Longitudinal Locomotor and Postural Control Following Mild Traumatic Brain Injury

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ABSTRACT

Millions of people sustain a mild traumatic brain injury (concussion) each year. While most clinical signs and symptoms resolve within 7-10 days for the majority of typical concussions, some gait and balance tasks have shown abnormalities lasting beyond the resolution of clinical symptoms. These abnormalities can persist after athletes have been medically cleared for competition, yet the implications of such changes are unclear. Most prior research has examined straight gait and standard measures of balance, yet there is a lack of knowledge regarding potential persistent effects on non-straight maneuvers or on indicators of motor control variability or complexity. To expand the knowledge of post-concussion locomotor and postural changes, this investigation examined the recovery of recently concussed athletes longitudinally, over the course of one year, in three domains: 1) path selection and body kinematics during turning gait, 2) non-linear local dynamic stability during straight gait, and 3) postural control complexity during quiet standing. Compared to matched health controls, concussed athletes exhibited significant and persistent differences in turning kinematics, local dynamic stability, and postural complexity over the initial six weeks following injury. These motor differences may increase the risk of injury to concussed athletes who are cleared to return to play. Given the persistent nature of these effects, future clinical tests may benefit from incorporating gait assessments before returning athletes to competition. Future research should prospectively and longitudinally monitor locomotor and postural control in conjunction with structural and functional changes within the brain to better understand the pathophysiology of concussions and potential rehabilitation strategies.

This dissertation is dedicated to my wife, Nora.

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AP:	anterior-posterior
ApEn:	approximate entropy
BESS:	balance error scoring system
BOS:	base of support
C:	concussed athletes
cDTC:	cognitive dual-task cost
COM:	whole-body center of mass
COM_{UB}:	center of mass of the upper body
COP:	center of pressure
DT:	dual-task
DTC:	dual-task cost
H:	healthy matched controls
IMU:	inertial measurement unit
LDS:	local dynamic stability
lmDTC:	locomotor dual-task cost
maxLE:	maximum Lyapunov exponent λ
ML:	medial-lateral
MV-CompMSE:	multi-variate composite multi-scale sample entropy
MV-SampEn:	multi-variate sample entropy
mTBI:	mild traumatic brain injury
NCAA:	National Collegiate Athletic Association
RTP:	return to play
SampEn:	sample entropy
SE:	standard error
SD:	standard deviation
ST:	single-task

LIST OF SYMBOLS

α:	power law exponent between velocity and radius of curvature
κ:	curvature
λ_S:	short term, finite-time Lyapunov exponent
θ:	mediolateral inclination angle
μ:	coefficient of friction
d_{min}:	minimum center-of-mass distance to the corner pylon
v_{com}:	velocity of the center-of-mass while turning
L_{path}:	length of the center-of-mass trajectory around the turn
A:	power law gain factor between velocity and radius of curvature
SLength:	stride length
SWidth:	stride width
STime:	stride time

CHAPTER ONE: INTRODUCTION

1.1 Motivation and Background

Traumatic brain injuries have become a public health concern with the Center for Disease Control estimating that 1.7 million people sustain a mild traumatic brain injury (mTBI), commonly referred as a concussion, every year (33). The prevalence of concussions in sports, particularly football and other contact sports, is even greater. It is estimated that each year 40,000 in 1.1 million high school football players sustain football related concussions (75). However, concussions are not isolated to football. Over a 16 year period encompassing male and female National Collegiate Athletic Association (NCAA) athletes, 5% of all reported injuries were concussions with women's ice hockey (18.3%) having the highest number of concussions relative to other injuries (24). The NCAA Injury Surveillance System has reported concussion rates per 1000 athlete-exposures (where 1 athlete-exposure is defined as one individual participating in one game or practice) of 0.45 for all men's sports and 0.38 for all women's sports in 2005-2006 (24), highlighting the pervasiveness of concussions throughout both men's and women's athletics. The prevalence of concussion in armed service personnel may be even greater; ~15% of combat veterans experience injuries consistent with mild to moderate traumatic brain injury while deployed (50, 81).

Despite this high rate of occurrence, the full extent of the short and long term medical sequellae of concussion are still largely unknown. Concussions are characterized as a "complex pathophysiological process affecting the brain, induced by biomechanical forces" (63) and that need not result in loss of consciousness. But, they are expressed over a wide-spectrum of symptoms (45), often with delayed or unclear onset (29). Short-term neuropsychological (4, 34, 54, 62, 80, 87) deficits are often observed for several days post-concussion, and long-term

cognitive and neurological ailments such as Alzheimer's disease, chronic traumatic encephalopathy, depression, and mild cognitive impairment have been associated with a history of concussions (43, 44). Yet, the effects of concussion on motor control, such as gait and balance, remain unclear. Clinical balance deficits are often observed post-concussion (1, 2, 5, 6, 38, 41, 42, 47, 48, 56, 67, 74, 78, 79) and tend to resolve within 3-5 days (46, 48). However, these clinical assessments, such as the Balance Error Scoring System (BESS) and other commonly used clinical balance tests, suffer from moderate to poor reliability (53) and provide little information about the neuromuscular control system.

Recent studies examining the effects of concussions on gait have provided more evidence of locomotor and balance changes following concussions. Compared to controls, decreased gait velocity (3, 10, 21, 61, 65, 68), a greater percentage of time spent in double support phase (3, 61), where both feet are in contact with the ground, and greater mediolateral (ML) movement of the center-of-mass (COM) (21, 71) have been reported in previously concussed athletes. However, several studies failed to report any group differences during normal gait (15, 64, 70, 72), suggesting that routine tasks such as standing balance and single-task straight gait may not exhibit noticeable effects from concussions.

When subjects performed a more challenging task, such as crossing an obstacle or performing a cognitive task while walking, the differences between concussed and healthy gait were more apparent (15, 21, 61, 68, 70, 72). Notably, these dual-task (DT) gait abnormalities persisted beyond the typical symptomatic recovery time of 7-10 days, with noticeable effects lasting months (10, 31, 52, 57, 72, 76) and years post-injury (61). Thus, while a recently concussed athlete may appear asymptomatic and healthy, it is likely there are lingering deficits during multi-tasking and challenging motor tasks (57, 77).

Given the prevalence of non-straight steps in every-day locomotion (40) and the significantly different kinematic (22, 23, 37, 49, 73) and kinetic (35, 85, 92) characteristics compared to straight gait, turning gait presents an opportunity to examine the effects of concussions on dynamic, transient maneuvers. However, turning gait post-concussion has received little attention thus far, with concussed athletes exhibiting greater segmental reorientation variability (76) and less COM clearance around obstacles during turns (31, 32).

The reported navigational and DT differences are concerning considering the large cognitive loads and high degree of mobility required in most sports (86). These findings suggest that athletes, who typically have enhanced navigational skills (39), may have altered performance that persists after return-to-play (RTP). The DT and navigational changes may represent a shift in the motor control system following a concussion. Decreased interhemispheric brain connectivity has been reported following concussions (55), which may limit the available cortical resources and the ability of recently concussed athletes to process multiple demands (93). These potential motor control changes can lead to concussed athletes trying to navigate the playing field in a compromised state and in turn may contribute to the increased frequency of musculoskeletal injuries following concussions (60, 69).

While kinematic measures of gait offer some insight into the locomotor changes post-concussion, nonlinear dynamics may provide information about changes to the neuromuscular control system. Approximate entropy (ApEn), a nonlinear measure of the complexity of a time-series, has identified differences in the postural control complexity between healthy controls and concussed athletes who were absent of other clinical postural instabilities (17-19). Similar differences in postural control complexity were found between controls and athletes with a previous concussion at least nine months earlier (25, 83), suggesting the neuromuscular control

system may experience detrimental effects that last longer than standard balance metrics. However, methodological variation and the influx of more sophisticated complexity measures (36) prompts a more thorough investigation of post-concussion changes in postural control complexity.

Similar nonlinear dynamic approaches can be used for gait analysis to analyze the neuromuscular response to local perturbations. One such tool, the maximum Lyapunov exponent (λ_S), has been widely applied to the quasi-periodic motion of human gait (7-9, 11, 26-28, 30, 58, 59, 66, 84, 88-90). The maximum Lyapunov exponent λ_S quantifies the average rate of exponential divergence of neighboring state-space trajectories and is a useful tool in assessing the stability of quasi-periodic motion within a state-space. As such, λ_S quantifies the system's response to small, local perturbations in the state-space and reflects an individual's ability to adapt to various local disturbances.

Despite the potential for neuromuscular control information, no study has yet characterized the local dynamic stability of gait post-concussion. However, λ_S has differentiated individuals with and without vestibular disruption (82). Therefore, local dynamic stability may have utility in concussion management given that concussions predominantly disrupt the vestibular system (48) during postural control.

Turning gait, postural control complexity, and local dynamic stability may be useful tools in detecting and assessing the effects of concussions on gait. These tools may provide more information about the neuromuscular control changes post-concussion than traditional clinical analyses. This sensitivity is important in characterizing the recovery of the athlete. If locomotor and balance deficits post-concussion persist beyond the typical RTP guidelines (10, 31, 32, 72),

then turning gait and nonlinear stability analyses may offer valuable information about persistent motor control effects and corresponding recovery times.

1.2 Specific Aims

The objective of this research was to prospectively examine motor control characteristics as they relate to gait and balance of concussed athletes before, during, and after the RTP decision. As the majority of previous literature examined straight gait in concussed athletes (12-16, 51, 52, 61, 65, 68, 70-72), the examination of turning gait was one primary focus. Additionally, while some nonlinear complexity analyses have been applied to concussions (17-19, 25, 83), there has yet to be a longitudinal, prospective analysis of the complexity of concussed balance or gait. Within this context of motor control, locomotion, and balance in concussed athletes, three specific aims were accomplished:

1. Determine whether kinematic differences exist in turning gait between healthy and recently concussed athletes, and whether such differences persist beyond the RTP decision.

Non-straight activities such as turning are inherent in nearly every sport and are prevalent in every-day activities. While changes in straight-gait have been well characterized post-concussion (12, 14, 20, 51, 52, 61, 68, 70-72), less is known about turning gait. Given the large COM excursions (37, 91), high frictional demand (35, 92), asymmetrical loading (85), and segmental reorientation (73) present in turning maneuvers, the increased complexity of turning gait relative to straight gait may reveal further task-complexity based changes post-concussion. Understanding potential biomechanical differences post-concussion during dynamic change of direction tasks is also important to understanding the competitive readiness of athletes after they RTP.

Hypothesis 1a: Recently concussed athletes will show center-of-mass kinematic differences during turning gait compared to matched controls throughout the RTP decision.

Hypothesis 1b: The addition of a cognitive dual-task will further increase the complexity of the task and exacerbate the kinematic differences between concussed and control athletes.

2. Using nonlinear dynamic analysis, determine if differences in gait stability are present between recently concussed athletes and health matched controls.

Traditional biomechanical gait measures can reveal gross changes in gait post-concussion (52, 61, 72), but they do not communicate any information about the motor control system or whether such biomechanical changes hinder or enhance the ability to respond to perturbations. Using nonlinear dynamics, it is possible to assess the neuromuscular control system's response to local perturbations during gait (26). Examining the local dynamic stability of gait may reveal neuromuscular control changes following concussion that are important considerations in the management and RTP care of recently concussed athletes. The reported gait differences in concussed athletes, combined with the associations between local dynamic stability, walking balance (7, 8, 26, 27, 58, 82) and gait speed (8, 30) suggest λ_S may detect neuromuscular control changes post-concussion.

Hypothesis 2a: Recently concussed athletes will exhibit less stable gait, defined through nonlinear analysis, compared to healthy, matched controls.

Hypothesis 2b: The addition of a cognitive dual-task will increase the local dynamics gait stability differences between concussed and control athletes.

3. Determine if postural control complexity of recently concussed athletes differs from healthy matched controls during the sub-acute and immediate post-RTP timeframe.

Previously concussed athletes have shown decreased postural control complexity compared to healthy controls during the acute (17, 18) and persistent (> 9 months) timeframes (25, 83). Yet, no study has investigated the postural control complexity changes during the sub-acute and RTP timeframe. Additionally, different methodologies for computing the postural control complexity may reveal conflicting results, and a more thorough investigation into post-concussion postural control will aid researchers in understanding complexity-based balance control changes post-concussion.

Hypothesis 3a: Postural control complexity will be reduced in previously concussed athletes compared to matched controls.

Hypothesis 3b: More sophisticated complexity algorithms will better distinguish the postural control differences between previously concussed and healthy control athletes.

1.3 Organization

This dissertation is organized into five chapters. This first chapter provides background information, a review of the pertinent literature, and the specific aims and hypotheses of this dissertation. It gives detail about the epidemiology and the potential dangers, both long and short term, of concussions. It focuses on the locomotor and balance deficits following concussions and introduces the areas this dissertation will cover.

Chapters 2-4 are structured in a journal manuscript format. These chapters address the specific aims of this dissertation by longitudinally evaluating the balance and gait characteristics of concussed athletes and matched controls over the course of one year, concentrated on the initial six weeks post-concussion. Chapter 2 examines the kinematic differences between concussed and healthy control athletes during turning. Chapter 3 compares the local dynamic gait stability of recently concussed athletes and matched controls to explore neuromuscular control

differences. Chapter 4 addresses postural stability complexity changes that occur post-concussion and explores the applicability of several entropy-based algorithms to concussion management. Chapter 4 also includes a brief supplemental study in non-athletes to interpret the postural stability complexity differences seen between the concussed and control groups.

The fifth and final chapter provides an overall conclusion to the dissertation. This chapter addresses the most salient results and implications from each study. Limitations from the dissertation on a whole are addressed. The chapter concludes with recommendations for future research within the combined domains of gait, balance, and concussions and overall conclusions.

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CHAPTER TWO: LOCOMOTOR DEFICITS IN RECENTLY CONCUSSED ATHLETES AND MATCHED CONTROLS DURING SINGLE AND DUAL-TASK TURNING GAIT

2.0 Abstract

There is growing evidence that mild traumatic brain injury (concussion) can affect locomotor characteristics for prolonged periods of time even when physical signs and symptoms are absent. While most locomotor deficits post-concussion have involved straight walking, turning gait has received little attention despite its pervasiveness is everyday locomotion and athletic competition. This study longitudinally examined kinematic characteristics during preplanned turning in a small sample of recently concussed athletes (n = 4) and healthy matched control athletes (n = 4) to examine potential deficits during single and dual-task turning gait over one year post-injury. Concussed athletes generally had slower turning speeds and longer stride times compared to healthy controls though no difference in path selection was apparent. When controlled for speed and turn curvature, the recently concussed athletes exhibited less inclination towards the inside of the turn. These differences in inclination angles may represent potential kinetic differences during other sport-related tasks which could be a factor in the higher rate of musculoskeletal injuries post-concussion. The kinematic differences between groups were estimated to recover to healthy levels between 100 and 300 days post-injury, suggesting future prospective longitudinal studies span at least six to twelve months post-injury.

2.1 Introduction

Sport-related mild traumatic brain injuries, commonly called concussions, affect between 1.6 and 3.8 million people in the United States annually (28). Within collegiate competitive athletics, concussions account for approximately 5% of all injuries (13), making concussion

treatment and rehabilitation an important public health and athletic training concern. While clinical signs and symptoms typically resolve within 10 days (33) there is growing evidence for an “atypical evolution” of symptoms (43) with neurological (27, 44) and locomotor (1, 2, 6, 9, 15, 25, 32, 35, 38, 40) changes that persist beyond two months.

The use of a dual-task (DT) paradigm has been particularly successful at identifying residual deficits post-concussion (42). During DT gait, simultaneous cognitive and motor tasks compete for limited cortical resources (52) and create gait modifications (dual-task costs) compared to single-task (ST) gait. While such dual-task costs (DTC) are present in healthy young adults (20), larger DTCs have been reported in asymptomatic, recently concussed athletes (4-6, 9, 14, 15, 24, 25, 29, 32, 38). The larger DTCs in recently concussed athletes suggest that a concussion may affect the available cortical resources or shift the prioritization from motor to cognitive tasks (5, 52). In athletics where gameplay, field, and environmental conditions (e.g., crowd noise) create substantial cognitive and physical loads (48), the complexity-dependent motor differences in medically cleared, recently concussed athletes could influence performance and / or injury risk. However, few studies have examined DTCs with complex non-straight movements to simulate the more dynamic demands of competition.

Some studies have reported greater cognitive DTC (cDTC) during obstacle circumvention (15) and hockey-specific skating and puck-handling drills (41), but provided little information about the locomotor DTCs (lmDTC), such as changes in gait speed or mediolateral (ML) sway. Similarly, while several studies have used a step-over obstacle as the secondary task to study lmDTC (6-8), few studies have combined a complex motor task with a complex cognitive task to elicit larger lmDTCs (9). No study has yet examined kinematic changes, with or without a cognitive task, in recently concussed athletes during preplanned turning.

Turning gait is a common (18) complex locomotor task (10-12, 46) applicable to change of direction and reorientation during dynamic athletic movements (15, 40). Yet, despite the high risk of slips and falls (16, 50, 51), large COM excursions (17), asymmetrical loading (47), and segmental reorientation variability (40) present during turning, the biomechanical lmDTCs during turning gait in recently concussed athletes remain largely unknown. Due to the high prevalence of turning gait, the segmental reorientation variability in athletes during unplanned turns (40), and the potential correlations to injury risk and performance, both single-task (ST) and DT turning gait in recently concussed athletes present important research areas in concussion management. We hypothesized that ST turning gait would identify kinematic differences between asymptomatic, recently concussed athletes and controls given the increased complexity of turning gait, compared to straight gait, (10-12) and the complexity-dependent effects of concussions on motor performance (9). The addition of a DT was expected to elicit greater lmDTC during turning in recently concussed athletes compared to healthy controls and further separate concussed and healthy athletes.

2.2 Methods

2.2.1 Participants

Eight Virginia Tech varsity athletes (four concussed, four matched controls) participated in this study. All concussed athletes were clinically diagnosed with a concussion by a Virginia Tech Sports Medicine physician and referred to this study by their athletic trainer. No concussed athlete had a prior concussion. Controls were recruited from teammates of the concussed subject and were individually matched based on sport, position, skill level, and stature. Exclusion criteria for both groups included any unresolved acute lower extremity injury, history of mental illness or diagnosed cognitive impairment. An additional exclusion criterion for controls was no

concussion or brain injury within the previous year. No control had any prior history of concussion. Table 1 provides demographic data of all participants. All participants gave informed written consent and all recruitment procedures and experimental protocols were approved by the Virginia Tech Institutional Review Board.

Table 1. Descriptive information of all eight participants in matched pairs. Full return-to-play (RTP) was identified as the first day the athlete was medically cleared for full athletic participation in practices and / or games.

ID	Gender	Age (years)	Height (cm)	Mass (kg)	Days Before Full RTP	Weeks Tested[a]
C1	Female	18	170	63.3	7	(2, 3, 4, 5, 6, 52)
H1		19	178	64.6	-	(2, 3, 4, 5, 6, 52)
C2	Male	20	178	78.9	14	(2, 3, 4, 5, 6)
H2		20	178	72.4	-	(3, 4, 5, 6)
C3	Female	19	163	58.7	11	(3, 4, 5, 6, 52)
H3		18	163	59.0	-	(3, 4, 5, 6, 52)
C4	Female	19	163	59.4	16	(3, 4, 5, 6, 52)
H4		21	170	66.6	-	(2, 3, 4, 6)

[a] One year follow-up session represented as 52 weeks post-concussion

2.2.2 Procedures

Recently concussed participants were tested weekly for six weeks, at mean (SD) 7 (0), 16 (2), 23 (2), 29 (1), 37 (2), and 45 (3) days post-concussion. Each control was also tested up to six times and, when possible, on the same days as their match to control for musculoskeletal afflictions from recent games or practices. A one year follow-up testing session occurred at an average (SD) of 363 (42) days post-concussion for three concussed and two control participants. Not all participants completed all sessions because of scheduling conflicts. For the purpose of this analysis in examining locomotor changes that remain unresolved after RTP, testing sessions which occurred before the concussed athletes were cleared for full athletic competition were excluded.

All testing sessions occurred in a basketball gymnasium with minimal distractions and clean hardwood floors. The participants were asked to walk barefoot around an 18 m by 3.5 m course marked with 1.5 m tall pylons (2.5 cm diameter PVC pipe) and consisting of several turns of ~90° (Figure 1). Participants completed seven laps in each direction around the course. For DT gait, participants were asked to serially subtract by sevens a given random number between 900 and 999 while walking around the course. The participants were not given explicit instruction to prioritize the cognitive or motor task, being instructed only to serially subtract "as fast as you can". For both ST and DT gait, the participants were instructed not to count the number of laps.

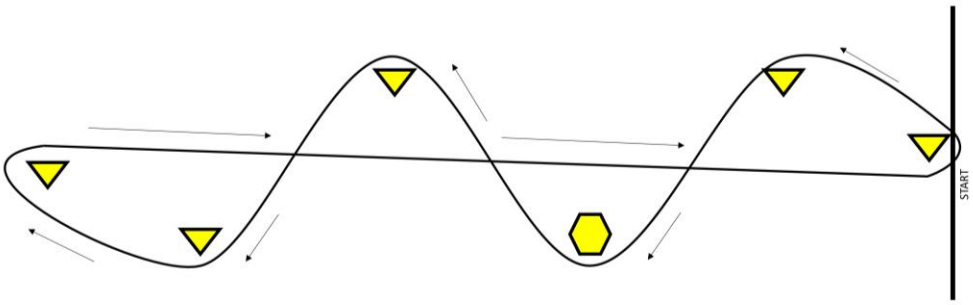

Figure 1. Depiction of the 18 m long course. Triangles indicate pylons. The hexagon indicates the pylon recorded with motion capture cameras.

Four motion capture cameras (ProReflex MCU 170 120, Qualisys Medical AB, Gothenburg, Sweden) were stationed around one ~90° turn on the course (Figure 1) to capture kinematics of the turn. Reflective markers were placed bilaterally on the calcaneus with additional markers placed over the xiphoid process and approximate T9 vertebra. One marker was placed on the pylon to mark its position in space. Data were collected at 120 Hz and filtered using a phaseless fourth order Butterworth filter with a 6 Hz cutoff frequency. A total of 14 turns, seven of both right and left turns, were captured for each condition corresponding to the 14 laps. Short durations with missing markers were spline interpolated. Trials with more than 60

consecutive missing frames (> 0.5 seconds) were discarded; 30 out of 1120 total trials were discarded.

2.2.3 Analysis

The upper-body center of mass (COM_{UB}) was estimated using the three-dimensional mean of the xiphoid process and T9 vertebral markers. The minimum horizontal distance from the COM_{UB} to the pylon (d_{min}) was used to represent the COM clearance. The length of the horizontal trajectory, L_{path}, was computed starting when the athlete proceeded below the pylon in the *y* direction, and ending when the athlete advanced above the pylon in the *y* direction (Figure 2). Quadratic fits to the horizontal trajectory of the COM_{UB} were applied to a moving window of three consecutive points. Instantaneous COM_{UB} curvature in the horizontal plane was calculated by taking the second derivative of each quadratic polynomial fits.

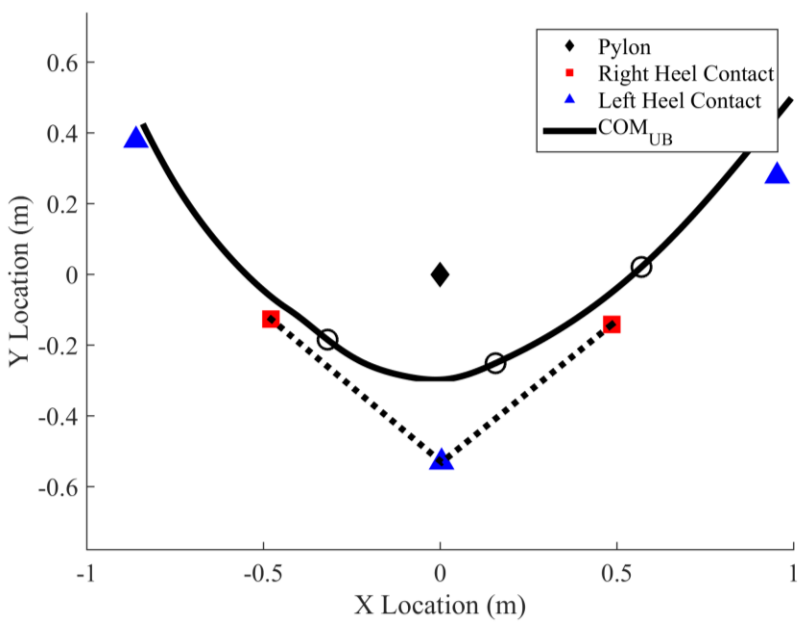

Figure 2. Example overhead view of the turning stride (dotted line) connecting the right-left-right heel contacts around the apex during a right turn. The movement tracks right to left. Black circles show the location of the COM_{UB} at each heel contact. The center of mass remains outside of the base of support throughout the entire duration of the turn.

Heel contacts corresponding to each step were identified using the local minima of the heel marker height. For each turn, a single stride was identified which encompassed the point of maximum curvature and therefore the greatest change in direction. This turning stride was comprised of three heel contacts and was selected such that the middle heel contact (apex limb) occurred closest to the point of maximum COM_{UB} curvature in the *x* direction (Figure 2). Stride length (SLength) and stride width (SWidth) were calculated based on the line of progression (26) shown in Figure 3, and stride time (STime) was calculated as the time from heel contact to heel contact on the same foot within the turning stride. Stride width was allowed to be negative if the apex limb crossed over the line of progression (Figure 3).

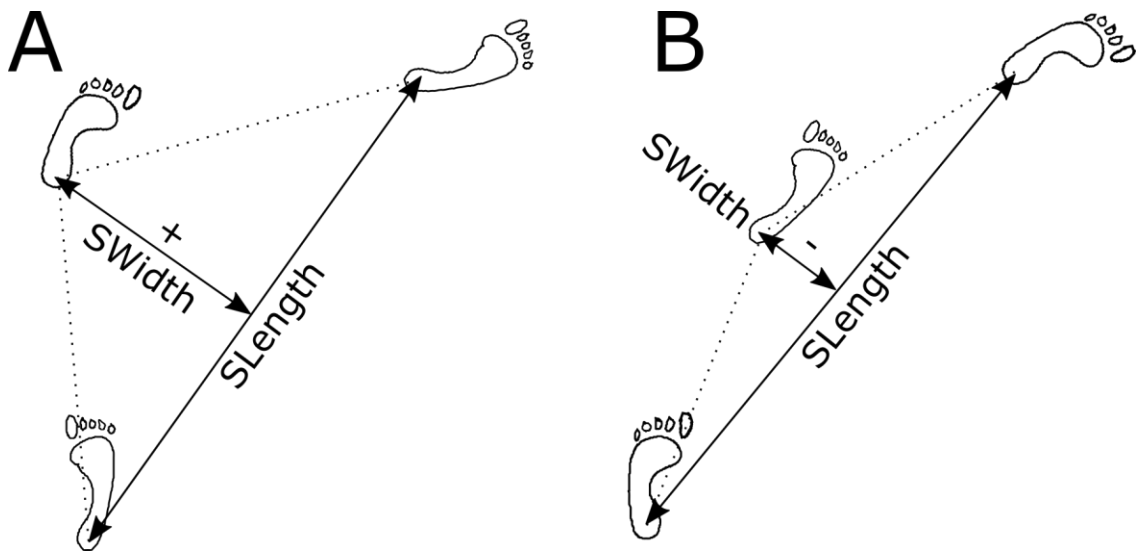

Figure 3. Illustration of step (A) and spin (B) turns to the right. The stride length, SLength, is the linear distance following the path of progression between successive heel contacts of the same limb. The stride width, SWidth, is the distance, normal to the path of progression, between the apex heel contact and the path of progression.

The strategy of the turn was characterized using the apex limb. A “step turn” was identified if the apex limb was contralateral to the turn direction (e.g., a right turn with the left leg in stance at the apex), whereas a “spin turn” was identified if the apex limb was ipsilateral to

the turn direction (e.g., a right turn with the right leg in stance at the apex). The mean velocity of the COM_{UB} (v_{com}) was calculated over the turning stride. The mean curvature, κ, of the COM_{UB} over the turning stride was calculated to compare the relationship between velocity and curvature.

2.2.3.1 Curvature – velocity power law relationship

A power law relationship between the radius of curvature, $R = 1/\kappa$, and velocity has previously been defined for turning gait in both outlined (23) and free paths (37). This power law relationship can be expressed linearly in logarithmic form

$$\log v_{com}(t) = \log A + \alpha \log R(t) \quad (1)$$

where R is the radius of curvature, α is a constant coefficient, and A is the piecewise velocity gain factor that can vary with shape (49). To compare whether the recently concussed athletes exhibited a different relationship between velocity and curvature, the relationships between mean velocities, v_{com}, and corresponding radii of curvature, $1/\kappa$, were compared for each group using the α and A coefficients determined from the linear statistical models.

2.2.3.2 Mediolateral inclination angles and centripetal force

The ML inclination angles θ_1, θ_2, and θ_3, corresponding to the three heel contacts in each turning stride, were estimated using the frontal plane angle between vertical and the vector connecting the heel to COM_{UB} at each heel contact. Because turning is a transient motion, the frontal plane was identified using the vertical plane normal to the instantaneous horizontal velocity of the COM_{UB} (19). Medial inclinations were defined positive as shown in Figure 4.

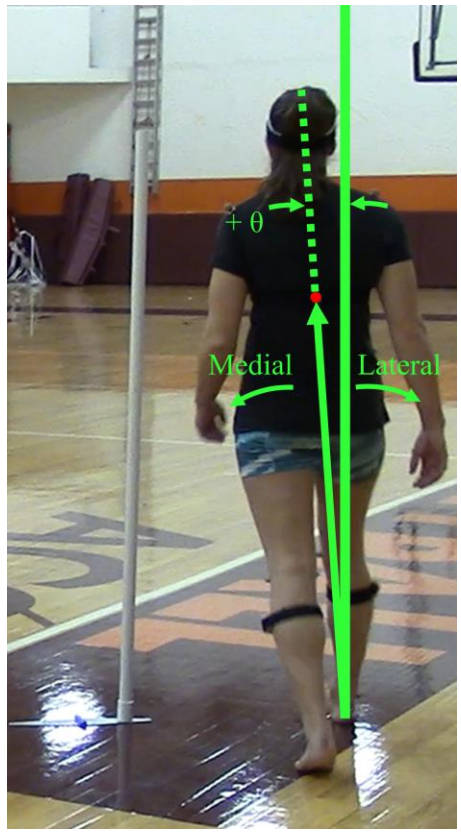

Figure 4. Depiction of the ML inclination angles, with medial inclinations presented as positive and lateral inclinations presented as negative values, regardless of stance limb.

The relationship between inclination angle, velocity v_{com}, and curvature κ during turning was accounted for using the relation between angle and coefficient of friction, μ (17). Using a simple model of a steady state turn of a banked object (mass m) at an angle of θ (Figure 5), balancing centripetal and gravitational moments, M_c and M_g, about the center of mass

$$M_c = M_g \tag{2}$$

$$F_f\, L \cos(\theta) = F_N\, L \sin(\theta) \tag{3}$$

$$\mu\, m\, g \cos(\theta) = m\, g \sin(\theta) \tag{4}$$

$$\mu = \tan(\theta) \tag{5}$$

and substituting for μ based on the centripetal force relationship

$$F_f = m\, v^2\, \kappa = \mu\, m\, g \tag{6}$$

$$\mu = \frac{v^2 \kappa}{g} \tag{7}$$

the inclination angles follow the proportional relationship

$$\theta \propto \tan^{-1}(v^2 \kappa) \tag{8}$$

when *g* is the gravitational constant. Though variability is introduced from the discontinuities of steps during human locomotion and the transient nature of the turns examined here, this proportional relationship was considered when comparing inclination angles.

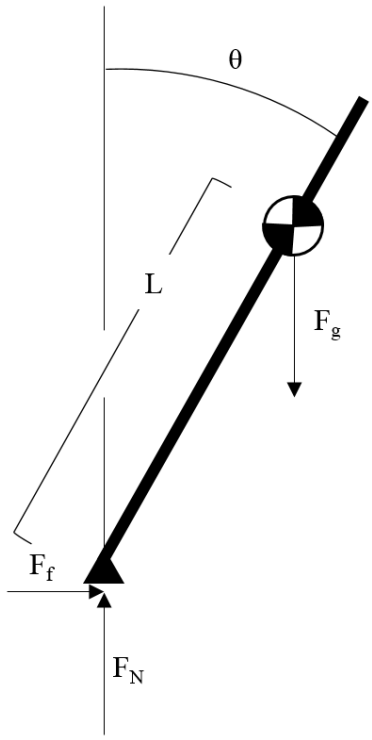

Figure 5. Simple banked rigid body with inclination angle *θ*, mass *m*, and length to center-of-mass L.

A total of 10 kinematic measures were analyzed (Table 2). Because many kinematic outcomes depend on the stance limb and gait speed (17), specific lmDTCs for each outcome were not calculated. Instead, lmDTCs were assessed in the statistical models, with group differences in lmDTC identified by significant group*task interaction effects.

Table 2. Description of all measured kinematic outcomes.

Outcome	Description
v_{com}	Mean velocity of the COM_{UB} throughout the turn
κ	Mean curvature of horizontal COM trajectory across turning stride
L_{path}	Total length of COM_{UB} path around the pylon
d_{min}	Minimum distance from the COM_{UB} to the corner pylon
SWidth	Distance from apex limb to the line of progression during the turning stride
SLength	Linear distance from first the third heel contact during the turning stride
STime	Time from first to third heel contact during the turning stride
θ_1	ML inclination angle at the first heel contact of the turning stride
θ_2	ML inclination angle at the second heel contact of the turning stride
θ_3	ML inclination angle at the third heel contact of the turning stride

2.2.4 Statistical Analysis

Pearson correlation coefficients between all pairs of outcomes were computed. Correlation coefficient magnitudes were compared, rather than inference values, given the violation of independence from repeated measurements. To determine the relationship between groups, linear mixed models with random intercepts and slopes for time, which account for the within-subject correlations among each participant's trials over time, were fit for each outcome. Given previous findings of decreased gait speed during DT gait and in concussed athletes (9), an initial mixed model was evaluated to compare the influence of group, day, task, as well as all interactions on v_{com}. Subsequent models for κ, L_{path}, and d_{min}, included v_{com} as a covariate to account for its influence on the kinematic variables during turning (17). Outcomes of SWidth,

SLength, and STime were stratified by turning strategy to account for the kinematic and kinetic differences between step and spin turns (47). Two-way interactions of group*day, group*task, and day*task, and three-way group*day*task interactions were included in the initial models. Interaction terms with p values less than 0.10 were retained in the final models. Kinematic lmDTC differences between groups were reflected with the group*task or group*day*task interaction.

The power law relationships between log v_{com} and log $1/\kappa$ given in Eq. 1 were compared between groups using a mixed model including group and a group*$1/\kappa$ interaction term. The exponent α and coefficient A were determined using the mixed model estimates for each group. The concussed one-year follow-up data were excluded from this comparison to prevent potential distortion of the immediate post-concussion relationship.

The inclination angles θ_1, θ_2, and θ_3 were stratified by turning strategy and compared using mixed models with covariates of group, day, task, and $\tan^{-1}(v_{com}^2\kappa)$ following from Eq. 8. Two way interactions of group*day, group*task, day*task, and a three-way interaction of group*day*task were included in initial models. Interactions with p values less than 0.10 were retained in the final models.

If significant group*day interaction effects were present in any final models, a linear recovery timeline was constructed using the predicted values from the mixed model. The time from injury to full recovery was then estimated using the time when the model estimates of the concussed and control groups crossed. For all models, assumptions were validated using residuals. All statistical analysis was performed in SAS 9.4 (SAS Institute Inc., Cary, NC, USA) using a two-tailed 0.05 significance level.

2.3 Results

Concussed athletes utilized spin turns at similar frequencies (45%) to controls (46%) during both ST and DT turning. A correlation matrix between all outcome variables is presented in Table 3. Strong positive associations (> 0.50) were found between L_{path} and d_{min}, v_{com} and SLength, SWidth and θ_2, and between θ_1 and θ_3. Curvature had strong negative correlations (< -0.50) with v_{com}, L_{path}, and SLength. Inclination angles θ_1 and θ_3 had strong negative correlations to θ_2.

Table 3. Correlation matrix of for each outcome. Correlations greater than 0.50 in magnitude are shown in bold. No inference was made because within-subject observations were not independent.

	κ	L_{path}	d_{min}	SWidth	SLength	STime	θ_1	θ_2	θ_3
v_{com}	**-0.56**	0.34	0.05	-0.06	**0.63**	**-0.55**	0.02	-0.05	-0.05
κ		**-0.65**	-0.29	0.05	**-0.59**	0.11	-0.06	0.03	0.14
L_{path}			**0.77**	-0.03	0.39	0.02	-0.04	-0.02	-0.03
d_{min}				0.02	0.07	0.07	-0.08	0.02	-0.09
SWidth					-0.46	0.04	**-0.88**	**0.98**	**-0.72**
SLength						-0.02	0.45	-0.42	0.24
STime							0.06	0.05	0.00
θ_1								**-0.91**	**-0.78**
θ_2									**0.67**

A significant group*day interaction for κ was found as the concussed group increased their curvature over time (Table 4), relative to controls when accounting for velocity. Controls maintained a constant curvature over time, when accounting for velocity. Path length, L_{path}, and COM clearance, d_{min}, did not differ between groups but did show significant effects from velocity. A three-way group*task*day interaction was found for v_{com} as the concussed group slowed more than the control group during the DT, and that lmDTC increased over time (greater difference between ST and DT over time) in the concussed group. Longitudinal results for v_{com}, κ, L_{path}, d_{min} are depicted in Figure 6 - Figure 9, respectively. Using the mixed model results, the

estimated time until concussed athletes recovered to healthy curvature levels, assuming a linear recovery, was 398 days.

Table 4. Model parameters and inference values from the final linear mixed models for v_{com}, κ, L_{path}, and d_{min}. Beta coefficients and standard errors (SE) are shown multiplied by 10^3. Higher order interactions terms with *p* values below 0.10 were retained in the final model. Significant terms are shown in bold.

	v_{com}		κ		L_{path}		d_{min}	
	β (SE) $*10^3$	*p* value	β (SE) $*10^3$	*p* value	β (SE) $*10^3$	*p* value	β (SE) $*10^3$	*p* value
Group	-70.0 (79.5)	0.3791	-139.2 (90.5)	0.1241	105.2 (130.6)	0.4209	11.1 (37.9)	0.7697
Day	0.80 (0.62)	0.2427	0.05 (0.13)	0.7357	**-0.28 (0.07)**	**0.0038**	0.04 (0.04)	0.3944
Task Condition	**-28.3 (6.86)**	**< 0.0001**	**-22.9 (10.5)**	**0.0297**	11.4 (9.6)	0.2340	-11.1 (3.47)	0.1833
v_{com}	-	-	**-1292 (73.3)**	**< 0.0001**	**876.5 (66.1)**	**< 0.0001**	**-32.4 (24.3)**	**0.0014**
Group*Day	0.33 (0.86)	0.7033	**0.35 (0.17)**	**0.0360**	-	-	-	-
Group*Task	-17.8 (9.51)	0.0614	-	-	-	-	-	-
Day*Task	-0.07 (0.05)	0.1980	-	-	-	-	-	-
Group*Day*Task	**-0.21 (0.07)**	**0.0028**	-	-	-	-	-	-

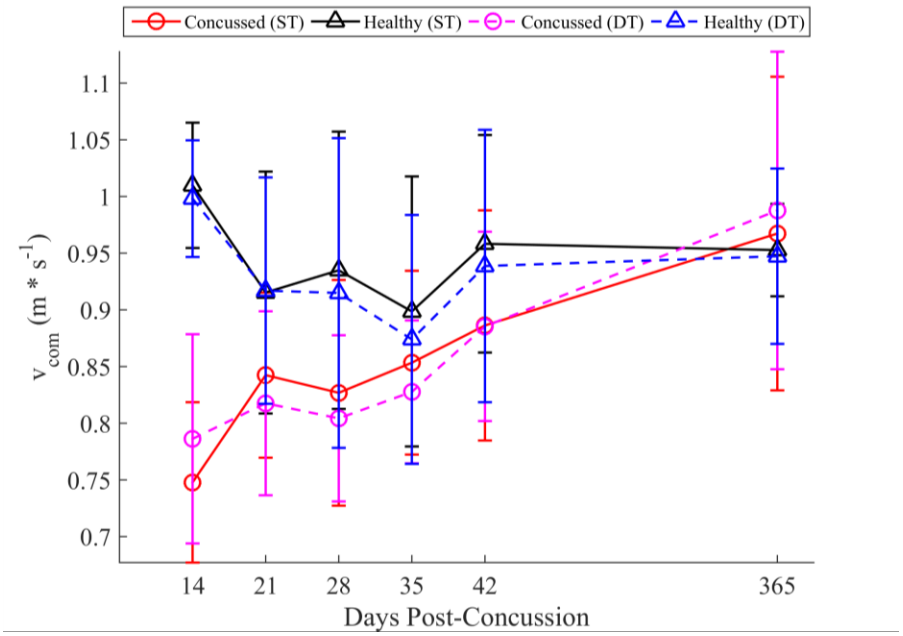

Figure 6. Longitudinal means (SD) of v_{com} for each week. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

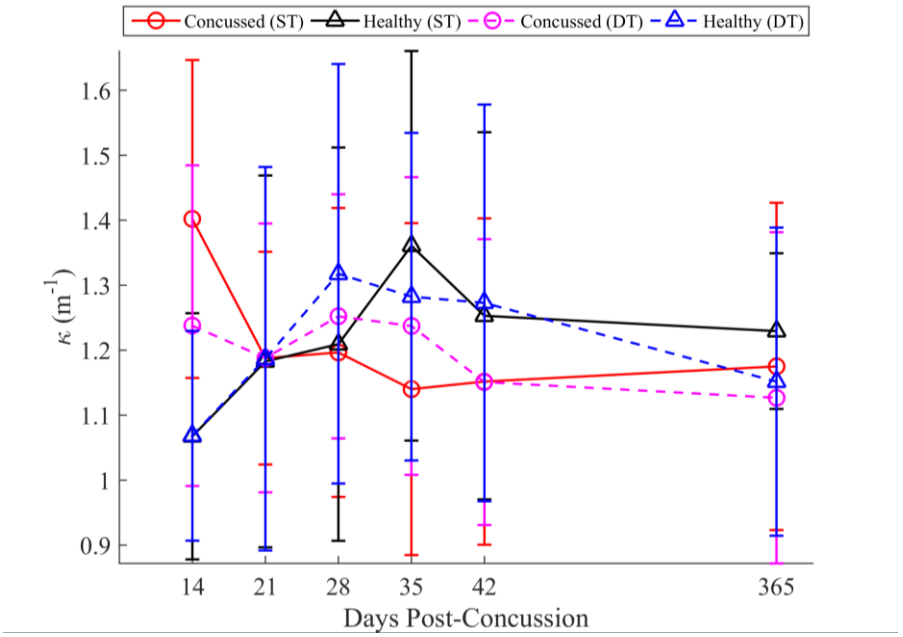

Figure 7. Longitudinal means (SD) of κ for each week. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

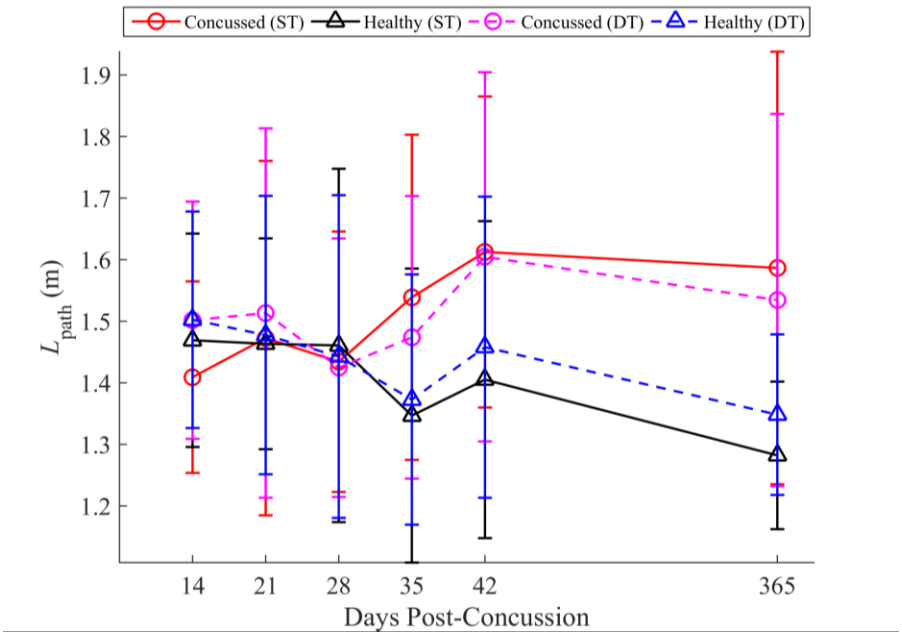

Figure 8. Longitudinal means (SD) of L_{path} for each week. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

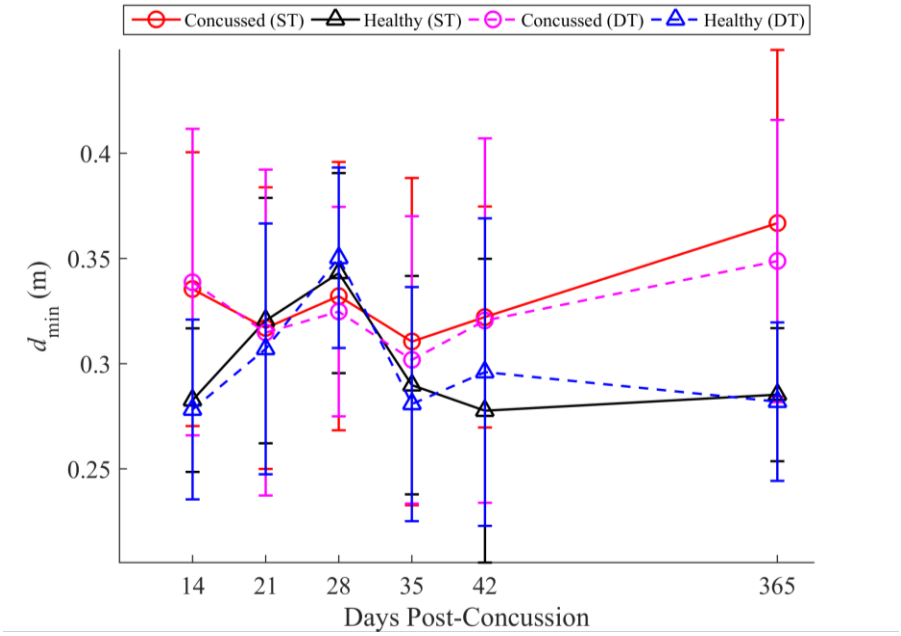

Figure 9. Longitudinal means (SD) of d_{min} for each week. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

Significant lmDTCs, indicated by significant group*task interactions, were found for STime during both step and spin turns and for SWidth during step turns (Table 5). Significant group*day interactions indicated the concussed group recovered back to healthy levels as time progressed in SWidth for step turns, STime for step turns, and STime for spin turns. The estimated time until concussed athletes recovered to healthy turning stride characteristics, assuming a linear recovery, was 391 and 144 days for SWidth and STime, respectively, during step turns, and 179 days for STime during spin turns. STime increased during the DT condition regardless of strategy. Though SLength increased over time (step turns) and decreased during DT turning (spin turns), no differences in SLength were found between groups. Longitudinal results for SWidth, SLength, and STime, stratified by strategy, are provided in Figure 10 - Figure 12.

Table 5. Model parameters and inference values from the final linear mixed models for SWidth, SLength, and STime. Beta coefficients and standard errors (SE) are shown multiplied by 10^3. Higher order interactions terms with *p* values below 0.10 were retained in the final model. Each outcome was stratified by turning strategy. Significant terms are shown in bold.

	SWidth		SLength		STime	
Step (*n* = 593)	β (SE) *10^3	*p* value	β (SE) *10^3	*p* value	β (SE) *10^3	*p* value
Group	-42.9 (33.5)	0.2009	-31.1 (71.6)	0.6640	51.5 (57.8)	0.3734
Day	-0.05 (0.02)	0.0883	**0.27 (0.10)**	**0.0347**	0.11 (0.07)	0.1932
Task Condition	-7.7 (4.3)	0.0763	-6.7 (6.3)	0.2961	**33.1 (9.1)**	**0.0003**
Group*Day	**0.11 (0.03)**	**0.0002**	-	-	**-0.36 (0.10)**	**0.0002**
Group*Task	**12.5 (5.9)**	**0.0344**	-	-	**37.5 (12.3)**	**0.0025**
Spin (*n* =497)						
Group	15.6 (26.6)	0.5564	-29.5 (77.0)	0.7017	67.8 (73.2)	0.3554
Day	**-0.06 (0.02)**	**0.0116**	0.47 (0.23)	0.0845	0.07 (0.07)	0.3408
Task Condition	-1.5 (3.9)	0.7007	**-18.0 (5.6)**	**0.0014**	**32.2 (11.8)**	**0.0067**
Group*Day	-	-	-	-	**-0.38 (0.09)**	**< 0.0001**
Group*Task	-	-	-	-	**38.5 (16.4)**	**0.0225**

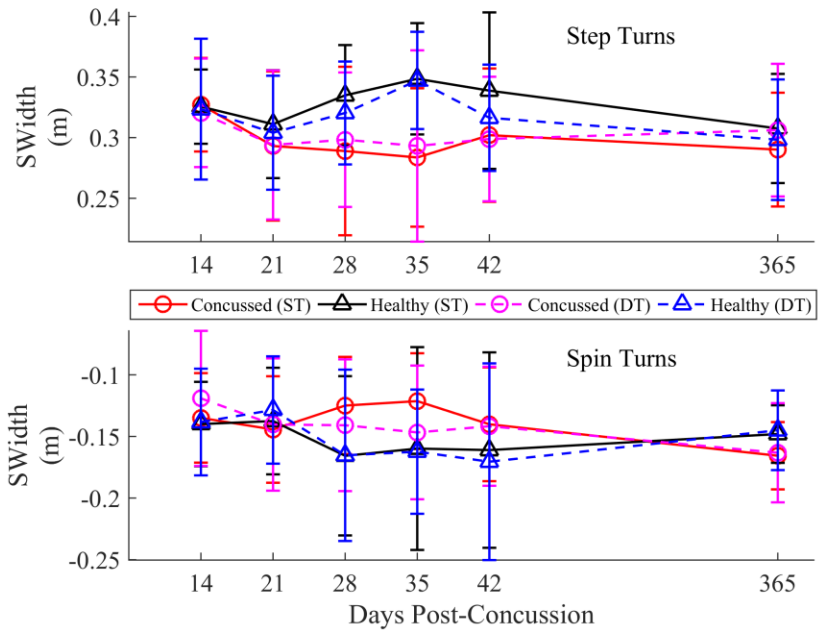

Figure 10. Longitudinal means (SD) of stride width, SWidth for each week stratified by step (top) and spin (bottom) turns. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

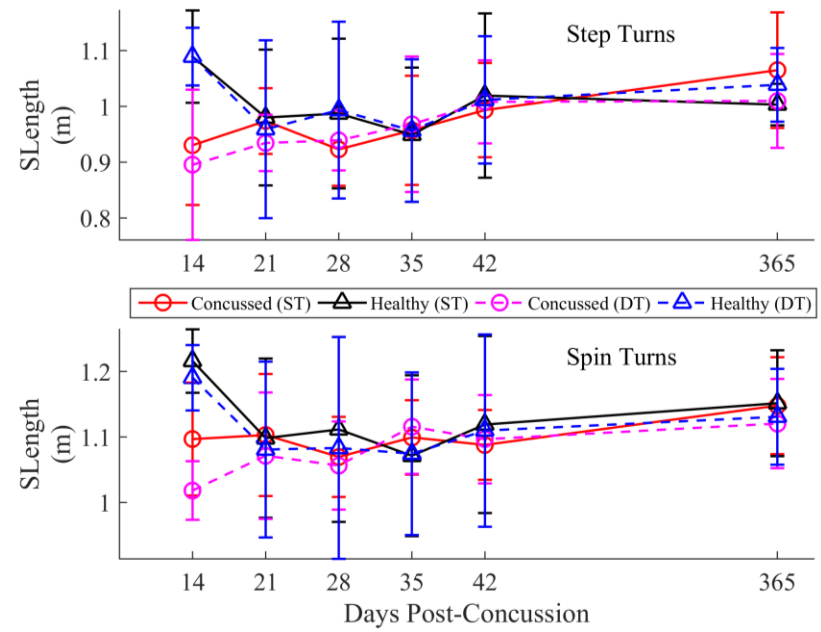

Figure 11. Longitudinal means (SD) of stride length, SLength for each week stratified by step (top) and spin (bottom) turns. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

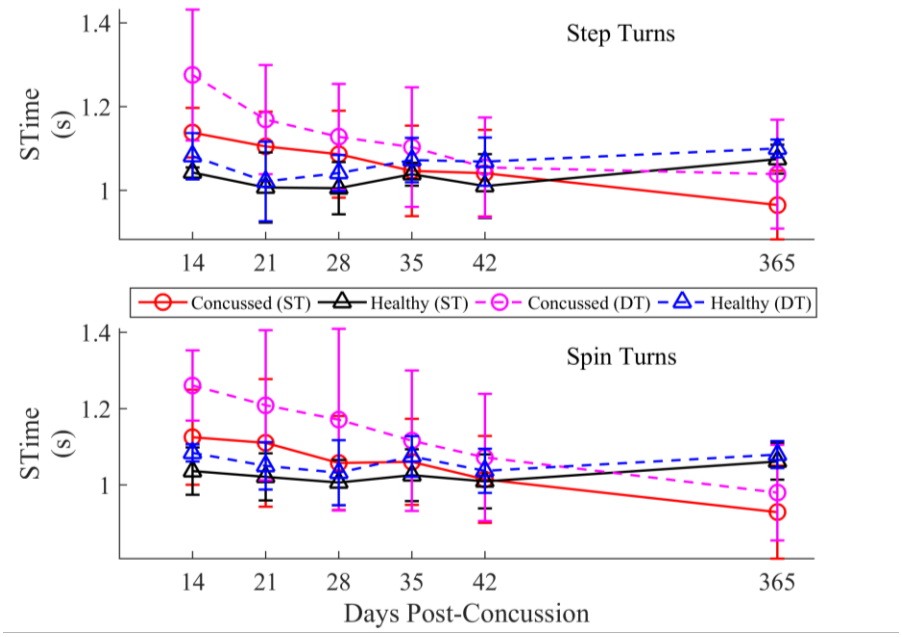

Figure 12. Longitudinal means (SD) of stride time, STime for each week stratified by step (top) and spin (bottom) turns. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

The linear fit between log v_{com} and log $1/\kappa$ yielded estimated parameters A (SE) of -0.24 (0.06) in concussed athletes within 6 weeks of their concussion, and -0.08 (0.04) in controls. A group difference was found (p = 0.009), indicating that A significantly decreased in concussed athletes during the first 6 weeks. No significant interaction was found between groups indicating no significant differences in α between concussed, α = 0.24 (0.03), or controls, α = 0.21 (0.02). The power law relationship explained more variability in controls than concussed athletes during the initial six weeks (Figure 13). However, after one year, the power law relationship explained the concussed group's variability similarly to the overall control group's variability.

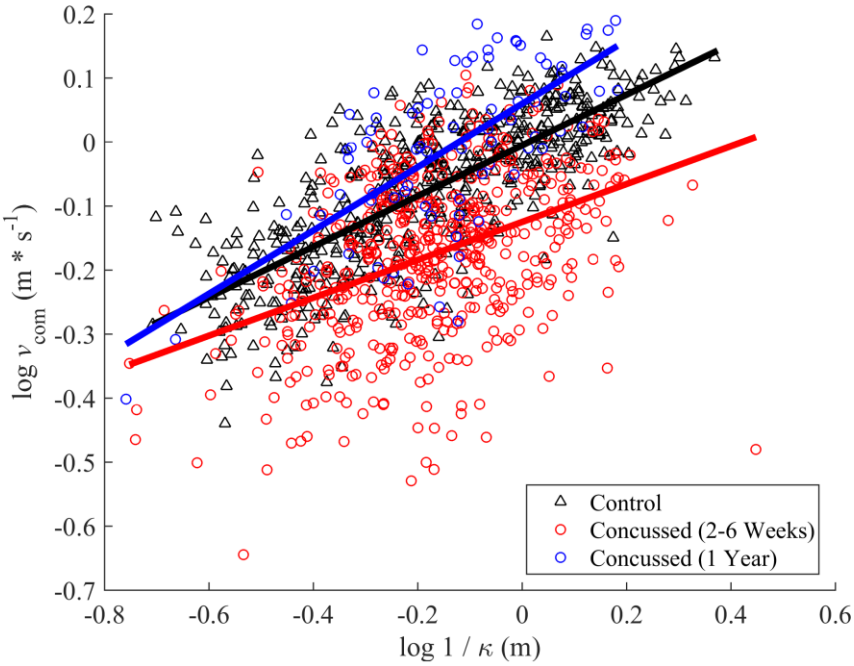

Figure 13. Linear fits of the power law relationship between velocity and radius of curvature for controls (black, $R^2 = 0.59$), concussed athletes within 6 weeks on injury (red, $R^2 = 0.18$) and concussed athletes after one year (blue, $R^2 = 0.45$). No difference in slopes were found between the fits, but a significant decrease in the velocity gain function *A* in concussed athletes 2-6 weeks post-concussion was found.

Significant group*day interactions were found for θ_1 during spin turns and θ_2 for both step and spin turns (Table 6) indicating the concussed group increased the magnitude of inclination as time got further from the injury. For ipsilateral stance limbs during the second half of the turning stride (θ_3 step turns), a group*day*task interaction was found indicating an *increase* in the lmDTC difference between concussed and control groups as time increased. The estimated time until concussed athletes recovered to healthy levels assuming a linear recovery was 115 and 123 days for θ_2 and θ_3, respectively, during step turns, 239 days for θ_1 during spin turns, and 222 days for θ_2 during spin turns. As the DTC differences increased for θ_3 over time,

no recovery date could be estimated. Longitudinal results for each angle are provided in Figure 14 - Figure 16.

Table 6. Model parameters and inference values from the final linear mixed models for the three ML inclination angles, θ_1, θ_2, and θ_3. Beta coefficients and standard errors (SE) are shown multiplied by 10^3. Higher order interactions terms with *p* values less than 0.10 were retained in the final model. Outcomes were stratified by turning strategy. Significant terms are shown in bold.

	θ_1		θ_2		θ_3	
Step (*n* = 593)	β (SE) *10^3	*p* value	β (SE) *10^3	*p* value	β (SE) *10^3	*p* value
Group	151.0 (653.6)	0.8174	-677.7 (623.6)	0.2777	806.0 (1056)	0.4457
Day	-0.09 (0.74)	0.9039	-2.2 (1.8)	0.2749	-3.2 (2.5)	0.2357
$\tan^{-1}(v_{com}^2\kappa)$	**-7241 (899.0)**	**< 0.0001**	**4605 (565.3)**	**< 0.0001**	**6754 (1441)**	**< 0.0001**
Task Condition	-94.6 (149.4)	0.5267	**-291.2 (90.0)**	**0.0013**	**1198 (394.9)**	**0.0025**
Group*Day	-	-	**5.9 (2.4)**	**0.0148**	**-6.6 (3.2)**	**0.0418**
Group*Task	-	-	-	-	203.0 (542.1)	0.7082
Day*Task	-	-	-	-	-3.1 (3.0)	0.2949
Group*Day*Task	-	-	-	-	**9.8 (4.0)**	**0.0156**
Spin (*n* = 497)						
Group	**-1338 (456.2)**	**0.0035**	1107 (1241)	0.3729	443.5 (474.8)	0.3507
Day	-2.7 (1.7)	0.1724	-0.53 (1.33)	0.7036	0.24 (0.81)	0.7750
$\tan^{-1}(v_{com}^2\kappa)$	**3738 (1009)**	**0.0002**	**-5344 (769.4)**	**< 0.0001**	**7490 (1109)**	**< 0.0001**
Task Condition	-77.4 (157.9)	0.6245	356.2 (194.6)	0.0678	**-412.1 (177.0)**	**0.023**
Group*Day	**5.6 (2.3)**	**0.0157**	**-5.0 (1.8)**	**0.0051**	-	-
Group*Task	-	-	-222.5 (272.1)	0.4139	-	-
Day*Task	-	-	-1.4 (1.5)	0.3703	-	-
Group*Day*Task	-	-	4.0 (2.1)	0.0537	-	-

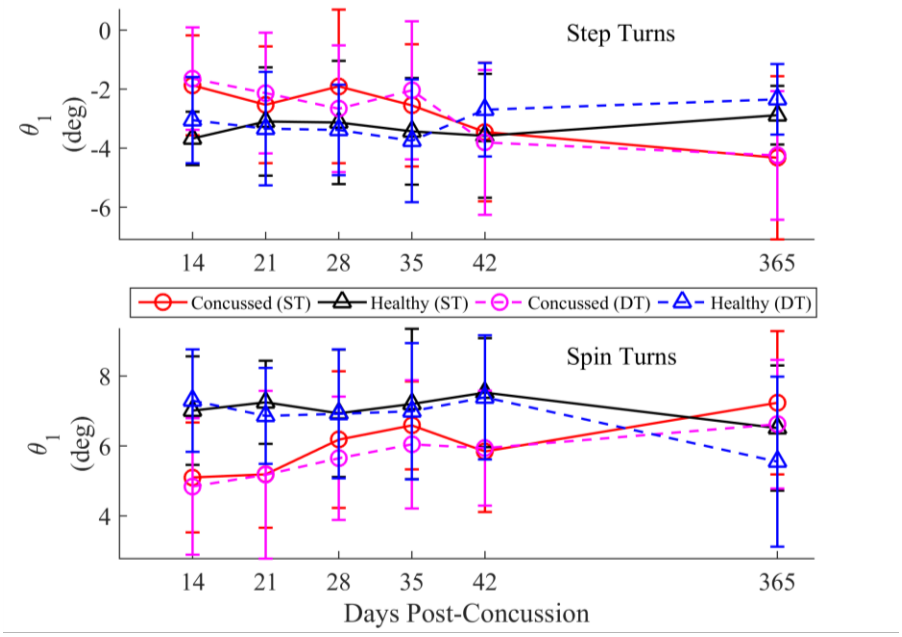

Figure 14. Longitudinal means (SD) of ML inclination angle θ_1 for each week stratified by step (top) and spin (bottom) turns. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

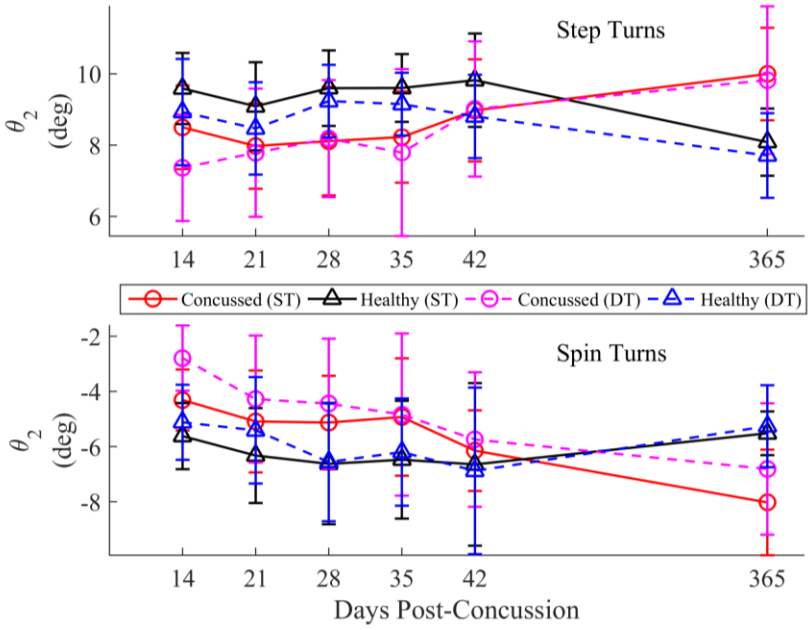

Figure 15. Longitudinal means (SD) of ML inclination angle θ_2 for each week stratified by step (top) and spin (bottom) turns. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

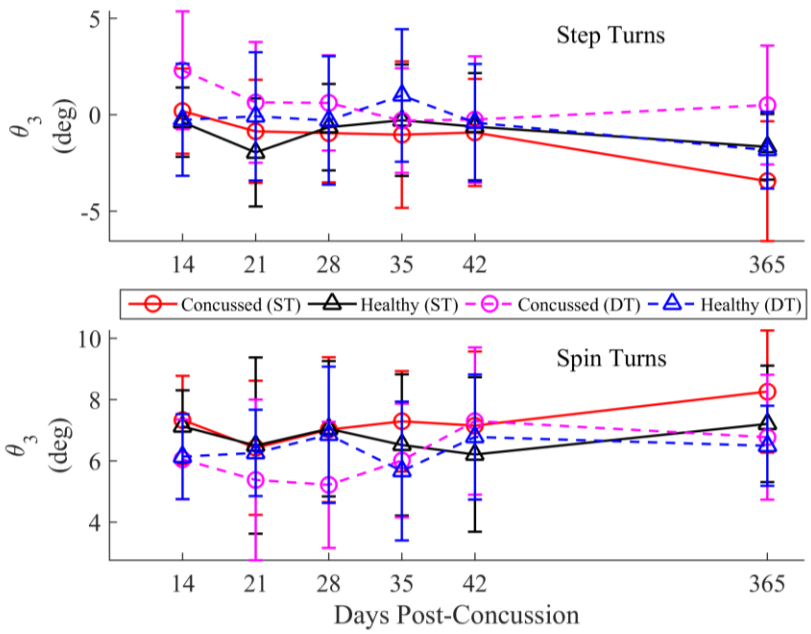

Figure 16. Longitudinal means (SD) of ML inclination angle θ_3 for each week stratified by step (top) and spin (bottom) turns. Concussed data are shown in red (ST) and magenta (DT) circles. Healthy controls are shown in black (ST) and blue (DT) triangles.

2.4 Discussion

This study prospectively examined single- and dual-task turning kinematics in recently concussed and healthy control athletes. It provides the first preliminary evidence of kinematic differences during ST and DT preplanned turning gait in a small sample of recently concussed athletes. Despite all athletes being medically cleared for competition, concussed athletes showed less curvature for a given turning speed and greater lmDTCs in SWidth, and STime compared to healthy controls. Additionally, several outcomes, including ML inclination angles θ_1, θ_2, and θ_3, showed different trends over time between concussed and control groups regardless of task, which suggests altered turning gait persists well beyond other concussion symptoms. Overall, these group differences describe the effects of concussion on turning gait in three domains: path trajectory (κ), stride characteristics (SWidth, SLength, STime), and body orientation (θ_1, θ_2, θ_3).

2.4.1 Path Trajectory

One key finding of this preliminary study is the decreased velocity gain factor A in concussed athletes relative to healthy controls. While both the concussed and control groups exhibited similar power law relationships between v_{com} and $1/\kappa$, the concussed group walked with slower velocities across the range of curvatures. Similar modulations of A have been reported as individuals transition to different outlined trajectories (23, 49), but only when the path is predefined. Though individuals follow stereotyped behavior without a predefined path (39), Olivier and Cretual (37) argued that the lack of a consistent power relationship over the course of one-turn indicated that a simple turn is not performed by first planning the trajectory and then by following the planned path. Instead, the motor control system is simplified by coupling velocity and curvature and minimizing the deviations in an orientation / velocity phase space from one trajectory to the next (37). The similar α values found here between the concussed and control groups suggest that this long term coupling remains present post-concussion. However, the lower A post-concussion indicates that the relationship has shifted to a slower speed, consistent with previously described decreases in gait speed post-concussion (9). Given the similar values of α between groups, it is possible that some of the navigational deficits reported in concussed individuals (15) are predominantly changes in gait speed and not differences in an intrinsic velocity – curvature relationship that may affect path characteristics. As such, gait speed during turning may be an important clinical indicator of recovery.

Our results do not indicate whether the path selection was quantitatively different between groups, but trajectories between groups and conditions were qualitatively similar (Figure 17) and there were no significant differences in path length, L_{path}, or minimum clearance, d_{min}. The present results, as evidenced by the low R^2 values from the velocity-curvature

relationship in concussed compared to healthy athletes, do agree with Baker and Cinelli (1) that concussed athletes are more variable in their path selection. It is likely that, on average, concussed athletes can exhibit healthy stereotyped path selection (22, 39) but do so with greater variability, similar to their increase joint-coordination (7) and step (35) variability.

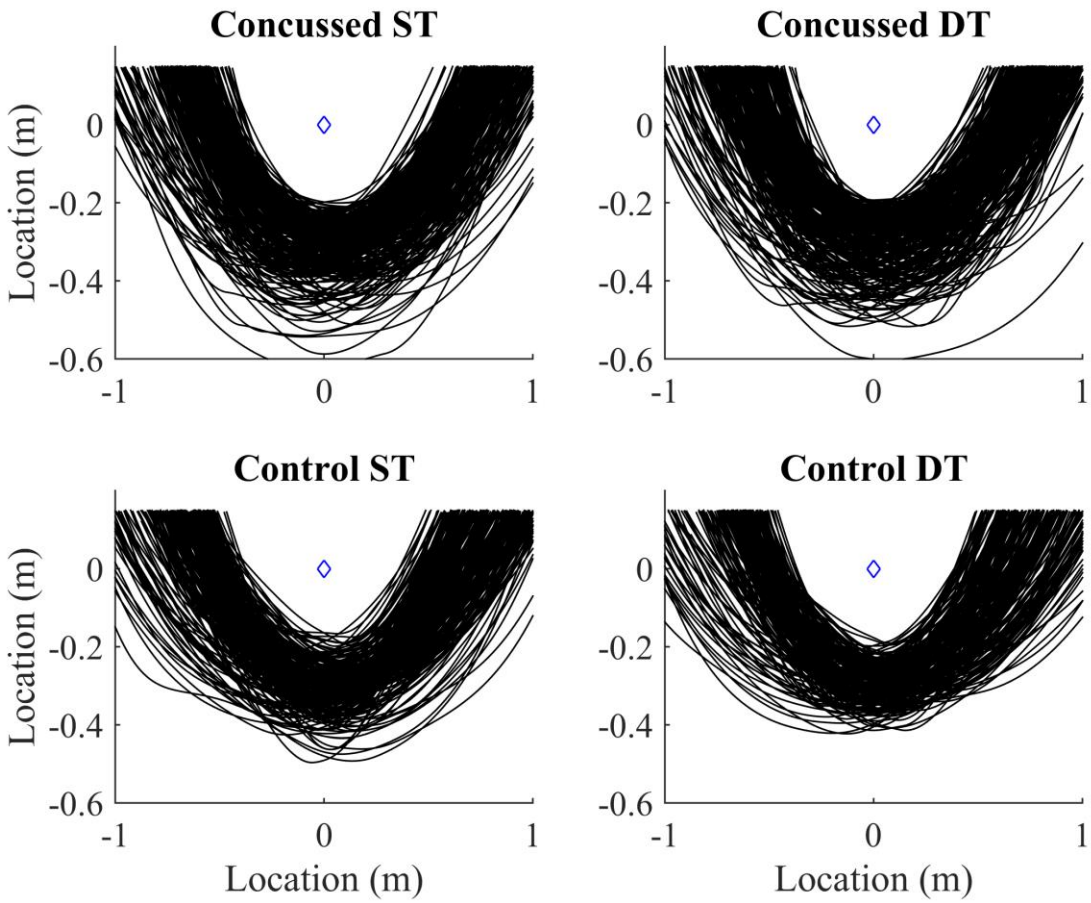

Figure 17. All trajectories for single- (left) and dual-task (right) turning. The concussed group's trajectories are shown in the top two figures; the controls on bottom. The pylon is represented with the blue diamond at the origin. All distances are in meters.

We did not find a significant difference between groups in minimum clearance, d_{min}, though overall conclusions are limited due to the small sample size. However, this appears to contradict Fait et al. (15), who descriptively reported greater minimum clearances in athletes post-concussion compared to controls. Upon further investigation, however, the apparent

differences in clearance reported by Fait et al. (15) may have been due to slightly slower gait speeds in the concussed group. Results presented here and by Fino et al. (17) showed a strong inverse relationship between the walking velocity and the minimum obstacle clearance. Additionally, Fait et al. (15) utilized an in-line obstacle circumvention paradigm, where the goal remained fixed and an obstacle was placed in the direct path to that goal. Conversely, this study used an obstacle navigation task, where participants were instructed to pass on a specific side of each obstacle in route to the next. It is possible these different tasks elicited distinct behaviors (45) considering the required change in heading angle was much larger in the current study.

2.4.2 Stride Characteristics

The concussed athletes exhibited consistent lmDTC increases in SWidth and STime, widening and slowing their strides during the DT more than the control group. This greater slowing of strides during DT concussed gait has previously been described in straight walking (3, 38), but no prior study has described this result during turning gait. The larger DTCs in recently concussed athletes are often attributed to a diminished multi-tasking capacity post-concussion, suggesting that a concussion may affect the available cortical resources or shift the prioritization from motor to cognitive tasks (5, 52).

Our results did not find differences in stride length between groups. Previous studies reported decreased or no change in stride length in recently concussed athletes during DT straight walking (3, 24, 34). Based on the longer STimes in the concussed group and similar SLengths between groups, the differences in gait speed between groups were controlled by increasing the STime and not by shortening the SLength during turning.

2.4.3 Body Orientation

Recently concussed athletes exhibited less inclination towards the turn for up to six weeks post-concussion. The inclination angle should be directly proportional to $\tan^{-1}(v_{com}^2\kappa)$ based on a simplistic model of a continuous rolling rigid body. Given that the concussed athletes generally walked slower across similar curvatures, lower magnitudes of θ in concussed athletes were expected. However, this group difference was present, though it resolved over time, even when adjusting the model for $\tan^{-1}(v_{com}^2\kappa)$. For a given velocity and curvature, concussed athletes exhibited less inclination towards the inside of the turn (less medial inclination during step turns, less lateral inclination during spin turns). One may interpret this decreased inclination as another cautious adaptation of the concussed group in order to preserve stability (3, 5, 32) and maintain the COM closer to the base-of-support (BOS). Yet, concussed individuals exhibit greater ML sway during straight walking and obstacle crossing (3, 4, 8, 24, 25). At this time, is unclear why this paradoxical change occurs between curved and straight walking.

While only limited kinematic data were collected in this preliminary study, the geometric relationship between inclination angle and centripetal force can be used as an initial step to examine potential kinetic differences between concussed and control athletes during turning. Using the simple banked rigid body model (Figure 5), a change in θ introduces a new coronal plane moment if velocity, curvature and mass are held constant based on Equation 3. For a given curvature and velocity, a decrease in θ induces a moment counteracting the centripetal force's moment on the COM; an increase in θ induces a moment acting with the centripetal force moment. Interestingly, similar placement of the COM closer towards the stance limb during cutting, accomplished through lateral trunk flexion, has been associated with ACL injuries (21), illustrating the potential risk of decreased inclination during sport-related maneuvers. While this

preliminary investigation was not designed to investigate how or where (ankle, knee, hip) these increased moments may manifest, it is likely that recently concussed athletes experience increased overall coronal plane moments due to their decreased inclination angles with respect to velocity and curvature during turning (31). These initial results encourage future research examining potential kinetic differences during faced-paced sport-related turning and cutting following concussion, which may be a contributing factors to the increased rate of acute musculoskeletal injuries, especially non-contact injuries, post-concussion (30).

2.4.4 Timeline of Effects and Recovery

Notably, all data presented here were collected after the concussed athletes had returned to full athletic participation. Thus, these results support the growing body of literature reporting locomotor deficits lasting well-beyond the clinical return to play date (1, 2, 4-6, 9, 14, 15, 24, 25, 29, 32, 38, 40, 41). Despite the absence of clinical signs or symptoms, gait deficits post-concussion persist and it is unclear if or when any such deficits resolve. Several locomotor abnormalities have been found to persist at least 30 days post-concussion (1, 15, 32, 35, 41), with some deficits lasting beyond 60 days (25). Our results of several kinematic abnormalities persisting beyond six weeks post-concussion agrees with this report. Yet, to the best of our knowledge, no previously published study has prospectively examined the recovery beyond 2 months to investigate a full recovery timeline. Approximations based on the present statistical models' group*day interaction terms suggest that complete recovery may not occur until after 200-300 days post-concussion. These linear approximations should be interpreted cautiously given the unlikelihood of a linear recovery, the reliance on the mean estimate only, the small sample size, and the general lack of data immediately surrounding the estimated recovery time. However, the results do give guidance about the duration of future prospective longitudinal

studies examining locomotor capabilities post-concussion, suggesting the need to follow participants for 6-12 months post-concussion. Interestingly, these estimates do fit within the same timeframe of reported increases in musculoskeletal and subsequent injuries post-concussion (30, 36), reinforcing the need for comprehensive 6-12 month longitudinal studies.

2.4.5 Limitations

The small sample size of this preliminary study is a primary limitation and prompts caution when generalizing these results to an entire population. Nevertheless, significant differences were found between concussed and healthy controls. Additionally, despite the small sample, this presents the first investigation into turning kinematics post-concussion and provides the only longitudinal data spanning up to one year. Variations in a few data points, such as the controls' one-year follow-up data, likely had particularly strong influence on the recovery timeline, potentially over or underestimated the recovery in some instances (θ_2 Step turns, Figure 15, may have been underestimated given the decrease in inclinations of the two controls subjects at one year).

Another limitation was the potential variation of cognitive task performance. The serial seven subtraction task was chosen based on its complexity and the ability to discriminate concussed and healthy individuals (24). While participants were directed to subtract as fast as possible, it was difficult to discern whether the priority was given to the cognitive or motor task as the response rate and number of errors were difficult to characterize. However, because this investigation was concerned with the kinematic deficits during locomotion that manifest in asymptomatic cleared athletes, potential differences in prioritization do not change the present conclusions about diminished dual-task capacity. Similar prioritization would likely occur during real-life situations outside the testing environment.

Finally, the use of a small set of reflective markers may have introduced small error into the COM_{UB} and inclination angle θ calculations compared to a full body marker set. Other markers were placed on the participants but proved unreliable due to the testing environment.

2.5 Conclusion

Kinematic differences were found during turning gait between concussed and healthy athletes that persisted despite the absence of clinical signs or symptoms. These results suggest that turning mechanics post-concussion, reflected in the mediolateral inclination angles during turning gait, may reflect potential kinetic differences during sport-related maneuvors (e.g., cutting) that could contribute to the higher musculoskeletal injury rate post-concussion (30, 36). Turning gait and change of direction tasks are recommended for inclusion in future locomotor studies of concussion-related kinematic differences and injury potential (40), and that such studies extend for six months to one year post-concussion to capture the full recovery timeline. Overall, these results reaffirm that concussions can present persistent gait modifications, and that when and how such gait disturbances subside should be an important area for future research.

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CHAPTER THREE: LONGITUDINAL DIFFERENCES IN LOCAL DYNAMIC STABILITY BETWEEN RECENTLY CONCUSSED AND HEALTHY ATHLETES DURING SINGLE AND DUAL-TASK GAIT

3.0 Abstract

Concussed individuals commonly exhibit locomotor deficits during dual-task gait that can last substantially longer than clinical signs and symptoms. Previous studies have examined traditional stability measures, but nonlinear dynamic stability may offer further information about the health of the motor control system post-concussion. For up to one year post-concussion, this study longitudinally examined the local dynamic stability of five concussed athletes and four matched healthy controls during single- and dual-task gait. Local dynamic stability was estimated using short-term, finite-time maximum Lyapunov exponents calculated from tri-axial accelerometers placed on the trunk and head. Significant group*task interactions were apparent for trunk stability and stride time variability. This interaction reinforces previous reports that concussions persistently affect dual-task processes even when single-tasks may be unaffected. The decreased local dynamic stability during dual-task gait indicates the concussed group attenuated local disturbances less than their healthy teammates. This difference in dual-task stability persisted beyond the athletes' return-to-play schedules and for up to six weeks post-concussion. The decreased dynamic stability during dual-task activities which was present after the athletes were cleared for competition may be a contributing factor in the higher rates of musculoskeletal injuries in athletes post-concussion. The results also strengthen the idea that decreased local dynamic stability during dual-task gait is indicative of a deviation from a healthy locomotor system.

3.1 Introduction

Recovery following concussion is commonly evaluated using self-reported symptoms (20), neuropsychological testing (24, 30, 36, 55), and balance assessments (25). These signs and symptoms for a standard athletic concussion typically resolve within 7-10 days post-concussion (38). However, recent studies have reported locomotor deficits that lasted beyond this typical recovery timeframe and persisted after the athlete had been cleared for competition (6, 11, 22, 23, 28, 33, 37, 43, 44).

Locomotor tasks with divided attention, in particular, have shown deficits lasting for up to two months in otherwise asymptomatic athletes (28). While walking with a simultaneous cognitive or attention-dividing task, recently concussed asymptomatic athletes have exhibited increased mediolateral (ML) sway (11, 28, 43) and increased step and joint-coordination variability (13, 39), despite single-task (ST) gait similar to a control group. These changes have been described as a conservative strategy to reduce anterior-posterior (AP) momentum and maintain balance (8, 10, 11), but these dual-task costs (DTCs) have not yet been translated to on-field performance or injury risk. It is possible that persistent gait modifications, such as decreased gait speed, reflect an altered motor control system (17) that is less suited for high-demand situations such as competitive athletics, and may be a contributing factor in the high rate of musculoskeletal injuries following concussions (35, 40).

One way to examine locomotor control is to assess an individual's response to small, local perturbations (2, 50), such as neuromuscular noise or surface irregularities. Traditional stride-to-stride variability measures (e.g., stride time variability, ML sway variability) only reflect the magnitude of such perturbations, but convey no information about the system's response to the disturbances (7, 56). Local dynamic stability (LDS) quantifies the neuromuscular

response and attenuation of these perturbations (17, 34) and therefore reflects an individual's ability to respond to various local disturbances and perturbations (50, 51). Decreased LDS has been associated with aging, recent fall-history, and mobility-impaired populations (7, 34, 54), suggesting that LDS may be an indicator of a system's response to global perturbations (e.g., slip, trip) as well.

An athlete's ability to quickly respond and adapt to local (e.g., irregular turf / surface conditions) and global (change of direction) perturbations is essential for performance and safety. Therefore, LDS may be a useful indicator of an athlete's competitive readiness and underlying locomotor control after a concussion. This study longitudinally examined the LDS of recently concussed and control athletes to determine the effect of concussion on the LDS of collegiate athletes. Based on the appearance of conservative gait strategies reported elsewhere (9-11, 23, 28, 37, 41-43), we hypothesized that recently concussed athletes would exhibit persistent decreases in LDS compared to matched controls, and that such deficits would be exacerbated during cognitive dual-task (DT) gait.

3.2 Methods

3.2.1 Participants

Nine Virginia Tech varsity athletes (five concussed, four matched control) participated in this study (Table 7). All concussed athletes were clinically diagnosed by a Virginia Tech Sports Medicine physician and referred to this study by their athletic trainer. Controls were recruited from among teammates of the concussed subject and individually matched based on sport position, skill level, and stature. Exclusion criteria for both groups included any unresolved acute lower extremity injury or any history of mental illness or diagnosed cognitive impairment. An additional exclusion criterion for controls was no concussion or brain injury within the previous

year. All participants gave informed written consent and all recruitment procedures and experimental protocols were approved by the Virginia Tech Institutional Review Board.

Table 7. Age, height, and mass of concussed (C) and healthy matched control (H) athletes and the weeks each athlete were tested. The number of days before the concussed athletes were cleared for full return-to-play (RTP) is also provided. The one year follow-up session is indicated as 52 week post-concussion.

ID	Gender	Age (years)	Height (cm)	Mass (kg)	Days Before Full RTP	Weeks Tested Post-Concussion
C1	Female	18	170	63.3	7	(2, 3, 4, 5, 6, 52)
H1		19	178	64.6	-	(2, 3, 4, 5, 6, 52)
C2	Male	20	178	78.9	14	(1, 2, 3, 4, 5, 6)
H2		20	178	72.4	-	(3, 4, 5, 6)
C3	Female	19	163	58.7	11	(3, 4, 5, 6, 52)
H3		18	163	59.0	-	(3, 4, 5, 6, 52)
C4	Female	19	163	59.4	16	(1, 3, 4, 5, 6, 52)
H4		21	170	66.6	-	(1, 2, 3, 4, 6)
C5	Male	18	165	67.8	11	(1, 2, 3, 5)

3.2.2 Procedures

Recently concussed participants were tested weekly for six weeks, at means (SD) of 7 (0), 16 (1), 23 (2), 29 (1), 36 (2), and 45 (3) days post-concussion. Each control was also tested up to six times, and on the same days as their match to control for musculoskeletal afflictions from recent games or practices. A one-year follow-up testing session occurred at a mean (SD) of 363 (42) days post-concussion for three concussed and two control participants. Not all participants completed all sessions because of scheduling conflicts (See Table 1).

At each testing session, the participants were fitted with two 6-axis inertial measurement units (IMUs) consisting of a MMA tri-axial accelerometer, an IDG-300 (x and y plane) gyroscope and an ADXRS300 z-plane uniaxial gyroscope, configured in a left-hand coordinate systems and aggregated in the Technology Enabled Medical Precision Observation (TEMPO) platform (1, 49). The IMUs were located over the participant’s xiphoid process and over the

forehead, such that the three measured *x*, *y*, *z* directions roughly corresponded to mediolateral, vertical, and anteroposterior directions, respectively. Data were sampled at 128 Hz.

All testing sessions occurred in a basketball gymnasium with minimal distractions and clean, hardwood floors. Participants were asked to walk barefoot around a course marked with 1.5 m tall pylons, and containing an 18 m long straight section denoted with a pylon at the start and end of the section. Participants completed 14 laps, resulting in 14 bouts of straight ST gait. Participants were then given a random number between 900 and 999 and repeated the procedure while serially subtracting by sevens, resulting in 14 bouts of straight, dual-task (DT) gait.

3.2.3 Analysis

Individual steps were identified using the local minima of the vertical accelerations, with strides identified as every two steps. Only the middle eight strides were analyzed for each bout to ensure the participants were walking straight and at a constant speed. The standard deviation of stride times, assessed from the trunk IMU in each condition (ST or DT), defined the traditional stride time variability. Gait speed, recorded with a stopwatch, was calculated using each participant's mean time to walk the 18 m straight section in each condition.

Local dynamic stability was estimated using short-term, finite-time maximum Lyapunov exponents (λ_s), calculated for each short bout of straight gait using the algorithm described by Rosenstein et al. (46). To remove the effect of data series length on LDS (5, 21), strides were time-normalized to 130 samples per stride, corresponding approximately to the participants' mean stride time of 1.05 seconds (48, 56, 57). A fixed state space was reconstructed using the accelerations in all three dimensions (AP, ML, vertical) and their twice time-delayed copies, resulting in a 9D state space (56, 57). A time delay of ¼ the average stride time was used to be consistent with previous methods that examined LDS over short bouts of gait (56, 57).

Within the state-space, the nearest-neighbor of each point was identified and the Euclidean distance between neighboring points was tracked over strides to produce multiple divergence curves (46). The LDS estimate, λ_s, was calculated from the slope of the log of the mean divergence curve between 0 and 0.5 strides (2), representing the ability of the locomotor system to attenuate small perturbations within one step (34). A positive λ_s indicates the rate of exponential divergence between points over time and signifies a locally unstable trajectory, while a negative λ_s indicates points converge over the interval and a stable system (46). Therefore, a *larger* positive λ_s indicates *less* LDS, while a smaller, more negative λ_s indicates a more stable system that is more robust to local perturbations.

The LDS estimate was calculated for both the trunk ($\lambda_{s\text{-Trunk}}$) and head ($\lambda_{s\text{-Head}}$) IMUs for each bout of eight strides, and means obtained over the 13 walking episodes (48, 56) for both ST and DT gait. The first episode for each condition was excluded based on missing data from some participants and to ensure the participants were adjusted to the task. The λ_s values were normalized to individual gait speed for each condition to control for the effect of gait speed on LDS (4, 18, 21). DTCs of the cognitive task on λ_s were calculated as

$$\mathrm{DTC}\,\lambda_S = \frac{(\mathrm{DT}\,\lambda_S - \mathrm{ST}\,\lambda_S)}{\mathrm{ST}\,\lambda_S} * 100\% \qquad 9$$

to represent the percent change in stability associated with the cognitive task. All analyses was performed in MATLAB (MATLAB and Statistics Toolbox Release 2014b, The Math-Works, Inc, Natick, Massachusetts, USA).

3.2.4 Statistical Analysis

Univariate descriptive statistics of gait speed, stride time variability, $\lambda_{s\text{-Trunk}}$ and $\lambda_{s\text{-Head}}$ were calculated for each condition (ST / DT) and group (concussed / control). Linear mixed models were fit to describe each outcome over time. Mixed models are similar to traditional

linear regression models but differ in that they allow for within-subject correlation. Models were adjusted for time, group, task, and two and three-way interaction effects. Higher order interactions with *p* values less than 0.10 were retained in the final model. Model assumptions were validated by examining the residuals. Stride time variability exhibited a non-normal distribution and was thus log-transformed. A two-sided significance level of 0.05 was used throughout, and all statistical analysis was performed in SAS 9.4 (SAS Institute, Cary, NC, USA).

3.3 Results

Univariate results at each week for gait speed, stride time variability, $\lambda_{s\text{-Trunk}}$ and $\lambda_{s\text{-Head}}$ are depicted in Figure 18 - Figure 21, respectively. Concussed athletes exhibited greater DTC than healthy athletes as indicated by significant group*task interactions for $\lambda_{s\text{-Trunk}}$ and stride time variability, with $\lambda_{s\text{-Head}}$ approaching significance (Table 8). No differences in ST stability or variability were found between groups as no significant main effects of group, week, or task were found for stride time variability, $\lambda_{s\text{-Trunk}}$, or $\lambda_{s\text{-Head}}$. The group*task interaction is represented in Figure 22 - Figure 23, which show the DTCs of the cognitive task on $\lambda_{s\text{-Trunk}}$ and $\lambda_{s\text{-Head}}$. Gait speed showed a significant group*day interaction and significant main effects of group and task. Concussed athletes walked slower than healthy athletes. Yet, concussed athletes increased their speed by the one-year follow up. Dual-task gait speed was slower than ST gait speed for both groups.

Table 8. Results from the final linear mixed models. Beta coefficients indicate the mean differences between each effect, with healthy ST gait used as the reference state. The significance level was set at 0.05, and significant terms are shown in bold.

	λs-Trunk		λs-Head		Gait Speed		log Stride Time Variability	
	Beta (SE) $*10^3$	*p* value	Beta (SE) $*10^3$	*p* value	Beta (SE) $*10^3$	*p* value	Beta (SE) $*10^3$	*p* value
Group	-4.0 (31.6)	0.8994	21.4 (23.9)	0.3734	**-154.3 (53.8)**	**0.0054**	58.7 (169.2)	0.7298
Days	-0.02 (0.08)	0.7867	-0.06 (0.07)	0.4318	-0.04 (0.20)	0.8601	0.16 (0.43)	0.7151
Task	-15.2 (15.1)	0.3169	-17.2 (14.6)	0.2431	**-66.7 (12.8)**	**< 0.0001**	-53.6 (69.3)	0.4420
Group*Task	**64.5 (19.9)**	**0.0018**	34.8 (19.3)	0.0753	-	-	**228.6 (91.5)**	**0.0146**
Group*Days	**-**	**-**	-	-	**0.67 (0.26)**	**0.0120**	**-**	**-**

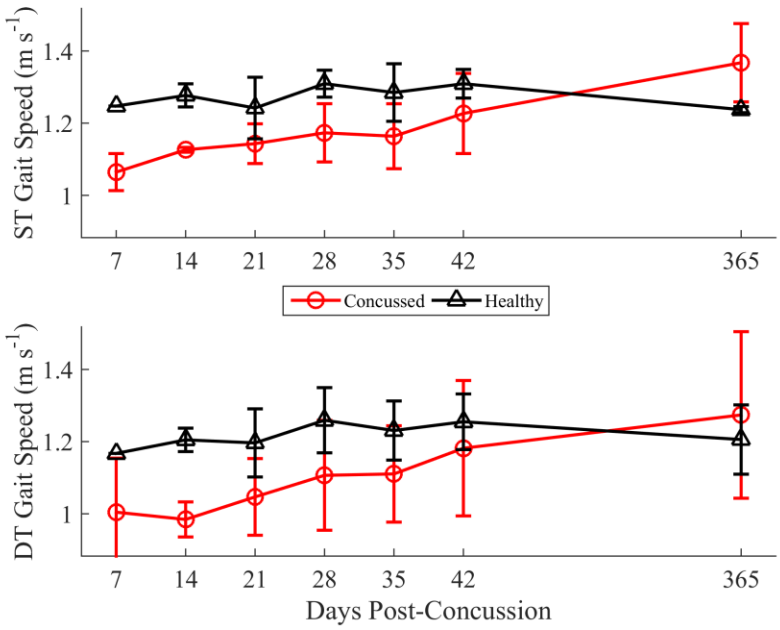

Figure 18. Single (top) and dual-task (bottom) gait speeds for concussed and healthy groups over each week of testing.

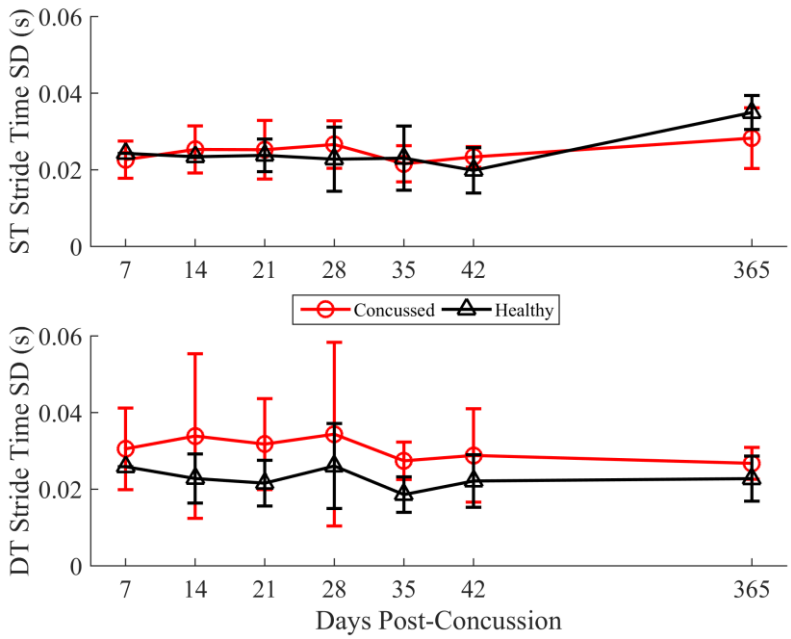

Figure 19. Single (top) and dual-task (bottom) stride time variability, represented as the stride time standard deviation, of concussed and healthy groups over each week of testing.

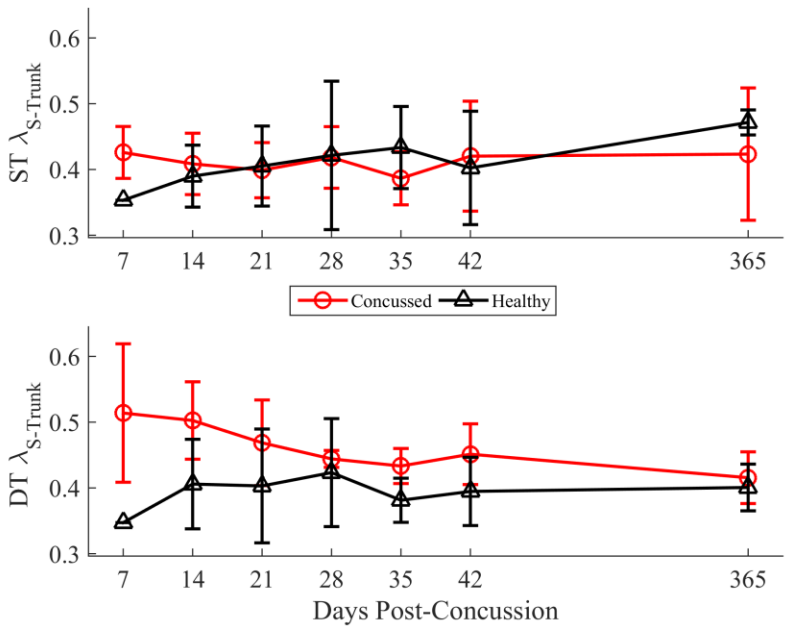

Figure 20. Single (top) and dual-task (bottom) $\lambda_{s\text{-Trunk}}$ for concussed and healthy groups over each week of testing. Larger $\lambda_{s\text{-Trunk}}$ values indicate less stability. All $\lambda_{s\text{-Trunk}}$ values have been normalized to gait speed.

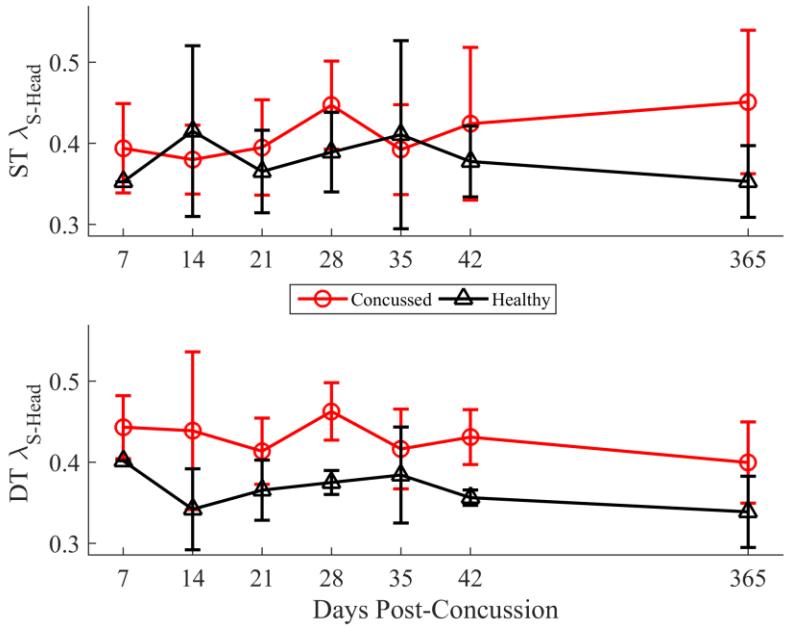

Figure 21. Single (top) and dual-task (bottom) $\lambda_{s\text{-Head}}$ for concussed and healthy groups over each week of testing. Larger $\lambda_{s\text{-Head}}$ values indicate less stability. All $\lambda_{s\text{-Head}}$ values have been normalized to gait speed.

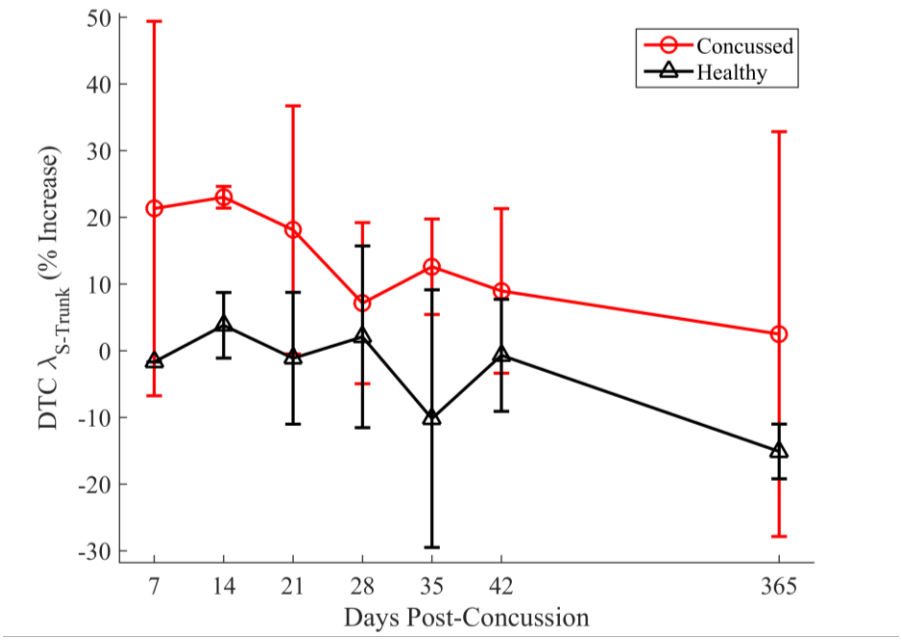

Figure 22. Dual-task costs of $\lambda_{s\text{-Trunk}}$, as a percentage increase from single-task $\lambda_{s\text{-Trunk}}$, for concussed and healthy groups over each week of testing. Positive DTCs indicate increases in $\lambda_{s\text{-Trunk}}$ for DT gait compared to ST gait. Larger $\lambda_{s\text{-Trunk}}$ values indicate less stability. All $\lambda_{s\text{-Trunk}}$ values have been normalized to gait speed.

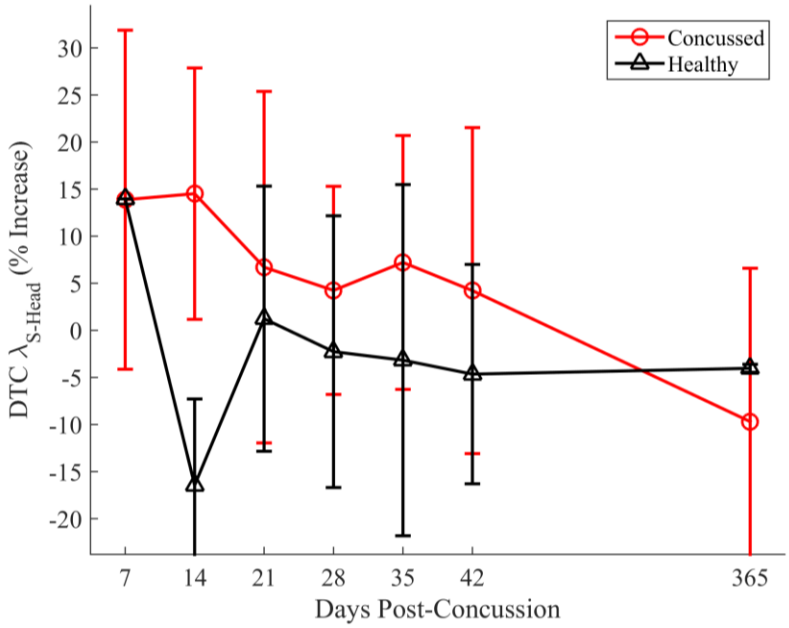

Figure 23. Dual-task costs of $\lambda_{s\text{-Head}}$, as a percentage increase from single-task $\lambda_{s\text{-Head}}$, for concussed and healthy groups over each week of testing. Positive DTCs indicate increases in $\lambda_{s\text{-Head}}$ for DT gait compared to ST gait. Larger $\lambda_{s\text{-Head}}$ values indicate less stability. All $\lambda_{s\text{-Head}}$ values have been normalized to gait speed.

3.4 Discussion

We examined the effect of a recent concussion on the LDS of gait through a prospective analysis of a sample of concussed athletes and matched controls. No significant differences were found between groups for any variability measure, supporting rejection of our initial hypothesis that concussed athletes would exhibit decreased LDS during ST gait. A significant group*task interaction supported our second hypothesis, however, and indicated that the addition of a cognitive dual-task influenced stability in the concussed group more than in the control group, despite similar ST gait stability. Recently concussed athletes also exhibited larger LDS DTCs for several weeks post-concussion. This DT instability was present after the concussed athletes were medically cleared and had returned to full athletic participation. A prolonged decrease in gait speed was also found in concussed athletes, which eventually resolved to normal levels, consistent with previous evidence (43).

In agreement with previous studies, the present results did not show any differences, other than gait speed, between concussed and control groups during ST gait (8, 11, 28, 43). However, DT gait involving a simultaneous arithmetic task elicited a disproportionate destabilizing effect on the concussed individuals, a result which is also consistent with recent literature (9, 10, 15, 23, 27, 28, 45). While previous studies assessed DTC using measures such as ML sway (8, 28, 43), the present investigation provides new evidence using nonlinear local dynamic stability. The larger DT λ_s in concussed athletes reflects diminished responses by the neuromuscular control system to local perturbations. In contrast, previous results quantified the magnitude of such perturbations, such as step time variability (39) and ML sway (8, 26, 28, 43), but did not reflect the neuromuscular response. Thus, these new LDS results confirm earlier suspicions of motor control differences occurring post-concussion.

The increased DTC in concussed individuals reported here and elsewhere may be associated with functional brain changes post-concussion. During DT gait, the cognitive task competes with gait for cortical resources (59) and this competition utilizes a large network of cortical areas (14). The ability to multi-task using cortical networks may be diminished after concussion, which may explain the DTC differences that manifested during gait. Decreased interhemispheric brain connectivity (31) and larger cortical networks (47) post-concussion may limit the available cortical resources or the organization of the cortical networks. During ST gait, there may still be sufficient cortical resources to exhibit healthy locomotion. However, as more demands are introduced the altered cortical resource structure may not have the capacity to multi-task. Conversely, increased attentional demands in healthy individuals did not produce consistent or noticeable changes in the LDS, consistent with results from a previous study (19). While both functional brain changes and DT gait abnormalities occur over the same sub-acute timescale post-concussion, more information from functional brain imaging and gait studies using the same populations are needed before any conclusions can be solidified.

A group*task interaction for stride time variability was found (Fig. 5), with a similar disproportionate increase in DT variability in the concussed group compared to controls. However, this temporal variability cannot account for the differences seen in λ_s because all strides were normalized to the same length of time. By normalizing stride time, the temporal variations caused by kinematic disturbances were removed, leaving perturbations in the movement patterns without stride-duration variability and simplifying the relationship between walking velocity and λ_s (21). Therefore, recently concussed athletes demonstrated greater temporal variability between strides, as indicated by stride-time variability, *and* attenuated those disturbances less effectively over the course of one step, as indicated by λ_s.

The current finding of decreased LDS post-concussion further strengthens the notion that a decrease in LDS (higher λ_s) during gait reflects a deviation towards an unhealthy locomotor system. Whereas previous authors concluded similarly in studies of elderly individuals (7, 34), and patients with Parkinson's disease (32) and multiple sclerosis (29), musculoskeletal deficiencies within those populations and / or long-term adaptations to their conditions may have contributed to observed decreases in LDS. Additionally, gait speed influences the stability estimate λ_s (4, 18, 21) and with varying gait speeds between mobility impaired populations and healthy adults, this influence of speed on λ_s further clouds the conclusions of previous work (2). Conversely, this study examined two populations in peak physical fitness without significant time for any adaptation to the neurological injury suffered by the concussed group. Given the differences in gait speed between groups and conditions, λ_s values were also normalized to gait speed to reflect the near-linear relationship of λ_s to gait speed over a narrow range (0.8 to 1.3 m/s). As such, the decreased LDS may represent functional neurologic changes that affect the neuromuscular control system without apparent musculoskeletal differences. This particular finding echoes results from van Schooten et al. (58) and Sloot et al. (48), who reported increased $\lambda_{s\text{-Trunk}}$ during gait when galvanic vestibular stimulation (GVS) was applied to disrupt vestibular feedback control.

Interestingly, whereas Sloot et al. (48) reported an increase of ~8-11% in $\lambda_{s\text{-Trunk}}$ when GVS was applied, concussed individuals exhibited a similar, and sometimes greater, increase in $\lambda_{s\text{-Trunk}}$ when asked to serially subtract by sevens. Even 4-6 weeks post-injury, the destabilizing effect of this cognitive arithmetic in concussed athletes had a similar magnitude to GVS in healthy young adults. While comparisons between LDS studies should be performed with caution, this similarity between the destabilizing effect of cognitive DT concussed gait and GVS

healthy gait further reinforces the need for in-depth research on locomotor recovery post-concussion, and encourages caution when returning recently concussed athletes to competition.

From a clinical perspective, the persistent decrease in DT LDS is concerning as it lasted beyond the athletes' typical symptom directed return-to-play progression. This prolonged neuromuscular control recovery timeline lasted over six weeks and confirmed previous findings of prolonged DT gait recovery post-concussion (6, 16, 23, 27, 28). The DT LDS results also raise concern about an athlete's well-being and risk of musculoskeletal injury after returning to competition from a concussion. Similar trends with decreased LDS have been associated with an increased risk of falls (3, 34) indicating that concussed athletes, while processing a cognitive load during practice or competition, may be at a greater risk of global destabilization. Given the high cognitive loads present during practices and competitions (52), this decrease in DT LDS may be a contributing factor in the high rate of musculoskeletal injuries in recently concussed athletes (35, 40). It is also worth noting that the LDS dysfunction post-concussion may resolve over the course of one year, indicating that these effects, though persistent, may be temporary. While no significant effect of time was found in the mixed models, the DTC appeared to resolve to healthy control levels in 2 of the 3 athletes tested after one year.

This study was primarily limited by a small sample size. Given the unpredictable nature of concussions and the longitudinal design, only five concussions and four controls were tested. Additionally, due to graduations and transfers of some participants, only three concussed athletes and two control athletes were available for a one-year follow-up. However, the group*task interactions were acutely apparent and detected even with this small sample size. Additionally, the un-normalized ST λ_s values for both concussed (0.48) and control (0.52) groups closely matched those reported by van Schooten et al. (56) of 0.44 (0.04) with near identical methods,

including short bouts of gait and state-space construction. The lack of footwear, which can be considered a compliant surface compared to barefoot, may have contributed to the slightly larger λ_s presented here (12, 53). Furthermore, given the unpredictable nature of sport-related concussions, no pre-concussion measurements were obtained which restricted the analysis to case-control comparisons. Future analysis should attempt to obtain baseline gait data prior to the athletes' competitive seasons. Secondly, this work focused on the initial six weeks post-injury, given established timelines of neuropsychological (38) and gait recovery (28). The results, though, suggest that this timeline was too short to capture the recovery of gait stability. Future studies should examine a time period between six weeks and one year, in addition to the acute and sub-acute clinical phases. Ultimately, larger, more comprehensive longitudinal studies are needed to examine gait, functional brain imaging, and neuropsychological performance congruently and in concussed athletes. Given the gait deficits occurring post-concussion, and the implications for injury and performance, gait and DT gait in particular (33, 45) should be considered important outcomes in future concussion research and clinical assessments.

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CHAPTER FOUR: DECREASED HIGH-FREQUENCY CENTER-OF-PRESSURE COMPLEXITY IN RECENTLY CONCUSSED ASYMPTOMATIC ATHLETES

4.0 Abstract

This study used two experiments to compare multiple postural stability entropy algorithms and processing techniques to identify 1) if postural control complexity changes exist in recently concussed athletes compared to healthy athletes during the subacute, post-return-to-play timeframe, 2) which entropy methodologies most reliably identify previously concussed individuals, and 3) an appropriate interpretation of the observed entropy changes. First, two-minute, eyes closed postural stability data from six recently concussed athletes was longitudinally collected over six weeks immediately following the concussion and compared to data from 24 healthy athletes. Second, 25 healthy non-athlete adults performed four eyes closed quiet standing tasks: normal, co-contracting their lower extremity muscles, performing a cognitive arithmetic task, and voluntary manipulation of their sway. Center of pressure (COP) complexity was calculated using approximate entropy (ApEn), sample entropy (SampEn), multi-variate sample entropy (MV-SampEn), and multi-variate composite multi-scale sample entropy (MV-CompMSE) for both high-pass filtered and low-pass filtered COP data. MV-CompMSE of the high-pass filtered COP signal identified the most consistent differences between groups, with concussed athletes exhibited less complexity over the high frequency COP time-series. In the healthy non-athletes, high-pass filtered MV-CompMSE increased during the co-contraction, but not during other conditions, suggesting the decrease in high frequency MV-CompMSE in concussed athletes may be due to more relaxed muscles or less complex muscle contractions. This decrease in entropy may correlate with previously reported increases in intra-cortical

inhibition. Furthermore, high frequency MV-CompMSE may be a useful clinical tool in the identification and assessment of concussions.

4.1 Introduction

Sport-related concussion has been recognized as a health concern in the United States and balance assessments have become an integral part of concussion assessments to ensure that athletes with previously disrupted balance are fit to return-to-play (RTP) (20, 34, 35). The National Collegiate Athletic Association (NCAA) requires baseline assessment for athletes who participate in sports at risk for concussion with a symptom checklist, standardized cognitive and balance assessments (47). While clinical measures of postural stability, such as the Balance Error Scoring System (BESS), typically resolve to baseline levels within 3-5 days (21-23, 37), there is growing evidence that postural and motor control abnormalities may persist for two months and beyond the standard recovery timeline (3, 12, 17, 24).

Compared with traditional clinical balance measures, application of nonlinear dynamic analyses to postural stability has shown greater sensitivity in identifying postural control differences post-concussion (7). Using approximate entropy (ApEn), which characterizes the regularity of a time-series by quantifying the repetition of patterns (38), Cavanaugh et al. (5) detected less postural control complexity (lower ApEn values) in concussed athletes 48 hours and 96 hours (6) after injury despite normal clinical balance performance. These changes in complexity appear to be persistent; less complexity in the center-of-pressure (COP) time-series has been reported in athletes greater than nine months post-concussion compared to individuals with no concussion history (12, 51). However, information on postural complexity changes that may occur in the intermediary period, between 96 hours and six months post-concussion, is lacking.

To date, only ApEn has been used to assess the complexity of postural stability in concussed athletes, but other entropy analyses may be more appropriate for COP complexity. Sample entropy (SampEn) is considered less biased than ApEn and better suited for short data sets (45). Examining complexity over multiple timescales can further improve the sensitivity to postural control changes (16). Because COP data are multi-dimensional, with both anteroposterior (AP) and mediolateral (ML) components, multivariate entropy methods may also be appropriate to account for potential correlations between the directional signals (54). Compared to ApEn, more recently developed entropy calculations may better identify complexity changes post-concussion.

In addition to the variety of complexity measures, data processing techniques can influence the calculation and interpretation of COP entropy (43, 44). Several studies have examined the entropy of the low-frequency (< 10 Hz) COP signal (6, 7, 12, 14, 19, 46, 52). However, this frequency spectrum is dominated by frequencies below 1 Hz (57), which can create non-stationarities and leave an insufficient number of dominant cycles if the data length is short (e.g., < 1 minute). Both of these technical issues can distort the entropy calculation (9, 17), and to address them other studies have removed the lower frequencies (< 1 Hz) through filtering (9, 33) or increment differencing methods (41, 44). The retained high frequency (> 1 Hz) content provides enough cyclic repetitions and reduced non-stationarities to provide valid entropy calculations. However, interpreting these high-frequency fluctuations is sometimes difficult. While COP fluctuations above 1 Hz may correspond to open-loop postural control (8) and / or ballistic muscle contractions (32), the analysis is based on less than 5% of the COP signal power and over a spectral band often associated with artificial noise (55). While past research has examined these independently, the efficacy of the low vs. high frequency approaches at

differentiating tasks or populations remains unclear (18). Therefore, any recommendation on entropy analysis for specific populations should also identify the appropriate bandwidth to analyze.

This study was completed to address three questions pertinent to future concussion research specifically, and postural control in general: 1) Are there prolonged postural control complexity differences between recently concussed and healthy athletes during the timeframe immediately post-RTP; 2) Which entropy methods can best detect any such differences; 3) How do the interpretations of such regularity differences change when analyzing the different frequency content of a COP signal.

Two separate experiments were conducted. Postural stability data of recently concussed athletes were recorded during the subacute timeframe post-concussion and compared to healthy athlete data. Second, the postural stability of non-athlete healthy young adults was recorded during four quiet standing tasks. The four conditions were chosen to address situations that may be undetectable using standard balance metrics and represented greater muscle stiffness, a greater demand on attentional and cortical resources, and a larger voluntary contribution.

4.2 Methods

4.2.1 Participants

Six Virginia Tech varsity athletes who had recently suffered a concussion participated in this study, along with 25 healthy athletes with no concussion history (Table 9). All concussed athletes were clinically diagnosed by a Virginia Tech Sports Medicine physician and referred to this study by their athletic trainer. Healthy athletes were recruited from various sports. Exclusion criteria for both groups included any unresolved acute lower extremity injury, history of mental illness or diagnosed cognitive impairment. An additional exclusion criterion for healthy athletes

was no concussion or brain injury within the previous year. Recently concussed participants were tested weekly for six weeks, at means (SD) of 7 (0), 16 (1), 23 (2), 29 (1), 36 (2), and 45 (3) days post-concussion, and after one year at a mean (SD) of 363 (42) days post-concussion. Not all participants completed all six sessions because of scheduling conflicts. Healthy athletes were tested one time.

Table 9. Demographic information of each participant group values are means (SD). Concussed athletes were cleared for full RTP after complete resolution of clinical symptoms and after completing the graduated RTP protocol. Full RTP here refers to full medical clearance to participate in games and practices.

Group	Gender	Height (cm)	Mass (kg)	Cleared Full RTP (days)
Concussed	2M / 4F	167.6 (6.5)	65.6 (8.3)	11 (4)
Healthy Athletes	1M / 23F	164.9 (10.3)	68.8 (9.0)	-
Healthy Non-Athletes	17M / 8F	173.7 (9.0)	71.7 (15.0)	-
H-C	1F	170.2	75.1	18

One healthy athlete was found, after testing had completed, to have suffered a concussion 23 days prior to testing and was therefore excluded from the healthy group. However, this single athlete's data (H-C) was retained to investigate the use of entropy as a clinical predictor of recent concussions.

A third sample of 25 healthy young adults, who were not varsity athletes, were recruited from Virginia Tech and the surrounding community to participate in a follow-up study. Exclusion criteria were the same as for the healthy control athletes. All participants gave informed written consent and all recruitment procedures and experimental protocols were approved by the Virginia Tech Institutional Review Board.

4.2.2 Procedures

Concussed and healthy athlete participants were instructed to stand barefoot with their feet together on a force platform (Model # BP400600-2K, AMTI, Watertown, MA, USA) while their postural stability was measured for two minutes. Participants were instructed to close their eyes and "stand still" with their arms at their sides for two minutes. Kinetic data from the force platform was sampled at 1000 Hz for 120 seconds.

Non-athlete participants performed the same quiet standing task described above for the concussed and athlete groups. Each of these participants, though, also performed three additional quiet standing trials with additional task instructions in each, to: 1) "contract all your leg muscles as hard as possible for the full two minutes while standing still" (co-contraction), 2) "silently count down from 800 by sevens" (cognitive task), and 3) "randomly control your body sway" (random sway). Participants were given further clarification for the random sway condition to sway in a manner that would be undetectable to an outside observer. An example was given by the researcher to demonstrate voluntary small amplitude sway that would not necessarily be noticeable to an untrained observer. All participants performed normal quiet standing first, with the following three tasks randomized in order across participants. Multiple practice attempts were allowed for each condition to ensure participants were accustomed to the tasks, and adequate rest was given for all participants following each condition.

4.2.3 Analysis

For each postural stability measurement, data were downsampled to 100 Hz to reflect the upper limit of postural control (26) and the AP and ML components of the COP were computed. Data were filtered two separate ways resulting in two COP time-series for each postural stability measurement. First, raw COP data were low-pass filtered by removing intrinsic mode functions

with frequencies above 10 Hz, using bivariate empirical mode decomposition (42). Raw COP data were also high-pass filtered removing intrinsic mode functions with frequencies below 1 Hz. This created a low-frequency COP time-series (LF) and a high-frequency COP time-series (HF) for each measurement. The resultant COP velocity and 95% ellipsoidal area were calculated from the LF COP time-series.

Regularities of both the HF and LF COP time-series were calculated using four unique entropy algorithms. For entropy analyses, the LF COP signal was further downsampled to 10 Hz, since higher frequency content had been removed. For each HF and LF COP time-series, ApEn and SampEn were calculated for AP and ML components separately. Details on the ApEn (38) and SampEn (27, 45) algorithms can be found in previous literature. Multi-variate SampEn (MV-SampEn) (54) and multi-variate multi-scale composite SampEn (MV-CompMSE) (54, 56) were calculated using the two-dimensional (AP and ML) COP time-series. MV-CompMSE uses a coarse-graining procedure to calculate the SampEn over several time-scales within the signal (56), whereas ApEn, SampEn, and MV-SampEn only calculate regularity over one specific time-scale. For all analyses, a template length $m = 2$ and radius $r = 0.2$ times the standard deviation of the time-series were used.

Because multi-scale entropy algorithms return a SampEn value for each time-scale, a single complexity index is often created to compare populations (9). For our analysis, a piecewise linear fit was applied to the SampEn values returned by MV-CompMSE over scales 1-10 (3.3 to 33.3 Hz) and scales 10-30 (1.1 to 3.3 Hz) for HF time-series, and over scales 1-20 (0.16 to 3.3 Hz) for LF time-series. The area under the entire curve, as well as the slope of each linear fit section, were used as complexity indices for MV-CompMSE (Figure 24). Each MV-CompMSE therefore had three HF CIs (one area, two slopes) and two LF CIs (area and slope).

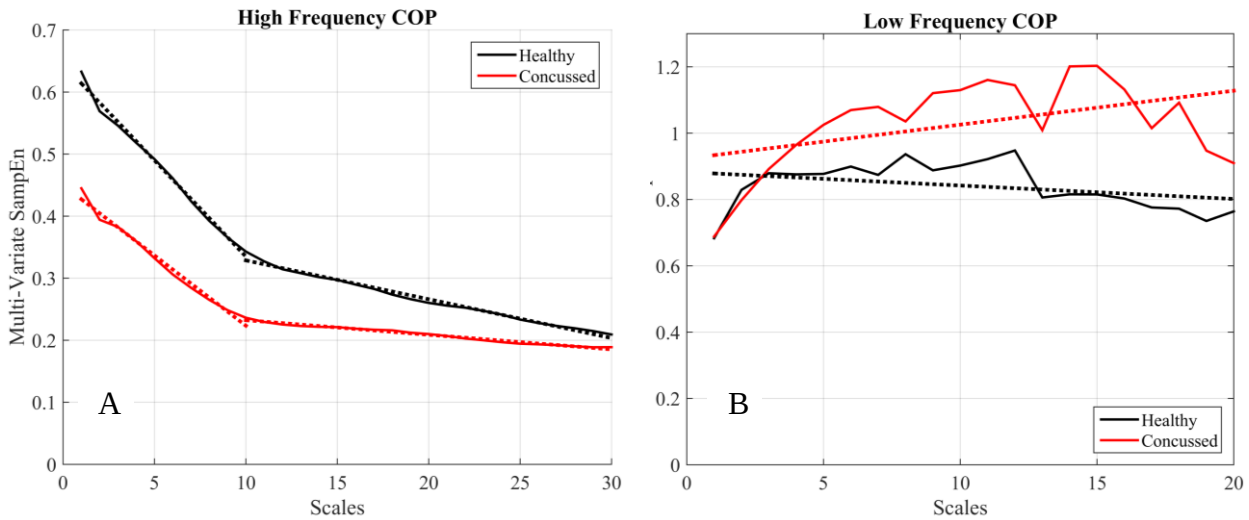

Figure 24. Representative MV-CompMSE curves to illustrate the area and slope complexity indices. A. Representative depiction HF MV-CompMSE curves for one healthy (black) and one concussed (red) athlete. The dotted lines show the linear fits to the two regions of the curve over scales 1-10, and scales 10-30. B. Representative depiction of the LF MV-CompMSE curves from the same healthy (black) and concussed (red) individuals as A. The linear fit was applied over the entire series from scales 1-20. Data in both A and B were obtained from the same raw COP time-series. The slopes of each piecewise linear fit, and the areas under the entire curves were used as complexity indices.

To identify changes in the COP frequency spectrum that contribute to differences in entropy, the power spectral densities of the raw COP series were computed using fast Fourier transforms. The power of the HF bandwidth (>1 Hz) was calculated relative to the overall power of the signal. All analyses were performed in MATLAB (MATLAB and Statistics Toolbox Release 2015a, The MathWorks, Inc., Natick, MA, USA).

4.2.4 Statistical Analysis

4.2.4.1 Concussed and Healthy Athletes

Univariate descriptive statistics of each entropy outcome, complexity index, COP area, and COP velocity were calculated for healthy and for concussed groups at each point in time. For each outcome, independent-sample *t* tests were used to compare the concussed group at each time to the single measurements from the healthy group. Methods that were unable to detect

significant differences between concussed and healthy groups in at least one time point were discarded from further analysis.

Logistic regression models were created for each retained outcome to classify concussed and healthy groups based on postural stability entropy. One combined regression model used all retained entropy outcomes as predictors. The logistic models were used to predict the probability that H-C (athlete whose concussion was discovered after data collection) had sustained a concussion. Since H-C had sustained a concussion, higher probabilities indicated more potential utility as a clinical assessment tool.

4.2.4.1 Non-athlete Healthy Adults

Univariate descriptive statistics for each standing condition were computed for each outcome. To compare the effect of standing condition on each outcome, generalized estimating equations (GEEs), which account for the correlation between measurements within subjects, were fit for all retained entropy outcomes. A compound symmetric covariance structure was used for all models, which assumes equal correlation between all within-subject measurements. Model assumptions were validated using residuals. A two-sided significance level of 0.05 was used throughout, and all statistical analysis was performed in SAS 9.4 (SAS Institute, Cary, NC, USA).

4.3 Results

4.3.1 Concussed and Healthy Athletes

Univariate descriptive statistics for each outcome in the concussed and healthy groups are given in Table 10 (LF outcomes) and Table 11 (HF outcomes). No significant differences in any outcome were found between groups at week 1. At week 2, the concussed group had significantly smaller HF MV-CompMSE areas ($p < 0.001$). This significant difference was also

observed at week 3 ($p = 0.043$), week 4 ($p = 0.005$), and week 6 ($p = 0.039$), with differences at week 5 approaching significance ($p = 0.055$) (Figure 25). Significantly larger HF MV-CompMSE slopes over scales 10 to 30 were found in concussed athletes at week 3 ($p = 0.015$) (Figure 26). The only LF outcomes that significantly differed between groups occurred at week 6 for AP ApEn ($p = 0.018$) (Figure 27) and AP SampEn ($p = 0.016$), with lower values of both among concussed athletes (Figure 28). No significant differences in COP area or resultant velocity were found at any time ($p > 0.09$). No significant differences in relative power were found between groups in weeks 1 through 5 or at one year ($p > 0.09$), but a significant increase in the relative power above 1 Hz was found in the AP direction at week 6 ($p = 0.014$).

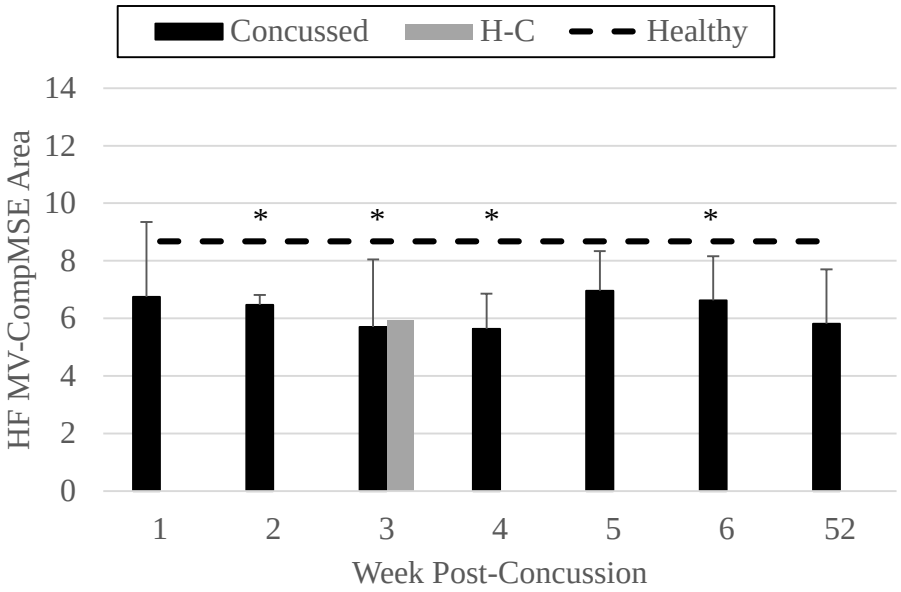

Figure 25. Longitudinal results of HF MV-CompMSE Area. Solid black bars show means and one SD error bars of concussed athletes as each week. The dotted black line indicates the mean of the healthy athletes. The athlete H-C is represented with a gray bar. Significant differences between the healthy group are shown with an asterisk (*). Significant differences between groups were found at weeks 2, 3, 4, and 6.

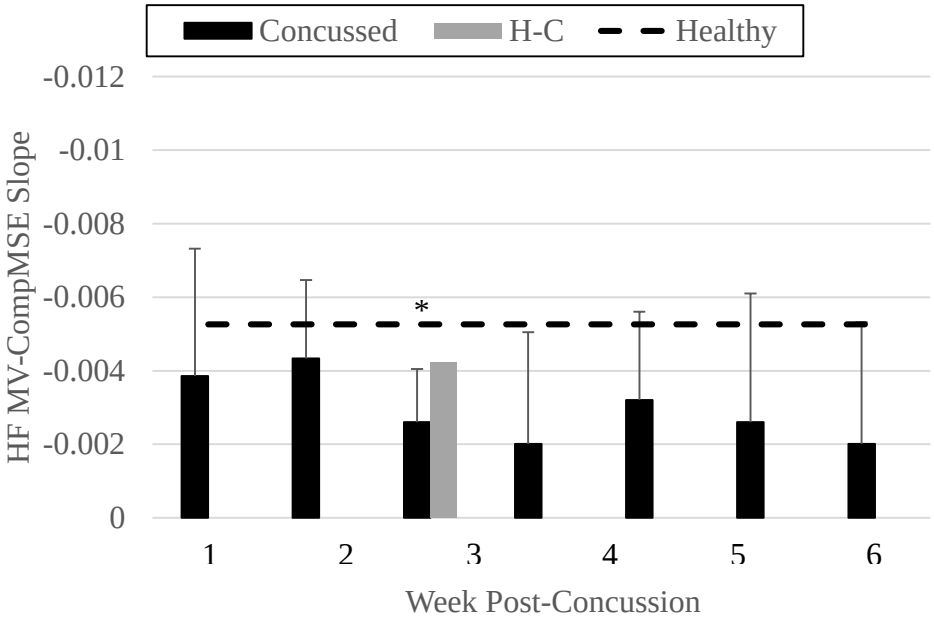

Figure 26. Longitudinal results of HF MV-CompMSE Slope (scales 11-30). Solid black bars show means and one SD error bars of concussed athletes as each week. The dotted black line indicates the mean of the healthy athletes. The athlete H-C is represented with a gray bar. Significant differences between the healthy group are shown with an asterisk (*). Significant differences between groups were found at week 3 only.

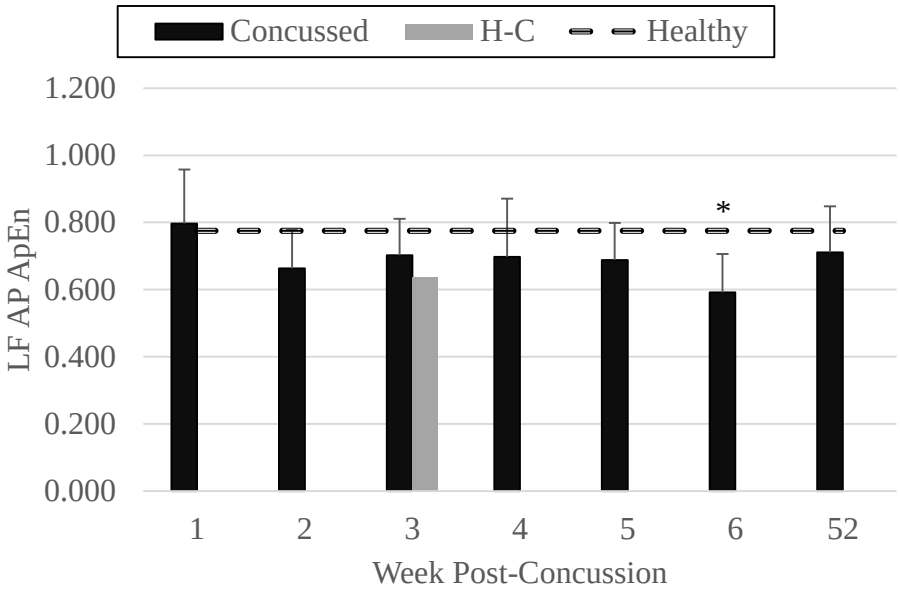

Figure 27. Longitudinal results of LF AP ApEn. Solid black bars show means and one SD error bars of concussed athletes as each week. The dotted black line indicates the mean of the healthy athletes. The athlete H-C is represented with a gray bar. Significant differences between the healthy group are shown with an asterisk (*). Significant differences between groups were found at week 6 only.

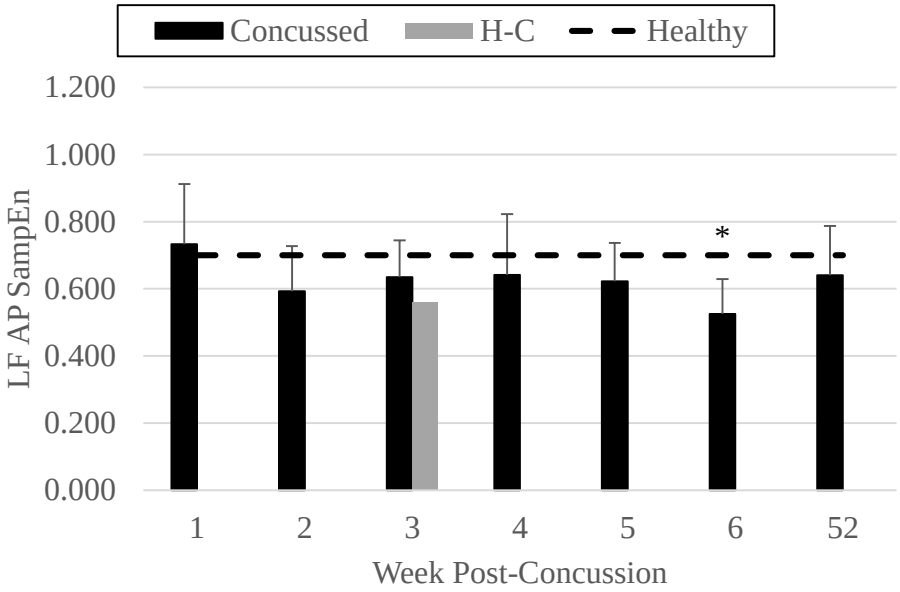

Figure 28. Longitudinal results of LF AP SampEn. Solid black bars show means and one SD error bars of concussed athletes as each week. The dotted black line indicates the mean of the healthy athletes. The athlete H-C is represented with a gray bar. Significant differences between the healthy group are shown with an asterisk (*). Significant differences between groups were found at week 6 only.

Table 10. Means (SD) for each entropy outcomes for the LF COP time-series and from concussed athletes at each week and healthy athletes, including COP area and COP velocity. Significant differences between groups ($p < 0.05$) are shown in bold.

		Concussed							
	Healthy (*n* = 24)	**Week 1 (*n* = 5)**	**Week 2 (*n* = 3)**	**Week 3 (*n* = 5)**	**Week 4 (*n* = 5)**	**Week 5 (*n* = 4)**	**Week 6 (*n* = 5)**	**One Year (*n* = 3)**	**H-C (*n* = 1)**
AP ApEn	0.78 (0.18)	0.80 (0.16)	0.66 (0.11)	0.70 (0.11)	0.70 (0.17)	0.69 (0.11)	**0.59 (0.11)**	0.71 (0.14)	0.64
ML ApEn	0.84 (0.12)	0.86 (0.23)	0.73 (0.10)	0.81 (0.18)	0.81 (0.25)	0.81 (0.21)	0.76 (0.17)	0.87 (0.22)	0.75
AP SampEn	0.70 (0.19)	0.73 (0.18)	0.59 (0.14)	0.64 (0.11)	0.64 (0.18)	0.62 (0.11)	**0.53 (0.10)**	0.64 (0.15)	0.56
ML SampEn	0.77 (0.13)	0.80 (0.28)	0.62 (0.09)	0.73 (0.19)	0.74 (0.29)	0.74 (0.25)	0.69 (0.18)	0.81 (0.27)	0.64
MV-SampEn	0.69 (0.09)	0.73 (0.10)	0.66 (0.05)	0.67 (0.08)	0.67 (0.11)	0.68 (0.10)	0.67 (0.07)	0.67 (0.15)	0.63
MV-CompMSE Area	18.25 (1.87)	18.05 (1.64)	18.91 (1.89)	18.53 (2.21)	18.06 (1.86)	18.18 (1.95)	18.16 (1.92)	17.64 (1.69)	16.63
MV-CompMSE Slope	0.004 (0.011)	0.000 (0.011)	0.011 (0.010)	0.007 (0.007)	0.004 (0.009)	0.006 (0.007)	0.004 (0.004)	0.009 (0.014)	0.005
COP Area (cm^2)	7.59 (3.18)	12.98 (14.59)	15.96 (14.16)	6.21 (1.32)	8.44 (4.64)	10.92 (8.80)	11.93 (8.45)	7.15 (2.58)	7.220
COP Vel (cm/s)	2.13 (0.53)	2.47 (0.72)	2.29 (0.93)	1.80 (0.31)	1.89 (0.23)	2.05 (0.30)	1.89 (0.33)	2.04 (0.42)	1.74

Table 11. Means (SD) for each entropy outcomes from the HF COP time-series for concussed athletes at each week and for healthy athletes, including the relative power of the HF bandwidth. Significant differences between groups ($p < 0.05$) are shown in bold.

		Concussed							
	Healthy (*n* = 24)	**Week 1 (*n* = 5)**	**Week 2 (*n* = 3)**	**Week 3 (*n* = 5)**	**Week 4 (*n* = 5)**	**Week 5 (*n* = 4)**	**Week 6 (*n* = 5)**	**One Year (*n* = 3)**	**H-C (*n* = 1)**
AP ApEn	1.07 (0.32)	0.98 (0.44)	1.08 (0.20)	1.03 (0.15)	1.11 (0.27)	1.15 (0.17)	1.17 (0.33)	1.20 (0.28)	1.30
ML ApEn	0.91 (0.29)	0.75 (0.33)	0.79 (0.37)	0.96 (0.20)	1.10 (0.21)	0.90 (0.08)	1.08 (0.33)	0.95 (0.17)	0.89
AP SampEn	0.95 (0.32)	0.84 (0.44)	0.93 (0.19)	0.90 (0.14)	0.96 (0.27)	1.00 (0.18)	1.04 (0.34)	1.06 (0.26)	1.16
ML SampEn	0.77 (0.28)	0.61 (0.31)	0.66 (0.36)	0.81 (0.19)	0.93 (0.19)	0.74 (0.08)	0.92 (0.31)	0.79 (0.15)	0.72
MV-SampEn	0.49 (0.17)	0.38 (0.20)	0.39 (0.12)	0.39 (0.09)	0.45 (0.07)	0.43 (0.07)	0.49 (0.19)	0.45 (0.06)	0.44
MV-CompMSE Area	8.67 (2.34)	6.75 (2.60)	**6.47 (0.34)**	**5.70 (2.34)**	**5.64 (1.23)**	6.95 (1.38)	**6.62 (1.53)**	5.81 (1.89)	5.93
MV-CompMSE Slope (1-10)	-0.020 (0.016)	-0.018 (0.016)	-0.017 (0.011)	-0.021 (0.009)	-0.28 (0.009)	-0.021 (0.006)	-0.028 (0.014)	-0.027 (0.012)	-0.025
MV-CompMSE Slope (10-30)	-0.005 (0.004)	-0.004 (0.003)	-0.004 (0.002)	**-0.003 (0.001)**	-0.002 (0.003)	-0.003 (0.002)	-0.003 (0.003)	-0.002 (0.003)	-0.004
AP Relative Power (%)	4.24 (3.74)	5.19 (3.42)	2.11 (1.30)	3.18 (1.93)	3.42 (3.56)	2.97 (2.07)	1.93 (0.95)	2.92 (1.91)	1.44
ML Relative Power (%)	6.65 (4.28)	9.08 (6.93)	3.46 (1.74)	6.44 (4.40)	6.84 (7.37)	7.04 (6.98)	4.90 (3.85)	8.61 (6.36)	5.76

From fitting logistic regression models, HF MV-CompMSE was the best individual predictor with a probability of 0.75 that H-C had suffered a recent concussion. The other retained outcomes – HF MV-CompMSE slope over scales 10 to 30, LF AP ApEn, and LF AP SampEn – yielded probabilities of 0.52, 0.61, and 0.59, respectively. The logistic regression model including HF MV-CompMSE area, HF MV-CompMSE slope (scales 10 to 30), LF AP ApEn,

and LF AP SampEn yielded a probability of 0.80, a slight improvement over the single predictor model using only HF MV-CompMSE.

4.3.2 Non-athlete Healthy Adults

Univariate descriptive statistics for selected measures are presented in Table 12 in each standing condition. The remaining measures were discarded as they failed to detect significant differences between concussed and healthy groups. Compared to normal quiet standing, the co-contraction condition produced significantly higher HF MV-CompMSE areas ($p < 0.001$) (Figure 29), higher LF AP ApEn ($p = 0.001$), higher LF AP SampEn ($p = 0.002$) (Figure 30), and larger COP sway areas ($p < 0.001$) and velocities ($p < 0.001$). Significant differences between normal standing and cognitive task conditions were only found in COP area ($p = 0.005$) and resultant velocity ($p = 0.002$). Similarly, the random sway condition only differed from normal standing in COP area ($p < 0.001$) and velocity ($p = 0.002$). The co-contraction condition produced a significant increase in relative power > 1 Hz in both the AP ($p < 0.001$) and ML ($p < 0.001$) directions compared to normal standing (Figure 31). The random sway condition decreased relative HF power in the ML direction only ($p < 0.001$), while the cognitive task condition did not produce any significant changes in the HF relative power.

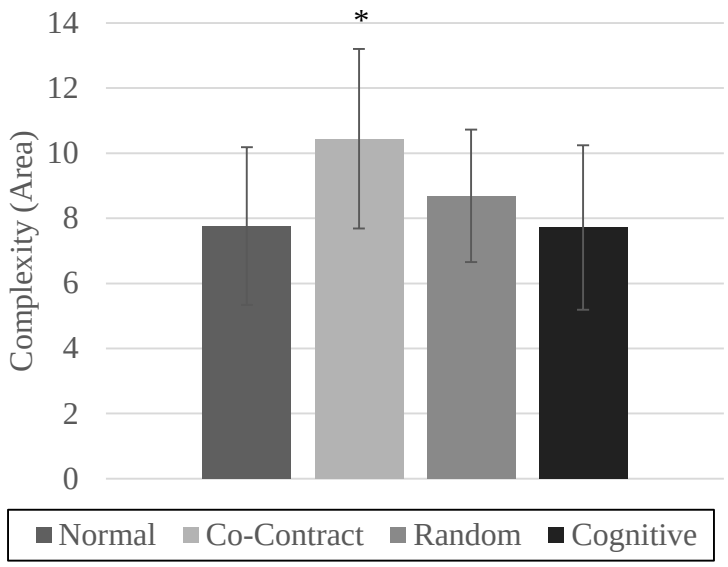

Figure 29. HF complexity from MV-CompMSE area in healthy non-athletes during normal, co-contraction, random, and cognitive standing conditions. Conditions significantly different than normal are indicated with an asterisk (*). The co-contraction condition has significantly greater complexity than normal.

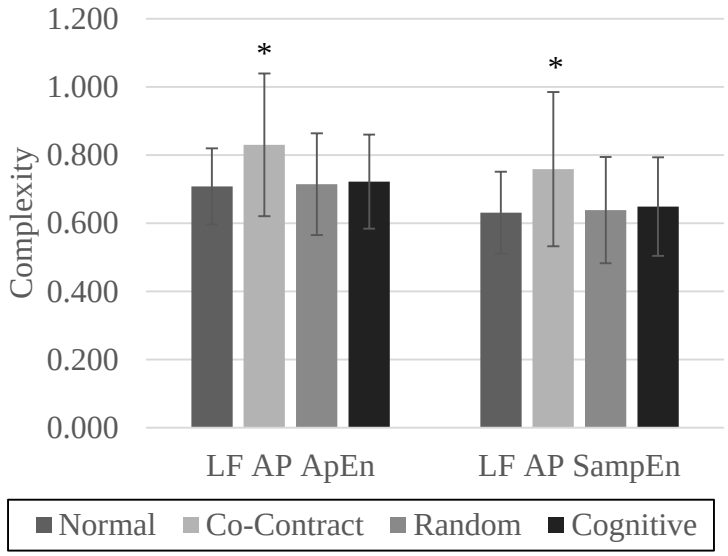

Figure 30. LF complexity from AP ApEn and AP SampEn in healthy non-athletes during normal, co-contraction, random, and cognitive standing conditions. Conditions significantly different than normal are indicated with an asterisk (*). The co-contraction condition has significantly greater complexity than normal.

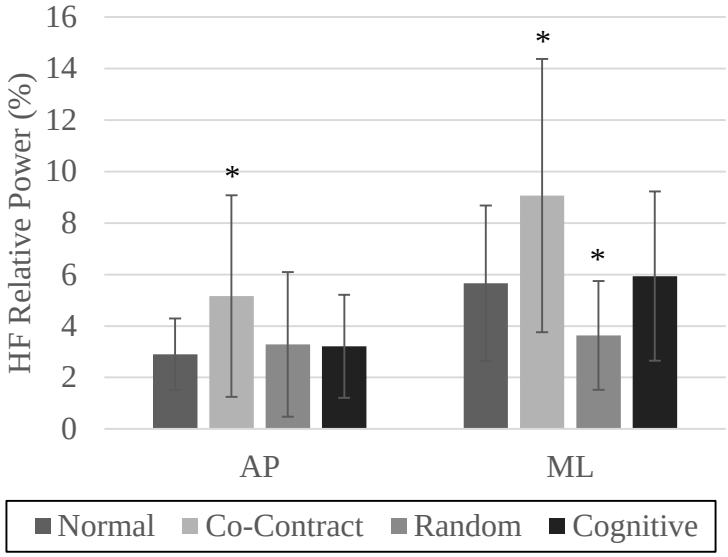

Figure 31. Relative power of the high frequency COP content >1 Hz in healthy non-athletes during normal, co-contraction, random, and cognitive standing conditions. Conditions significantly different than normal are indicated with an asterisk (*). The co-contract condition had significantly greater power than normal in both the AP and ML directions, while the random condition had significantly less relative high frequency power compared to normal.

Table 12. Means (SD) for select entropy outcomes for each standing condition in healthy non-athletes, including the relative power of the HF bandwidth. Conditions significantly different ($p < 0.05$) than normal are shown in bold.

	Normal	**Co-contraction**	**Random Sway**	**Cognitive Task**
LF AP ApEn	0.71 (0.11)	**0.83 (0.21)**	0.72 (0.15)	0.72 (0.14)
LF AP SampEn	0.63 (0.12)	**0.76 (0.23)**	0.64 (0.16)	0.65 (0.14)
HF MV-CompMSE Area	7.76 (2.43)	**10.44 (2.76)**	8.69 (2.04)	7.72 (2.53)
HF MV-CompMSE Slope (10-30)	-0.006 (0.004)	-0.005 (0.003)	-0.006 (0.004)	-0.006 (0.004)
HF AP Relative Power (%)	2.91 (1.38)	**5.16 (3.91)**	3.28 (2.81)	3.21 (2.00)
HF ML Relative Power (%)	5.66 (3.02)	**9.06 (5.31)**	**3.63 (2.12)**	5.94 (3.29)

4.4 Discussion

This study evaluated the postural control complexity of recently concussed athletes, healthy athletes, and healthy non-athletes to identify and help interpret postural control abnormalities that are present 1-6 weeks post-concussion. In agreement with previous evidence (5, 6, 12, 51), concussed athletes exhibited more regularity (less entropy, less complexity) compared to healthy athletes. While prior studies had identified a decrease in complexity among athletes 48-96 hours post-concussion (5, 6) and over nine months post-concussion (12, 51), this study presents the first direct evidence that postural stability complexity remains affected by concussion in the timeframe surrounding RTP (1-6 weeks post-concussion). This result strengthens the growing body of evidence documenting post-concussion postural control deficits that persist beyond the recovery of clinical signs and symptoms (12, 13, 25, 40, 50, 51).

While identifying decreased postural control regularity in recently concussed athletes was one goal of this study, the existence of several entropy algorithms (16) and data processing techniques (43, 44) prompted a comparison to identify the most appropriate postural complexity method for concussion management. When analyzing the low-frequency content of the COP signal, ApEn and SampEn were the only methods to detect significant differences between concussed and healthy groups, and these were only able to do so at week 6. Our LF ApEn results are partially supported by prior work, in that previous reports of postural stability entropy in concussed athletes used ApEn on LF COP signals (5, 6, 12, 51). However, LF ApEn only identified significant differences between groups here at one point in time (week 6), suggesting LF ApEn may not be the most sensitive entropy algorithm to use in concussion management. Conversely, HF MV-CompMSE identified significant increases in regularity (lower complexity)

in concussed athletes over several points in time and provided the most consistent differentiation of the concussed and healthy groups.

This change in complexity over a HF spectrum (1.1-33.3 Hz) can be challenging to interpret as it is often associated with artificial noise when analyzing human movement (55). However, ballistic muscle contractions in the soleus and gastrocnemius occur over a portion of this relatively high frequency spectrum (1-3 Hz) (31, 32) and comprise a critical active component for maintaining upright stance (2, 58). Additionally, COP spectral content up to 9 Hz can be affected by visual feedback, indicating postural control mechanisms at these higher frequencies (49). Given the observed increase in entropy during the co-contraction condition in non-athlete adults, changes in application of these ballistic impulses and joint stiffness may account for some of the entropy changes in the HF bandwidth. Yet, the MV-CompMSE slope over scales 10 to 30 corresponding to the appropriate bandwidth (1.1-3.3 Hz) did not detect any complexity differences. The overall complexity of the HF COP signal increased, but the shape of the curve did not, indicating that constant muscle activity may increase the complexity of the COP over multiple scales but does not alter the scale-dependent shape.

The contribution of ballistic contractions to the HF COP complexity is further supported by the LF ApEn and SampEn results. Though potential long-range correlations and non-stationarities within the raw COP data were retained in the LF COP signal, to replicate the methods of prior investigations on post-concussion COP complexity (5, 6), the parameter selection of $m = 2$ resulted in an examination of patterns 3 points long. Thus, the single-scale LF entropy metrics actually quantified the regularity over 333 ms time scales (3.3 Hz), consistent with the time scale of unidirectional changes in muscle length (32). The lack of significant differences in the LF multi-scale entropy measures indicates that the overall regularity of the LF

content did not differ between concussed and healthy groups between 0.16-3.3 Hz, even though healthy athletes exhibited more complexity at 3.3 Hz.

While it is possible that the decreased complexity in concussed HF COP is due to more ordered and / or less frequent ballistic muscle contractions, such movements only explain a small portion of the HF spectrum analyzed here. Monosynaptic reflex responses, which occur over faster timescales, 53 to 65 ms (15-18 Hz) (15, 48), also contribute to the regulation of ankle angle with short-latency torques opposing stretch (1). Manor et *al*. (33) found multi-scale complexity changes in human COP time-series over even shorter time scales of 8-33 ms (30-125 Hz), suggesting joint stiffness may also contribute to multi-scale complexity over HF bands. Postural control over these high frequency bands (11-33 Hz) has also been supported by the presence of an anti-persistent fractal scaling process, suggesting an "ON/OFF" intermittency model for the ballistic muscle contractions (26), and the presence of a "proprioceptive dead zone" (30). Combined, this suggests that recently concussed athletes experience a decline in HF postural control complexity caused by a degradation of one or more multi-scale processes, or a change in the interaction between processes.

There is physiological evidence to support this decreased HF postural control complexity post-concussion. Particularly, increased intra-cortical inhibition in the motor cortex (39) and longer cortical silent periods following transcranial magnetic stimulation (36) are indicative of alterations to the corticospinal excitability and / or pathways following concussion (29). This increased cortical inhibition has been associated with decreased muscle stiffness (4, 28, 53) and inhibition of prepared actions (10), which would be consistent with the present results. Though previous authors suggested that decreased COP complexity post-concussion may be due to increased co-contraction or stiffened musculature as an adaptation to decreased balance control

(6, 11), we found an increase in complexity when non-athlete participants were instructed to co-contract their lower extremity muscles. Thus, co-contraction and muscle stiffness appear positively correlated with HF multi-scale entropy and complexity. Based on these results and prior cortical excitability studies (12, 29, 36, 39), the increase in COP regularity post-concussion may be caused by decreased co-contraction, more relaxed musculature, and potentially less cortical excitability to initiate corrective ballistic contractions. However, future investigations that include measures of muscle activation and length are needed to confirm this result.

This study had three primary limitations. First, the relatively small sample size limits the generalizability of this study to broader populations and does not provide adequate normative data for future clinical use. However, this study presents the first evidence of persistent decreases in complexity over the subacute timeframe post-concussion. Second, baseline (pre-concussion) measurements were not obtained from the concussed group. It is possible that the lower complexities evident in the concussed group were natural differences in those individuals, though that concern is partially mitigated by the significant differences between groups. Future studies can address this limitation by including baseline measurements prior to concussion. Third, while this study drew conclusions from prior evidence regarding muscle contraction, stiffness, and cortical excitability, such outcomes were not measured here. Therefore, greater muscle co-contraction and stiffness could only be assumed from the non-athlete co-contraction condition using the relative power of the COP.

While the clinical implications of decreased HF postural complexity post-concussion are not clear, the current results do encourage further study of HF postural control complexity as a potential clinical test in RTP decisions. Athletes may be medically cleared for competition, but residual postural deficits can still be present (12, 40). Interestingly, postural complexity deficits

were apparent for several weeks post-concussion, matching the timeframe of reported cortical hypoexcitability (36, 39). Thus, HF postural complexity may serve as a useful clinical tool in concussion diagnosis or the assessment of RTP suitability, especially given the relatively low material and time-demands compared to transcranial magnetic stimulation. A single case-study presented here supported this utility, with postural complexity offering moderate clinical utility in identifying an athlete with a previous concussion, without the use of any other clinical tools. It may be possible to improve the diagnostic accuracy of concussion screenings or return-to-play decisions by incorporating postural complexity into current concussion protocols. Ultimately, larger and longer studies are needed to identify normative data and the appropriate recovery timeline before postural complexity can be used in a clinical setting. However, these results indicate postural complexity presents promise and is worthy for inclusion in future longitudinal studies.

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CHAPTER FIVE: SUMMARY OF FINDINGS, CONCLUSIONS, AND FUTURE WORK

Concussions affect each individual uniquely (13) and, while the knowledge of concussions continues to improve, disagreements regarding the definition, duration, and severity of the injury are present within the clinical and scientific community (21, 30). In particular, while a recent consensus statement on concussion in sport concluded that "the majority (80-90%) of concussions resolve in a short (7-10 day) period" (21), multiple studies have reported that motor dysfunction can last substantially longer and up to two months (8-11, 15, 16, 20, 23, 26-29, 31). Most locomotor abnormalities have been characterized during straight-line gait and with clinical measures of stability (4-8, 15, 16, 20, 23-25). This dissertation added to existing evidence in addressing three specific aims that examined potential effects of concussion on 1) turning gait, 2) nonlinear local dynamic gait stability, and 3) postural stability complexity after clinical signs and symptoms had subsided. This work found significant motor abnormalities in turning gait, dynamic gait stability, and postural complexity that persisted well beyond the clinical recovery timeline, supporting evidence of prolonged motor changes in previous studies (3-8, 15, 16, 20, 23-25).

5.1 Effects of concussion on turning kinematics

While paths around the corner obstacle did not differ between groups, concussed athletes walked slower while turning. When controlled for gait speed, recently concussed athletes exhibited less abrupt turns (lower curvatures) and less ML inclination compared to matched controls, regardless of task condition (Chapter 2). During straight gait, less ML inclination is typically an indication of higher stability as the COM is located closer to the stance limb and within the base of support (25). During turning, however, inclination away from the stance limb and towards the direction of the turn is used to compensate for the lateral forces that are present

(12). The concussed athletes, by leaning into the turn less, likely experienced less stable body orientations and greater coronal plane moments compared to the healthy controls. Whether these kinematic differences produced any substantial injury risk has yet to be determined, but similar decreases in inclination towards the turn during maneuvers such as cutting have been associated with ACL injuries (14).

5.2 Effects of concussion on dual-task local dynamic stability

Contrary to the initial hypothesis, concussed athletes' LDS did not differ from healthy athletes during single-task gait (Chapter 3). However, the addition of a cognitive task elicited a decrease in LDS in the concussed group, while the controls exhibited no change or an increase in stability. This significant group*task interaction may be indicative of functional brain changes post-concussion and a decreased ability to complete multiple simultaneous tasks, each competing for cortical resources. During dual-task gait, concussed athletes exhibited more temporal variability (stride time) and attenuated disturbances less effectively (LDS) than healthy controls. These changes in temporal variability and LDS likely represented an altered motor control system and a greater risk of destabilizing perturbations (1, 18).

5.3 Effects of concussion on postural stability complexity

Consistent with previous studies (10, 31), a decrease in postural complexity was found in concussed athletes that persisted beyond the recovery of clinical signs and symptoms (Chapter 4). High-frequency MV-CompMSE exhibited consistent differences between groups, with decreased COP complexity in concussed athletes over several weeks post-concussion. The changes in the complexity of the high-frequency COP content may correspond to physiological changes to corticospinal excitability after concussion, which have been associated with decreased muscle stiffness and inhibition (2, 17, 32). Compared to normal standing, the co-contraction

condition in healthy non-athletes produced an increase in high frequency MV-CompMSE complexity further supporting the idea that the decrease in high-frequency COP complexity in concussed athletes is due to relaxed musculature and / or changes to the corrective ballistic muscle contractions potentially caused by abnormalities in the motor cortex. However, since muscle activity, muscle length, and motor cortex inhibition were not measured, these conclusions are necessarily speculative.

5.4 Effects of concussion on locomotor and balance as a whole

The current evidence supports a conclusion that concussions produce important effects on locomotion and balance in a generally destabilizing manner. The connection between these long-lasting gait and balance abnormalities and concussions likely lies in changes to the corticospinal system and the functional reorganization of the brain. The longitudinal effects of concussion on obstacle circumvention and balance control during straight gait have previously been defined, especially during dual-task gait (11, 15, 16), but the direct pathways that cause these deficits are unclear. The correlations between COP fluctuations presented in Chapter 4 and the intra-cortical inhibition reported in the literature (22, 26) provide a potential cause. Weaker maximum voluntary contractions and longer cortical silent periods (22, 26) may also contribute to the lower dual-task LDS post-concussion (Chapter 3), since signals for corrective movements may degrade in amplitude as they travel from the brain and are thus less effective at attenuating perturbations. While no concrete conclusions regarding this causal pathway can be made without further studies and direct associations, this work illustrates that concussions disrupt locomotor and balance capabilities beyond the traditional clinical recovery time-frame.

5.5 Clinical implications

Three clinically-relevant conclusions can be drawn from these results: 1) gait and balance control remains affected for several weeks to months post-concussion, 2) these motor control abnormalities may increase the risk of injury for recently concussed athletes, and 3) current clinical concussion assessments do not identify these motor dysfunctions. Kinematic differences during turning persisted for weeks and may not have fully recovered until 6-12 months post-injury (Chapter 2). Similarly, the decreased dual-task LDS (Chapter 3) and decreased postural complexity (Chapter 4) persisted for up to six weeks post-concussion. As all concussed participants were cleared to return to full athletic competition within three weeks of their concussion, all of these deficits were present as the athletes were in full athletic participation. The decreased ML inclinations (Chapter 2) and decreased LDS (Chapter 3) are particularly concerning from a clinical standpoint, considering their respective associations with ACL injuries (14) and falls (1, 18). An increased rate of musculoskeletal injuries in recently concussed athletes has already been reported (19), and these specific deficits are plausible causative factors. Unfortunately, current concussion assessments do not incorporate gait and dynamic balance measurements and therefore remain blind to many of the motor control dysfunctions reported here and elsewhere. Inclusion of quantified, specific motor tasks, such as dual-task gait LDS, may improve the clinical return-to-play decision and prevent athletes from returning to competition in compromised states.

5.6 Contributions to concussion research

This work presents the first prospective, longitudinal study to examine pre-planned turning kinematics or local dynamic gait stability. It is also the first to report on postural complexity changes in the 1-6 weeks post-concussion. While the timeframe of symptom

resolution drove many initial balance studies, the evolving discussion about the duration and severity of concussions will undoubtedly lead to increased research on the sub-acute timeframe. These results support previous investigations that motor control, particularly gait, should be a critical component to these future studies. Furthermore, this work provides the first information incorporating one year post-injury data to inform the duration of such future longitudinal gait studies. This work also suggests that clinical rehabilitative techniques to enhance neuromuscular control may need to be incorporated into return-to-play protocols.

5.7 Contributions to gait and balance research

Beyond the concussion domain, this dissertation also makes several contributions to the domains of gait and balance. Local dynamic stability results in Chapter 3 support growing evidence that decreased LDS represents an unhealthy system. Previous investigations were limited by potential musculoskeletal differences or adaptations between populations. As all participants were physically similar and in peak physical condition, it is unlikely any differences were due to musculoskeletal differences not attributable to motor control. Additionally, while previous postural complexity analyses struggled to interpret complexity changes in high-frequency COP signals, results from Chapter 4 suggest such changes are related to muscle activity, with greater stiffness and contraction associated with greater complexity.

5.8 Limitations

The most important potential limitation to this dissertation is the small sample sizes employed. Only six concussed athletes participated in at least one session, with only five completing two or more of the longitudinal time points. Delayed recruitment, logistical complications, scheduling, and availability also contributed to at least one missing test session in every participant. Additionally, while the experiments included time points throughout six weeks

and at one year post-concussion, no measurements were taken between six weeks and one year resulting in a long period of time with no data. Therefore, conclusions about the recovery or resolution of any abnormalities will not be accurate within this timeframe and serve merely as a rough guide for future research.

More generally, this work suffered from a limitation inherent in most concussion research. Concussions, as a clinical diagnosis and unique to individuals, cannot be cleanly parsed into groups based on the clinical diagnosis alone. Variations in the severity, affected areas of the brain, and other individual factors can all influence the presentation and resolution of signs and symptoms. Therefore, while a clinical diagnosis can be used as a guide, future research should attempt to directly associate neuroimaging metrics, brain biomarkers, and motor dysfunction without relying solely on clinical diagnoses.

5.9 Future research directions

Concussion research is a rapidly expanding and evolving field covering a wide range of domains from neuroscience to biomechanics to athletic training. As a multi-faceted and individualized injury, going forward it is necessary to combine these multiple disciplines in cohesive, longitudinal studies to draw direct comparisons between motor deficits, neuroimaging abnormalities, cognitive decline, and neuropsychological dysfunction. However, given the enormity of that challenge, specific directions in the locomotion domain should also be targeted.

This dissertation provided some evidence of an increased risk of musculoskeletal injury post-concussion using kinematic data, but future studies should specifically monitor kinetics to identify different joint loading characteristics, particularly during dynamic movements such as turning and cutting. Response to global perturbations such as slips and trips is also worthy of future investigation.

Concussions are unique injuries in that, while there are a return-to-play guidelines, there is no evidence-based rehabilitative protocol. Return-to-play protocols ensure symptoms will not worsen with exercise, but do not evaluate the competitive readiness of the athlete. Rehabilitation, through targeted vestibular, obstacle avoidance, and dual-task paradigms may help restore normal functioning faster. Given that some of the locomotor and balance deficits reported here persist for weeks and months post-concussion, a concentrated effort to create, and examine the efficacy of, various rehabilitation protocols is necessary to ensure a timely, accurate, and risk-averse return-to-play decision.

5.10 Overall Conclusions

In conclusion, the present research focused on the longitudinal effects of concussion on turning kinematics, local dynamic stability, and postural control complexity. Recently concussed athletes exhibited persistent locomotor deficits during both non-straight and straight gait and decreased postural complexity, and that such deficits persisted for over six weeks. The application of non-linear dynamics to concussed gait and postural control, in particular, highlighted the motor control abnormalities in concussed athletes. While these changes may be due to structural and functional changes within the brain, more research is needed to directly connect brain changes and motor dysfunction in concussed athletes.

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APPENDIX A: CONCUSSED AND MATCHED CONTROL CONSENT FORM

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants in Research Projects Involving Human Subjects

Title of Project: The Effects of Mild Traumatic Brain Injury on Turning Biomechanics and Non-Linear Gait Measures in Collegiate Athletes (Concussed and Control Athlete)

Investigators: Per Gunnar Brolinson, D.O., Principal Investigator
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Peter Fino, Graduate Student
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I. Purpose of this Research/Project

Concussions have been shown to affect how people walk. However, some subtle differences between healthy and concussed people may not be visible with traditional tools. More advanced tools such as non-straight gait analysis and non-linear stability analysis may be able to identify concussed populations and examine the recovery timeline. Because athletes exert themselves both physically and mentally while playing, the changes reported during walking while performing a mental task such as counting may indicate the athlete is not fully recovered. If the athlete is not fully recovered from the concussions, he/she may be at risk of both injury and poor performance during competitions. To address this concern, this study seeks to figure out the recovery timeline of various balance and gait characteristics.

B. Time Requirements

The study will consist of up to 7 sessions, with each session lasting approximately 35 minutes. The six sessions will be spaced out as follows: 48 Hours post-concussion, 7 days post-concussion, 14, 21, 28, 35, and 42 days post-concussion. The post-concussion sessions for the teammate serving as the healthy comparison will be conducted at the same intervals as the concussed teammate.

C. Study Procedures

A marked course will be arranged using a series of pylons for you to walk through. A series of 1.5 m (or taller) pylons will be set up in a slalom pattern across the ground. The pylons will be set up for you to make approximately 90° turns whenever you reach a pylon. The straight section will be 18 m long. This course will be arranged inside the Jamerson Athletic Center. Reflective tape or markers will be attached on your suprasternale, left and right acromioclavicular joints, sacrum,

head, left and right ankles, and left and right heels. Inertial measurement unit sensors (IMUs) will be placed over your sternum, on your head, and on your left and right legs to measure the accelerations and velocities of those body segments. You will be videotaped to record your movements.

After completing the consent form, your balance will be measured. Once outfitted with reflective markers and the inertial measurement unit sensors (IMUs), you will be instructed to remove your shoes, stand on the force plate with your feet together, and look straight ahead. You will then be instructed to close your eyes and remain as still as possible for 2 minutes, at which time your balance will be recorded with the force plate while you count backwards by sevens. Following the balance measurement, you will jump as high as you can three separate times.

Next, you will be introduced to the walking and turning task. You will begin at the starting line of the pylon course. You will be asked to walk around each pylon and return to the starting line, repeating the pattern continuously while barefoot. You will be given time to practice the motion through several laps. For data collection, you will be asked to continuously walk around the course 15 times. A researcher on hand will count each lap so you will not need to count the laps. The researcher will also record each individual lap time based on when you cross the starting line and record each time.

Next, you will walk through the course again, but while performing a counting task in addition to walking. While walking, you will be asked to serially subtract a set of numbers given by the researcher. Next, you will be asked to repeat the procedure while walking "as fast as possible without running or jogging".

The total time commitment of each testing sessions is approximately 35 minutes total, broken up as follows:
Instrumentation with markers and IMUs: 2 minutes
Balance: 2 minutes
Gait and turning course: ~7.5 minutes per condition = 30 minutes

III. Risks Involved in Participation

There are only minimal risks to the participants in this study. Although you may have recently suffered a concussion, no activity outside of normal everyday activities (walking, standing) will be required to participate. Standard concussion symptoms of headache and dizziness may occur during the fast walking in the testing sessions immediately following the injury. A Virginia Tech athletic trainer or medical staff will be present at each testing session. You are free to discontinue participation at any time. Results of this study will not affect any athletic scholarship you receive.

Neither Virginia Tech nor the research team have funds set aside for any medical costs associated with injury during this study. Therefore, should injury occur, you will be responsible for any medical costs associated with injury directly related to this study.

IV. Benefits from Participation

You are not promised any specific/direct benefits for your participation in this study. You may learn about changes in your own walking patterns when walking faster or while performing a mentally challenging task while walking.

Results from this study have the potential to redefine the standard concussion return-to-play (RTP) guidelines. The results of this study may also lead to the development of a wireless IMU system to detect or monitor the recovery from concussions.

V. Extent of Anonymity and Confidentiality

You will be assigned a unique individual code number. The code number will be used on all of your study documents and data files. The Principal Investigators (PIs), Dr. Lockhart or Dr. Brolinson, will maintain a code key list to link your personal information to the code number used on your data. The code key list will be kept locked in a filing cabinet in the PI's office and will not be accessible to anyone who is not a project staff member. Coded data will be stored on a computer with password-protected access, and hard copies of data will be kept in a locked filing cabinet in the lab or in the PI's office. At the conclusion of the study, the data will be analyzed and will be submitted for publication in scientific journals. You will not be identified in the publications, and your anonymity and confidentiality will be maintained. As required by federal law and Virginia Tech IRB Policy, study records will be maintained for 3 years after the conclusion of the study, after which time they will be destroyed.

VI. Compensation

You will not be compensated.

VII. Freedom to Withdraw

You are free to withdraw from the study at any time and for any reason. Should the researchers determine that you should be removed from the study, you will be thanked.

VIII. Subject Responsibilities

You agree to participate in this study and confirm that you have no restrictions to your participation. You agree to assert an honest effort for each task asked of you.

IX. IRB Review of Research

The Virginia Tech Institutional Review Board (IRB) for projects involving human subjects, has reviewed this proposed study, and has determined that it is in compliance with federal laws and Virginia Tech policies governing the protection of human subjects in research. However, you should recognize that the review does not constitute an endorsement of the research, and that it is up to you to determine whether you are willing to participate in the study after having been informed of the risks, benefits, and procedures involved in this study. IRB# 14-677.

X. Authorization to Use or (Release) Health Information that Identifies You for a Research Study

If you sign this document, you give permission to Virginia Tech Sports Medicine trainers and Virginia Tech Athletics team physicians at Virginia Tech Sports Medicine, Virginia Tech Athletics, and the Edward Via College of Osteopathic Medicine to use or disclose (release) your health information that identifies you for the research study described in this consent form.

The health information that may be used or disclosed (released) for this research includes:
Portions of your electronic medical record pertaining to the effects of concussions on gait and balance, which includes, but is not limited to, past musculoskeletal injuries (e.g. prior lower extremity injury), orthopedic conditions, medical conditions (e.g. inner ear infections, cardiac conditions), past concussion history, new concussions, height, weight, and age.

The health information listed above may be used by and/or disclosed (released) to:

- Researchers listed in the approved IRB submission or in the consent form for this project.

Virginia Tech Athletics and Virginia Tech Sports Medicine is required by law to protect your health information. By signing this document, you authorize Virginia Tech Athletics and Virginia Tech Sports Medicine to use and/or disclose (release) your health information for this research. Those persons who receive your health information may not be required by Federal privacy laws (such as the Privacy Rule) to protect it and may share your information with others without your permission, if permitted by laws governing them.

Please note that:

- Virginia Tech Athletics and Virginia Tech Sports Medicine may not withhold or refuse treating you on whether you sign this Authorization.

- You may change your mind and revoke (take back) this Authorization at any time, except to the extent that Virginia Tech Athletics and Virginia Tech Sports Medicine has already acted based on this Authorization. To revoke this Authorization, you must write to: Virginia Tech Athletics and Sports Medicine, c/o Brett Griesemer, bgriese@vt.edu.

This Authorization expires at the conclusion of the attached research study.

XI. Subject / Participant's Permission

I have read the Consent Form and conditions of this project and have discussed it with the research staff or PI. I have had all my questions answered to my satisfaction. I hereby acknowledge the above and give my voluntary consent to participate in this study:

_____
Printed Name

_____ Date_____
Signature

Subject's Project Identification Code: _____ Sport_____

Should you have any questions about this research or its conduct, research subjects' rights, and whom to contact in the event of a research-related injury to the subject, you may contact:

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APPENDIX B: ONE-YEAR FOLLOW UP ADDENDUM

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants in Research Projects Involving Human Subjects

Title of Project: The Effects of Mild Traumatic Brain Injury on Turning Biomechanics and Non-Linear Gait Measures in Collegiate Athletes (One Year Follow Up)

Investigators: Per Gunnar Brolinson, D.O., Principal Investigator
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Department of Mechanical Engineering, Virginia Tech

Addendum

A one year follow-up session will occur approximately one year after the concussion or initial testing session. The time requirements, study procedures, risks, benefits, anonymity and confidentiality, compensation, freedom to withdraw, responsibilities and authorization will remain unchanged from the initial testing sequence as described in the attached original consent form.

Subject / Participant's Permission

I have read the Consent Form and conditions of this project and have discussed it with the research staff or PI. I have had all my questions answered to my satisfaction. I hereby acknowledge the above and give my voluntary consent to participate in this study's one year follow-up session:

_____
Printed Name

_____ Date_____
Signature

Subject's Project Identification Code: _____ Sport_____

Should you have any questions about this research or its conduct, research subjects' rights, and whom to contact in the event of a research-related injury to the subject, you may contact:

Peter Fino PhD Candidate
248-622-3637 (cell) fino@vt.edu (email)
Department of Mechanical Engineering,
Virginia Tech
Blacksburg, VA 24060

Maury Nussbaum Principal Investigator
540-231-6053 (office) nussbaum@vt.edu (email)
Grado Department of Industrial and Systems Engineering,
Virginia Tech
Blacksburg, VA 24060

Per Gunnar Brolinson Principal Investigator
540-231-4981 (office) pbrolinson@vcom.vt.edu (email)
Edward Via College of Osteopathic Medicine
Blacksburg, VA 24060

Brett Griesemer Assistant Athletic Trainer
540-231-2526 (office) bgriese@vt.edu (email)
Virginia Tech Sports Medicine,
Virginia Tech
Blacksburg, VA 24060

David M. Moore 540-231-4991 (office) moored@vt.edu (e-mail)
Chair, Virginia Tech Institutional Review
Board for the Protection of Human Subjects
Office of Research Compliance
300 Turner St. NW, Suite 4120
Blacksburg, VA 24060

APPENDIX C: CONTROL BASELINE CONSENT FORM

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants in Research Projects Involving Human Subjects

Title of Project: The Effects of Mild Traumatic Brain Injury on Turning Biomechanics and Non-Linear Gait Measures in Collegiate Athletes (Baseline Testing)

Investigators: Per Gunnar Brolinson, D.O., Principal Investigator
Edward Via College of Osteopathic Medicine

Maury Nussbaum, Ph.D.,
Grado Department of Industrial and Systems Engineering, Virginia Tech

Peter Fino, Graduate Student
Department of Mechanical Engineering, Virginia Tech

I. Purpose of this Research/Project

Concussions have been shown to affect how people walk. However, some subtle differences between healthy and concussed people may not be visible with traditional tools. More advanced tools such as non-straight gait analysis and non-linear stability analysis may be able to identify concussed populations and examine the recovery timeline. Because athletes exert themselves both physically and mentally while playing, the changes reported during walking while performing a mental task such as counting may indicate the athlete is not fully recovered. If the athlete is not fully recovered from the concussions, he/she may be at risk of both injury and poor performance during competitions. To address this concern, this study seeks to figure out the recovery timeline of various balance and gait characteristics.

B. Time Requirements

This first component of the study will consist of one testing session lasting approximately 35 minutes.

C. Study Procedures

A marked course will be arranged using a series of pylons for you to walk through. A series of 1.5 m (or taller) pylons will be set up in a slalom pattern across the ground. The pylons will be set up for you to make approximately 90° turns whenever you reach a pylon. The straight section will be 18 m long. This course will be arranged inside the Jamerson Athletic Center. Reflective tape or markers will be attached on your suprasternale, left and right acromioclavicular joints, sacrum, head, left and right ankles, and left and right heels. Inertial measurement unit sensors (IMUs) will be placed over your sternum, on your head, and on your left and right legs to measure the accelerations and velocities of those body segments. You will be videotaped to record your

movements.

After completing the consent form, your balance will be measured. Once outfitted with reflective markers and the inertial measurement unit sensors, you will be instructed to remove your shoes, stand on the force plate with your feet together, and look straight ahead. You will then be instructed to close your eyes and remain as still as possible for 2 minutes, at which time your balance will be recorded with the force plate while you count backwards by sevens. Following the balance measurement, you will jump as high as you can three separate times.

Next, you will be introduced to the walking and turning task. You will begin at the starting line of the pylon course. You will be asked to walk around each pylon and return to the starting line, repeating the pattern continuously while barefoot. You will be given time to practice the motion through several laps. For data collection, you will be asked to continuously walk around the course 15 times. A researcher on hand will count each lap so you will not need to count the laps. The researcher will also record each individual lap time based on when you cross the starting line and record each time.

Next, you will walk through the course again, but while performing a counting task in addition to walking. While walking, you will be asked to serially subtract a set of numbers given by the researcher. Next, you will be asked to repeat the procedure while walking "as fast as possible without running or jogging".

The total time commitment of each testing sessions is approximately 35 minutes total, broken up as follows:
Instrumentation with markers and IMUs: 2 minutes
Balance: 2 minutes
Gait and turning course: ~7.5 minutes per condition = 30 minutes

III. Risks Involved in Participation

There are only minimal risks to the participants in this study. You are free to discontinue participation at any time. Results of this study will not affect any athletic scholarship you receive.

Neither Virginia Tech nor the research team have funds set aside for any medical costs associated with injury during this study. Therefore, should injury occur, you will be responsible for any medical costs associated with injury directly related to this study.

IV. Benefits from Participation

You are not promised any specific/direct benefits for your participation in this study. You may learn about changes in your own walking patterns when walking faster or while performing a mentally challenging task while walking.

Results from this study have the potential to redefine the standard concussion return-to-play (RTP) guidelines. The results of this study may also lead to the development of a wireless IMU system to detect or monitor the recovery from concussions.

V. Extent of Anonymity and Confidentiality

You will be assigned a unique individual code number. The code number will be used on all of your study documents and data files. The Principal Investigators (PIs), Dr. Lockhart or Dr. Brolinson, will maintain a code key list to link your personal information to the code number used on your data. The code key list will be kept locked in a filing cabinet in the PI's office and will not be accessible to anyone who is not a project staff member. Coded data will be stored on a computer with password-protected access, and hard copies of data will be kept in a locked filing cabinet in the lab or in the PI's office. At the conclusion of the study, the data will be analyzed and will be submitted for publication in scientific journals. You will not be identified in the publications, and your anonymity and confidentiality will be maintained. As required by federal law and Virginia Tech IRB Policy, study records will be maintained for 3 years after the conclusion of the study, after which time they will be destroyed.

VI. Compensation

You will not be compensated.

VII. Freedom to Withdraw

You are free to withdraw from the study at any time and for any reason. Should the researchers determine that you should be removed from the study, you will be thanked.

VIII. Subject Responsibilities

You agree to participate in this study and confirm that you have no restrictions to your participation. You agree to assert an honest effort for each task asked of you.

IX. IRB Review of Research

The Virginia Tech Institutional Review Board (IRB) for projects involving human subjects, has reviewed this proposed study, and has determined that it is in compliance with federal laws and Virginia Tech policies governing the protection of human subjects in research. However, you should recognize that the review does not constitute an endorsement of the research, and that it is up to you to determine whether you are willing to participate in the study after having been informed of the risks, benefits, and procedures involved in this study. IRB# 14-677.

X. Future Participation

This study also contains a portion of testing after athletes have sustained a concussion. Teammates of these concussed athletes are needed for this portion of the testing for a healthy comparison. If you would like to be contacted to be a healthy comparison for your teammates should they suffer a concussion, please check the box below.

☐ Yes, I would like to be contacted to serve as a healthy comparison for my teammates
Phone Number_____ Email_____

XI. Authorization to Use or (Release) Health Information that Identifies You for a Research Study

If you sign this document, you give permission to Virginia Tech Sports Medicine trainers and Virginia Tech Athletics team physicians at Virginia Tech Sports Medicine, Virginia Tech Athletics, and the Edward Via College of Osteopathic Medicine to use or disclose (release) your health information that identifies you for the research study described in this consent form.

The health information that may be used or disclosed (released) for this research includes:
Portions of your electronic medical record pertaining to the effects of concussions on gait and balance, which includes, but is not limited to, past musculoskeletal injuries (e.g. prior lower extremity injury), orthopedic conditions, medical conditions (e.g. inner ear infections, cardiac conditions), past concussion history, new concussions, height, weight, and age.

The health information listed above may be used by and/or disclosed (released) to:

- Researchers listed in the approved IRB submission or in the consent form for this project.

Virginia Tech Athletics and Virginia Tech Sports Medicine is required by law to protect your health information. By signing this document, you authorize Virginia Tech Athletics and Virginia Tech Sports Medicine to use and/or disclose (release) your health information for this research. Those persons who receive your health information may not be required by Federal privacy laws (such as the Privacy Rule) to protect it and may share your information with others without your permission, if permitted by laws governing them.

Please note that:

- Virginia Tech Athletics and Virginia Tech Sports Medicine may not withhold or refuse treating you on whether you sign this Authorization.

- You may change your mind and revoke (take back) this Authorization at any time, except to the extent that Virginia Tech Athletics and Virginia Tech Sports Medicine has already acted based on this Authorization. To revoke this Authorization, you must write to: Virginia Tech Athletics and Sports Medicine, c/o Brett Griesemer, bgriese@vt.edu.

This Authorization expires at the conclusion of the attached research study.

XII. Subject / Participant's Permission

I have read the Consent Form and conditions of this project and have discussed it with the research staff or PI. I have had all my questions answered to my satisfaction. I hereby acknowledge the above and give my voluntary consent to participate in this study:

_____
Printed Name

_____ Date_____
Signature

Subject's Project Identification Code: _____ Sport_____

Should you have any questions about this research or its conduct, research subjects' rights, and whom to contact in the event of a research-related injury to the subject, you may contact:

Peter Fino PhD Candidate
248-622-3637 (cell) fino@vt.edu (email)
Department of Mechanical Engineering,
Virginia Tech
Blacksburg, VA 24060

Maury Nussbaum Principal Investigator
540-231-6053 (office) nussbaum@vt.edu (email)
Grado Department of Industrial and Systems Engineering,
Virginia Tech
Blacksburg, VA 24060

Per Gunnar Brolinson Principal Investigator
540-231-4981 (office) pbrolinson@vcom.vt.edu (email)
Edward Via College of Osteopathic Medicine
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Brett Griesemer Assistant Athletic Trainer
540-231-2526 (office) bgriese@vt.edu (email)
Virginia Tech Sports Medicine,
Virginia Tech
Blacksburg, VA 24060

David M. Moore 540-231-4991 (office) moored@vt.edu (e-mail)
Chair, Virginia Tech Institutional Review
Board for the Protection of Human Subjects
Office of Research Compliance
300 Turner St. NW, Suite 4120
Blacksburg, VA 24060

APPENDIX D: NON-ATHLETE POSTURAL STABILITY CONSENT FORM

VIRGINIA POLYTECHNIC INSTITUTE AND STATE UNIVERSITY

Informed Consent for Participants in Research Projects Involving Human Subjects

Title of Project: Influences of Behaviors on Postural Control

Investigators: Maury Nussbaum, Ph.D., Principal Investigator
Department of Industrial and Systems Engineering, Virginia Tech

Peter Fino, Graduate Student
Department of Mechanical Engineering, Virginia Tech

I. Purpose of this Research/Project

The purpose of this study is to determine how various balance measurements are influenced by a person's actions.

II. Procedures

A. Participant Selection

Healthy young (18-30 years old) participants will be tested. There are no restrictions on gender or ethnicity on this study. For the purposes of this study, however, we cannot include people with a history of balance problems, ankle and knee injuries within the past year, or an inability to maintain a standing posture for 2 minute.

B. Time Requirements

The study will last for approximately 15-20 minutes.

C. Study Procedures

You will be asked to complete a brief survey regarding your physical activity in the past two weeks. You will then be asked to remove your shoes, socks, and any electronics from your person. You will then be asked to stand on a force plate with their feet together, looking straight ahead and instructed to close your eyes and stand still for two minutes. During that two-minute period, your balance will be measured. Following this, you will be asked to repeat the process three more times while either silently counting down by sevens from a random number, contracting your muscles in your legs, arms, and torso as hard as you can, or trying to randomly control your body sway pattern. Multiple practice attempts will be giving for each of these three conditions, and the order will be random. You will be given time to rest between conditions.

III. Risks Involved in Participation

There are no more than minimal risks for participating in this study, consistent with a normal day of activity. You may feel fatigued from the muscle contraction depending on your typical physical activity level.

IV. Benefits from Participation

You are not promised any specific/direct benefits for your participation in this study. The results of this study may yield benefits in understanding how effort and participant actions play a role in balance.

V. Extent of Anonymity and Confidentiality

You will be signed a unique individual code number. The code number will be used on all of your study documents and data files. The lead graduate student, Peter Fino, will maintain a code key list to link your personal information to the code number used on your data. The code key list will be kept locked in a filing cabinet in the graduate student's locked office and will not be accessible to anyone who is not a project staff member. Coded data will be stored on a computer with password-protected access, and hard copies of data will be kept in a locked filing cabinet in the lab or in the graduate student's office. At the conclusion of the study, the data will be analyzed and will be published in scientific journals. You will not be identified in the publications, and your anonymity and confidentiality will be maintained. As required by federal law and Virginia Tech IRB Policy, study records will be maintained for 3 years after the conclusion of the study, after which time they will be destroyed.

VI. Compensation

You will be compensated $10 for your time.

VII. Freedom to Withdraw

You are free to withdraw from the study at any time and for any reason. There may also be cases where the researched determine that you need not complete the study.

VIII. Subject Responsibilities

You are expected to provide accurate information on your survey form. You agree to participate in this study and confirm that you have no restrictions to your participation.

IX. IRB Review of Research

The Virginia Tech Institutional Review Board (IRB) for projects involving human subjects, has reviewed this proposed study, and has determined that it is in compliance with federal laws and Virginia Tech policies governing the protection of human subjects in research. However, you

should recognize that the review does not constitute an endorsement of the research, and that it is up to you to determine whether you are willing to participate in the study after having been informed of the risks, benefits, and procedures involved in this study.
IRB# 15-795

X. Subject / Participant's Permission

I have read the Consent Form and conditions of this project and have discussed it with the research staff or PI. I have had all my questions answered to my satisfaction. I hereby acknowledge the above and give my voluntary consent to participate in this study:

_____ Date_____
Subject's Signature
Subject's Project Identification Code: _____

Should you have any questions about this research or its conduct, research subjects' rights, and whom to contact in the event of a research-related injury to the subject, you may contact:

Peter Fino — PhD Candidate
248-622-3637 (cell) fino@vt.edu (email)
Department of Mechanical Engineering,
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Maury Nussbaum — Principal Investigator
540-231-6053 (office) nussbaum@vt.edu (email)
Grado Department of Industrial and Systems Engineering,
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David M. Moore — 540-231-4991 (office) moored@vt.edu (e-mail)
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