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BALANCE RECOVERY TO IMPROVE CLINICAL CONCUSSION PROTOCOL

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Abstract— **Current clinical concussion assessments do not evaluate patients in a manner that mimics the demand of the real world. Balance is impaired after concussion, but it is assessed in static, unrealistic conditions. Failure to assess real-world balance demands may not detect subtle deficits that could increase the risk of subsequent injury to the lower extremities. Through more realistic balance assessments that include dynamic perturbations, turning maneuvers, and increased walking speeds, we aim to increase the physical demand of the task and better reflect to the patients' vulnerability in the real world.**

The aim of this research is to analyze the turning gait of healthy control subjects and concussion patients in response to underfoot perturbations delivered by a custom-designed pair of shoes while performing continuous turning tasks. The shoes allow for both reaction time and postural stability to be analyzed during locomotion by means of inertial measurement units that measure centripetal acceleration. The results of one healthy control subject show a difference in the balance recovery response at a self-set normal and fast speed. By observing patterns in how healthy and concussed individuals respond to the perturbations while turning, we intend to better detect persistent neuromuscular control deficits that may be linked to subsequent injuries to the lower extremities in concussion patients.

Index Terms— **Biomechanics, Gait, Locomotion, Mild Traumatic Brain Injury (mTBI), Stability**

I. Introduction

Annually, 1.4 million Americans sustain a concussion, many of whom continue to suffer from the consequent symptoms after returning to their daily lives, sports, or active duty [1]. Currently, there is no good objective measure of when one is ready to return to play, work, or duty post-concussion. Recovery from a concussion is determined from clinical evaluations, but these clinical assessments do not evaluate patients in a manner that mimics the demand of the real world. Although much of the concussion's cognitive symptoms typically resolve within days or weeks of the head injury, subtle balance deficits may persist for years post-concussion [2]. In a clinical setting, balance is assessed in static, sterile conditions, which do not properly detect the subtle balance deficits that persist longer [3]. In the real world, individuals are exposed to higher demands than in static testing. Individuals must be able to navigate uneven ground, respond to unexpected obstacles, and change directions, all of which challenge one's balance.

Those who sustain a concussion are more likely to incur a future injury to their lower extremities (LE) [4]. It is possible that failure of clinical concussion protocol to assess realistic balance demands may lead to undetected balance impairments post-concussion that could increase the risk of LE injury. It remains unknown why, exactly, concussion patients have an increased susceptibility to LE injury [2], [4]–[7], but examining gait kinematics following perturbations while performing dynamic tasks may provide insight as to why the risk is greater [8].

The aim of this research is to collect more realistic balance assessments to provide a better assessment of the neuromuscular control deficits that may associate with future injury. To accomplish this aim, we analyzed the gait of healthy control subjects and concussion patients in response to unexpected perturbations (postural disturbances) while performing dynamic turning tasks at various walking speeds. The underfoot perturbations are delivered by a custom-designed, mechanized shoe device throughout a walking trial, during which acceleration values are collected via inertial sensors on the feet and trunk. This process allows for both reaction time and postural stability to be analyzed during locomotion.

The dynamic nature of this study allows for a better understanding of the underlying postural deficits in concussion patients, as the static balance tests and symptom reports used in clinical practice are not sufficient in providing a clear picture of the subtle deficits that may impact daily life. Furthermore, adding a turning maneuver and increased walking speed to the dynamic perturbations escalates the physical demand of the task and risk for further injury, making it more applicable to the patients' vulnerability in the real-world. In mimicking the demands of the real world, we hope to better detect persistent balance deficits that may be linked to subsequent injuries to the lower extremities in concussion patients. In the future, more realistic balance assessments could be implemented in clinical settings to better reflect recovery in individuals with concussions. Better reflections of the recovery of concussed individuals may allow for more development in personalized physical therapy and treatment of concussions in the future.

II. Background

Concussion symptoms include cognitive deficits, as well as impaired neuromuscular control [1]. Current clinical evaluations assess the presence and severity of these deficiencies through computer-based cognitive assessments and physical balance tests [9]. Some symptoms resolve quickly, as determined from baseline scores on cognitive assessments, but subtle deficits can persist for years after injury [2]. Current clinical evaluations lack the sensitivity to detect these subtleties [3]. Static balance tests can detect balance deficits due to concussion within the first few days post-injury [10]. But, these static balance tasks aren't sufficient in detecting deficits that last longer the acute phase (1-2 weeks) of a concussion [11].

The deficits in neuromuscular control that concussion patients experience may contribute to their susceptibility to subsequent injuries to the lower extremities. Altered or diminished postural stability has been shown to increase one's odds of injury to the lower extremities by seven times [6]. In studying professional and collegiate athletes with concussion history, it was found that a concussion increased a player's odds of sustaining an injury to their lower extremities by 60% or more depending on the time since the head injury [7]. While it is unknown why, exactly, concussion patients have an increased susceptibility to injury to the lower extremities, examining gait kinematics following perturbations may provide insight as to why the risk is greater [8].

Dynamic tests of balance are better than static tests at detecting subtle deficits post-concussion [12]. Concussion patients appear to adopt a more conservative gait pattern in order to account for their lack of balance [13]. This conservative gait pattern is indicated by a decrease in velocity, an increase in mediolateral sway [14], as well as an increase in weight-shifting time, and changes in step length, step time, and step width [13]. This gait pattern can persist over six years after the acute, or symptomatic, phase of a concussion that lasts only a few weeks [9]. Additionally, those who walk slower, regardless of the underlying cause, are more prone to LE injury [15]. Although dynamic tests are better at detecting deficits, neither static nor straight-line walking tasks adequately represent daily life.

In the real world, our daily activities are not confined to straight paths. Rather, people must change directions often. We perform between 800 and 1000 turns per day [16] and 35-45% of our steps are not straight [17]. Turning and tasks that require changing direction require rapid reorientation of the head, trunk, pelvis, knees and ankles [18]; these adjustments require a shift in center of mass and, therefore, whole-body coordination. Such movements can worsen symptoms of concussion [19]. Therefore, individuals with a concussion may limit their reorientation to minimize the exacerbation of symptoms by turning at a slower speed or adjusting their gait kinematics [18]–[21].

Only a few small, preliminary studies have analyzed the gait kinematics of turning in the post-concussion population. Powers and colleagues demonstrated variability in limb shifting at the start of turning in concussion patients and showed this variability persisted after the patients were deemed clinically recovered [12]. Additionally, a study by Fino et al. showed less curvature and bodily reorientation in those with concussions; this pattern remained for up to one-year post-injury [22]. The turns in this study were methodical in that they were prompted to be at specific angles (45°, 60°, or 90°) and at specific points in dynamic tasks. Therefore, the turns examined in previous studies may not be representative of daily life. A later study by Fino et al. compensated for this lack of association to the real world by examining the gait kinematics of concussion patients on a less-structured turning course. This course left much of the turning mechanics (such as trajectory and speed) to individual subject discretion. In this study, the concussion group exhibited slower readjustment of the pelvis and trunk, slower turning velocity, and overall slower completion of the designated course [5]. The allowance of variability in turning strategies makes this study the most applicable to a real-world setting; however, it does not account for recurrent turning steps over longer distances. Longer turning maneuvers are frequent in daily life and may prompt further challenges to balance, particularly in balance-impaired populations such as those who have experienced concussions.

By examining locomotion, turning, and perturbations in a single balance assessment, we hope to better reflect the demands of a patient's daily life, sport, and duty in order to identify subtle balance deficits that persist post-concussion and that may be linked to the risk of subsequent injury to the lower extremities.

III. Methods

A. Subjects

One healthy control subject was recruited for the aim of determining the balance recovery response during turning. Initially, only healthy controls will be recruited in order to finalize protocol that shows to be unlikely to exacerbate symptoms in concussion patients. In the future, healthy control will be age- and gender-matched to each concussion subject. Inclusion criteria are: 18-50 years of age, able to walk unassisted, no major injuries or surgeries to the lower extremities, have a shoe size that matches one of 10 pre-built pairs of perturbing shoes (EU sizes 37- 46), and be willing and able to provide informed written consent. Exclusion criteria for control subjects are a concussion within the past year or self-reported symptoms consistent with concussion or imbalance. Recruitment of subjects came from the University of Utah and the surrounding Salt Lake City area. Recruitment procedures and experimental protocols were approved by the University of Utah Institutional Review Board.

B. Questionnaires

After obtaining informed consent, participants completed several self-reported questionnaires to assess self-reported symptoms, physical activity level, quality of life, and concussion history.

C. Innovation

Subjects wore shoes capable of delivering an unexpected underfoot perturbation based on the design utilized by Kim et al. [8]. The shoes allow for normal gait, but when triggered, an aluminum flap is deployed from the shoe creating an ankle eversion of ~12° when unloaded (Figure 1). The perturbation is designed to be ecologically valid; it resembles stepping on a small rock. The flap is deployed during the swing phase of the foot, remains deployed through stance, and recedes back into the shoe during the following swing phase. The magnitude (12°) and timing (through stance) of the perturbation were constant across trials and subjects, allowing controlled intra- and inter-subject comparisons.

D. Protocol

Two concentric circles with diameters of approximately 2.4 meters and 3.6 meters were marked on the ground, creating a distinct path for the subjects to walk around. Subjects walked clockwise around the circle. Prior to the completion of the block trials, subjects performed two baseline speed trials wherein they will walk around the circle for four laps at a normal and fast pace, respectively. The second and third laps at each speed were recorded for the purpose of setting a metronome to reflect each walking speed. Subjects were expected to cover one-quarter of the circle with each beat from the metronome in the turning block trials. Trial blocks were
randomized between normal clockwise, normal counter-clockwise, fast clockwise, fast counter-clockwise, expected clockwise and expected counter-clockwise. Each trial block consisted of an unperturbed 1-min washout trial, followed by a 3-minute perturbed trial and an additional unperturbed 1-min washout trial at the appropriate speed and in the appropriate direction. Throughout the 3-minute perturbed trial, subjects were exposed to 16-20 perturbations on the left foot. The expected trial blocks incorporated an audible beep that warned of the perturbation that would occur on the following step. Kinematics for each trial were recorded via inertial sensors (APDM, Inc, city) placed on the head, sternum, trunk, and feet. Perturbation triggers were timestamped within the inertial sensor acceleration data.

E. Data Analysis

The primary kinematic outcome was the change in centripetal acceleration of the feet with the underfoot perturbation. The secondary kinematic outcome was the change in centripetal acceleration of the feet in preparation for and in recovering from the perturbation step. Analysis of inertial sensor data included examining both the local and global reference frames calculated based on the head, trunk, and foot sensors. In MATLAB, gait events were identified using a continuous wavelet transform. Each trial was then filtered, and each stride was time-normalized. Statistical analysis is to be determined upon testing additional subjects.

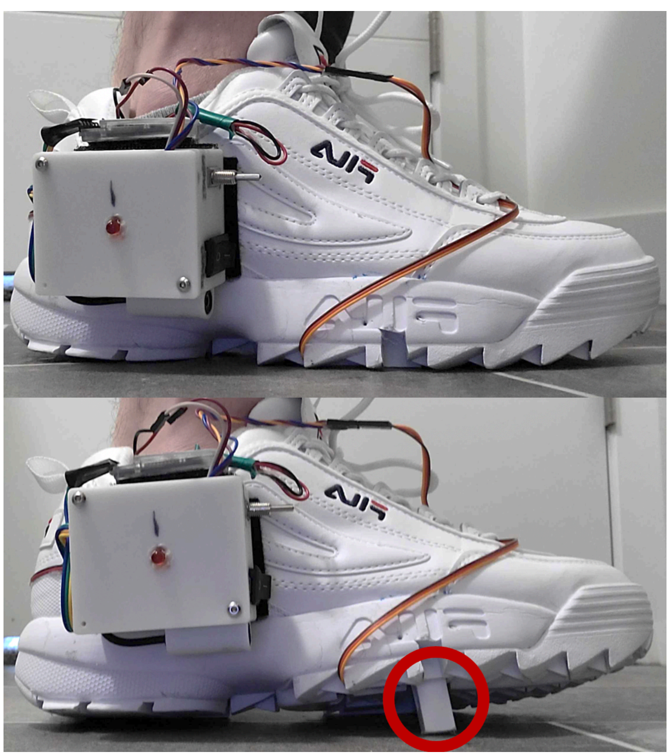

Fig. 1. Custom-built shoes capable of delivering unexpected and expected perturbations during locomotion. Top: Shoe shown without flap deployed and weighted. Bottom: Shoe shown with flap deployed and unweighted.

IV. Results

A. Normal, Preparatory, Perturbed, and Recovery Steps at Normal and Fast Speeds

Centripetal acceleration data was analyzed for one subject (n=1) to yield preliminary results for the balance recovery response to underfoot perturbations during turning. To first quantify the balance recovery response, the unperturbed steps (3 or more steps prior to the perturbation and 4 or more steps following the perturbation) were compared to the two preparatory steps on the left and right feet, respectively, perturbed steps on the left foot, and the following three recovery steps on the right, left, and right feet, respectively.

At a normal walking speed, unperturbed steps on the left foot had a median centripetal acceleration value of 0.12g with a minimum value of 0.08 g and maximum value of 0.147g (See Figure 2A). In the same trial, unperturbed steps on the right foot had a median centripetal acceleration value of 0.05g with a minimum value of 0.01g and maximum value of 0.10g (See Figure 2A).

The preparatory step on the left foot (-2) had a median centripetal acceleration value of 0.13g with a minimum value of 0.08g and maximum value of 0.15g (See Figure 2A). The preparatory step on the right foot (-1) had a median centripetal acceleration value of 0.035g with a minimum value of 0.01g and maximum value of 0.08g (See Figure 2A). Thus, in preparation for the perturbation step at a normal walking speed, there was an increase in centripetal acceleration on the left foot and a decrease in acceleration on the right foot in comparison to the normal steps.

The perturbed step on the left foot (PTB) had a median centripetal acceleration value of 0.09g with a minimum value of 0.08g and maximum value of 0.14g (See Figure 2A). The perturbed step at a normal walking speed, then, displayed a decrease in acceleration compared to normal steps on the left foot.

The first recovery step on the right foot (+1) had a median centripetal acceleration value of 0.06g with a minimum value of 0.01g and maximum value of 0.09g (See Figure 2A). The second recovery step on the left foot (+2) had a median centripetal acceleration value of 0.13g with a minimum value of 0.08g and maximum value of 0.14g (See Figure 2A). The third recovery step on the left foot (+3) had a median centripetal acceleration value of 0.06g with a minimum value of 0.01g and maximum value of 0.08g (See Figure 2A). Therefore, there was an increase in acceleration for the recovery steps on both feet.

At an increased walking speed, unperturbed steps on the left foot had a median centripetal acceleration value of 0.13g with a minimum value of -0.02g and maximum value of 0.19g (See Figure 2B). In the same trial, unperturbed steps on the right foot had a median centripetal acceleration value of 0.08g with a minimum value of -0.06g and maximum value of 0.13g (See Figure 2B).

The preparatory step on the left foot (-2) had a median centripetal acceleration value of 0.14g with a minimum value of -0.01g and maximum value of 0.17g (See Figure 2B). The preparatory step on the right foot (-1) had a median centripetal acceleration value of 0.09g with a minimum value of -0.02g and maximum value of 0.12g (See Figure 2B). Thus, in preparation for the perturbed step at a faster pace, there is, instead, an increase in centripetal acceleration for both feet.

The perturbed on the left foot (PTB) had a median centripetal acceleration value of 0.13g with a minimum value of -0.01g and maximum value of 0.18g (See Figure 2B), which is almost no change in acceleration from the normal steps.

The first recovery step on the right foot (+1) had a median centripetal acceleration value of 0.06g with a minimum value of -0.04g and maximum value of 0.12g (See Figure 2B). The second recovery step on the left foot (+2) had a median centripetal acceleration value of 0.13g with a minimum value of 0.05g and maximum value of 0.18g (See Figure 2B). The third recovery step on the left foot (+3) had a median centripetal acceleration value of 0.04g with a minimum value of -0.02g and maximum value of 0.11g (See Figure 2B). There was, then, a decrease in acceleration on the right foot but, again, no change in acceleration on the left foot to recover from the perturbation at a faster walking speed.

B. *Normal and Perturbed Trials at Normal and Fast Speeds*

To determine the efficacy of the unperturbed washout trials, the steps of these trials at both walking speeds (normal and fast) were compared to the unperturbed steps of the perturbation trial. At a normal walking speed, the pre-perturbation washout trial had a median centripetal acceleration value of 0.11g, a minimum value of 0.08g, and maximum value of 0.16g on the left foot and a median value of 0.06g, a minimum value of 0.02g, and a maximum value of 0.11g on the right foot (See Figure 3A). At a normal walking speed, unperturbed steps on the left foot had a median centripetal acceleration value of 0.12g with a minimum value of 0.08g and maximum value of 0.15g (See Figure 3A). In the same trial, unperturbed steps on the right foot had a median centripetal acceleration value of 0.05g with a minimum value of 0.01g and maximum value of 0.10g (See Figure 3A). At a normal walking speed, the post-perturbation

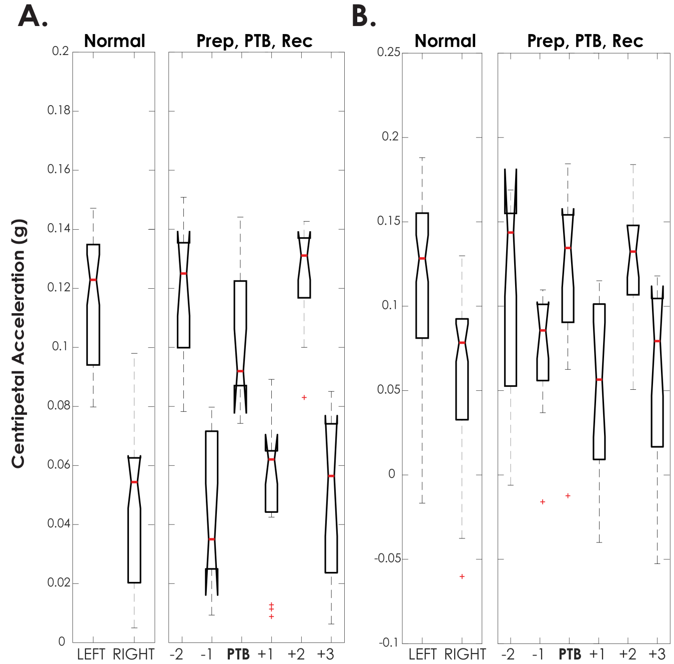

Fig. 2. Difference in centripetal acceleration (g) between the normal, two preparatory (-2,-1), perturbed (PTB), and three recovery steps (+1, +2, +3) of a perturbation trial in the clockwise direction (A) at a normal walking speed and (B) at an increased walking speed for one healthy subject (n=1).

washout trial had a median centripetal acceleration value of 0.12g, a minimum value of 0.06g, and maximum value of 0.15g on the left foot and a median value of 0.06g, a minimum value of 0.01g, and a maximum value of 0.10g on the right foot (See Figure 3A). At a normal speed, then, the normal steps of the perturbed trial and both washout trials had no change in acceleration for either foot.

At an increased walking speed, the pre-perturbation washout trial had a median centripetal acceleration value of 0.13g, a minimum value of 0.04g, and maximum value of 0.23g on the left foot and a median value of 0.08g, a minimum value of -0.03g, and a maximum value of 0.15g on the right foot (See Figure 3B). At an increased walking speed, unperturbed steps on the left foot had a median centripetal acceleration value of 0.13g with a minimum value of -0.02g and maximum value of 0.19g (See Figure 3B). In the same trial, unperturbed steps on the right foot had a median centripetal acceleration value of 0.08g with a minimum value of -0.06g and maximum value of 0.13g (See Figure 3B). At an increased walking speed, the post-perturbation washout trial had a median centripetal acceleration value of 0.15g, a minimum value of 0.01g, and maximum value of 0.23g on the left foot and a median value of 0.08g, a minimum value of -0.05g, and a maximum value of 0.11g on the right foot (See Figure 3B). At an increased walking speed, no change in acceleration was observed for either foot between the normal steps of the perturbed trial and both washout trials.

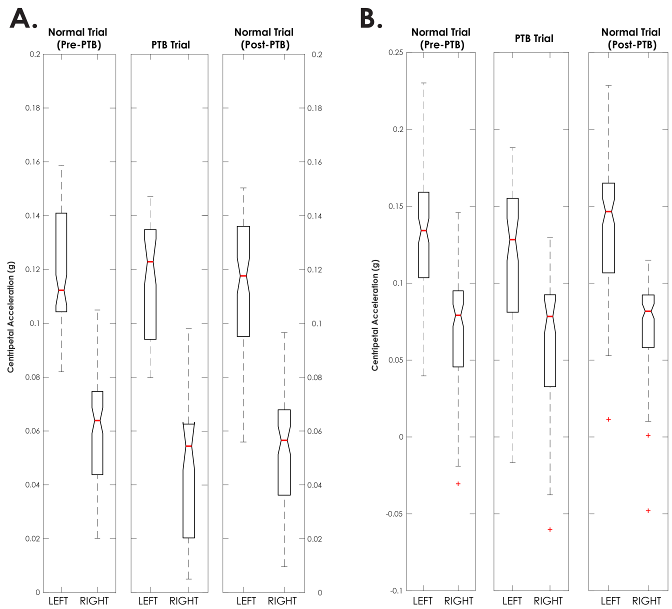

Fig. 3. Difference in centripetal acceleration (g) on the left and right feet between the unperturbed trial prior to the perturbed trial, the unperturbed steps of the perturbation trial, and the unperturbed trial following the perturbed trial at (A) a normal walking speed and (B) an increased walking speed for one

C. Perturbed and Recovery Steps over Time at Normal and Fast Speeds

Perturbation steps on the left foot and first recovery steps on the right foot over the course of the 3-minute perturbed walking trial were analyzed for the purpose of determining whether the subjects adapt to the perturbations at both walking speeds. At a normal walking speed, the first perturbed step has a centripetal acceleration value of 0.09g and the last perturbed step has a centripetal acceleration value of 0.12g (See Figure 4). The second perturbed step of the trial holds the minimum centripetal acceleration value of 0.07g, while the sixth perturbed step of the trial holds the maximum centripetal acceleration value of 0.14g (See Figure 4). At a normal walking speed, the first recovery step has a centripetal acceleration value of 0.05g and the last recovery step has a centripetal acceleration value of 0.06g (See Figure 4). The recovery step with the minimum centripetal acceleration value is the eighth at 0.01g, while the recovery step with the maximum centripetal acceleration value is the seventh at 0.09g (See Figure 4). Both the perturbed steps and recovery steps become more regular in acceleration value after 100 seconds.

At an increased walking speed, the first perturbed step has a centripetal acceleration value of 0.18g, which is the maximum value throughout the course of the trial, and the last perturbed step has a centripetal acceleration value of 0.16g (See Figure 5). The twelfth perturbed step of the trial holds the minimum centripetal acceleration value of -0.0123g (See Figure 5). At an increased walking speed, the first recovery step has a centripetal acceleration value of 0.10g and the last recovery step has a centripetal acceleration value of 0.10g. The recovery step with the minimum centripetal acceleration value is the fourteenth at -0.04g, while the recovery step with the maximum centripetal acceleration value is the third at 0.12g. Both the perturbed and recovery steps remain irregular in acceleration value for the entirety of the 3-minute trial.

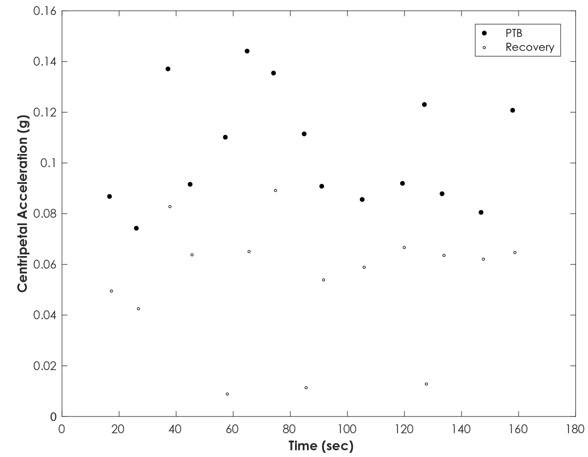

Fig. 4. Centripetal acceleration (g) of the perturbed step on the left foot and first recovery step on the right foot over the course of a 3-minute clockwise turning trial at a normal walking speed for one healthy subject (n=1).

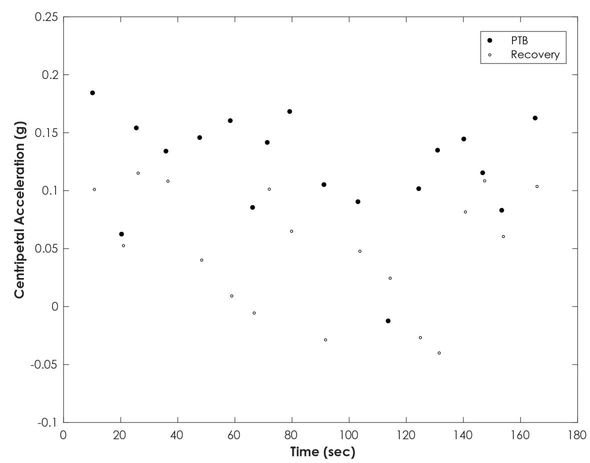

Fig. 5. Centripetal acceleration (g) of the perturbed step on the left foot and first recovery step on the right foot over the course of a 3-minute clockwise turning trial at an increased walking speed for one healthy subject (n=1).

V. Discussion

Current clinical concussion evaluations fail to match the demands of the real world and do not detect the subtle balance deficits that persists for years post- injury [1-3]. The aim of this research was to quantify the balance recovery response in subjects exposed to underfoot perturbations throughout a dynamic turning trial and the effect of increasing walking speed on said response. The subject appeared to be capable of acknowledging and adjusting to the perturbation at a normal walking speed but was not able to adjust when they were asked to walk at a quicker pace. Once we are able of quantifying a larger healthy population, this protocol will be applied to a concussed population and the balance recovery response.

At a normal speed, the subject displayed a decrease in centripetal acceleration to prepare for the disturbance, as well as on the step of the disturbance. For the first three steps after the disturbance, the subject displayed an increase in centripetal acceleration to recover. Whereas, at a faster speed, the subject displayed an increase in acceleration to prepare for the disturbance, but very minimal change in acceleration non the step of the disturbance and to recover from it. Thus, this particular healthy control subject was able to adjust to the perturbation in terms of acceleration at a normal walking speed, but was incapable of implementing the adjustment strategy at a faster pace, which is most likely due to the greater balance challenge the increase in speed provides.

Similarly, the acceleration values over the course of the entire trial show a greater balance challenge at the faster pace. Roughly halfway through the 3-minute trial at a normal pace, the acceleration values of the perturbed (PTB) and first recovery steps become more constant and regular. Over the course of the faster-paced trial, however, the acceleration values of the perturbed and first recovery steps remain sporadic and irregular. This trend displays the ability of the subject to adjust and adapt to the perturbation at a normal pace, once again, and the inability of the subject to do so at a faster pace.

Other biomechanics research displays that, even for healthy individuals, turning challenges one's balance more than straight-line walking [23]. Turning has shown to be a combination of ground reaction time and joint kinetics and kinematics, both of which alter the center of mass trajectory and trunk orientation [23]. Because of the greater balance challenge, it is to be expected that subjects would attempt to adjust their walking strategy to maintain stability while turning. This adjustment was apparent in the subject we tested at a normal walking speed, but not at a faster walking speed. It is possible that the subject was not able to react as quickly to the perturbation at the faster pace and was, thereby, unable to adjust in time.

A noteworthy adjustment that the subject made at the normal walking speed was the decrease in centripetal acceleration to prepare for the perturbation step. One aspect of the conservative gait strategy observed in individuals with concussions during straight-line gait is a decrease in velocity [14]. When turning, one is changing direction so this decrease in velocity is comparable to the decrease in centripetal acceleration. If healthy individuals decrease their acceleration in preparation for the perturbation, it is likely that balance impaired populations with a more conservative gait strategy, such as those with concussions, will exhibit an amplified balance recovery response.

It is important to note that data was only collected and interpreted for one healthy subject. The responses this subject exhibited must be compared to that of other healthy subjects in order to better understand the balance recovery response to perturbations while turning at both speeds. Once the response of healthy individuals is understood, this protocol can then be applied to the concussion population in order to assess whether the balance recovery response differs in individuals that are balance impaired. The response may also be further exacerbated if the individuals were exposed to perturbations on both limbs that caused either an inversion or an eversion, as this study was only capable of inducing eversion perturbations to the left foot. Additionally, measuring the balance recovery response in terms of centripetal acceleration does not allow for a spatial representation of the balance recovery response. Methods that display the orientation of both limbs, as well as the trunk and head, may be beneficial in further quantifying the balance recovery response while turning.

Though this research is still in its early phases, it presents a new outlook on assessing the deficits of balance impaired populations in clinical settings. Currently, clinics assess concussed individuals with static and straight-line gait tasks only. In the real world, much of one's daily life consist of tasks that require greater balance from the individual than what is required in static and straight conditions. Individuals must change direction, navigated uneven surfaces, and react to unexpected events. All three of these aspects are assessed in observing the balance recovery response to underfoot perturbations while turning. If this balance recovery research shows promise, it is possible for similar evaluations to be assessed in a clinical setting to better understand the balance deficits that are left undetected with current clinical protocol.

If we find that individuals improve the postural responses over time, it encourages future research into the potential for perturbation-based balance training. If we find that increased walking speed exacerbates the kinematic responses to postural perturbations, it will warrant testing whether the responses are heightened in individuals with concussions. If we find individuals with a recent concussion exhibit different kinematic responses to postural perturbations, it will support our hypothesis that concussion affects the reactive restoration of balance. If we find that individuals with concussion take longer gain balance after the perturbations, particularly during turning and at increased speeds, it will support our hypothesis that individuals with concussion are unable to attenuate perturbations as quickly as controls during complex activities that require anticipatory control. If we find that postural responses improve slower during turning compared to straight walking, it suggests perturbation-based balance training needs to incorporate more diverse movements and activities. If we fail to find any differences between concussion and control groups, it suggests 1) some balance domains are not affected by concussion and prompts research examining why, or 2) the perturbations were too small to elicit differences and prompts research using larger perturbations.

Balance impairment is one of the major symptoms of concussion. Though, with current clinical concussion protocol, balance is assessed in unrealistic conditions that do not properly translate to the demands of daily life, which may contribute to the risk of further injury. The aim of this research is to collect more realistic balance assessments, to provide a better reflection of the neuromuscular control deficits that persist in individuals with concussion by observing a balance recovery response. If promising, it is possible that evaluations that incorporate balance recovery could be applied in a clinical setting to provide a clearer picture of the deficits that endure in those with concussions, as well as other balance-impaired populations.

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