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Abdominal Bracing Increases Ground Reaction Forces and Reduces Knee and Hip Flexion During Landing

Exercises that target activation of the abdominal wall musculature to enhance spine stability have been a major focus of exercise interventions for the prevention and management of low back, pelvic, and lower-limb disorders.[18,25,32] Abdominal wall training has been adopted by rehabilitation specialists and athletic trainers throughout the world, due to a perceived benefit to on-field performance and reduced low back pain risk.[2,16]

There are different approaches for teaching abdominal wall activation. Some authors propose abdominal bracing (AB), which involves synchronous coactivation of the abdominal wall,[12] while others have advocated targeting the recruitment of the transversus abdominis while "drawing in" the abdominal wall.[17] There is continued debate as to which approach is superior; however, a number of studies have reported that AB is more effective in increasing spine stability than abdominal hollowing.[12,29]

The underlying hypothesis of AB is that contracting the abdominal wall muscles at preset levels provides increased rigidity and stability of the spine, thereby reducing the incidence of aberrant spinal movements and potential for injury.[28,32] A number of studies have confirmed that AB can reduce spinal movements during horizontal perturbation tasks.[4,12,29] Vera-Garcia et al[28] studied various abdominal wall–bracing strategies (no brace and 10%, 20%, and 30% maximum voluntary isometric contraction [MVIC]) and the effects on spine stability during a rapid horizontal perturbation task in a seated position. The authors reported that 30% MVIC of the external oblique muscles was optimal for providing spine stability. Brown and colleagues[4] used a similar method, with participants undergoing rapid perturbations during sitting, and reported that AB significantly increased spine stability and reduced movement of the lumbar spine after perturbation, but at the cost of increased spinal compression.

While there is evidence that AB can result in increased stability of the spine, there may be detrimental consequences.

STUDY DESIGN: Controlled laboratory study.

BACKGROUND: Abdominal bracing (AB) is a widely advocated method of increasing spine stability, yet the influence of AB on the execution of sporting movements has not been quantified. Landing is a common task during sporting endeavors; therefore, investigating the effect of performing AB during a drop-landing task is relevant.

OBJECTIVE: To quantify the effect of AB on kinematics (ankle, knee, hip, and regional lumbar spine peak flexion angles) and peak vertical ground reaction force (vGRF) during a drop-landing task.

METHODS: Sixteen healthy adults (7 female, 9 male; mean ± SD age, 27 ± 7 years; height, 170.6 ± 8.1 cm; mass, 68.0 ± 11.3 kg) were assessed using 3-D motion analysis, electromyography (EMG), and a force platform while performing a drop-landing task with and without AB. Abdominal bracing was achieved with the assistance of real-time internal oblique EMG feedback. Lower-limb and regional lumbar spine kinematics, peak vGRF, and normalized EMG of the left and right internal obliques and lumbar multifidus were quantified. Paired-samples *t* tests were used to compare variables between the AB and no-AB conditions.

RESULTS: Abdominal bracing resulted in significantly reduced knee and hip flexion and increased peak vGRF during landing. No differences in lumbar multifidus EMG or lumbar spine kinematics were observed.

CONCLUSION: Abdominal bracing reduces impact attenuation during landing. These altered biomechanics may have implications for lower-limb and spinal injury risk during dynamic tasks. *J Orthop Sports Phys Ther 2016;46(4):286-292. Epub 8 Mar 2016. doi:10.2519/jospt.2016.5774*

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A more rigid lumbar system increases spinal compressive forces,[28] which may increase injury risk during dynamic tasks.[22] Furthermore, motor control theorists propose that contracting specific muscles during dynamic tasks may cause motor overflow to other muscles.[1] For example, it has been proposed that facilitating activation of the abdominal wall will result in cocontraction of the lumbar multifidus (LM), further enhancing trunk stability.[28] Similarly, increased cocontraction of lower-limb muscles may further increase system stiffness by limiting lower-limb movement during tasks that require impact attenuation, such as when landing from a jump.[10]

Landing is a task required by athletes participating in a number of sports, including gymnastics, volleyball, football, basketball, and running.[11,21] There is evidence that the high prevalence of lower-limb and lumbar spine pain and injury among these sports[11,24,26] is associated with high-impact forces, including significant ground reaction forces during landing.[10,20,23] To date, the effect of AB on lower-limb kinematics and ground reaction forces during functional tasks such as landing from a jump has not been investigated.

The primary aim of this study was to investigate the effect of AB on lower-limb and regional lumbar spine kinematics and vertical ground reaction forces (vGRFs) during a drop-landing task. The secondary aims were to (1) investigate whether AB can be maintained during a drop-landing task, (2) determine whether AB results in increased activation of the LM, and (3) determine whether AB reduces lumbar spine movement during a drop-landing task.

METHODS

Participants

Sixteen healthy adults (7 female, 9 male) with a mean ± SD age of 27 ± 7 years, height of 170.6 ± 8.1 cm, and mass of 68.0 ± 11.3 kg participated in this study. Participants had no history of low back, trunk, or lower-limb pathology during the 6 months prior to testing (musculoskeletal, neurological, or vascular) and/or any condition precluding physical activity the week prior to testing. The Curtin University Human Research Ethics Committee approved this study, and all participants provided written informed consent prior to participation.

Data Collection

Participants were required to attend a single data-collection session at the Curtin University biomechanics laboratory. Lower-limb and lumbar kinematics were collected using a 14-camera Vicon 3-D motion analysis system at 250 Hz (OMG plc, Oxford, UK). Ground reaction forces (AMTI force plate; Advanced Mechanical Technology, Inc, Watertown, MA) and trunk muscle activation (8-channel, octopus-cable telemetric surface electromyography [EMG] system; Bortec Biomedical Ltd, Calgary, Canada) were collected simultaneously and sampled at a rate of 1000 Hz using Vicon Nexus software (OMG plc).

Participant Preparation

Participants were prepared for surface EMG data collection using standard procedures, which included cleaning the required areas of skin with alcohol and light sandpaper abrasion.[14] Pairs of 12-mm-diameter, Ag/AgCl, self-adhesive, disposable, wet-gel disc electrodes (Ambu A/S, Ballerup, Denmark) were then placed parallel to the muscle fibers at an interelectrode distance of approximately 2.5 cm. The electrodes for transverse fibers of the internal oblique (TrIO) and the underlying transversus abdominis (representing the transverse abdominal wall)[8] were placed bilaterally 1 cm medial to the anterior superior iliac spines, below a line connecting the left and right anterior superior iliac spines.[8,19] The electrodes for the LM were placed bilaterally and aligned parallel to the line between the posterior superior iliac spine and the L1-2 interspinous space at the L5 level.[9] Skin impedance was assessed using an impedance meter, and a value less than 5 kΩ was considered acceptable.[8] Snap leads connected the pairs of electrodes to the preamplifiers, and were taped to the skin surface to minimize movement artifact.[8] The EMG cable telemetry system utilized analog differential amplifiers (frequency response, 10-1000 Hz; common-mode rejection ratio, 115 dB).

Participants were required to perform a series of MVICs.[8] To generate an MVIC of the TrIOs, the participants were asked to lie supine, with their shoulders and lower limbs secured to a plinth using an adjustable belt. Participants were asked to lift their shoulders using as much force as possible and maintain this position for a period of 3 seconds. For the LM muscles, the participants were prone, with their shoulders and lower limbs securely fastened to a plinth using adjustable straps. Participants were asked to lift their shoulders off the table, using as much force as possible, and to maintain this for a period of 3 seconds. Alternate LM and TrIO MVIC trials were recorded using the Vicon Nexus software (OMG plc), with 2 minutes of rest between trials. Three MVIC trials for each muscle group were obtained.

A physical therapist then instructed the participants in the technique of AB[18] with the assistance of a customized real-time EMG feedback program (LabVIEW; National Instruments Corporation, Austin, TX). Training was performed until the participant could contract the TrIO to 30% MVIC without difficulty.[28] This generally required 20 ± 5 minutes. The real-time EMG feedback program utilized a 100-millisecond moving window to calculate the root-mean-square (RMS) of the left TrIO EMG signals for each 3-second trial. The average of the RMS across the 3 TrIO MVIC trials was used to represent overall TrIO MVIC. The program then produced a line graph of the level of the left TrIO contraction (ie, percent MVIC) in real time. The *y*-axis of the graph ranged from 0% to 100% MVIC, with the target contraction area of 30% ± 5% MVIC being highlighted. Participants were led to believe that the feedback was generated from both left and right TrIO muscles to ensure that bilateral contractions were performed.

Retroreflective markers were attached to each participant's lumbar region and lower limbs, according to previously outlined marker sets[3,5,30] recommended by the International Society of Biomechanics.[31] In short, the upper lumbar region (y-axis created from markers placed over the first and third lumbar vertebrae; z-axis from the orthogonal plane created from markers placed 5 cm bilaterally from the second and third lumbar vertebrae; x-axis was the cross-product of the z-axis and y-axis) movement was calculated relative to the lower lumbar region (y-axis created from markers placed over the third and fifth lumbar vertebrae; z-axis from the orthogonal plane created from markers placed 5 cm bilaterally from the fourth and fifth lumbar vertebrae; x-axis was the cross-product of the z-axis and y-axis). The lower lumbar region movement was output relative to the pelvis segment. This marker set has been used in several investigations, including one that evaluated regional lumbar postures during gymnastic landings.[30] Prior to dynamic-trial data collection, a static trial was obtained to establish local coordinate systems.[3]

TABLE 1 AVERAGE ELECTROMYOGRAPHY THROUGHOUT THE 3 PHASES OF DROP LANDING WITH AND WITHOUT ABDOMINAL BRACING (N = 16)

	Braced*	Not Braced*	Difference†	t‡	P Value
Phase 1 (take-off until landing)					
TrIO	25.1 ± 9.1	16.3 ± 8.2	–11.7, –5.8	6.39	<.001
LM	16.7 ± 7.6	15.8 ± 8.1	–4.5, 2.8	0.50	.621
Phase 2 (landing until peak force)					
TrIO	75.3 ± 29.9	50.5 ± 21.0	–35.1, –14.5	5.15	<.001
LM	72.2 ± 29.7	59.6 ± 25.2	–31.4, 6.2	1.43	.174
Phase 3 (peak force until stability)					
TrIO	23.3 ± 12.5	15.7 ± 9.5	–14.6, –0.5	2.28	.038
LM	35.8 ± 19.5	35.7 ± 12.9	–8.9, 8.7	0.02	.988

Abbreviations: LM, lumbar multifidus; TrIO, transverse fibers of internal oblique.
**Values are mean ± SD percent maximum voluntary isometric contraction.*
†Values are 95% confidence interval.
‡df = 15.

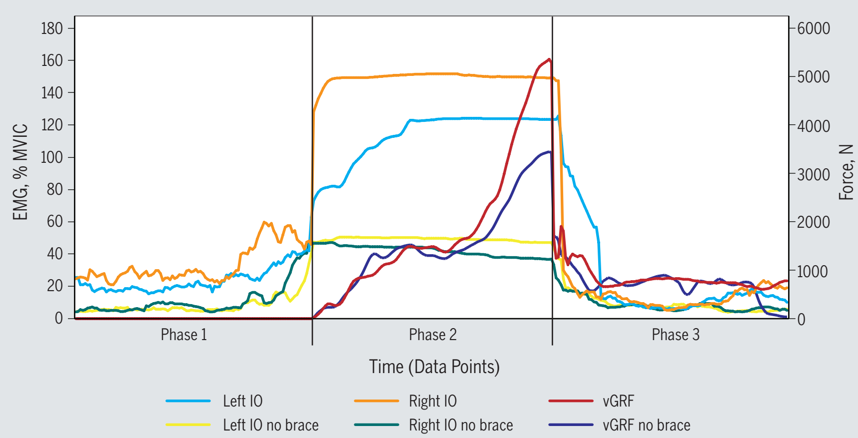

FIGURE 1. Vertical ground reaction force and left and right IO muscle activity across the 3 phases of drop landing, braced and not braced, during a single representative trial. Abbreviations: EMG, electromyography; IO, internal oblique; MVIC, maximum voluntary isometric contraction; vGRF, vertical ground reaction force.

Dynamic-Trial Data Collection

Participants were asked to perform a series of drop-landing trials from a 40-cm bench. From a standing position, participants were instructed to step off the bench (leading with their dominant leg) and land with both feet simultaneously on the force platform, and hold the landing position for a period of 3 seconds. Practice trials allowed participants to become familiar with the task. Three trials were performed with AB (visual biofeedback provided via a large computer screen placed slightly to the right side of the force plate and in front of the participant), followed by 3 trials with no AB (and no visual biofeedback). The no-AB trials were always performed prior to the AB trials to mitigate the potential influence of the bracing training/feedback on the participant's natural technique. The AB trials began with 20 seconds of "prebracing" to achieve the required 30% level of contraction.

Data Processing

Utilizing the vGRF data, a customized LabVIEW program (National Instruments Corporation) divided each drop-landing trial into 3 phases of interest: phase 1, take-off until the instant of ground contact/landing; phase 2, landing until peak vGRF; phase 3, peak vGRF until landing stability (when vGRF force was equivalent to body weight). While the main focus of the study was on phase 2, a description of the muscle activity in phase 1 was considered essential to ensure that the bracing condition had been met.

Kinematic data were processed using Vicon Nexus software (OMG plc). Following a residual analysis, the trajectories were filtered using a recursive,

TABLE 2 PEAK LUMBAR AND LOWER-LIMB FLEXION ANGLES DURING PHASES 2 AND 3, AND PEAK VERTICAL GROUND REACTION FORCE, OF DROP LANDING WITH AND WITHOUT ABDOMINAL BRACING (N = 16)

	Braced*	Not Braced*	Difference†	*t*‡	*P* Value
Phase 2 (landing until peak force)					
Upper lumbar	13.7 ± 6.2	13.6 ± 5.5	–1.2, 1.2	0.05	.958
Lower lumbar	7.1 ± 8.1	8.2 ± 7.0	–0.3, 2.3	1.68	.115
Ankle	6.5 ± 12.7	9.4 ± 11.0	–0.1, 6.0	2.10§	.056
Knee	46.4 ± 10.6	49.8 ± 10.6	0.7, 6.1	2.66	.018
Hip	39.3 ± 8.3	41.3 ± 7.8	0.4, 3.8	2.60	.020
Phase 3 (peak force until stability)					
Upper lumbar	9.4 ± 6.0	8.0 ± 6.6	...	...	...
Lower lumbar	14.8 ± 8.3	15.5 ± 7.2	...	...	...
Ankle	29.4 ± 9.1	27.3 ± 10.3	...	...	...
Knee	95.6 ± 15.4	95.6 ± 12.3	...	...	...
Hip	72.8 ± 12.3	75.6 ± 9.6	...	...	...
Peak force, $N \cdot kg^{-1}$	5.0 ± 0.9	4.3 ± 0.9	–0.4, –0.9	6.41	<.001

**Values are mean ± SD degrees unless otherwise indicated.*
†Values are 95% confidence interval.
‡df = 15 unless otherwise indicated.
§df = 13.

low-pass Butterworth filter with a 20-Hz cutoff frequency. A mathematical model then utilized a z-x-y Euler angle decomposition to calculate relevant lower-limb and lumbar region angles.[3,6,30] Kinematic variables of interest included peak flexion of the upper and lower lumbar regions, hip, knee, and ankle.

The EMG data were demeaned, and a 100-millisecond moving window was used to calculate the mean RMS (percent MVIC) across the entire trial. The maximum value of each kinematic variable of interest was identified. As preliminary analysis of the data indicated similar patterns across the right and left sides for both EMG and lower-limb kinematics, the data from both limbs were averaged. Ground reaction forces were normalized to each participant's body mass.

Statistical Analysis

Electromyography, kinematics, and peak vGRF data were averaged across each participant's brace and no-brace trials and used for statistical analysis (SPSS Version 19.0; IBM Corporation, Armonk, NY). Paired-samples *t* tests were used to compare muscle activity, kinematics, and peak vGRF data between the brace and no-brace trials during phase 2. Muscle activity also was compared during phase 1 to determine whether participants had met the AB condition. A *P* value of less than .05 was used for statistical significance. Descriptive statistics also are presented for phase 3 to provide the full context for jump preparation, landing, and recovery.

RESULTS

THE TrIO MUSCLES WERE SIGNIFICANTLY more active during each phase of the drop-landing task performed with AB compared to the no-brace trials (**TABLE 1**, **FIGURE 1**). The TrIO activation levels during phase 1 were within the target range of 30% ± 5% MVIC, indicating that participants achieved the goal of AB prior to contacting the ground. There was no effect of AB on LM activation, regardless of phase (**TABLE 1**).

Comparison of the landing kinematics between brace and no-brace trials indicated that bracing resulted in reduced peak knee and hip flexion during landing (**TABLE 2**, **FIGURE 2**). There was no statistical difference in peak ankle dorsiflexion or in regional lumbar spine kinematics between the brace and no-brace conditions.

On average, the peak vGRF was signficantly greater during the drop-landing trials performed with AB in comparison to the trials performed without AB (**TABLE 2**, **FIGURE 3**). This pattern was consistent across all participants.

DISCUSSION

THE CURRENT STUDY QUANTIFIED the effect of AB on a drop-landing task. Specifically, we compared the influence of a brace versus a no-brace strategy on lower-limb and regional lumbar spine flexion, the vGRF, and TrIO and LM activation. The primary finding of this study was that when participants performed AB at an average of 25% of their MVIC, significantly reduced knee and hip flexion and increased vGRF during the landing phase of a drop-landing task were observed compared to the no-brace condition. There were no differences in lumbar spine movement between the brace and no-brace conditions.

Our results demonstrate that it is possible to maintain AB with an average TrIO contraction of 25% MVIC during the execution of a drop-landing task with the assistance of visual feedback. These findings are in agreement with the findings of several studies demonstrating that participants could activate their internal oblique muscles at 10%, 20%, or 30% MVIC during AB using EMG feedback when sitting upright and subjected to perturbation.[4,28,32] In the current study, AB appeared to have no influence on

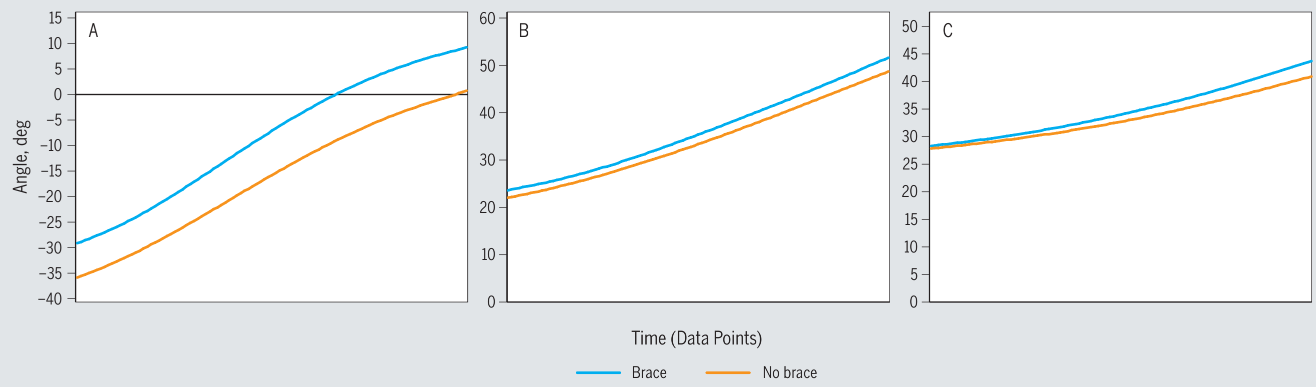

FIGURE 2. Group mean (n = 16) angle of the (A) ankle, (B) knee, and (C) hip through the second phase of landing (from the instant of ground contact until peak vertical ground reaction force) during a drop-landing task performed with and without abdominal bracing.

lumbar kinematics during landing. This is in contrast to previous studies demonstrating that AB reduced lumbar spine movements during sagittal and lateral trunk perturbations.[4,29] However, these differences are not particularly surprising, given the very different nature of the tasks assessed: largely vertical loading associated with a drop-landing task compared to the horizontal loading in the aforementioned comparative study by Vera-Garcia et al.[29]

The results of our study also demonstrated reduced peak knee and hip flexion and increased vGRF during drop landing while performing AB. This finding may be clinically relevant, given that reduced lower-limb flexion during the landing phase, combined with an increased magnitude of vGRF, has been associated with an increased risk of lower extremity and lumbar spine injury and/or low back pain in athletes across different sports.[13,15,21] While not directly comparable and speculative, an increase in abdominal activation may be one potential mechanism placing athletes at greater risk of lumbar spine/lower-limb injury during loading tasks. This hypothesis is supported by previous research in which runners with a history of low back pain were found to have significantly greater knee joint stiffness than a control group during the shock-attenuation phase of the stance phase during running.[13] Further

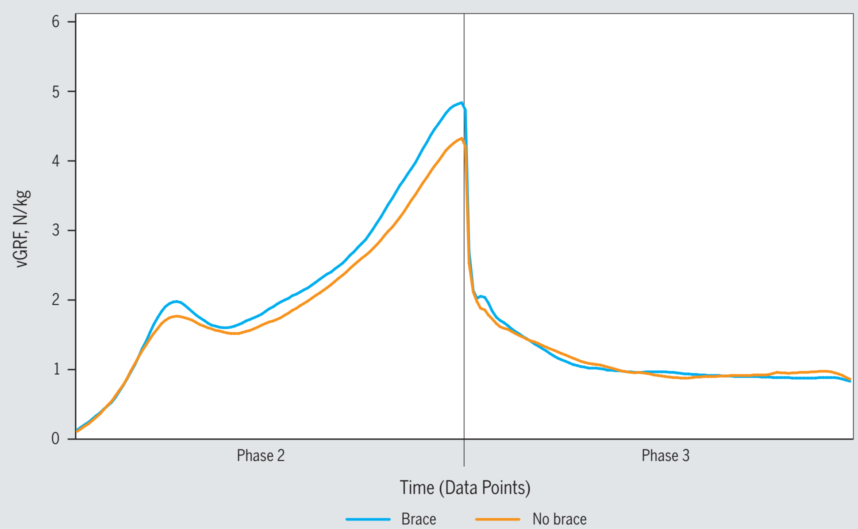

FIGURE 3. Group mean (n = 16) vGRF across phases 2 (from the instant of ground contact until peak vGRF) and 3 (peak vGRF until body-weight stability) during a drop landing performed with and without abdominal bracing. Abbreviation: vGRF, vertical ground reaction force.

analysis of these results indicated that this change in stiffness was largely the result of reduced knee joint range of motion (5.9°).[13] These findings support the premise that the reduced kinematics reported in the current study (3.4° at the knee and 2.0° at the hip) may be clinically meaningful. The 2.9° apparent difference at the ankle might have been nonsignificant owing to inadequate statistical power (estimated power was 55%). Furthermore, the relevance of the difference in peak vGRF reported in this investigation (435.6 N) between AB and no-brace drop landings is supported by a prospective study that found that landing from a similar drop-landing task (from a box 90 mm lower than the current study) with 209 N greater vGRF was significantly associated with anterior cruciate ligament injury.[15] Given the exploratory nature of this and previous research, further investigation is required to confirm a relationship between AB during loading tasks and lower extremity injury.

Our findings raise important questions regarding the approach of teaching athletic and nonathletic populations to perform AB prior to impact tasks.[12,18]

While a more stable spine may be advantageous when the spine is exposed to bending moments to prevent end-range strain,[18] AB may have negative consequences during impact tasks when the role of the motor system is to effectively dampen loading forces. While the current study presents no direct evidence that AB increases injury risk, there is evidence that increased ground reaction forces and reduced flexion of the lower limbs during loading are factors associated with increased injury risk in the lower limb and lumbar spine.[13,15,21] Further, recurrent back pain has been associated with an increase in trunk stability rather than reduced trunk stability.[7,27] Whether AB during loading tasks (at least at the level of 25% MVIC) is provocative of musculoskeletal pain in sports that involve landing requires further investigation. However, in light of these findings, it may be prudent to carefully consider when AB should be recommended, rather than promoting it indiscriminately.

The current study was conducted in pain-free young adults and, therefore, our findings cannot be directly generalized to other age groups and populations. Further, the results of this study relate to AB performed at an average of 25% MVIC. Future research should consider investigating different AB levels, as well as other bracing strategies, such as the commonly taught abdominal hollowing technique. Instrumentation utilizing motion-capture analysis may be affected by soft tissue artifact and marker placement error, although utilizing a within-participant design minimized the effects of these factors on the present study's results.

The current study only reported motion in the sagittal plane; therefore, while it is possible that AB reduced aberrant motion and thus enabled a higher tolerance for vGRF, this cannot be confirmed. Our study did not measure upper trunk, upper limb, or head motion, which may have influenced the vGRF during landing. Further, given that participants were provided visual feedback during the AB trials, and not during the no-brace trials, there is a possibility that attentional bias may have influenced the results.

CONCLUSION

This study found that AB during drop landing resulted in reduced knee and hip flexion and increased peak ground reaction force, suggesting that AB may reduce impact attenuation during loading. There were no changes to lumbar kinematics or LM activation with AB. The observed altered biomechanics may have implications for training and rehabilitating the lower limb and spine during loading tasks and highlights the need for clinicians to carefully consider whether or not to advocate AB during these tasks. ◉

KEY POINTS

FINDINGS: Abdominal bracing resulted in reduced knee and hip flexion and increased vGRF during a drop-landing task.

IMPLICATIONS: Abdominal bracing during dynamic trunk-loading tasks may not be appropriate for some athletes who have musculoskeletal pain provoked by landing tasks.

CAUTION: This study was limited to a young, pain-free population.

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