciate ligament deficiency and repair. *Clin Biomech (Bristol, Avon)*. 2003;18(2): 132-141.
18. Ferber R, Osternig LR, Woollacott MH, Wasielewski NJ, Lee JH. Reactive balance adjustments to unexpected perturbations during human walking. *Gait Posture*. 2002;16(3):238-248.
19. Fleming BC, Renstrom PA, Ohlen G, et al. The gastrocnemius muscle is an antagonist of the anterior cruciate ligament. *J Orthop Res*. 2001; 19(6):1178-1184.
20. Friston KJ. *Statistical Parametric Mapping: The Analysis of Functional Brain Images*. 2nd ed. Elsevier; 2007.
21. Griffin L. Understanding and preventing noncontact anterior cruciate ligament injuries. *Am J Sports Med*. 2006;34(9):1512-1532.
22. Haddara R, Harandi VJ, Lee PVS. Anterior cruciate ligament agonist and antagonist muscle force differences between males and females during perturbed walking. *J Biomech*. 2020;110:109971.
23. Hangalur G, Brenneman E, Nicholls M, Bakker R, Laing A, Chandrashekar N. Can a knee brace reduce the strain in the anterior cruciate ligament? A study using combined in vivo/in vitro method. *Prosthet Orthot Int*. 2016;40(3):394-399.
24. Hartigan E, Axe MJ, Snyder-Mackler L. Perturbation training prior to ACL reconstruction improves gait asymmetries in non-copers. *J Orthop Res*. 2009;27(6):724-729.
25. Heintz S, Gutierrez-Farewik EM. Static optimization of muscle forces during gait in comparison to EMG-to-force processing approach. *Gait Posture*. 2007;26(2):279-288.
26. Hewett T, Lindenfeld T, Riccobene J, Noyes F. The effect of neuromuscular training on the incidence of knee injury in female athletes. A prospective study. *Am J Sports Med*. 1999;27(6):699-706.
27. Hnat SK, van den Bogert AJ. Inertial compensation for belt acceleration in an instrumented treadmill. *J Biomech*. 2014;47(15): 3758-3761.
28. Hurd WJ, Chmielewski TL, Davis IS, et al. Differences in normal and perturbed walking kinematics between male and female athletes. *Clin Biomech (Bristol, Avon)*. 2004;19(5):465-472.
29. Junge A, Rösch D, Peterson L, Graf-Baumann T, Dvorak J. Prevention of soccer injuries: a prospective intervention study in youth amateur players. *Am J Sports Med*. 2002;30(5):652-659.
30. Kain CC, McCarthy JA, Arms S, et al. An in vivo analysis of the effect of transcutaneous electrical stimulation of the quadriceps and hamstrings on anterior cruciate ligament deformation. *Am J Sports Med*. 1988;16(2):147-152.
31. Koga H, Nakamae A, Shima Y. Mechanisms for noncontact anterior cruciate ligament injuries knee joint kinematics in 10 injury situations from female team handball and basketball. *Am J Sports Med*. 2010; 38(11):2218-2225.
32. Kristianslund E, Krosshaug T, van den Bogert AJ. Effect of low pass filtering on joint moments from inverse dynamics: implications for injury prevention. *J Biomech*. 2012;45(4):666-671.
33. Krosshaug T, Nakamae A, Boden BP. Mechanisms of anterior cruciate ligament injury in basketball: video analysis of 39 cases. *Am J Sports Med*. 2007;35(3):359-367.
34. Landry SC, McKean KA, Hubley-Kozey CL, Stanish WD, Deluzio KJ. Neuromuscular and lower limb biomechanical differences exist between male and female elite adolescent soccer players during an unanticipated run and crosscut maneuver. *Am J Sports Med*. 2007; 35(11):1901-1911.
35. Li G, Rudy TW, Sakane M, Kanamori A, Ma CB, Woo SLY. The importance of quadriceps and hamstring muscle loading on knee kinematics and in-situ forces in the ACL. *J Biomech*. 1999;32(4):395-400.
36. Lin CF, Liu H, Garrett WE, Yu B. Effects of a knee extension constraint brace on selected lower extremity motion patterns during a stop-jump task. *J Appl Biomech*. 2008;24(2):158-165.
37. Lu T, Connor JJO. Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. *J Biomech*. 1999;32:129-134.
38. Malfait B, Staes F, De Vries A, et al. Dynamic neuromuscular control of the lower limbs in response to unexpected single-planar versus multiplanar support perturbations in young, active adults. *PLoS One*. 2015; 10(7):1-17.
39. Markolf KL, Gorek JF, Kabo JM, Shapiro MS. Direct measurement of resultant forces in the anterior cruciate ligament. *J Bone Joint Surg Am*. 1990;72(4):557-567.
40. Moeller J, Lamb M. Anterior cruciate ligament injuries in female athletes: why are women more susceptible. *Phys Sportsmed*. 1997;25(4): 31-47.
41. Mokhtarzadeh H, Ewing K, Janssen I, Yeow CH, Brown N, Lee PVS. The effect of leg dominance and landing height on ACL loading among female athletes. *J Biomech*. 2017;60:181-187.
42. Mokhtarzadeh H, Yeow CH, Hong Goh JC, Oetomo D, Malekipour F, Lee PVS. Contributions of the soleus and gastrocnemius muscles to the anterior cruciate ligament loading during single-leg landing. *J Biomech*. 2013;46(11):1913-1920.
43. More RC, Karras BT, Neiman R, Fritschy D, Woo SL, Daniel DM. Hamstrings—an anterior cruciate ligament protagonist. An in vitro study. *Am J Sports Med*. 1993;21(2):231-237.
44. Morgan KD, Donnelly CJ, Reinbolt JA. Elevated gastrocnemius forces compensate for decreased hamstrings forces during the weight-acceptance phase of single-leg jump landing: implications for anterior cruciate ligament injury risk. *J Biomech*. 2014;47(13): 3295-3302.
45. Myers JB, Riemann BL, Hwang JH, Fu FH, Lephart SM. Effect of peripheral afferent alteration of the lateral ankle ligaments on dynamic stability. *Am J Sports Med*. 2003;31(4):498-506.
46. Myklebust G, Engebretsen L, Brækken IH, Skjølberg A, Olsen OE, Bahr R. Prevention of anterior cruciate ligament injuries in female team handball players: a prospective intervention study over three seasons. *Clin J Sport Med*. 2003;13(2):71-78.
47. Noonan B, Wojtys EM. Gender differences in muscular protection of the knee. *J Bone Joint Surg Am*. 2003;85(5):782-789.
48. Noyes FR, Barber Westin SD. Anterior cruciate ligament injury prevention training in female athletes: a systematic review of injury reduction and results of athletic performance tests. *Sports Health*. 2012; 4(1):36-46.
49. Olsen OE, Myklebust G, Engebretsen L, Bahr R. Injury mechanisms for anterior cruciate ligament injuries in team handball: a systematic video analysis. *Am J Sports Med*. 2004;32(4):1002-1012.

50. Osternig LR, Robertson RN. Effects of prophylactic knee bracing on lower extremity joint position and muscle activation during running. *Am J Sports Med*. 1993;21(5):733-737.
51. Sell TC, Ferris CM, Abt JP, et al. The effect of direction and reaction on the neuromuscular and biomechanical characteristics of the knee during tasks that simulate the noncontact anterior cruciate ligament injury mechanism. *Am J Sports Med*. 2006;34(1):43-54.
52. Shultz SJ, Carcia CR, Gansneder BM, Perrin DH. The independent and interactive effects of navicular drop and quadriceps angle on neuromuscular responses to a weight-bearing perturbation. *J Athl Train*. 2006;41(3):251-259.
53. Shultz SJ, Perrin DH, Adams JM, Arnold BL, Gansneder BM, Granata KP. Assessment of neuromuscular response characteristics at the knee following a functional perturbation. *J Electromyogr Kinesiol*. 2000;10(3):159-170.
54. Sitler M, Ryan J, Hopkinson W, et al. The efficacy of a prophylactic knee brace to reduce knee injuries in football. A prospective, randomized study at West Point. *Am J Sports Med*. 1989;18(3):310-315.
55. Teixeira Da Fonseca S, Silva PLP, Ocarino JM, Guimaràes RB, Oliveira MTC, Lage CA. Analyses of dynamic co-contraction level in individuals with anterior cruciate ligament injury. *J Electromyogr Kinesiol*. 2004;14(2):239-247.
56. Thelen DG, Anderson FC, Delp SL. Generating dynamic simulations of movement using computed muscle control. *J Biomech*. 2003;36(3):321-328.
57. Umberger BR, Rubenson J. Understanding muscle energetics in locomotion: new modeling and experimental approaches. *Exerc Sport Sci Rev*. 2011;39(2):59-67.
58. Weinhandl JT, Earl-Boehm JE, Ebersole KT, Huddleston WE, Armstrong BSR, O'Connor KM. Anticipatory effects on anterior cruciate ligament loading during sidestep cutting. *Clin Biomech (Bristol, Avon)*. 2013;28(6):655-663.
59. Yu B. Immediate effects of a knee brace with a constraint to knee extension on knee kinematics and ground reaction forces in a stop-jump task. *Am J Sports Med*. 2004;32(5):1136-1143.