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Speeding Up or Slowing Down: The Effect of Decoupling Speed on Knee Biomechanics and Limb Loading During Split-Belt Treadmill Training in Persons With ACL Reconstruction

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Background: Surgical limb underloading is a common biomechanical adaptation after anterior cruciate ligament reconstruction (ACLR) and has been linked to early degenerative changes in knee cartilage, which are considered precursors to posttraumatic osteoarthritis. Split-belt treadmill training is an emerging rehabilitation approach that modifies load through asymmetric gait behavior, in which one limb walks faster than the other. While previous research has indicated that split-belt treadmill training can modify limb loading, its effects on post-ACLR biomechanics remain unexplored.

Purpose/Hypothesis: The purpose of this study was to examine the effects of decoupling speed on knee biomechanics and limb loading and model their relationship. It was hypothesized that at faster decoupling speeds, knee loads would increase, and at slower speeds, they would decrease.

Study Design: Controlled laboratory study.

Methods: Knee joint biomechanics were evaluated while 24 participants (15 females; mean age, 23.5 ± 6.5 years; mean height, 1.72 ± 0.08 m; mean mass, 75.61 ± 13.83 kg; mean postoperative time, 7.94 ± 1.74 months) with ACLR walked on an instrumented treadmill (2000 Hz) synced with a 12-camera motion capture system (200 Hz). Participants completed 5 minutes of baseline walking at 1.1 m/s on tied treadmill belts. Afterward, the authors manipulated the speed of the belt under the ACL-reconstructed leg with 8 randomized 5-minute decoupled speed configurations (ACL-reconstructed limb at 30%-170% of 1.1 m/s in 20% increments) with 2 minutes of tied-belt walking between each speed. Bilateral sagittal plane knee moments, angles, and vertical ground-reaction forces were calculated. Statistical parametric mapping was used to evaluate the effects of decoupling speed on the dependent variables during the stance phase of the gait.

Results: A clear dose-response relationship between decoupling speed and knee/limb loading was found. Specifically, faster decoupling speeds (130%-170% of 1.1m/s) generally increased peak loads and slower speeds (30% and 50% of 1.1 m/s) decreased peak loads during early stance compared with tied-belt walking for both limbs. In contrast, slower decoupling speeds (30%-90% of 1.1 m/s) led to higher bilateral knee/limb loads at midstance, while faster decoupling speeds (130%-170% of 1.1 m/s) resulted in less knee/limb loading during midstance.

Conclusion: These findings suggest that split-belt treadmill training offers a promising method for modulating knee/limb loading post-ACLR.

Clinical Relevance: Split-belt treadmill training could be a viable intervention to target the loading asymmetry that is prevalent in persons post-ACLR.

Keywords: knee; rehabilitation; gait retraining; asymmetry

Anterior cruciate ligament (ACL) injuries are both common and costly, with about 300,000 tears occurring each year in the United States,[9] resulting in an estimated annual lifetime financial burden of approximately \$8 billion.[15] The standard of care treatment is surgical reconstruction followed by approximately 9 months of aggressive rehabilitation to restore knee stability and return patients to activity. Despite this arduous treatment regimen, the outcomes from ACL reconstruction (ACLR) are suboptimal as >50% of ACL-reconstructed limbs show signs of posttraumatic osteoarthritis as early as 5 years after the injury.[13]

The American Journal of Sports Medicine
2025;53(11):2591–2599
DOI: 10.1177/03635465251365035
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Abnormal gait biomechanics are present early after ACLR and persist beyond one's return to activity. Of particular concern is the underloading behavior (ie, decreases in sagittal plane knee moments and peak vertical ground-reaction forces [vGRFs]) that occurs in the ACL-reconstructed limb after surgery.[3-5,12,17,22] This underloading has been linked to early deleterious changes in knee cartilage, which are considered precursors to posttraumatic osteoarthritis.[16,21,24,25] As such, interventions targeting underloading in individuals with ACLR are needed, as they have the potential to preserve long-term knee joint health.

Split-belt treadmill training is an emerging rehabilitation/gait retraining approach in which treadmill belts are decoupled so that one limb walks faster than the other, resulting in an asymmetric locomotor behavior during training. In healthy individuals, split-belt treadmill walking has been shown to significantly increase (57% from baseline) knee moment impulses during the braking phase of gait, suggesting that it may be an effective method to increase knee loading.[19] Similarly, a study on those who survived stroke with asymmetric loading patterns showed a 25% improvement in knee loading after split-belt training.[1] Limited research has assessed how split-belt training can improve gait biomechanics after ACLR; however, a recent study revealed that loading rates become more symmetrical after about 8 minutes of split-belt walking about 3.5 years after ACLR.[10] Additionally, this study showed that after completing the split-belt training, when the treadmill belts were returned to the same speed (ie, tied-belt walking), participants with ACLR were able to retain symmetrical loading rates for the full 2-minute measurement period (ie, the amount of time the aftereffects were measured). While these results are promising and support that split-belt training may be beneficial, more research is necessary to support its efficacy in persons with ACLR. In particular, research is needed to determine what decoupling speed is required to result in increased knee/limb loading in the ACL-reconstructed leg.

A review of the literature shows that most studies arbitrarily implement split-belt treadmill training with a 2:1 belt ratio[19,20] (one limb walking at 50% of the speed of the other limb) that was developed in neurological populations.[1,14] This approach, however, fails to consider that there is likely a dose-response effect of decoupling speed (ie, as the percentage of decoupling between limbs changes, the magnitude of the effect on knee/limb loading likely changes). In fact, recent work in individuals with ACLR has shown that having the overloading limb walk at a slower speed (1.0 m/s) can lead to more symmetrical loading rates, when compared with having the overloading limb walk at a faster speed (1.5 m/s).[10] However, this work and that of others still arbitrarily selected their belt ratio and, as such, failed to consider a dose-response effect. The current study aimed to fill this gap by systematically examining different belt decoupling speeds in individuals with ACLR to determine which speed is most effective at improving underloading in the ACL knee/limb. Testing a variety of belt speeds would also allow us to assess the feasibility of implementing the studied decoupling speeds into ACL rehabilitation (eg, are patients able to comfortably and safely walk at the speeds that are ideal for improving knee loading?).

Therefore, the purpose of this study was to determine if decoupling speed affected knee/limb loading and to establish a model that described the relationship between decoupling speed ratios and knee/limb loading in individuals with ACLR. We hypothesized that there would be a dose-response relationship between decoupling speed ratios and knee/limb loading such that early-stance knee/limb loading in the ACL-reconstructed limb would increase as the speed on the decoupled/ACL-reconstructed limb belt increased. Furthermore, we expected that slower decoupling speeds would lead to a more sustained, less dynamic loading pattern in midstance. Lastly, we hypothesized that faster speeds would result in larger knee flexion angles in early stance. Understanding how split-belt treadmill training affects loading metrics in individuals with ACLR is important for clinicians, as it may provide preliminary evidence supporting its potential as a gait retraining approach to combat the common issue of underloading—a factor that has been theorized to contribute to the posttraumatic osteoarthritis in this population.

METHODS

Study Participants

A total of 24 participants who had undergone ACLR at a single medical center were recruited to participate in this study

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Submitted January 9, 2025; accepted July 2, 2025.
One or more of the authors has declared the following potential conflict of interest or source of funding: Research reported in this publication was supported by the National Institute of Arthritis and Musculoskeletal and Skin Diseases of the National Institutes of Health under award No. 5R21AR082643. The content is solely the responsibility of the authors and does not necessarily represent the official views of the National Institutes of Health. 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.

Figure 1. Retroreflective marker placement. The left photograph shows the anterior view, and the right photograph shows the posterior view.

(sex, 15 females, 9 males [all cisgender]; mean age, 23.5 ± 6.5 years; mean height, 1.72 ± 0.08 m; mean mass, 75.61 ± 13.83 kg). Additional demographic information about these participants can be found in Appendix 1 (available in the online version of this article). Participants were required to have had their reconstruction using an autograft (22 patellar tendon, 1 hamstring tendon, and 1 quadriceps tendon) and needed to be between 6 and 12 months after surgery (mean postoperative time, 7.94 ± 1.74 months) to participate. Potential participants were excluded from taking part if they had any previous ACL injury, any previous knee surgery, or a bony fracture accompanying the ACL injury; had sustained a knee dislocation; and/or had a staged or multiligamentous ACLR. The sample size for this study was determined using an a priori power analysis (GLIMMPSE Version 3.0) based on pilot data (n = 5), which showed that the peak knee extension moment (pKEM) varied from 0.23 N·m/kg·m (at 30% decoupling speed) to 0.45 N·m/kg·m (at 170% decoupling speed), with a grand mean of 0.32 N·m/kg·m and a standard deviation of 0.09 N·m/kg·m. Assuming a conservate effect size of 50% of the mean differences observed in the pilot data (mean scale factor, 0.5), a conservative correlation in repeated measures (r) of 0.5, an adjusted P value of .00625 to account for 8 post hoc comparisons (ie, baseline speed compared with the 8 decoupling speeds), and a power ($1 - \beta$) of 80%, our analysis indicated that a minimum of 21 participants would be required to detect a significant effect of decoupling speed on pKEM. This study was approved by the university's medical institutional review board, and all participants provided written informed consent before taking part.

Study Design

A cross-sectional study design was utilized in which participants reported for a single data collection session. During this session, the ACL-reconstructed limb was trained (ie, the speed of the belt under the ACL-reconstructed leg was modified) with 8 different decoupling speeds, described below.

Participant Preparation

Before any training, participants were outfitted with 39 retroreflective markers (diameter, 14 mm) placed bilaterally on the shoulders (acromion processes), hip/pelvis (anterior and posterior superior iliac spine, iliac crests, and greater trochanters), thigh (anterior and lateral), knee (medial and lateral epicondyles and tibial tubercles), shin (lateral and distal tibias), and ankle/foot (medial and lateral malleoli, calcanei, first and fifth metatarsal heads, and bases of the second metatarsals). Additional markers were placed on the trunk at the manubrium of the sternum and at the C7 and T10 spinous processes (Figure 1). After participants were outfitted with these reflective markers, a single standing trial was performed with participants positioned on a single belt of our instrumented split-belt force treadmill (Bertec; sampling rate, 2000 Hz) so that our 12-camera motion capture system (Qualisys; sampling rate, 200 Hz) could record static marker position.

Split-Belt Training Procedures

After participants were prepared for testing, they completed a baseline 5-minute tied-belt walking trial in which they walked at a fixed speed of 1.1 m/s. Afterward, they walked on the treadmill for eight 5-minute decoupled speed configurations (30%-170% of the 1.1 m/s or 0.738-4.18 mph) in 20% increments. The ACL-reconstructed limb walked on the decoupled belt, while the non–ACL-reconstructed limb walked on the belt set at 1.1 m/s. After each decoupled speed configuration, participants walked with both belts tied at 1.1 m/s for 2 minutes to wash out any aftereffects. This speed was chosen as the baseline walking speed because all participants could walk comfortably at 1.1 m/s, and it allowed for systematic decoupling of belt speeds without any issues. From a practical standpoint, it was impossible to use self-selected speed as the baseline speed for participants and still decouple the belts using the systematic 20% increments. In this scenario, participants whose self-selected speed was on the faster side would have had to jog/run on the ACL-reconstructed leg when reaching the 130% to 170% decoupling speeds.

Gait Biomechanics

Kinematics and ground-reaction force (GRF) data were recorded during walking trials using the equipment and parameters described above for the static trial. Marker trajectories and GRF data were then imported into Visual3D (HAS-Motion) and filtered at 12 Hz using a fourth-order, zero phase–lag lowpass Butterworth filter.[2,12] Sagittal plane knee angles were calculated using a Cardan rotation sequence and expressed relative to the static standing trial. Sagittal plane knee moments were calculated using inverse dynamics, expressed as internal moments, and normalized to mass and height (N·m/kg·m). Vertical GRFs were expressed as a times body weight. All biomechanical data were time normalized to the stance phase of the gait cycle, with heel-strike and toe-off identified using a vGRF threshold of 50 N. The stance phase data of all recorded steps for the last 3 minutes of the 5-minute split-belt training conditions were ensemble averaged separately for each of the 3 variables (sagittal plane angles, moments, and vGRFs) and each limb and were later used in statistical analyses.

Statistical Analysis

To test our hypothesis that knee/limb loading and knee angles in the ACL-reconstructed limb increase as the speed of the decoupled/ACL-reconstructed limb belt increased, we utilized statistical parametric mapping (SPM) using MATLAB R2022b (The MathWorks Inc; SPM1d Version M0.4.10, www.spm1d.org). SPM allowed us to compare our outcome measurements across the entirety of the stance phase via 2 (limb) by 8 (decoupling speeds) repeated-measures analyses of variance (ANOVAs). Before these ANOVAs were executed, discrete outcomes (ie, peak moments and GRFs) were assessed for normality using Shapiro-Wilk tests, and the data were found to be normally distributed. When limb $\times$ speed interactions were significant ($P \leq .05$), SPM 1-way repeated-measures ANOVAs were run separately for each limb to examine for speed differences ($P \leq .05$). In instances in which the 1-way ANOVAs were significant, post hoc pairwise comparisons with SPM paired t tests were used to compare baseline tied-belt walking with individual decoupled speeds. Speeds were only compared in relation to baseline tied-belt walking to preserve statistical power as we were only interested in how the biomechanics during decoupled speeds compared with tied-belt walking. The alpha level for post hoc tests was adjusted using a Bonferroni correction to account for multiple comparisons (ie, 8 comparisons, each decoupled speed vs baseline) ($P = .00625$). We also created a best-fit model for the relationship between the ACL-reconstructed limb sagittal plane knee moment and the decoupled speed ratios using TableCurve 2D (Grafiti Inc).

RESULTS

Sagittal Plane Knee Moment

The SPM analysis indicated a significant interaction of limb and speed between initial ground contact (0%) and 97.4% of stance ($P < .0001$) in the ACL knee moment (Figure 2). The 1-way ANOVA assessing the ACL-reconstructed limb knee moment, with decoupling speed as a within-person factor, was significant from 0.83% to 33.8% of stance ($P < .0001$) and from 37.5% to 100% of stance ($P < .0001$). Post hoc t tests indicated that the ACL-reconstructed limb pKEM typically occurring between 20% and 30% of stance was significantly higher than that of baseline tied-belt walking in the 130%, 150%, and 170% speeds and significantly lower than that of baseline in the 30% and 50% speeds (see Appendix 2 [available online] for t test results highlighting specific bands of significance). There was a higher pKEM in midstance for the 30%, 50%, 70%, and 90% decoupling speeds compared with tied-belt walking, while there was a smaller pKEM (ie, moment was approaching or became a flexion moment) for the faster (130%-170%) decoupling speeds. The change in the ACL-reconstructed limb knee extension moment (KEM) as a function of decoupling speed is plotted in Figure 3 and fits well with simple linear models. The models show that as the decoupling speed increases, the pKEM increases and the midstance KEM decreases. The average peak and minimum midtance values depicted in Figure 3 and the associated standard errors of the means can be found in Appendix 3 (available online).

The 1-way ANOVA assessing the non–ACL-reconstructed limb knee moment, with decoupling speed as a within-participant factor, was significant from 0% to 62.6% of stance ($P < .0001$) and from 72.2% to 100% of stance ($P < .0001$). Post hoc analyses indicated that the non–ACL-reconstructed limb pKEM occurring in early stance was significantly higher than that in baseline tied-belt walking in the 130%, 150%, and 170% speeds and significantly lower than that in baseline in the 30%,

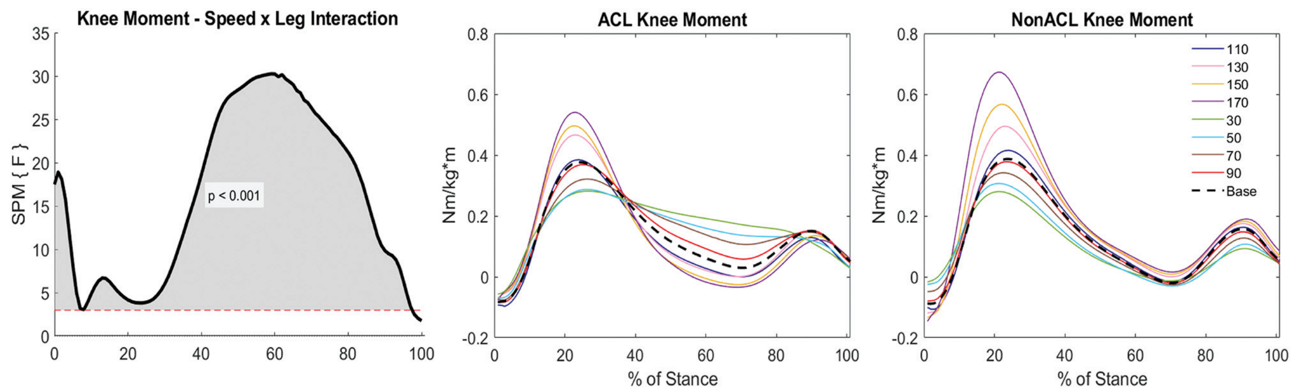

Figure 2. Left graph: Statistical parametric mapping (SPM) (*F*) speed × leg interaction results for the sagittal plane knee moment. Any portion of the waveform that exceeds the critical *F* threshold (shaded in gray) indicates a significant difference. Center and right graphs: Ensemble mean curves of the sagittal plane knee moment for the anterior cruciate ligament (ACL) leg (center) and non-ACL leg (right) for the baseline speed and the 8 split-belt decoupling speeds.

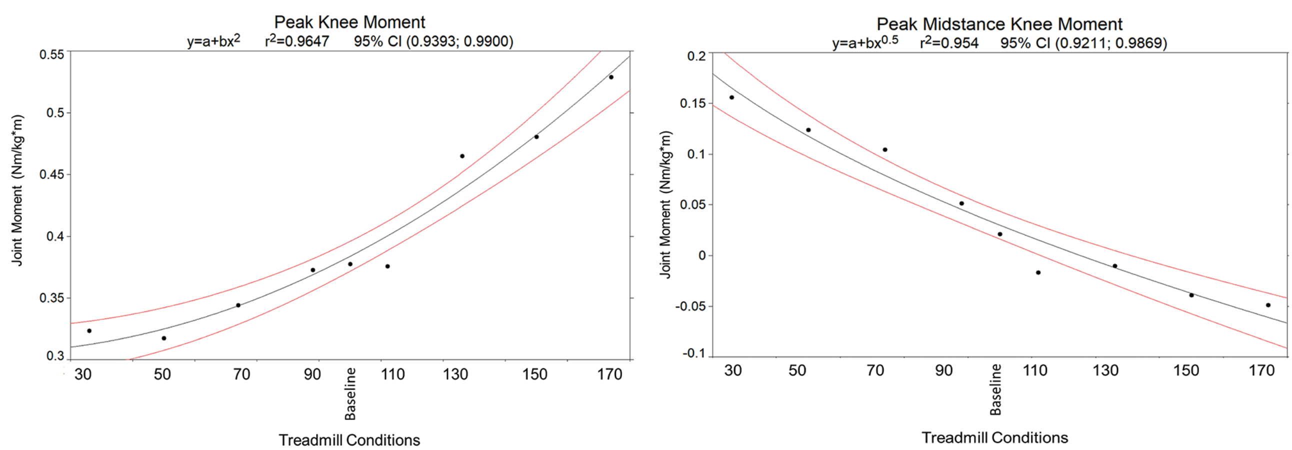

Figure 3. Left graph: Peak knee moment model illustrating that as decoupling speed increases, the peak early-stance sagittal plane knee moment for the anterior cruciate ligament (ACL) limb increases. Right graph: Peak midstance knee moment model showing that as the decoupling speed increases, the peak sagittal plane moment for the ACL limb during midstance decreases. For both graphs, the black line is the line of fit, the red lines are the 95% confidence interval for the line of fit, and the filled circles are the mean peak moments at each speed. For color interpretation, refer to the online version of the figure.

50%, and 70% speeds. KEMs around the time of ground contact (approximately 0%-9% of stance) were higher and late-stance knee moments (approximately 80%-95% of stance) were lower at the 30% and 50% decoupling speeds (Appendix 2, available online).

Vertical GRF

There was a significant leg × speed interaction during stance from 0.11% to 32.1% ($P < .0001$) and from 40.1% to 99.7% ($P < .0001$) in the vGRF (Figure 4). The 1-way ANOVA for the ACL-reconstructed limb indicated a significant decoupling speed effect on vGRF from 0% to 30.7% ($P < .0001$) and from 32.3% to 95.6% ($P < .0001$) of stance, with post hoc *t* tests identifying all speeds (30%, 50%, 70%, 90%, 110%, 130%, 150%, and 170%) being different from baseline tied-belt walking (see Appendix 2 [available online] for *t* test results highlighting specific bands and directions of significance). Generally speaking, the slower decoupling speeds (30%, 50%, and 70%) resulted in a less dynamic or flatter vGRF compared with the tied-belt vGRF, which was more dynamic (ie, typical early and late peaks with a trough in the middle). This pattern resulted in the vGRF in the middle of a stance being significantly higher, but it being significantly lower in the earlier and later part of a stance in the slow split-belt conditions compared with the tied-belt condition (Appendix 2, available online). For the faster decoupling speeds (130%, 150%, and 170%), in general, the first peak vGRF was of greater magnitude and came earlier in stance (ie, steeper slope) when compared with the tied-belt condition. Also, the vGRF trough in the middle of the stance became deeper or significantly lower in magnitude in the split-belt

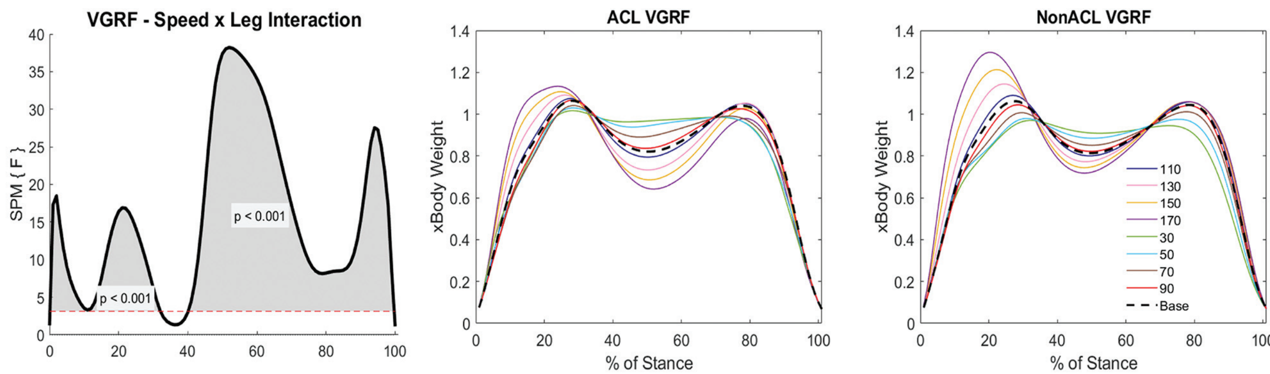

Figure 4. Left graph: Statistical parametric mapping (SPM) (F) speed $\times$ leg interaction results for the vertical ground-reaction force (vGRF). Any portion of the waveform that exceeds the critical F threshold (shaded in gray) indicates a significant difference. Center and right graphs: Ensemble mean curves of the vGRF for the anterior cruciate ligament (ACL) leg (center) and non-ACL leg (right) for the baseline speed and the 8 split-belt decoupling speeds.

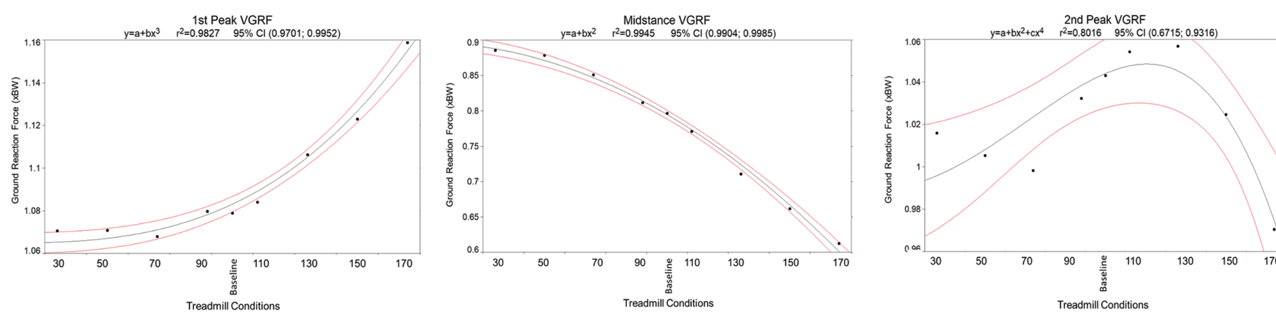

Figure 5. Left graph: First peak vertical ground-reaction force (vGRF) model illustrating that as decoupling speed increases, so does the first peak vGRF. Center graph: Minimum midstance knee moment model illustrating that as decoupling speed increases, the minimum midstance vGRF decreases. Right graph: Second peak vGRF model illustrating a nonlinear relationship between the second peak vGRF and decoupling speed that best fit using a fourth-order polynomial. For all graphs, the black line is the line of fit, the red lines are the 95% confidence interval for the line of fit, and the filled circles are the mean peak moments at each speed. For color interpretation, refer to the online version of the figure. BW = body weight.

compared with the tied-belt condition (Appendix 2, available online). The change in the ACL-reconstructed limb vGRF as a function of decoupling speed is plotted in Figure 5 and shows a good fit with simple linear models for the first peak and midstance vGRF. The models depict the relationship that as decoupling speed increases, the first peak vGRF increases and the midstance vGRF decreases. The second peak vGRF model is more complex, suggesting a nonlinear relationship that fits best with a fourth-order polynomial (Figure 5). The average peak and minimum midtance values depicted in Figure 5 and the associated standard errors of the means can be found in Appendix 3 (available online).

The 1-way ANOVA for the non–ACL-reconstructed limb indicated a significant decoupling speed effect on vGRF from 0% to 63.7% ($P < .0001$) and from 67.7% to 99.9% ($P < .0001$) of stance. The post hoc t tests showed significant differences in all speeds from baseline ($P < .05$) (Appendix 1, available online). The slower decoupling speeds (30%, 50%, and 70%) tended to result in similar changes to the non–ACL-reconstructed limb as those reported above for the ACL-reconstructed limb. The non–ACL-reconstructed limb displayed a less dynamic vGRF pattern, whereby the magnitudes of the first and second peaks were lower and the midstance trough or minimum was higher in comparison with the tied-belt condition. For the faster decoupling speeds (130%, 150%, and 170%), the non–ACL-reconstructed limb had a higher first peak vGRF and the midstance minimum was lower compared with the tied-belt condition.

Sagittal Plane Knee Angle

A significant leg $\times$ speed interaction was also noted from 0% to 24.8% ($P = .019$) and from 80.8% to 100% ($P = .028$) of stance in the knee angle (Figure 6). The 1-way ANOVA for the ACL-reconstructed limb indicated a significant decoupling speed effect on knee angle from 0% to 50.5% ($P = .0016$), from 59.7% to 69.0% ($P = .044$), and from 79.3% to 100% ($P = .0281$) of stance. Specifically, post hoc t tests identified that speeds 30%, 50%, 70%, 130%, 150%, and 170% were different from baseline (see Appendix 2

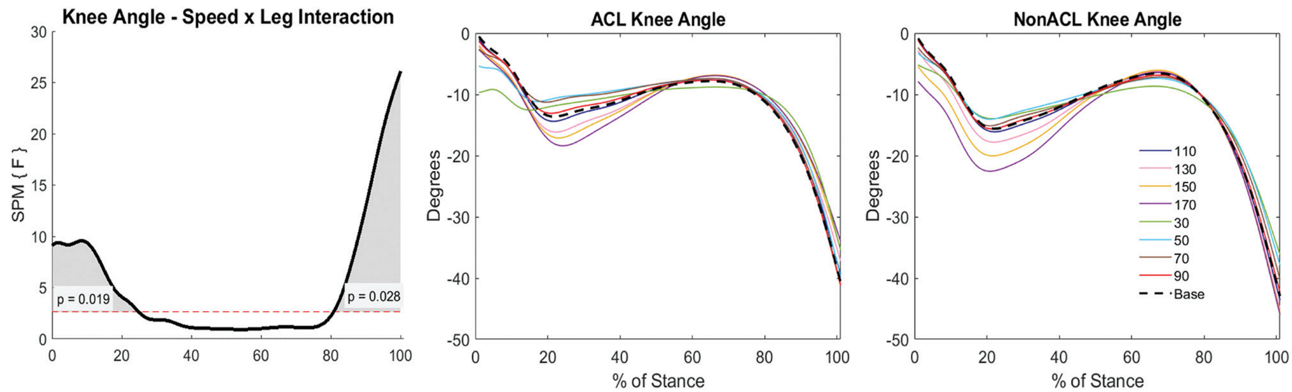

Figure 6. Left graph: Statistical parametric mapping (SPM) (*F*) speed × leg interaction results for the sagittal plane knee angle. Any portion of the waveform that exceeds the critical *F* threshold (shaded in gray) indicates a significant difference. Center and right graphs: Ensemble mean curves of the sagittal plane knee angle for the anterior cruciate ligament (ACL) leg (center) and non-ACL leg (right) for the baseline speed and the 8 split-belt decoupling speeds.

[available online] for SPM paired *t* test results highlighting specific bands and directions of significance). In general, for the slower decoupling speeds the ACL-reconstructed limb had more knee flexion around the time of ground contact (30% and 50% decoupling speeds) and less flexion in late stance (30% decoupling speed). At the 50% and 70% decoupling speeds, the ACL-reconstructed limb displayed less knee flexion around 20% of stance. At the faster decoupling speeds (130%, 150%, and 170%), the ACL-reconstructed limb displayed more knee flexion around 15% to 40% of stance and less knee flexion in late stance (approximately 80%-100%).

The 1-way ANOVA for the non–ACL-reconstructed limb indicated a significant decoupling speed effect on knee angle from 0% to 48.3% ($P < .0001$), from 55.5% to 75.7% ($P = .0234$), and from 82.9% to 100% ($P = .0291$) of stance. The post hoc *t* tests identified that speeds 30%, 50%, 70%, 130%, 150%, and 170% were different from baseline tied-belt walking ($P < .05$) (Appendix 1, available online). The non–ACL-reconstructed limb at slower decoupling speeds (30%, 50%, and 70%) generally had more knee flexion around the time of ground contact (0%-1.5% of stance) and less flexion in late stance (approximately 85%-100%). At the faster decoupling speeds, particularly at 150% and 170%, the non–ACL-reconstructed limb displayed more knee flexion for approximately the first 40% stance.

DISCUSSION

The purpose of this study was to comprehensively evaluate the effects of decoupling speeds on ACLR and non–ACL-reconstructed limb biomechanics during split-belt walking while training the ACL-reconstructed limb, and to establish a model that described the relationship between decoupling speed ratios and knee and limb loading in individuals with ACLR. As hypothesized, a dose-response relationship was observed between decoupling speed ratios and knee/limb loading such that faster decoupling speeds led to bilateral increases in the pKEM during early stance and first and second peak vGRFs during early and late stance, respectively. Faster speeds also led to lower pKEMs (or a knee flexion moment in some participants) and lower vGRFs during midstance. These bilateral limb loading changes were also accompanied by bilateral changes in the knee flexion angles, with peak angles in early stance getting larger at faster decoupling speeds and smaller at lower decoupling speeds for both limbs with ACL-reconstructed leg training. Interestingly, these trends reversed, and opposite effects to those of the fast speeds were generally observed for the slower decoupling speeds. Overall, these results indicate that the decoupling speed chosen for split-belt treadmill training is important and should be considered when attempting to modify loading patterns in individuals with ACLR.

Training the ACL-reconstructed leg at faster decoupling speeds led to bilateral increases in the pKEM and in the first and second peak vGRFs when compared with baseline tied-belt walking. We attribute the higher loading metrics on the ACL-reconstructed leg to the limb moving at a faster speed compared with the baseline speed (1.1 m/s). It is well established that increasing walking speed leads to increases in lower limb kinetics in healthy people[6] and individuals with ACLR.[7,8] Walking faster leads to greater forces applied by the body to the ground (F = mA) and thereby greater forces transmitted back from the ground to the body and can explain the higher loads observed on the faster leg. Interestingly, the non–ACL-reconstructed limb that was moving at a constant speed (1.1 m/s) also experienced higher knee/limb loading with increases in decoupling speed ratios. The higher loading metrics observed on the non–ACL-reconstructed limb, which maintained a speed of 1.1 m/s while the ACL-reconstructed limb moved faster, may be attributed to the non–ACL-reconstructed limb attempting to "catch up" with the faster leg. This could result in a harder and faster ground impact, leading to a higher breaking GRF impulse, as has been previously reported,[19] compared with when both limbs move at the same speed. While faster decoupling speeds led to greater loading during early and late stance for both limbs, they led to a smaller vGRF and pKEMs (actually achieving a peak knee flexion moment in some participants) during midstance for the ACL-reconstructed limb. We attribute this finding to the faster overall movement of the body during stance, which results in a quicker heel-strike, a rapid transition over the support limb during midstance, and greater propulsion at push-off. This faster body movement causes the center of mass to move more quickly up and down, leading to more pronounced peaks in vGRF during early and late stance, and a more pronounced trough during midstance. The reduced vGRF during midstance also explains the lower knee moments observed at that point. In other words, it is like an inverted pendulum moving faster. Overall, training the ACL-reconstructed leg at the faster decoupling speeds led to changes in walking gait mechanics that would be preferred to overcome the reduced loading that is often evident after ACLR in early stance and to drive the loading toward an internal knee flexion moment during midstance.

Opposite effects to those of the faster decoupling speeds tended to appear when the ACL-reconstructed leg was trained at slower decoupling speeds. The slowest decoupling speeds led to bilaterally lower pKEM and vGRF in early and late stance. Additionally, these slower speeds also led to greater magnitudes of and more sustained midstance loading (ie, higher pKEM and vGRF over a longer portion of midstance) for the ACL-reconstructed limb. The non–ACL-reconstructed leg did not display sustained knee loading (measured via pKEM) during midstance, but it did show higher and sustained limb loading (measured via peak vGRF) during midstance. The changes to the ACL training limb are not unexpected. Speed is known to affect walking kinetics as discussed above.[7,8] As such, we expected that having the ACL-reconstructed limb move at 0.33 m/s (30% decoupling) rather than 1.1 m/s (baseline speed) would reduce our peak kinetic measures (ie, pKEM and peak vGRF). The sustained higher loading during midstance at the slower decoupling speeds is also logical, given the longer stance duration that would be probable at the slower speed for the ACL-reconstructed leg. As such, the ACL-reconstructed leg was forced to support body weight for an extended period of time compared with the non–ACL-reconstructed leg, resulting in more sustained and higher limb loading. Overall, training the ACL-reconstructed leg at slower decoupling speeds led to a less dynamic loading pattern that is not typical of a normal walking gait, exacerbating peak weight acceptance loading deficits seen in participants with ACLR and increasing limb loading magnitude and duration during midstance.

Based on our findings, it might seem as if we would want to recommend training at faster decoupling speeds to achieve a more dynamic loading pattern (ie, less sustained loads) and higher loads where underloading of the ACL-reconstructed limb is prevalent (ie, early stance). However, this is not necessarily the case. The outcome we see during training could, in fact, be opposite of that which we see when the split-belt stimulus is removed and participants resume normal tied-belt or overground walking (ie, the aftereffects that occur once training is over could be different from the training effect itself). In those who survive stroke, for example, step length asymmetries are augmented with split-belt treadmill training, but when the belts become tied again, step length becomes more symmetrical, achieving the desired gait result for this clinical population.[18] This finding suggests that when we amplify the error we wish to correct during training, it may lead to the desired improvements in gait mechanics after the training is over. In the case of split-belt walking for the participants with ACLR, we aim to increase peak knee loading in early stance and decrease it during midstance given that these are commonly identified gait deficits[3-5,12,17,22] and are linked to imaging markers of osteoarthritis.[16,21,24,25] As such, it may be that training the ACL-reconstructed leg at slower decoupling speeds where error is augmented during early stance and midstance may lead to our desired outcome. We did not pursue this in the current work, as our primary question was focused on examining the training effects of decoupling speed on knee/limb loading and establishing a dose-response relationship. While we plan to evaluate the aftereffects of short-duration split-belt treadmill training in a separate paper, our design may not be optimal for understanding the sustained aftereffects of split-belt training when used as a rehabilitation intervention. Hence, more research is necessary to better understand what aftereffects are produced with longer-duration slow and fast split-belt treadmill training in individuals with ACLR.

While knee flexion angle was not our target gait variable for split-belt treadmill training, decoupling speed did influence knee flexion angle in the participants with ACLR. The finding likely to be of greatest clinical relevance in the ACLR population is that peak knee flexion angle during early stance increased with the faster decoupling speeds and decreased with slower decoupling speeds. As knee flexion angle is altered after ACLR (ie, reductions in peak knee flexion are common between 10 and 40 months after surgery),[23] split-belt training may be an effective gait retraining approach to restore knee flexion in this population. The increase in knee flexion angle on the training limb with faster speeds is not unexpected, as it has been reported previously and is thought to be an indication of the need for greater shock absorption when the limb is moving faster.[11]

This work is not without limitations. First, the participants in this study did not have the loading asymmetries during baseline tied-belt walking that we expected (>10% reduction in knee/limb loading based on the existing literature[22,23]), which could have influenced our outcomes. However, despite a lack of baseline loading asymmetries, our work supports the notion that a split-belt training intervention can be effective in inducing alterations in knee/limb loading in individuals with ACLR and could be an effective gait retraining approach after ACLR. Second, a short washout period of 2 minutes was used between each decoupling speed, which may not have been long enough to remove the effects on our outcome measures from prior speeds. However, decoupling speeds were randomized, which should minimize systematic effects on the data. Third, our study was designed to examine the acute effects of a variety of decoupling speeds. It is possible that training for a longer time within a day and over a greater number of days could influence our findings. Lastly, most participants in our study received patellar tendon grafts for ACLR, and as such, we cannot ascertain if other graft types would respond to the split-belt training in a similar manner.

CONCLUSION

Our results show for the first time that there is a clear dose-response relationship between split-belt treadmill training decoupling speed and knee/limb loading. Faster decoupling speeds increased peak knee/limb loading bilaterally, while slower speeds led to a bilateral decrease. Slower speeds led to higher and more sustained midstance loading, whereas faster speeds generated lower midstance loading. These findings indicate that individuals with ACLR are responsive to split-belt treadmill training and can modify knee/limb loads, suggesting that split-belt treadmill training at slower (30%-50%) or faster (150%-170%) decoupling speed could be an effective gait retraining approach to restore knee biomechanics after ACLR. However, more research to better understand the aftereffects of split-belt treadmill training and whether the training effects transfer to overground activities is necessary before any conclusions about its gait retraining capabilities can be made.

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