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ORIGINAL RESEARCH

Modulation of Vertical Ground Reaction Impulse With Real-Time Biofeedback: A Feasibility Study

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Given its apparent representation of cumulative (*vs* peak) loads, this feasibility study investigates vertical ground reaction impulse (vGRI) as a real-time biofeedback variable for gait training aimed at reducing lower limb loading. Fifteen uninjured participants (mean age = 27 y) completed 12 2-min trials, 1 at each combination of 4 walking speeds (1.0, 1.2, 1.4, and 1.6 m/s) and 3 targeted reductions in vGRI (5, 10, and 15%) of the assigned ("target") limb, with the latter specified relative to an initial baseline (no feedback) condition at each speed. The ability to achieve targeted reductions was assessed using step-by-step errors between measured and targeted vGRI. Mean (SD) errors were 5.2% (3.7%); these were larger with faster walking speeds but consistent across reduction targets. Secondarily, we evaluated the strategy used to modulate reductions (ie, stance time or peak vertical ground reaction force [vGRF]) and the resultant influences on knee joint loading (external knee adduction moment [EKAM]). On the targeted limb, stance times decreased ($P < .001$) with increasing reduction target; first and second peaks in vGRF were similar ($P > .104$) across all target conditions. While these alterations did not significantly reduce EKAM on the target limb, future work in patients with knee pathologies is warranted.

Keywords: rehabilitation, gait training, biomechanics, knee

Abnormal joint mechanics are often associated with an elevated risk for joint pathology. During ambulation, increases in peak knee joint loads have been related to the severity or progression of existing joint degeneration[1] and, to some extent, its initiation.[2] Regarding the latter, noting that larger peak knee joint contact forces which occur in running compared to walking[3,4] do not necessarily increase the risk for joint injury, it has been suggested that characterization of other waveform features, such as the shape or duration, be considered when assessing general injury risk.[5] Moreover, the mechanical and biological response of joint tissues are also modulated by other (nonpeak) loading factors such as rate and duration. Given this, one might presume that such consideration for joint loading characterization would likely also apply in the context of movement-based rehabilitation, particularly with application to specific populations wherein increased and/or prolonged musculoskeletal loads are common (eg, persons with unilateral lower extremity trauma).[6] Yet, interventions that target or capture features of both the amplitude and temporal nature of loading are lacking.

Rehabilitation focused on gait modification has been used to control or alter mechanical loads during ambulation and ultimately minimize the risk for longer-term complications. Existing implementation of such strategies, particularly those involving (bio) feedback, is commonly based on discrete parameters (ie, peak of a given kinetic or kinematic variable). For example, kinetic-based biofeedback from a force-measuring treadmill improved symmetry in peak limb loading for patients with total hip arthroplasty (THA) and transtibial amputation,[7,8] and kinematic-based biofeedback strategies such as increasing lateral trunk motion[9] or toe-out angle[10,11] have been used to reduce frontal plane moments at the knee. As noted previously, approaches that focus exclusively on peak values may preclude effective biofeedback-based therapies to address injury risk more closely associated with accumulated loads. Furthermore, prior work indicates concomitant alterations in temporal characteristics of gait during biofeedback of peak loading; a result that may ultimately offset any perceptible reduction in injury risk.[12] Although the optimal selection of biofeedback variable(s) for given rehabilitation goals is presently unclear,[13] given the important interaction between peak and temporal loading characteristics on the mechanical environment of a joint,[14] biofeedback based on metrics reflecting both the magnitude and duration of applied load may be an effective strategy.

We anticipate that as the primary mechanisms of various lower extremity injuries are elucidated, gait retraining paradigms will adapt accordingly. For example, tibial stress fractures are thought to result from high magnitude *vs* high frequency loads,[15] suggesting feedback of purely kinetic variables may be most effective (eg,[8]). However, debate surrounds the pathomechanics of knee osteoarthritis (OA), where the extent to which individual peaks or accumulated loads are the primary driver of injury is unclear. However, larger knee adduction angular impulses have been shown to effectively distinguish between symptomatic and asymptomatic knee OA patients,[16] and radiologically determined severity of OA.[17] With the long-term goal of developing a biofeedback intervention for treatment of unilateral limb pathologies initiated or exacerbated by cumulative loads, we set out to evaluate the

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feasibility of utilizing a biofeedback strategy based on vertical ground reaction impulse (vGRI). This variable represents the cumulative load applied vertically to the lower limb throughout stance, but does not require real-time inverse dynamics calculations, establishing a low barrier to adoption. We theorize that vGRI could be an effective biofeedback parameter for controlling limb loading during gait and, thus, ultimately support its implementation as part of a gait (re)training paradigm. In this study, we hypothesized that uninjured individuals, when presented with visual vGRI-based biofeedback, would be able to accurately reduce their vGRI in response to a specified target. We also sought to (1) determine the predominant strategy used to modulate such reductions (ie, peak or temporal characteristics of applied load), and (2) describe any associated biomechanical changes linked to joint pathology (ie, knee joint moments, proximal movements) or influences on the contralateral (nontarget) limb. Of note is that overall compensatory strategies were of interest because unintended biomechanical consequences have been exhibited in previous biofeedback studies, specifically the onset of acute low back pain[9,13] and potential for long term injury.[18]

Methods

Fifteen uninjured adults (11 male/4 female) participated after completing informed consent procedures approved by the local Institutional Review Board. Mean (SD) age, stature, and body mass were 27 (6) years, 175.7 (10.8) cm, and 80.7 (17.1) kg, respectively. All participants reported being free of current or recent injuries, illnesses, and musculoskeletal disorders within the past 12 months, as well as major joint surgeries (such as ACL reconstruction) at any time. Participants also reported no dizziness or vestibular concerns, as these may affect gait or balance. Mean (SD) self-selected walking velocities were 1.10 (0.10) m/s, determined by increasing treadmill speed until participants felt they were walking at a "comfortable" pace.

A repeated measures design was used, in which biofeedback of vGRI was provided while walking within an instrumented virtual reality environment (Computer Assisted Rehabilitation Environment (CAREN); Motekforce Link, The Netherlands). Participants completed 12 2-minute walking trials representing each combination of 4 speeds (1.0, 1.2, 1.4, and 1.6 m/s) and 3 targeted reductions in vGRI (5, 10, and 15%), with the latter specified relative to mean vGRI obtained in a 2-minute baseline (no feedback) condition. The baseline condition was always performed first at a given speed; presentation order of walking speeds and targeted reductions were individually randomized to reduce potential order-related confounding effects. Each participant was informed of their target limb (left or right) prior to actual testing based on which presented with the larger vGRI during initial acclimation trial. After assignment, the target limb was consistent across all trials within each individual.

Prior to all trials, retro-reflective markers were placed in the mid-sagittal plane over the T10 and C7 spinous processes, sternal notch, and xiphoid; and bilaterally over the left and right acromion, anterior/posterior superior iliac spines, and lower extremities (1 tracking marker on the thighs and shanks in addition to medial and lateral axis markers at the ankles and knees). Participants subsequently walked within a virtual environment consisting of an endless (level) walkway surrounded by grass, trees, and other objects designed to mimic an outdoor, park-like scenario. Participants first completed a 4-minute acclimation period to the virtual reality environment at their self-selected walking speed. A baseline (no feedback) trial was then completed to obtain reference vGRI values at the first assigned walking speed. In all subsequent targeting trials, feedback was provided as an arrow that moved continuously along a vertical scale displayed in the middle of the scene (Figure 1); the vertical position of the arrow reflected the vGRI of the target limb (updated by the subsequent step) relative to the previously determined baseline value (for a given speed). Briefly, vGRI of the target limb was calculated within software (D-Flow, Motekforce Link, The Netherlands) that synchronizes the motion platform, treadmill, motion capture, image projection, and other data systems as the time integral of measured vertical ground reaction force (vGRF) during stance. Heel strike and toe-off events were determined using an ascending and descending threshold (80 N) in vGRF signals. A larger force threshold was required to improve reliability of detection due to noise in the force signals introduced by the hydraulic platform. vGRI feedback presented on the screen was thus from 1 gait cycle previous but displayed during contralateral stance in advance of the subsequent target limb step. When targeting, participants were simply instructed that impulse is "the amount of force applied over the period of time your foot is in contact with the ground"; importantly, no additional information or verbal cues were given as to not bias potential strategies used to achieve targeted reductions.

Throughout all trials, full-body kinematics were obtained by tracking (120 Hz) the positions of the reflective markers using a 12-camera motion capture system (T40 series, Vicon, Oxford, UK); vGRFs were simultaneously sampled (1200 Hz) from bilateral force platforms within the dual-belt treadmill (approximately 1.8 m L by 1.0 m W; Bertec, Columbus, OH). Full-body kinematic and vGRF data were recorded in the final 30 seconds of each condition for subsequent analyses.

To obtain the dependent measures, kinematic and GRF data were processed in Visual3D (C-Motion, Germantown, MD). First, raw GRFs and marker trajectories were low-pass filtered with a zero-lag, 4th order Butterworth filter at a cut-off frequency of 50 Hz and 6 Hz, respectively. vGRI were computed by integrating the time-varying vGRF during stance. To evaluate ability to achieve

Figure 1 — Virtual environment with biofeedback display: an arrow moved continuously along the vertical scale displaying the step-by-step vertical ground reaction impulse (vGRI) of the target limb (as a percentage relative to previously recorded baseline vGRI at a given speed). Values were reflective of the prior step of the target limb.

targeted reductions (objective 1), step-by-step errors were computed between measured vGRI and the targets estimated relative to recorded baseline values [$vGRI_{measured} - vGRI_{target}/vGRI_{target}$]; both absolute and true errors were calculated with and without the absolute value of the numerator, respectively. Additionally, 3 characteristics of vGRI (ie, first and second peaks of vGRF magnitude and stance time) were individually evaluated to assess predominant strategies used to obtain such reductions (objective 2). Several other factors, potentially influenced by/contributing to the chosen strategy, were also evaluated: step rate, step length, stride width, peak external knee adduction moment (EKAM), knee adduction angular impulse, and peak lateral trunk flexion (objective 3). External knee adduction moments (EKAM) were calculated from raw marker and force data, then low-pass filtered at 6 Hz using a 4th order Butterworth filter. Time varying EKAM values were integrated during stance to generate knee adduction angular impulses. Where appropriate, dependent measures were separately analyzed for both the target and nontarget limb.

A set of 2-way (speed × target) repeated measures analyses of variance (ANOVAs) were used to compare all dependent measures, using SPSS (version 21.0; IBM SPSS Inc., Chicago, IL, USA), with significance determined at $P < .05$. For variables with significant main effects, post hoc comparisons were made using Tukey's honestly significant difference test. Trial order significantly influenced target limb knee adduction angular impulse ($P = .025$), but no other dependent measure ($P > .07$). Data are presented as means (standard deviations). Across all 30-second collections, the number of recorded (clean) target/nontarget steps = 28 (5)/29 (4), respectively.

Results

During baseline trials, vGRI on the target limb were 0.59 (0.14), 0.55 (0.13), 0.52 (0.13), and 0.50 (0.12) BW-s at 1.0, 1.2, 1.4, and 1.6 m/s, respectively; these were similar ($P = .14$) between target and nontarget limb (ie, bilaterally symmetric).

When targeting, absolute errors across all conditions were 5.2 (3.7) %; these were larger ($P = .02$) with increasing speed, but consistent ($P = .36$) across reduction targets. True errors, however, increased with both walking speed ($P = .04$) and reduction target ($P = .01$); Figure 2A). Absolute errors tended to decrease ($P = .05$) with trial order (Figure 2B).

On the targeted limb, first peaks in vGRF were similar ($P = .95$) across all 3 target conditions, though they did increase ($P < .001$) with walking speed (Table 1; Figure 3). On the contralateral

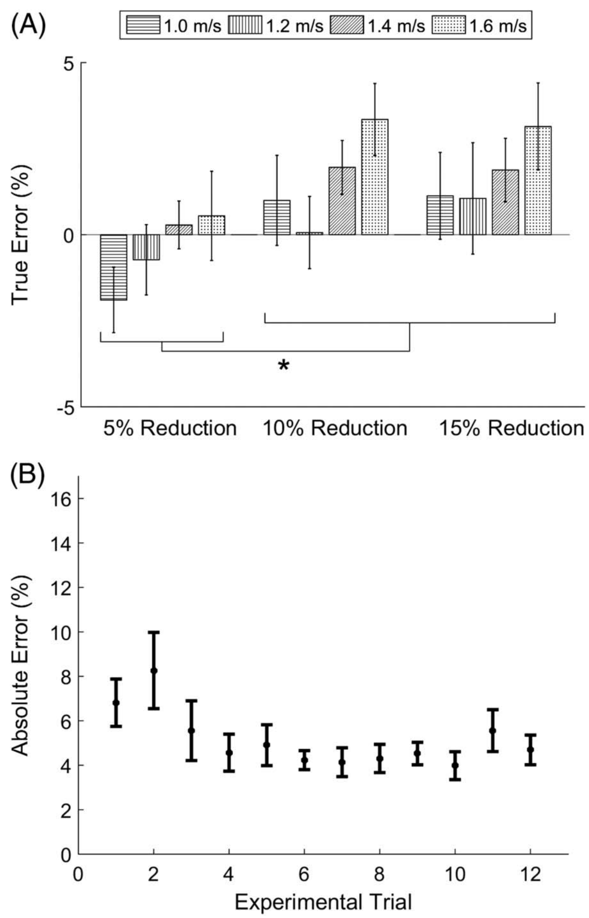

Figure 2 — Mean targeting errors. (A) True error in measured *vs* estimated vertical ground reaction impulse (vGRI) by targeted reduction and speed. Post hoc comparisons are indicated by brackets (* = significant difference between reduction targets). (B) Absolute errors in vGRI by experimental trial (independent of reduction target/walking speed). Error bars indicate standard error.

Table 1 Mean ± Standard Deviation First Peak of Vertical Ground Reaction Force (vGRF) and Stance Times of the Targeted and Nontargeted Limbs, by Reduction and Walking Speed

	Speed (m/s)			
Reduction	1.0	1.2	1.4	1.6
First Peak vGRF (BW)				
Target Limb				
Baseline	1.16 ± 0.27	1.22 ± 0.32	1.31 ± 0.33	1.41 ± 0.24
5%	1.14 ± 0.26	1.21 ± 0.29	1.30 ± 0.33	1.39 ± 0.23
10%	1.17 ± 0.26	1.20 ± 0.27	1.28 ± 0.32	1.39 ± 0.24
15%	1.16 ± 0.24	1.20 ± 0.25	1.27 ± 0.32	1.38 ± 0.23
Nontarget Limb				
Baseline	1.16 ± 0.28	1.23 ± 0.31	1.32 ± 0.32	1.41 ± 0.27
5%	1.18 ± 0.28	1.26 ± 0.29	1.37 ± 0.30	1.45 ± 0.27
10%	1.21 ± 0.29	1.32 ± 0.29	1.40 ± 0.32	1.49 ± 0.28
15%	1.25 ± 0.30	1.36 ± 0.30	1.47 ± 0.29	1.55 ± 0.28
Stance Time (s)				
Target Limb				
Baseline	0.78 ± 0.05[a]	0.72 ± 0.04[a]	0.67 ± 0.04[a]	0.63 ± 0.06[a]
5%	0.74 ± 0.04[b]	0.68 ± 0.04[b]	0.64 ± 0.05[ab]	0.61 ± 0.05[ab]
10%	0.72 ± 0.05[bc]	0.67 ± 0.04[bc]	0.63 ± 0.04[ab]	0.59 ± 0.05[ab]
15%	0.69 ± 0.05[c]	0.64 ± 0.04[c]	0.60 ± 0.04[b]	0.57 ± 0.06[b]
Nontarget Limb				
Baseline	0.78 ± 0.04[a]	0.72 ± 0.04[a]	0.67 ± 0.04[a]	0.63 ± 0.05[a]
5%	0.74 ± 0.04[b]	0.68 ± 0.04[ab]	0.64 ± 0.03[ab]	0.61 ± 0.05[a]
10%	0.73 ± 0.04[b]	0.68 ± 0.04[ab]	0.63 ± 0.03[b]	0.60 ± 0.06[a]
15%	0.71 ± 0.06[b]	0.65 ± 0.05[b]	0.62 ± 0.04[b]	0.58 ± 0.08[a]

Abbreviation: vGRF, vertical ground reaction force.
Note. Peak vGRF are normalized to body weight (BW). Superscripted letters by individual values indicate post hoc comparisons between reduction targets (within each speed individually), whereby those not connected by the same letter are significantly different.

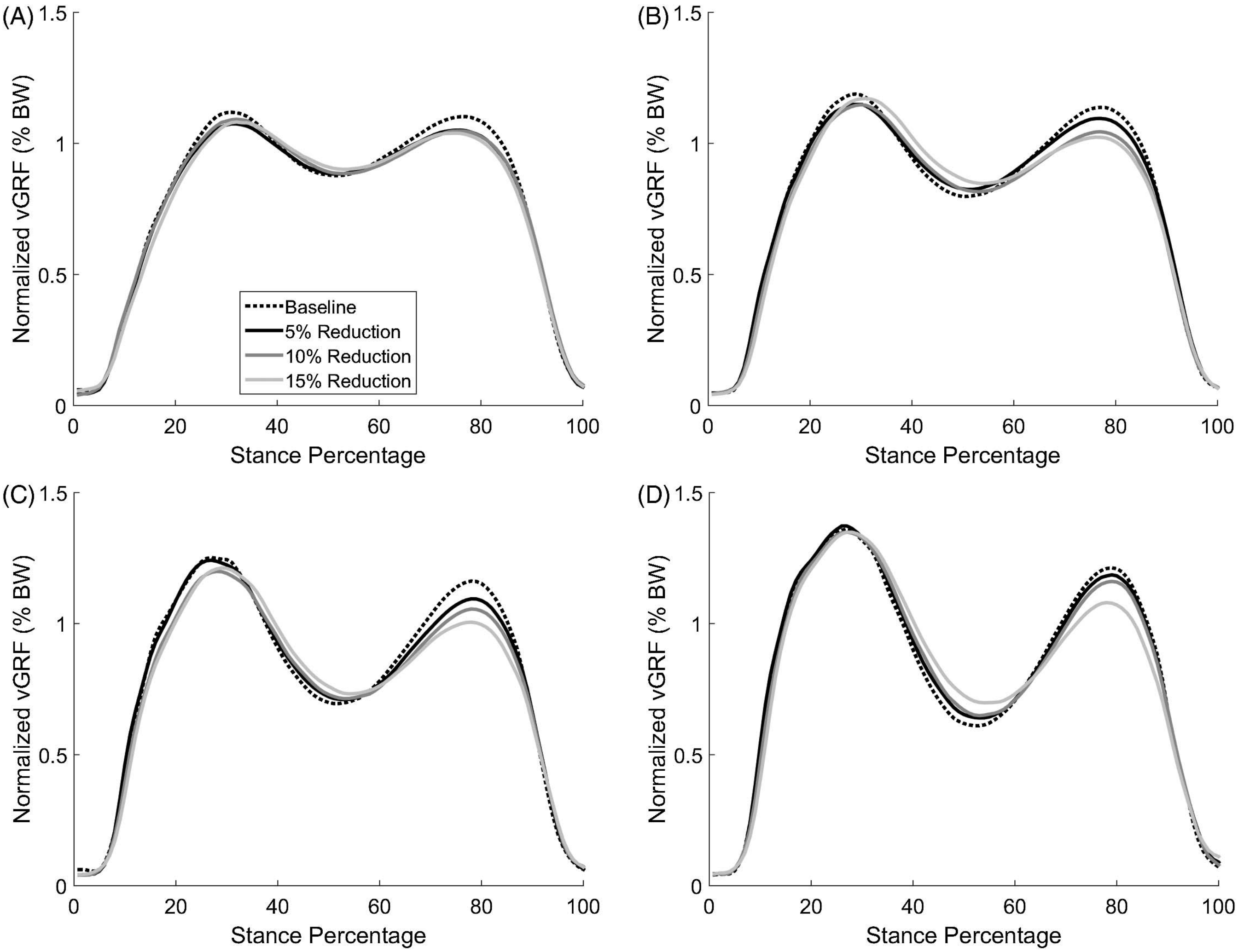

Figure 3 — Vertical ground reaction forces (vGRF) time-normalized to stance of the target limb. Curves represent ensemble averages by reduction target for 1.0, 1.2, 1.4 and 1.6 m/s walking speeds (A–D, respectively).

(nontarget) limb, first peaks in vGRF tended ($P = .08$) to increase with larger reduction targets; however, these increases did not appreciably ($P = .80$) alter contralateral vGRF impulses. For both target (Figure 3) and nontarget limbs, second peaks in vGRF were similar ($P > .10$) across walking speeds and reduction targets. Stance times on both the targeted and nontargeted limbs decreased with walking speed ($P < .001$) and reduction target ($P < .001$; Table 1).

Speed and target were not significant effects for peak EKAM on the targeted limb ($P > .20$), though mean values generally decreased slightly with increasing speed and reduction target—a trend which was consistent for target and nontarget peak EKAM and angular impulse (Table 2). On the nontargeted limb, peak EKAM decreased significantly with speed ($P < .001$), but not reduction target ($P = .53$). Similarly, knee adduction angular impulse on the target limb decreased significantly with speed ($P < .001$), but not target ($P = .10$), while knee adduction angular impulse on the nontarget limb decreased significantly with both speed ($P < .001$) and target ($P = .02$).

For both the target/nontarget limbs, step rates increased ($P < .001$) with target reduction at 119.6 (12.0)/118.5 (12.0), 122.2 (11.5)/121.0 (13.7), and 125.6 (14.1)/127.3 (12.7) steps/min for 5, 10, and 15% reductions, respectively. Similarly, step lengths on both the target/nontarget limbs decreased ($P < .001$); 5, 10, and 15% reductions resulted in step lengths of 0.65 (0.09)/0.66 (0.11), 0.63 (0.10)/0.65 (0.13), and 0.62 (0.10)/0.61 (0.10) meters, respectively. Peak trunk lateral flexion angles were similar ($P > .796$) across speeds and target reductions. Importantly, no significant (all $P > .89$) speed × target interactions were observed for any dependent measure.

Discussion

This study evaluated the ability of uninjured individuals to modulate vertical limb loading in response to visual feedback of vGRI while walking. Participants were able to reduce vGRI toward requested targets, primarily via minimizing the duration of stance rather than peak loading. The chosen strategies did not appear to affect proximal joint motions, specifically lateral trunk lean; however, the first peak of vGRF tended to increase on the contralateral limb with increasing target reductions (despite no appreciable change in vGRI; likely related to simultaneous reductions in stance time). Additionally, metrics linked to the presence and severity of knee osteoarthritis did generally decrease with targeted reduction,

Table 2 Mean ± Standard Deviation Overall Peak of External Knee Adduction Moment (EKAM) and Knee Adduction Angular Impulse of the Targeted and Nontargeted Limbs, by Reduction and Walking Speed

	Speed (m/s)			
Reduction	1.0	1.2	1.4	1.6
*Peak EKAM (N*m/BW)*				
Target Limb				
Baseline	0.37 ± 0.18[a]	0.36 ± 0.17[a]	0.34 ± 0.15[a]	0.32 ± 0.16[a]
5%	0.35 ± 0.19[ab]	0.34 ± 0.19[ab]	0.33 ± 0.16[a]	0.31 ± 0.16[a]
10%	0.34 ± 0.20[ab]	0.35 ± 0.20[ab]	0.32 ± 0.16[a]	0.31 ± 0.17[a]
15%	0.34 ± 0.21[b]	0.33 ± 0.20[b]	0.31 ± 0.17[a]	0.30 ± 0.18[a]
Nontarget Limb				
Baseline	0.48 ± 0.12[a]	0.45 ± 0.12[a]	0.45 ± 0.12[a]	0.43 ± 0.11[a]
5%	0.47 ± 0.11[a]	0.45 ± 0.10[a]	0.43 ± 0.12[a]	0.43 ± 0.11[a]
10%	0.46 ± 0.11[a]	0.44 ± 0.11[a]	0.42 ± 0.11[a]	0.42 ± 0.10[a]
15%	0.46 ± 0.11[a]	0.44 ± 0.11[a]	0.44 ± 0.11[a]	0.42 ± 0.09[a]
*Knee Adduction Angular Impulse (N*m*s/BW)*				
Target Limb				
Baseline	0.13 ± 0.11[a]	0.10 ± 0.10[a]	0.08 ± 0.09[a]	0.05 ± 0.09[a]
5%	0.11 ± 0.11[b]	0.09 ± 0.10[ab]	0.07 ± 0.09[ab]	0.05 ± 0.09[a]
10%	0.10 ± 0.11[b]	0.09 ± 0.11[ab]	0.07 ± 0.09[bc]	0.04 ± 0.10[a]
15%	0.10 ± 0.11[b]	0.08 ± 0.10[b]	0.06 ± 0.09[c]	0.04 ± 0.10[a]
Nontarget Limb				
Baseline	0.20 ± 0.07[a]	0.17 ± 0.06[a]	0.14 ± 0.06[a]	0.11 ± 0.05[a]
5%	0.19 ± 0.07[ab]	0.15 ± 0.05[ab]	0.13 ± 0.06[ab]	0.10 ± 0.05[ab]
10%	0.18 ± 0.07[b]	0.15 ± 0.06[ab]	0.12 ± 0.05[b]	0.10 ± 0.05[ab]
15%	0.17 ± 0.07[b]	0.14 ± 0.05[b]	0.12 ± 0.05[b]	0.09 ± 0.05[b]

Abbreviation: EKAM, External Knee Adduction Moment.
Note. Both metrics are normalized to body weight (BW). Superscripted letters by individual values indicate post hoc comparisons between reduction targets (within each speed individually), whereby those not connected by the same letter are significantly different.

though this trend was not significant for the majority of such measures.

The concept of using vGRI as a biofeedback metric has been of interest for some time,[7] but to our knowledge has not been previously implemented due to existing hardware or software limitations. Here, we were able to calculate vGRI of the target limb such that it was displayed on the screen during contralateral stance and thus provided sufficient time to observe and modify the subsequent step accordingly. In agreement with our hypothesis, participants were able to modulate vGRI using visual feedback within 5% of the target, which suggests at least a 10% reduction is needed. A trend toward decreasing error over the course of the experiment also suggests participants should be provided a minimum of 3 to 4 training trials with this particular biofeedback. Further, the significant increases in absolute and true error of vGRI with increased walking speed indicate that reconciling the strategies used to maintain faster walking speeds, such as larger first and second peaks in vGRF,[19] with those of impulse reduction on a single limb is particularly challenging. Nevertheless, the absence of speed × target interactions suggests that walking speed did not differentially affect targeting ability between targeted reduction levels.

When able-bodied individuals target unilateral reductions in vGRI, our results indicate temporal characteristics (stance time and step rate) dominate over changes in the magnitude of (peak) load, consistent with prior work.[20] Such a preference has also been observed, albeit indirectly, in prior work providing feedback based on peak vGRF, whereby patients with THA improved symmetry in vGRI but not peak vGRF.[7] With increasing walking speed, persons with unilateral transtibial and transfemoral amputation exhibit less kinetic symmetry, but greater temporal symmetry,[21] suggesting that a preference for temporal over kinetic modulation may be present not only in patients with THA, but also persons with unilateral lower limb amputation. Our results suggest that vGRI reduction, which appears to be primarily accomplished by reducing stance duration more so than kinetic magnitudes, does not significantly reduce peak EKAM nor knee adduction angular impulse on the target limb, though both of these metrics did tend to decrease with targeted reduction. Further study may be warranted to evaluate whether this general trend becomes significant for individuals with existing knee joint pathologies (eg, osteoarthritis). Additionally, despite increasing trends in the magnitude of vGRF contralaterally, reductions in stance time of the nontarget limb further support the use of vGRI (*vs* peak vGRF) to modify/reduce knee-specific loading metrics (Table 2).

When evaluating efficacy of a given biofeedback metric, compensatory (unintended) adaptations should also be considered. Among kinematic compensatory responses to vGRI reduction, specifically, trunk angle was of particular interest given prior indications of acute onset of low back pain during biofeedback,[9,13] and the potential for longer term injury.[18] No significant relationship between trunk angle and target reduction was observed in this study, though marginal increases in the base of support were observed. However, in offloading the targeted limb, compensatory kinetic changes were observed contralaterally; specifically, a tendency toward a larger first peak in vGRF. When taken together with both the decreased stance time and maintained impulse on the contralateral side, such a finding suggests impulse and temporal symmetry are preferred at the expense of kinetic symmetry among able-bodied individuals. Interestingly, faster step rates and shorter step lengths with increasing impulse reduction are consistent with prior step rate biofeedback associated with reduced absorption of energy at the knee[12]; impulse modulation training may thus have unforeseen therapeutic benefits. To decouple the influence of impulse modulation and changes in step rate here, subsequent studies may look to combine vGRI biofeedback with additional cueing of stride frequency.[22]

Several limitations should be noted when interpreting the current results. First, given the preliminary nature of the study, participants only completed a single session; thus, dose-response relationships and potential retention or application to overground walking cannot be determined. Second, increasing the force threshold to improve reliability of event detection in real time ultimately reduced the magnitude of vGRI, though such an effect was consistent across all trials. Third, although participants were able to reduce vGRI, the metabolic and cognitive demands associated with the targeting strategies and different types of feedback have only begun to be explored.[23] Regarding the former, improvements in kinetic and temporal-spatial symmetry with real-time biofeedback have resulted in improved metabolic efficiency for stroke patients,[24] while changes from preferred temporal-spatial parameters in able-bodied individuals can increase energetic cost.[25,26] Regarding the latter, although participants indicated some difficulty in targeting based on walking speed, they self-reported clear perception and control of the metric during feedback; additional work can attempt to improve vGRI-based feedback given other important neural correlates of feedback and learning.[27] Finally, given that mean self-selected ("comfortable") walking speeds were on the lower end of the experimental walking speeds tested here, slower (<1.0 m/s) walking speeds should also be evaluated, particularly if eventually applied to various patient populations who may generally walk slower.

In summary, this feasibility study supports the use of vGRI as a biofeedback variable toward the reduction of vertical loads applied to the limb while walking. Feedback based on the impulse of other gait parameters (eg, at a specific joint) and/or in the mediolateral and anteroposterior directions may be warranted, though the relatively smaller magnitude of such forces or moments may preclude such an approach. Additionally, an important consideration throughout the development of future iterations of biofeedback training is the resources necessary for implementing biomechanical feedback (eg, the need for motion capture and/or instrumented treadmill systems for metrics derived from inverse dynamics). Avenues for future study also include identification of the appropriate targeting resolution and evaluating the longer-term effects of vGRI biofeedback training including retention of changes in daily life. However, vGRI may be a promising focus of gait training paradigms for specific patient populations to minimize musculoskeletal loads during gait and mitigate risk for secondary complications.

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